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Accelerometery and Force Plate Systems for Measurement of Orthostatic Postural Stability in Older Adults

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Posters, Presentations and Publications related to this thesis

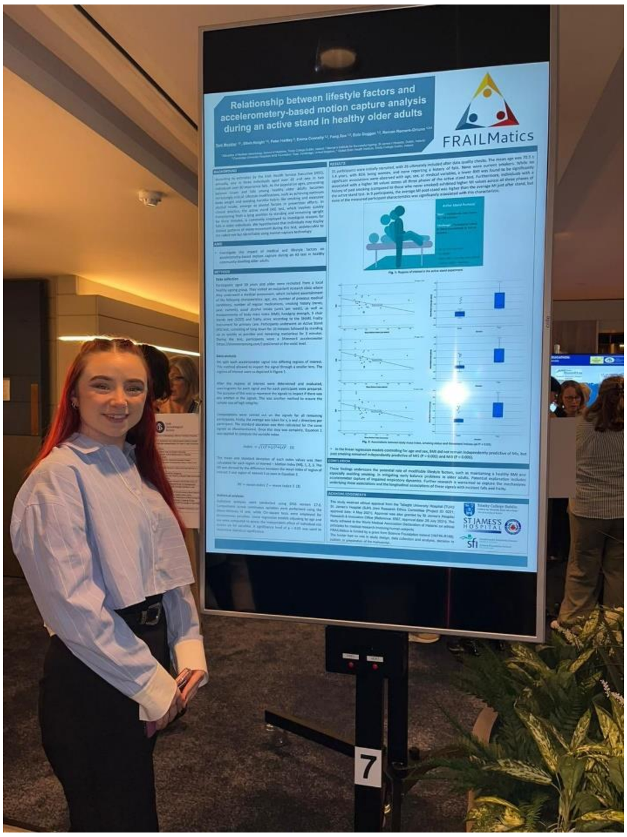
- Annual Meeting of the Clinical Movement Analysis Society UK and Ireland 2024 (Leeds, April 18-19th), https://cmasuki.org/annual-meeting-of-the-clinical-movement-analysis-society-2024-leeds/

    - First presentation of a poster titled: "*Relationship between lifestyle factors and accelerometery-based motion capture analysis during an Active Stand in healthy older adults*" (unpublished).

- Discipline of Medical Gerontology Postgraduate Research Day 2024 (Stanley Quek Hall, Trinity Biomedical Sciences Institute, Trinity College Dublin, April 25th), https://www.tcd.ie/news_events/events/event/discipline-of-medical-gerontology-postgraduate-research-day-2024.php

    - Presented a platform presentation titled: "*Relationship between modifiable lifestyle factors and accelerometery-based motion capture analysis during an Active Stand in older adults*".

- Irish Gerontological Society Annual Scientific Meeting 2024 (The Sheraton Hotel, Athlone, October 3-5th), https://www.irishgerontology.com/events/igs-71st-annual-and-scientific-meeting
    - Second presentation of the poster "*Relationship between lifestyle factors and accelerometery based motion capture analysis during an Active Stand in healthy older adults*", leading to abstract publication in *Age and Ageing*: https://doi.org/10.1093/ageing/afae178.350



Abstract

In this research-based Master's thesis, I investigated the potential utility of implementing accelerometry and force balance plate (FBP) systems for assessing orthostatic postural stability in older adults undergoing the Active Stand (AS) test, which is a standard clinical tool for evaluating autonomic cardiovascular function. Departing from its conventional use, I applied the AS test from a novel perspective, focusing on orthostatic motion as an indicator of postural control in older adults. The study was conducted in two distinct phases, each with a specific methodological emphasis, but unified by the overarching aim of improving the detection of postural instability, which is a key contributor to falls in the ageing population.

In Phase One, I conducted a secondary analysis of existing accelerometry data collected from 26 older volunteers attending a specialist Falls and Syncope Clinic. An anteriorly positioned accelerometer at the waist was used to capture postural motion during AS tests. The analysis indicated that individuals with a history of smoking demonstrated significantly elevated postural motion during both the initial supine rest phase and the later post-stand recovery phase (ROIs 1 and 3), even after adjusting for age. These results suggest a potential link between smoking history and altered neuromuscular or postural control, possibly mediated by impaired respiratory or compensatory mechanisms. This may be relevant given the anterior accelerometer placement, which may be more sensitive to thoracoabdominal movement.

In Phase Two, I designed and conducted another small pilot study, in which I collected original data from 10 older patients using dual force balance plates during AS testing in a clinical gait laboratory. I focused on two key postural sway metrics: sway area and 95% prediction ellipse area, measured across early (ROI 2: 20–50s) and late (ROI 3: 60–90s) upright intervals. Although smoking-related patterns identified in Phase One were not replicated using FBP metrics (potentially due to differing biomechanical sensitivities between modalities), I did observe a notable increase in postural sway among participants with orthostatic hypotension at 40 seconds post-stand (OH40), particularly

during ROI 3. Statistical analysis revealed a significant interaction between OH40 status and ROI in Plate 2 for ellipse area, suggesting that this metric may be sensitive to postural instability during more prolonged upright posture.

Studies in this MSc are exploratory and inherently limited by their small sample size and pilot design. Nonetheless, it demonstrates the feasibility of implementing wearable sensors and force plates to investigate postural stability during the AS test in a falls clinic. These modalities capture complementary biomechanical phenomena (upper-body motion via accelerometry and lower-limb dynamics via force plates), which may enrich our understanding of compensatory mechanisms during orthostasis.

Through this project, I significantly developed my skills in data processing, statistical analysis, and clinical research. I became proficient in tools such as SPSS and MATLAB, and gained valuable hands-on experience working with patients in a real-world clinical setting. This Master's programme allowed me to grow both personally and professionally, providing an opportunity to contribute preliminary but interesting new data to a complex and clinically relevant research question. The results presented here offer a foundation for future studies with larger samples and more diverse participant groups, which will be essential for validating and extending these preliminary findings.

Abbreviations

Anterior-Posterior	AP
Body Mass Index (kg/m^2)	BMI
Centre Of Mass	COM
Centre of Pressure	COP
Euclidean Distance of all the data points (x, y and z) collected by the accelerometer in the first region of interest	Index_ROI1
Euclidean Distance of all the data points (x, y and z) collected by the accelerometer in the second region of interest	Index_ROI2
Euclidean Distance of all the data points (x, y and z) collected by the accelerometer in the third region of interest	Index_ROI3
Falls And Syncope Unit	FASU
Force Balance plates	FBP
Health Service Executive	HSE
Medio-lateral	ML
Movement Index	MI
Orthostatic Hypotension	OH
Patient Information Leaflet	PIL
Region Of Interest 1	ROI1
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Region Of Interest 3	ROI3
The Irish Longitudinal Study on Ageing	TILDA

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1. Introduction

Ireland's ageing population is expanding rapidly, with the number of people aged 65 and older projected to nearly double, reaching approximately 1.6 million in the Republic of Ireland by 2051, according to the Institute of Public Health (IPH, 2020). This demographic shift presents significant challenges, particularly in preventing balance problems and falls among older adults. The Health Service Executive (HSE) estimated that one in three people over the age of 65, and one in two people over the age of 80, experience a fall each year (HSE, 2008). The Irish Longitudinal Study on Ageing (TILDA) has also found that falls are common even among adults over the age of 50, occurring in 29% of men and 34% of women aged 50–64 years; furthermore, falls in all adults can result in reduced mobility, decreased social participation, and an increased risk of mental health issues such as isolation and depression (McGarrigle et al., 2017).

Many falls in older adults occur after sudden postural changes, such as standing up quickly (Komisar et al., 2023). This happens because the body needs to compensate rapidly to achieve postural balance. This process is complex and involves multiple physiological systems, including the musculoskeletal, neurological, and cardiovascular systems (Kenny et al., 2017). Hence, the measurement of postural stability during a standard standing test is of medical interest, as the act of standing is itself a stressor to the body. As such, it challenges compensatory mechanisms and can reveal subtle motion abnormalities in older adults, especially those who live with frailty. Indeed, frailty reflects a reduced ability to compensate for stressors due to dysregulation across multiple physiological systems (Clegg et al., 2013, Dent et al., 2019), and it has been previously shown that in those who live with frailty, challenging tasks (even cognitive distractions) can lead to degradation of complex balance dynamics and lead to an increased risk of falls (Kang et al., 2009). It is not surprising, therefore, that the prevalence of frailty is higher among older patients attending falls clinics compared to the general population (O'Donoghue et al., 2021).

To address the challenges of measuring postural stability for falls risk assessment in older populations, the field of biomedical engineering has advanced rapidly, introducing innovative motion capture technologies capable of providing detailed analyses of movement and balance under various experimental conditions. This includes the use of wearable sensors such as accelerometers (Chen et al., 2022) and force balance plates (FBP) (Piirtola and Era, 2006). Both systems have advantages and disadvantages (Andres and Anderson, 1980).

Accelerometers are portable, relatively inexpensive, and widely accessible, making them suitable for use in various settings and experimental conditions (Marchetti et al., 2013). Accelerometery is a technique that measures the acceleration of body segments to assess postural stability. These small, wearable devices detect changes in velocity and direction, capturing body-segment motion through linear acceleration in three-dimensional space (x, y, and z). This method can be particularly useful in clinical settings for evaluating fall risk and postural control in older adults (Yang and Hsu, 2010). To assess postural instability, the device is typically placed at the participant's centre of mass (COM), traditionally located at the lower back. However, this placement may not be feasible if the participant is required to lie on their back prior to standing quickly. In such scenarios, positioning the accelerometer at the front may introduce artefacts unrelated to postural balance, such as movements caused by respiration. Furthermore, accelerometers lack the capability of FBPs to measure ground reaction forces and provide detailed pressure distributions, which limits their precision in more comprehensive postural stability assessments.

FBPs are considered the gold standard for assessing postural stability by capturing high-resolution ground reaction force data and center of pressure (COP) trajectories. They are used in gait laboratories and some clinical settings to support the assessment and treatment of patients with gait and balance-related issues. FBPs operate based on Newton's Third Law of Motion: when one body exerts a force on another, the second body exerts an equal and opposite force on the first. These devices measure ground reaction forces and the COP through foot platforms embedded in the floor, using piezoelectric sensors and strain gauges to capture forces along the three main axes: x, y, and z (Chen et al., 2021). Despite their higher precision, FBPs have notable limitations. They typically assume the human body functions as a single rigid mass, which oversimplifies the complex, multi-segmental dynamics of human movement. This assumption limits their ability to accurately capture joint-specific movements or distinguish the contributions of individual body segments, such as the legs, hips, and torso, to postural sway. Additionally, force plates are relatively expensive and require dedicated laboratory setups, making them less accessible than wearable alternatives like accelerometers.

More advanced three-dimensional motion capture systems, which use optical tracking technologies, can offer richer insights by capturing detailed kinematic data. These systems are often integrated with force plates and inertial measurement units to provide a comprehensive assessment of posture and movement. However, they also present practical challenges, including high cost, complex setup, and the need for considerable physical space (Longo et al., 2022).

1.1 Aims/Objectives

This MSc (Research) thesis was undertaken within the broader context of the FRAILMatics programme (18/FRL/6188), a Research Ireland-funded programme aimed at improving the early detection of frailty through the application of digital and sensor-based technologies. My thesis specifically investigated the utility of implementing accelerometry and FBP systems for measuring orthostatic postural stability during the Active Stand (AS) test in older adults. While the AS test is typically used to evaluate cardiovascular autonomic function, my research explored its novel application in assessing orthostatic postural motion as a potential surrogate for balance and fall risk.

Falls remain a major clinical concern in ageing populations, frequently leading to injury, hospitalisation, reduced quality of life, and an increased risk of premature mortality. Identifying subtle impairments in postural control, particularly during transitions such as standing, may offer opportunities for earlier detection of fall risk and targeted intervention. My thesis aimed to contribute feasibility and pilot data on the use of two complementary sensor modalities: accelerometry and force balance plates (FBPs), for assessing orthostatic postural stability in a clinical setting. The research was structured in two sequential phases:

- **Phase One** involved secondary analysis of accelerometry data previously collected by FRAILMatics during AS tests in the Falls and Syncope Unit (FASU) at St. James's Hospital (https://www.stjames.ie/services/medicinefortheelderlymedel/fallssyncopeunit/). In this phase I focused on extracting motion-based indices and exploring their clinical associations.

- **Phase Two** comprised a pilot study in which I collected original data from older adults attending FASU. In this phase, participants underwent AS testing in a gait laboratory setting using gold-standard FBPs, from which I derived standard metrics to explore associations with clinical characteristics.

My research addressed the following key questions:

- Accelerometry analysis (Phase One): *Are there clinically relevant associations between accelerometer-derived motion indices and patient characteristics, and how might anterior sensor placement affect these measurements?*

- Force plate analysis (Phase Two): *Can FBPs provide additional or complementary insights into postural stability during AS testing, and do they capture changes that align with clinical markers?*

This thesis is organised as follows:

Chapter 2 critically reviews the literature on accelerometers and FBPs as tools for assessing postural control in adults. It situates these technologies within the broader clinical context of falls risk assessment in older adults and frames their relevance in relation to the aims and methodology of this thesis.

Chapter 3 outlines the methods and findings from Phase One, focusing on accelerometry signal processing, metric development, and clinical associations.

Chapter 4 presents the methodology and results from Phase Two, including the design and feasibility of dual force plate analysis and comparison with clinical characteristics.

Chapter 5 synthesises key findings, discusses limitations and pilot nature of the work, and outlines implications for future research and clinical practice.

2. Literature review

This literature review examines key academic sources related to motion capture, synthesising current knowledge within this evolving and interdisciplinary field. Motion capture technologies have advanced considerably and now extend beyond traditional disciplinary boundaries, with growing applications in clinical research and medical practice. Particular emphasis is placed on methodologies that directly informed or were incorporated into the data analysis for this thesis.

A critical appraisal of the literature is then presented, specifically in the context of the aims and scope of this Master's research.

2.1 The use of accelerometers to quantify postural sway

Accelerometers have been shown to be suitable alternatives to the more expensive and stationary force plates (Saunders et al., 2015). This advancement has provided researchers with accessible, portable, and efficient tools for assessing balance and postural control in various contexts.

Building on this technological shift, Alkathiry et al. conducted a detailed investigation into the magnitude of postural sway in adolescents with concussion, focusing on different balance tasks (Alkathiry et al., 2018). Their study aimed to not only quantify the postural sway induced by these tasks but also explore the associations between postural sway and the presence of concussion symptoms, which are often challenging to assess using traditional subjective measures. To gather data, the researchers employed an accelerometer (model ADXL213AE; Analog Devices, Inc, Norwood, MA), which was strategically placed on the participants' lower backs at the level of the iliac crest to ensure consistent and accurate measurement of body sway. This placement is significant because it captures the COM movement, offering a reliable indicator of overall postural stability.

The accelerometer was configured with a sampling rate of 50 Hz, specifically targeting acceleration in the anterior-posterior (AP) direction, a common axis of sway in balance control studies. This choice aligns with previous research emphasising the importance of AP sway in detecting balance impairments, especially in individuals recovering from concussion. Participants were instructed to perform a series of balance tasks, including standing with their feet together on a firm surface (FT-firm), standing in tandem on a firm surface (tandem-firm), and standing with feet together on a foam surface, both with their eyes open and eyes closed. These tasks were designed to progressively challenge the participants' balance by introducing more unstable conditions, with the foam surface and closed eyes scenarios being particularly demanding. The use of the Romberg method in this context is notable, as it has been widely recognised as a standard clinical procedure in balance testing, particularly for identifying balance deficits that may not be apparent in simpler conditions. The Romberg method is a clinical test used to assess postural stability, in which a person stands with feet together and eyes closed to evaluate their ability to maintain balance without visual input, thereby identifying potential sensory or neurological deficits (Khasnis and Gokula, 2003).

To ensure the accuracy of the data, the authors applied a low-pass Butterworth filter (fourth-order zero-phase with a cutoff frequency of 2 Hz) to clean the recorded acceleration signals. This filtering process removed high-frequency noise that could distort the results, ensuring that the data reflected true postural sway rather than artifacts of measurement. The normalized path length (NPL) of the sway was then calculated using equation 1:

$$NPL = \frac{1}{t}\sum(P_j + 1 - P_j) \qquad (1)$$

where t is time duration, N is the number of time samples, and p_j is the acceleration data at time sample j in the AP direction.

This metric is particularly relevant, as it provides a quantifiable measure of the total distance the body moves while maintaining balance, normalised for the duration of the trial. The use of NPL as an outcome measure offers a more objective approach to assessing balance impairments, as it accounts for both the frequency and magnitude of sway.

The study's findings suggest that accelerometers, when used in combination with challenging balance tasks, may offer a reliable, objective method for quantifying balance impairments in adolescents with concussions. This represented a significant advancement, as traditional clinical balance assessments often rely on subjective observations or basic static postures that may not fully capture the subtle impairments caused by concussions.

Similarly, Kamen et al. conducted a study analysing an accelerometery-based system to assess postural sway (Kamen et al., 1998). They focused their study on comparing the postural sway measured between younger and older adults. They used the Romberg method as above, with 10 younger (range 18–32 years) and 10 older (range 69–86 years) individuals. They completed the test under the three conditions as per Romberg's guidelines. For each trial, sway amplitude was computed using the root mean square (RMS) area of the acceleration time series. The frequency of sway was displayed by computing the median power frequency obtained from the power frequency spectrum. These participants attended the lab over three days. They tested the reliability of the of amplitude and frequency measures obtained from the accelerometer by using an intraclass correlation analysis of variance model. After conducting these tests and analysing the data, Kamen et al. reported obtaining preliminary evidence indicating that this accelerometery system can effectively differentiate between healthy older adults and older individuals who are frequent fallers.

On the other hand, Alberts et al. took on a different approach to the measuring of postural sway (Alberts et al., 2015). Instead of using a standalone accelerometer device, they opted to use an iPad with an embedded accelerometer. They conducted an experiment to investigate an alternative method to measure postural stability and compared against the gold standard FBP (Neurocom balance master). Their alternative method was the use of the accelerometer that is embedded in an iPad 2 (Apple Inc, Cupertino, CA). The iPad was attached to ensure that its sensors were as close as possible to the participants' COM. They also utilised the Romberg method during their data acquisition. These researchers made use of the derived parameters such as AP centre-of-gravity (COG) sway, and this value was then used to calculate an Equilibrium score as shown in equation 2:

$$Equilibrium\ Score = \frac{12.5° - [Max\ (Sway_{AP}(\emptyset)) - Min\ (Sway_{AP}(\emptyset))\ ]}{12.5°}\ x\ 100\% \qquad (2)$$

The scores from the Neurocom and the iPad 2 were then compared using the Bland-Altman assessment for agreement. Mean absolute percentage error and median difference between the Neurocom and iPad 2 measurements were used to evaluate how closely the real-time COG sway measured by the two systems tracked each other. They concluded that the iPad 2 hardware provided data of sufficient precision and accuracy to quantify postural stability.

Other researchers also took the smart device approach. For example, Dewan et al. performed an experiment drawing a comparison between the gold standard motion capture system (VICON Motion Systems Ltd) and a smartphone app (Dewan et al., 2019). An iPhone was secured to the participant's chest and captured their thoracic movement. They set their Vicon systems sampling rate to 100 Hz and their accelerometer to 10 Hz. The participants were asked to wear minimal clothing, and they had 39 reflective markers. The Romberg method was utilised to collect the data for the study. A custom MATLAB script was developed to process their data. They COM balance indices using equations 3 and 4:

$$COM\ sway\ index\ (COM\ SI)(m) = \sqrt{\frac{1}{N}\left(\sum_{i=1}^{N}((x_i - x_i)^2 + (y_i - y)^2)\right)} \quad (3)$$

$$COM\ mean\ displacement\ velocity\ (COM\ velocity)\left(\frac{m}{s}\right) = \frac{\frac{\sum_{i=1}^{N}((x_i - x_i)^2 + (y_i - y)^2)}{(t_i - t_{i-1})}}{N} \quad (4)$$

where N = number of frames, xi = location of COM in x-axis at frame i, x_m = mean COM x-coordinate value, y_i = location of COM in y-axis at frame i, y_m = mean COM y-coordinate value, and t_i = time at frame i.

Statistical analysis was performed using SPSS to investigate a linear relationship for their test conditions; a combination of the phone embedded accelerometer, the Vicon and the force plate. Their results displayed a linear increase in postural sway with increases in condition difficulty (different conditions were taken as per the guidelines of the Romberg method). The study found that the force plate was valid for the assessment of postural sway in progressively challenging double leg stances in the Romberg method and the phone application (accelerometer) was valid for the assessment of postural sway in a balance assessment protocol consisting of progression from double leg stance to single leg stance.

Following the footsteps of other researchers, Hsieh et al. conducted a study to gain insight into the use of smartphones to measure postural stability to estimate fall risk in older adults (Hsieh et al., 2019). They recruited 30 older adults – with a mean age of 66 to undergo seven balance tests as per Romberg guidelines. They were asked to step onto a force plate whilst holding a smartphone to their chest. The force plate had a sampling frequency of 1000 Hz and the smartphone's accelerometer had a sampling rate of 200 Hz. COP data from the force plate for each trial was exported and processed with a 4^{th} order, low pass Butterworth filter at a cutoff frequency of 10 Hz using a custom MATLAB script. The validity between measures derived from the force plate and measures derived from the smartphone was assessed with Spearman rank-order correlations for all balance conditions. Their participants were asked to complete a physiological profile assessment (PPA), which was used to analyse the ability of smartphone measures to distinguish individuals with high fall risk. From all this, Hsieh et al. concluded that the smartphone provides a valid measure of postural stability in older adults.

Table 1 shows a synopsis of the above research concerning the use of accelerometery to measure postural instability.

Table 1. Summary of accelerometery studies.

Author	Method	Objective	Sensor placement	Data Analysis method
(Alkathiry et al., 2018)	Romberg	To investigate the magnitude of postural sway induced by different balance tasks in adolescents with concussion and to examine the associations of postural sway with concussion symptoms.	lower back	4th order, low pass Butterworth filter – cut off 2 Hz (sampling @ 50 Hz)
(Kamen et al., 1998)	Romberg	To develop an inexpensive, efficient system for the clinical assessment of static and dynamic balance and postural sway using accelerometery-based measurements.	Lower back	The input signal from the transducer was typically bandwidth-filtered at 0.01–20 Hz (–3 dB). (sampling @ 50 Hz)
(Alberts et al., 2015)	Romberg	To determine whether accelerometer and gyro-scope data sampled from a consumer electronics device (iPad2) provide sufficient resolution of centre-of-gravity (COG) movements to accurately quantify postural stability in healthy young people.	Lower back	4th order, low pass Butterworth filter – cut off 1.25 Hz (sampling @ 100 Hz)

(Dewan et al., 2019)	Romberg	To investigate the concurrent validity of standing postural sway measurements by using the portable Biosway and SBMA systems with kinematic measurements of the whole-body Center of Mass (COM) derived from a gold-standard reference, a motion capture system.	Chest	4th order, two pass, zero lag Butterworth filter – cut off 6 Hz (Sampling @ 100 Hz)
(Hsieh et al., 2019)	Romberg	Smartphone technology can measure postural stability and discriminate fall risk in older adults	Chest	4th order, low pass Butterworth filter – cut off 10 Hz (Sampling @ 1000 Hz)

2.2 The use of the gold standard method - Force balance plate

Balance is the ability to control body coordination when moving or maintaining a loading posture. Balance involves coordinating the transfer of the COM and the COP. The COM represents the point where the body's mass is evenly distributed, while the COP reflects the point of application of the ground reaction force, measured through systems like FBPs. Maintaining balance requires continuous adjustments to keep the COP within the base of support relative to the COM. These adjustments involve a dynamic interplay of sensory, musculoskeletal, cardiovascular, and neurological systems to ensure stability during both static and dynamic postures. If the COP fails to compensate effectively for shifts in the COM, it can result in postural instability and an increased risk of falls, especially in older adults. Indeed, posture control in clinical populations is very important and involves maintaining and adjusting the body's alignment and balance while standing (Lafond et al., 2004). Furthermore, older adults with a history of falls are likely to have worse compensatory responses to minor postural stresses (Wolfson et al., 1986). As in the present master's investigation, previous studies have evaluated postural instability with COP movement (Merlo et al., 2012).

Lo et al. conducted a study to evaluate the reliability of postural sway measurements in relation to the effects of ageing, utilising a low-cost FBP (Lo et al., 2022). Their study included 40 participants (20 younger adults and 20 older adults). Everyone's height was taken and used to place a marker 2m from their eyeline. The marker is described as circular with a diameter of 10 cm. The researchers also asked their participants to step into the centre of their custom-built force plate. The sampling frequency was set to 512 Hz. The researchers made use of the single axis force sensor to measure the balance performance. Their analysis of the data consisted of the use of 30-second data (including the removal of 5 seconds at the start and at the end). Their protocol included standing on both legs, with eyes opened and closed three times for 40 seconds, along with one minute rest between each measurement session. During the examination, participants looked directly at the 2 m mark and stood on the marked spot on the force plate, with both arms naturally placed by their sides. They analysed their data in terms of average velocity (V), medio-lateral average velocity (V-ML), antero-posterior average velocity (V-AP), mean distance (MD), medio-lateral mean distance (MD-ML) and antero-posterior mean distance (MD-AP). They also used the following equations as part of their data analysis:

$$The\ mean\ COP_{AP} = \frac{1}{N}\sum_{n=1}^{N} AP_0\ n = 1{,}2{,}3 \ldots N \quad (5)$$

$$The\ mean\ COP_{ML} = \frac{1}{N}\sum_{n=1}^{N} ML_0\ n = 1{,}2{,}3 \ldots N \quad (6)$$

The AP and ML time series were referenced to the mean COP per the following equations:

$$AP(n) = \sum_{n=1}^{N} AP_0(n) - COP_{AP} \quad (7)$$

$$ML(n) = \sum_{n=1}^{N} ML_0(n) - COP_{ML} (8)$$

Resultant distance (RD) time series represented a distance between the mean COP and the COP in AP0 and ML0 time series as follows:

$$RD(n) = \sum_{n=1}^{N} [AP\ (n)^2 +\ ML\ (n)^2]^{\frac{1}{2}} \quad (9)$$

The mean distance was the mean of the RD time series and was calculated as:

$$The\ mean\ distance\ (mm) = \frac{1}{N}\sum_{n=1}^{N} RD(n) \quad (10)$$

Their statistical analyses consisted of Standard Error of the Mean (SEM) and Intraclass Correlation Coefficient (ICC). This also included a reliability evaluation of the results. Lo et al. concluded from their analyses that there was excellent reliability across all their tests for their hypotheses. They claimed that their study "provided suggestions for the selection of reliable COP-related parameters for a static standing task to measure the ability to balance for different age groups" (Lo et al., 2022).

On the other hand, Seimetz et al. carried out a study to compare two methods of measuring postural sway: force plates versus accelerometers (Seimetz et al., 2012). Their sample consisted of 6 subjects (3 male and 3 female) and asked their participants to stand barefoot for the study. The participants were wearing an accelerometer as well. Their method consisted of four trials: visually impaired (with the participants with their eyes closed) and proprioceptive feedback was impaired using a thick pad over the force plate. The co-ordinates of the COP were found with the use of equations 11 and 12:

$$x = \frac{-hF_x - M_y}{F_z} \quad (11)$$

$$y = \frac{-hF_y - M_x}{F_z} \quad (12)$$

For the first two trials, h (pad thickness) was 0 as the pad was not required. The values for x, y, and z were normalised to their mean. Then the sway path of the accelerometer was computed using the equation from Mayagoitia et al. (Mayagoitia et al., 2002). The resultant acceleration was calculated using equation 13:

$$A = \sqrt{+\, a_x^2 + a_y^2 + \, a_z^2} \quad (13)$$

The angles between the components and the resultant were calculated using equations 14-16:

$$\alpha = cos^{-1}(\frac{\alpha_x}{A}) \quad (14)$$

$$\beta = cos^{-1}(\frac{\alpha_y}{A}) \quad (15)$$

$$\gamma = cos^{-1}(\frac{\alpha_z}{A}) \quad (16)$$

Next, to find the magnitude of the vector D, a vector collinear to the resultant acceleration, the height of the accelerometer above the floor, dz, was divided by −cosγ. The x- and y-components of D could then be found using the equations below. Sway diagrams were produced by plotting dx versus dy, and the mean velocity of the sway was calculated in the same way as the COP's mean velocity:

$$d_x = D\ x\ cos\alpha \quad (17)$$

$$d_y = D\ x\ cos\beta \quad (18)$$

Stabilograms and inclinometer sway diagrams were plotted for each subject and each condition using the force plate data and accelerometer data and the researchers noted an increase in the COP path in both diagrams and it was also highlighted that the data acquired from each method of balance assessment was not directly comparable, although the trends they exhibited were comparable.

Similarly, Blaszczyk et al. conducted a study looking at the use of force plates in the measurement of postural instability (Blaszczyk, 2016). They focused on three main groups of participants; young adults, *N* = 60, mean age 21 ± 2 years; older adults, *N* = 54, mean age 64 ± 8 years; and patients with idiopathic Parkinson's disease (PD) without dyskinesia and motor fluctuations above grade 3, *N* = 54, mean age 65 ± 9 years. Their data was collected by the participants standing on the FBP (QFP Medicapteurs, France) and standing for six 26.5-second trials: 3 with eyes open (EO) and 3 with eyes closed (EC). The signals were sampled at 40 Hz and to remove the noise they were filtered off-line with the Chebyshev II 10^{th} order low-pass filters (MATLAB v. 6.0, The MathWorks, Inc, USA), with a cutoff frequency set at 0.4 Hz. They measured sway by calculating a sway vector (this is characterised by the average COP velocity) as well as the sway directional index (DI) and the sway ratio (SR). Statistical analysis showed highly significant differences in the outcome variables between the groups of participants. They concluded that the sway vector may be recommended as a useful standard in static posturography.

Karlsson and Frykberg carried out a study to compare different force plate measures for assessment of postural stability during normal standing (Karlsson and Frykberg, 2000). During this study, five main measures were investigated: (1) the standard deviation of the horizontal ground reaction force; (2) the standard deviation of the COP; (3) the mean velocity of the COP; (4) movement strategy measures; and (5) the standard deviation of the vertical ground reaction force. Their sample of participants were those affected by a cerebral vascular accident. Their sample included 12 men and 8 women, mean age 50 years (range: 21–65 years). A piezoelectric force plate (Kistler type 9284, Kistler AG, Winterthur, Switzerland) as well as two video cameras, connected to a computer, were used to collect their data. The sampling frequency was set to 50 Hz. The participants were asked to stand normally on the force plate, with arms hanging loosely by their sides, and to look at a spot in front of them. It is notable that the participants kept their shoes on during this study. The mean of three measurements was taken for improved accuracy. The participants rested in between tests. After this, the Berg balance test was then evaluated. The COP and force data were lowpass filtered with a cut-off frequency of 15 Hz to reduce the measurement noise. Then the COP data was high pass-filtered with a cut-off frequency of 0.1 Hz to remove slow drifts in the COP, which are not directly associated with the spontaneous sway. The acceleration of the COM was estimated using an inverted pendulum model matching the participant's mass and height. This estimated COM acceleration was then compared to the measured acceleration of the participant's COM. The researchers found a significant correlation between the model's COM acceleration and the standard deviation of the measured COM acceleration, suggesting that the inverted pendulum model is a valid representation for assessing postural stability in older adults.

2.3 The use of two force plates to measure postural instability

Dual force plates offer richer insights into asymmetrical loading and postural control between limbs, especially relevant in older populations or those with neurological conditions. Matłosz et al. conducted a study to compare three different force plates for the assessment of postural stability (Matlosz et al., 2020). Three different devices were used to evaluate vertical ground reaction forces: (1) the Alfa platform designed by AC International East, Knurów, Poland; (2) The CQStab2P device, a dual platform stabilometric device developed by CQ Elektronik System, Czernica, Poland; and (3) the AccuGait system from Advanced Mechanical Technology, Inc., Watertown, MA, USA. They enlisted 111 volunteers to complete the Romberg test on each device, with each test lasting 30 seconds to avoid tiredness of the volunteers. The subjects were asked to be barefoot on a thin cellulose sheet, orienting their feet freely and naturally, with both hands hanging loose along the trunk. The outline of the study participant’s feet was drawn on the cellulose sheet. The cellulose sheet with the outline of the feet was placed on each of the analysed force platforms to ensure identical arrangement of the feet during each test. Significant differences were found among the platforms for all measures, including the Romberg ratio, path length, and COP area. The highest Spearman’s rank correlation was between Alfa and CQStab2P (0.20 to 0.38), and the lowest between Alfa and AccuGait (−0.19 to 0.09). Bland-Altman analysis indicated low agreement levels, especially between AccuGait and Alfa. The study concluded that caution is necessary when comparing COP values from different devices analysed by various software, and it questioned the validity of unperturbed stance tests among young healthy adults as a postural control parameter.

Lee and Park carried out a study on the development of a more robust tool for postural stability analysis of laparoscopic surgeons (Lee and Park, 2008). They began their study by recruiting six surgeons to undergo the tasks from the fundamentals of laparoscopy (FLS™) skill set. Endoscope images from a 0° scope were displayed on a standard cathode-ray tube (CRT) monitor positioned at eye level in front of the participants. Thirty-nine markers were placed on each participant's anatomical landmarks. These researchers made use of the ViconPeak™ (Lake Forest, CA) motion capture system. The sampling rate was set to 100 Hz. To reach the optimal table height for the participants to complete their tasks adequately, Lee and Park relied on the findings of Berquer et al., which define the optimal height as that at which the laparoscopic instrument handles are at elbow height or 10 cm below elbow height (Berquer et al., 2002). During each task, participants stood with one foot on each force plate. A 4^{th} order Butterworth low-pass filter with a cut-off frequency of 6 Hz was applied to the raw force plate data to remove high-frequency noise. During data analysis, COP_{Left} and COP_{Right} were calculated under the left and right foot, and an overall COP ($COP_{overall}$) was defined as the weighted average of each force plate COP based on vertical ground reaction forces. They then characterised the excursions by sway area, by using a principal component analysis (PCA) technique. Overall, the study developed a tool for analysing surgeons' postural stability during minimally invasive surgery by integrating COM, COP, and a novel metric termed postural stability demand (PSD), revealing that variations in postural adjustments correlate with skill level and task complexity, and that PSD strongly correlates with performance time, indicating its potential as a predictor of surgical proficiency.

2.4 Critical appraisal of the literature review

Table 2 provides a synthesis of the key studies reviewed in this chapter, highlighting their methodologies, strengths, and limitations. A narrative critical appraisal of the literature follows below.

Table 2. Critical appraisal summary of previous studies using accelerometery and force plates for postural assessment.

Study	Methodology	Population	Pros	Cons
Alkathiry et al. (2018)	Accelerometer + Romberg	Adolescents with concussion	Focused on AP sway; sensitive to concussion-induced instability	Static tasks only no transitional metrics
Kamen et al. (1998)	Accelerometer + Romberg	Young vs older adults	Age-related sway differences; high test-retest reliability	Static stance only; small sample
Alberts et al. (2015)	iPad + NeuroCom FBP	Healthy young adults	Consumer device feasible; reasonable agreement with FBP	Static testing only; no transitional data
Dewan et al. (2019)	iPhone + Vicon/Force Plate	Healthy adults	Multimodal integration (camera, FP, accelerometer)	No data from older adults; low phone sampling rate; static tasks
Hsieh et al. (2019)	Smartphone + Force Plate	Older adults	Fall risk assessment; strong correlation with COP data	Only static tasks; Romberg method only
Lo et al. (2022)	Custom Force Plate	Young and older adults	Assessed reliability across age groups	No accelerometry; static conditions only

Seimetz et al. (2012)	Accelerometer vs FBP	Healthy adults	Compared both sensor types; included vector decomposition	Very small sample (n=6); no clinical group
Blaszczyk et al. (2016)	Single Force Plate	Young, older adults, Parkinson's patients	Introduced sway vector and directional index; group comparison	Eyes-open/closed only; no movement phase assessed
Karlsson & Frykberg (2000)	FBP + COM model	Stroke patients	Used inverted pendulum model to estimate sway; validated approach	Simplified model; limited segmental insight; static task
Matłosz et al. (2020)	Three FP systems	Healthy adults	Platform comparison; used dual FP setups	Results not generalisable to older/frail; stance-based only
Lee & Park (2008)	Dual FP + Motion Capture	Laparoscopic surgeons	Developed postural stability demand (PSD) metric; real-time kinematics	Not ageing/frail sample; task-specific postural responses

The literature reviewed in this chapter provides an overview of methodologies for assessing postural stability using both accelerometry and FBPs. While the studies examined offer robust methodological examples in clinically relevant applications, several critical gaps remain with particular relevance to this Master's thesis, most notably, the limited attention to transitional balance phases (e.g., standing up from lying down), the segmentation of sensor modalities, and the underrepresentation of older or fall-prone populations.

Most reviewed studies focus predominantly on static postural assessments, often using variations of the Romberg test to evaluate sway under altered sensory conditions (e.g., eyes open vs. closed, firm vs. foam surfaces), as demonstrated by Alkathiry et al. (2018), Kamen et al. (1998), and Hsieh et al. (2019). While such methods provide important insights into sensory integration and standing balance, they do not capture the dynamic demands of everyday movements, particularly transitions such as rising from a supine position, which are clinically important in fall-prone older adults.

Moreover, while consumer-grade technologies such as smartphones and tablets have shown promise for balance assessment (e.g., Alberts et al., 2015), limitations related to sampling resolution, sensor accuracy, and applicability in high-risk older populations restrict their clinical utility. This thesis contributes to the literature by including older adults attending a specialist falls clinic, a population underrepresented in many prior studies.

Studies using force plates, such as Blaszczyk et al. (2016) and Karlsson & Frykberg (2000), have established COP metrics as a gold standard. However, these systems often treat the body as a single rigid mass and may not account for segmental contributions to sway. Furthermore, force plate research tends to focus on quiet standing, with little emphasis on real-world postural transitions such as standing up from lying or sitting, limiting ecological validity. The literature also rarely investigates mediolateral asymmetries or tracks sway evolution across postural phases, despite evidence that lateral instability is a significant predictor of falls in older adults (Siragy et al., 2024).

Another key limitation in the literature concerns the segmentation and scope of sensor modalities. Many studies employ accelerometers or force plates in isolation (e.g., Kamen et al., 1998; Seimetz et al., 2012). A few have attempted to integrate both technologies (e.g., Dewan et al., 2019; Lee & Park, 2008), but such efforts have been confined to static conditions or younger, healthier populations. Given the complexity of integrating data from multiple sensors and the absence of established protocols for older adults during AS transitions, this thesis analysed accelerometry and FBP data in two separate phases and independent samples. This approach allowed for a clearer appraisal of each modality's specific contributions and limitations when applied to orthostatic postural control during the AS.

In summary, while the existing literature supports the use of both accelerometry and force plates for assessing balance, it often overlooks dynamic transitions and older or frailer clinical populations. This thesis addresses these gaps by applying sensor-based analyses to a clinically meaningful transition in older adults and offers pilot evidence to inform future, larger-scale research in this area.

3. Phase one: A pilot study measuring orthostatic motion in older adults using accelerometery data

The first phase of this Master's research involved an ethically-approved secondary analysis of data previously collected by the FRAILMatics team. The dataset included participants aged 50 years and older who were recruited from a local healthy ageing group. These participants visited the FASU clinic at St. James's Hospital, Dublin, where they underwent a medically-led assessment to document their demographic, anthropometric, and clinical characteristics. They also completed a standard AS test, which involved lying down for 10 minutes, standing up as quickly as possible, and remaining motionless for three minutes. During the test, participants wore an accelerometer positioned anteriorly at the waist. As discussed in previous chapters, accelerometers are typically placed at the participant's COM, traditionally at the lower back (Ahmed et al., 2017). However, this placement was not feasible in this research, as participants were required to lie on their back before standing, and a posterior placement of the accelerometer would have caused discomfort.

The specific aims of this first phase were to develop the capacity to process accelerometry-based data collected during the AS test, compute a Motion Index (MI) reflecting postural movement intensity, and segment the data into three predefined Regions of Interest (ROIs): pre-stand (ROI 1), early stand (ROI 2), and late stand or recovery phase (ROI 3). Additionally, this phase introduced and explored a novel metric, the Early Micromovement Index (EMI). The analysis aimed to examine associations between accelerometry-derived metrics (MI and EMI) and a range of demographic, anthropometric, and clinical variables. Two primary hypotheses guided this phase: first, that the MI/EMI might exhibit varying associations with participant characteristics, potentially capturing micromovements not visible to the naked eye; and second, that the accelerometry signal could reflect indirect influences such as respiration, possibly attributable to the anterior placement of the sensor at the waist.

3.1 Ethics approval

Ethics approval for this secondary analysis of FRAILMatics data was obtained from the Tallaght University Hospital/St. James's Hospital Joint Research Ethics Committee (Project ID: 0221; initial approval date: 4 May 2021; amendment approval date: 2 August 2022). Additional approval was granted by the St. James's Hospital Research and Innovation Office (Reference: 6567; approval date: 26 July 2021). All participants provided written informed consent prior to participation. The study adhered to the ethical principles outlined in the World Medical Association Declaration of Helsinki for medical research involving human subjects (Duggan, 2024).

3.2 Participants and setting

This study involved a secondary analysis of a pre-existing FRAILMatics dataset of volunteers aged 50 years and older, recruited from a local healthy ageing group. Participants were eligible if they could provide written informed consent, mobilise independently (with or without assistance), and transfer from lying to standing with minimal support. Exclusion criteria (for the purpose of the primary purposes for which the data was collected) included a history of confirmed or presumed neurogenic orthostatic hypotension (OH) or contraindications to bioelectrical impedance analysis (BIA), such as the presence of an indwelling electronic device (e.g., a cardiac pacemaker). Full details of the sample and methodology are reported elsewhere (Duggan, 2024).

Participants attended the FASU at St. James's Hospital, Dublin, for a dedicated medically-led research assessment, which included an AS test involving participants lying supine for 10 minutes, standing up as quickly as possible, and remaining motionless in an upright position for three minutes. During the AS test, accelerometery data was collected using a single accelerometer placed anteriorly at the level of the waist.

3.3 Active Stand

The AS test, as comprehensively outlined by Finucane et al., involves a standardised procedure to assess cardiovascular responses to orthostatic stress (Finucane et al., 2019). Participants rest supine for at least 5 minutes to establish baseline cardiovascular measurements, with continuous beat-to-beat blood pressure and heart rate monitoring. They are then instructed to stand up promptly, achieving an upright position usually within 3 seconds. Once standing, they remain still for at least 3 minutes while continuous monitoring continues to observe hemodynamic responses. This protocol is designed to evaluate autonomic cardiovascular regulation and is commonly used to diagnose conditions such as orthostatic hypotension (OH) and other autonomic dysfunctions. Whilst the AS test is primarily used to assess cardiovascular responses to standing, in the FRAILMatics study, participants also wore an accelerometer to capture additional data, as described earlier. A hypothesis was that the accelerometer would detect subtle micromovements during the transition and re-stabilisation phases, particularly in relation to more adverse clinical characteristics.

For the purpose of this research phase, the AS test was divided into three distinct regions of interest (ROIs), as illustrated in Figure 1. These ROIs were defined as follows: **pre-stand (ROI 1)**, spanning from 60 to 30 seconds before standing, to capture baseline postural dynamics; **stand (ROI 2)**, covering 10 to 40 seconds after standing, focusing on the early post-stand phase; and **recovery (ROI 3)**, from 120 to 150 seconds post-standing, representing the late stabilisation phase. Notably, ROI 2 begins 10 seconds after the initiation of standing to minimise the influence of artefacts associated with the physical act of standing and its immediate biomechanical aftermath, which are not representative of steady-state postural control.

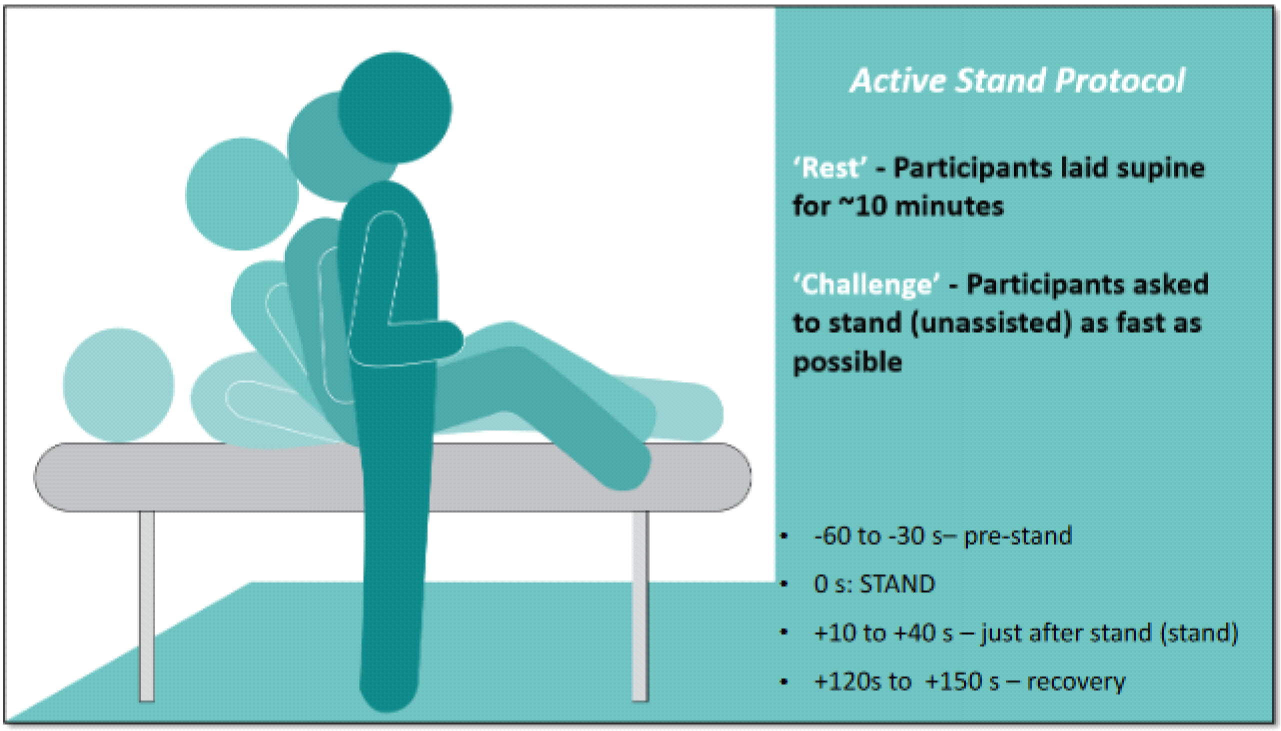


Figure 1. Regions of interest in the active stand experiment.

3.4 Participant Characteristics

The measures used in this study included: age (in years); sex (male or female); Body Mass Index (BMI, kg/m^2); the number of self-reported medical conditions; the number of medications regularly taken; current smoking status (never smoked, ex-smoker, current smoker); weekly alcohol consumption (measured in units); maximum handgrip strength (in kilograms); time to perform the 5-Times Sit-to-Stand Test (5CST) (in seconds); and the physical frailty score assessed using the SHARE Frailty Instrument for Primary Care (SHARE-FI) (Romero-Ortuno et al., 2010).

3.5 Accelerometer Equipment

The Shimmer3 EMG accelerometer (Shimmer Sensing, Dublin, Ireland, https://www.shimmersensing.com/product/shimmer3-emg-unit/) was the model used in this study, with a consistent sampling rate of 512 Hz maintained throughout. Due to the supine starting position, the accelerometer was positioned anteriorly at the waist rather than at the standard location on the lower back, ensuring participant comfort.

3.6 Signal quality test

A data integrity report was created to thoroughly assess the quality of the collected data. This process involved visualising the data for each participant by plotting it. Following this initial review, the data was further analysed using the digital signal processor in MATLAB (The MathWorks, Inc., Natick, MA, USA). The spectrogram function was employed to identify any significant background noise, ensuring the accuracy of the dataset.

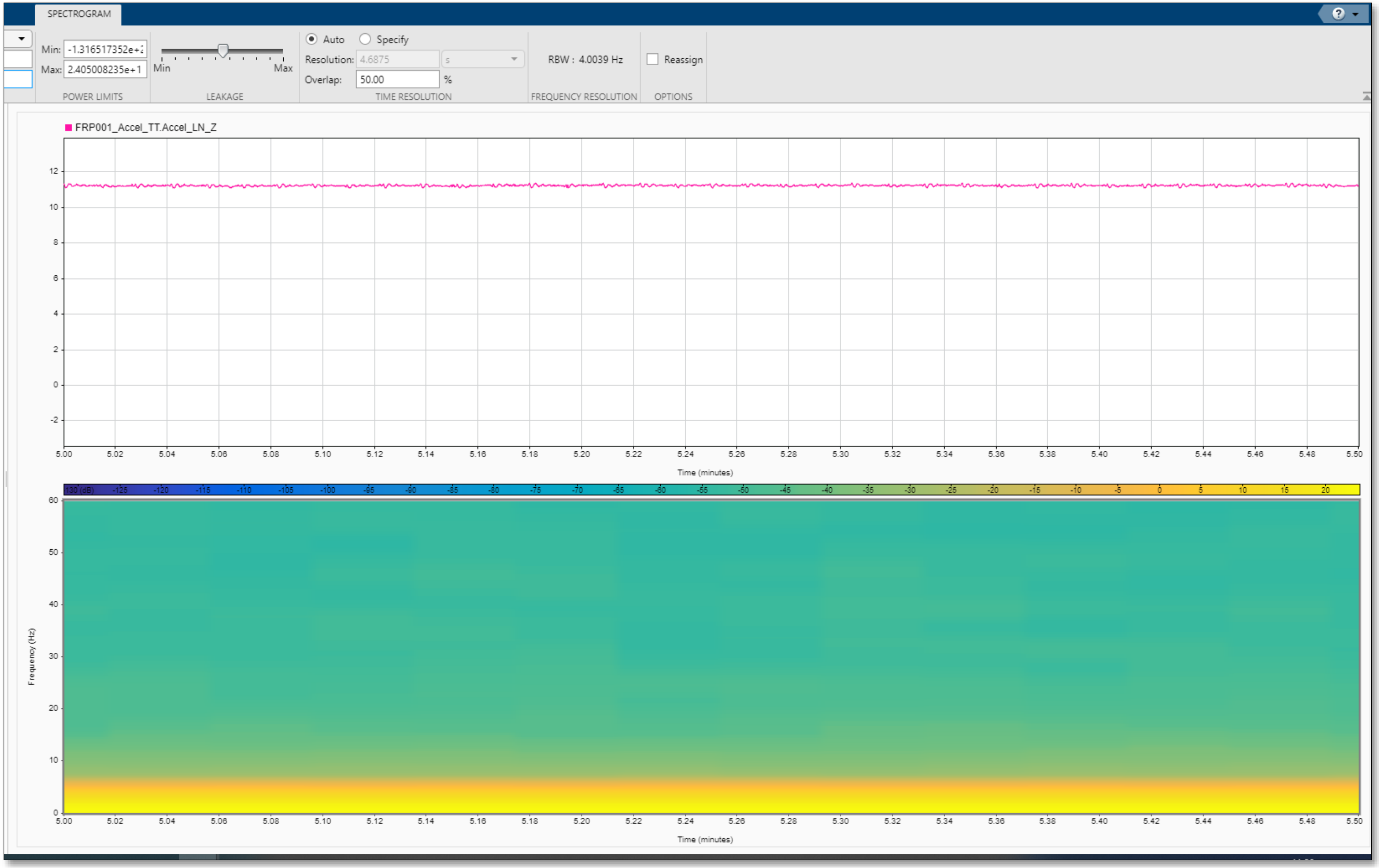
3.7. Data analysis

3.7.1 MATLAB

MATLAB was extensively used for the analysis of the acquired data. Multiple custom scripts were developed, each tailored to a specific analytical function, and were systematically organised based on their purpose to streamline the workflow. Given that this study involved secondary data analysis, early visualisation of the dataset was essential to facilitate understanding of its structure and inform subsequent analytical steps. To achieve this, a custom MATLAB script was created to generate initial visual outputs, allowing me to interpret the data effectively and gain a clear overview of the tasks required.

Firstly, a MATLAB script was developed to automatically identify the three ROIs in the AS. ROI 1 (spanning from 60 to 30 seconds before standing) corresponded to 300–330 seconds in the data, capturing baseline postural dynamics before the transition. ROI 2 (spanning from 10 to 40 seconds after standing) corresponded to 370–400 seconds in the data, focusing on the early post-stand phase during the transition to upright stability. ROI 3 (spanning from 120 to 150 seconds after standing) corresponded to 480–510 seconds in the data, representing the late stabilisation phase as participants regained steady posture.

After defining the ROIs, spectrograms were generated for each signal and participant. The purpose of this step was to visually represent the signals and assess their artefact level or amplitude, to identify any potential anomalies or noise. This served as an additional method to ensure the integrity and reliability of the sample data. Figures 2 and 3 provide examples of the data quality check process, demonstrating a comparison that highlights artifacts present in the raw dataset.



SPECTROGRAM
Min: -1.316517352e+2
Max: 2.405008235e+1
POWER LIMITS
Min
Max
LEAKAGE
Auto
Specify
Resolution: 4.6875
Overlap: 50.00 %
TIME RESOLUTION
RBW : 4.0039 Hz
FREQUENCY RESOLUTION
Reassign
OPTIONS
FRP001_Accel_TT.Accel_LN_Z
Time (minutes)
Frequency (Hz)
Time (minutes)

Figure 2. Example of a participant with minimal artefact in the accelerometer signal.

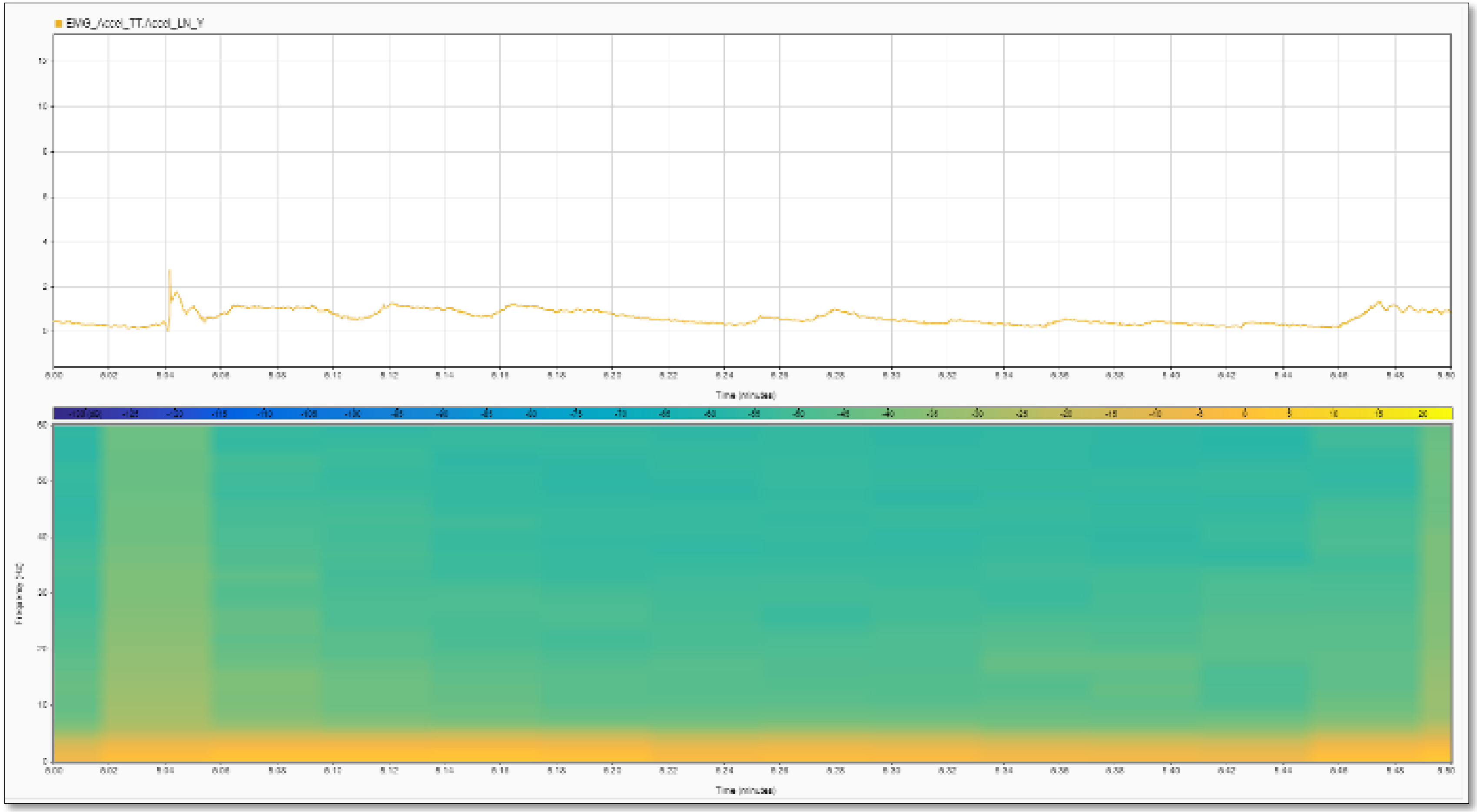


Figure 3. Example of a participant with visible artefact in the accelerometer signal.

After excluding major data artefacts, a low-pass Butterworth filter (fourth-order, zero-phase) with a cutoff frequency of 10 Hz was applied to the data of the included participants. Computations were then carried out on the valid signals for all participants. Firstly, the average was taken for x, y and z directions per participant. The standard deviation was then calculated for the same signals as aforementioned. Once this step was complete, the equation for the Euclidean distance was applied to compute the variable Index:

$$Index = \sqrt{(x)^2 + (y)^2 + (z)^2} \qquad (19)$$

The mean and standard deviation of the Index values were calculated for each ROI, resulting in three distinct mean indices: Mean Index 1 (MI ROI 1), Mean Index 2 (MI ROI 2), and Mean Index 3 (MI ROI 3). To further quantify postural adjustments and dynamic stability, the Early Micromovement Index (EMI) was derived. The EMI was calculated as the difference between the mean index of ROI 2 (early stand phase) and ROI 3 (recovery phase):

$$\mathrm{E}MI = mean\ index\ 2 - mean\ index\ 3 \qquad (20)$$

This metric provides a focused measure of the magnitude of micromovements during the early post-stand phase (ROI 2) relative to the recovery phase (ROI 3). The EMI was introduced as a novel parameter to identify and quantify subtle postural instabilities that may be indicative of frailty or impaired compensatory mechanisms, thereby offering a meaningful way to differentiate between varying levels of postural stability.

When the MI in ROI 2 (early post-stand phase) is greater than the MI in ROI 3 (recovery phase), the Early Micromovement Index (EMI) will have a positive value. This indicates that the magnitude of micromovements is higher during the early phase after standing (ROI 2) compared to the recovery phase (ROI 3). It was hypothesised that a positive EMI would mean greater instability or more pronounced adjustments immediately after standing, potentially reflecting delayed or impaired stabilisation, which could be associated with conditions like frailty.

3.8 Statistical Analyses

Statistical analyses were conducted using SPSS version 27.0 (IBM Corp., Armonk, NY, USA).

Descriptive statistics for continuous variables were presented as either mean with standard deviation (SD) or median with interquartile range (IQR). For categorical variables, data were summarised using counts and percentages (%). Visualisations of continuous variables across groups were created using grouped box-and-whisker plots, generated with the Explore function. To visualise the relationships between two continuous variables, simple scatter plots were generated with a linear regression line superimposed. The statistical significance of these bivariate associations was tested using the simple linear regression procedure in SPSS.

Bivariate statistical comparisons across continuous variables were performed using the independent samples t-test or Mann-Whitney U test, while the Chi-square or Fisher's exact test were employed for dichotomous variables.

To examine whether participant characteristics were associated with postural sway during the AS test, two Multivariate General Linear Models (MANOVA) were conducted. In both models, mean index (MI) values from the three defined regions of interest (ROIs)—pre-stand (ROI 1), early stand (ROI 2), and recovery (ROI 3)—served as the dependent variables. The first model included age (continuous) and Body Mass Index (BMI; continuous) as independent variables. The second model included age and smoking status (ex-smoker vs. never smoker) as predictors. Each MANOVA tested for overall multivariate effects using Wilks' Lambda, followed by univariate tests of between-subjects effects to assess the contribution of each variable to individual ROIs. Partial eta squared values were calculated to estimate effect sizes, and model fit was evaluated with adjusted R^2 values. Estimated marginal means, adjusted for age, were generated to summarise average MI values across ROIs.

A significance level of $p < 0.05$ was used throughout to determine statistical significance.

3.9 Results for Phase One

Thirty-one participants were initially recruited, with 26 ultimately included in the analysis following data quality checks. The mean age was 70.5 ± 5.4 years (range: 59–82), with 85% identifying as female, and none reporting a history of falls. No participants were current smokers, although four were former smokers. The mean BMI for the sample was 25.9 ± 3.8 kg/m². While no significant bivariate associations were observed with age, sex, or other medical variables, a lower BMI was statistically significantly associated with higher MI values across all three ROIs in the AS test (Figures 4–6). However, the corresponding R^2 values were low, indicating a very small explanatory power. Additionally, individuals with a history of smoking exhibited higher MI values across all three AS phases compared to those who had never smoked (Figures 7–9).

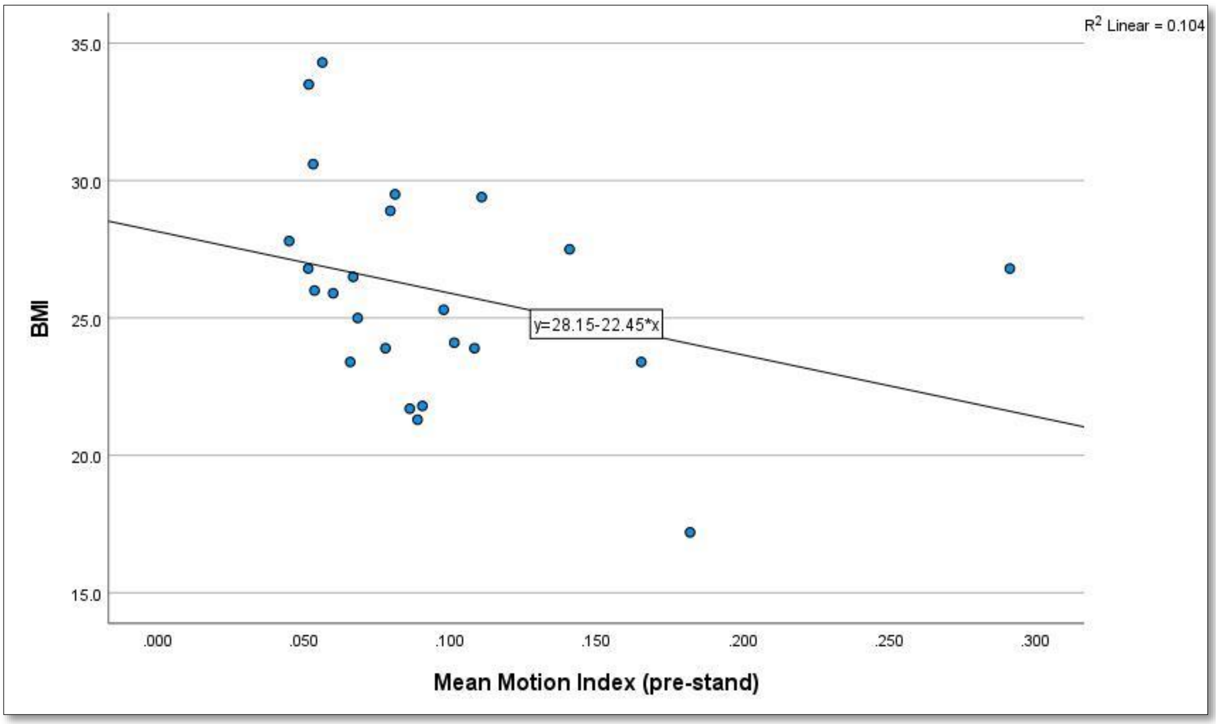


Figure 4. Scatter dot plot between mean Motion Index in ROI 1 (pre-stand) and Body Mass Index (linear regression $p < 0.05$).

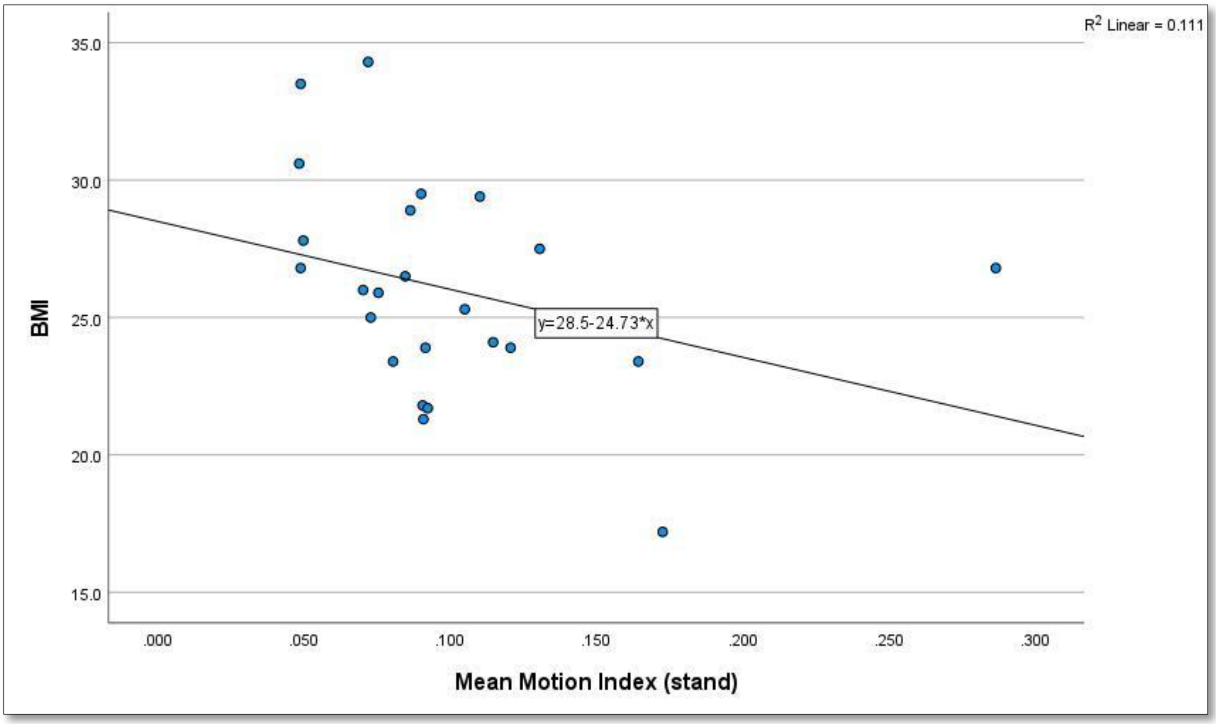


Figure 5. Scatter dot plot between mean Motion Index in ROI 2 (stand) and Body Mass Index (linear regression $p < 0.05$).

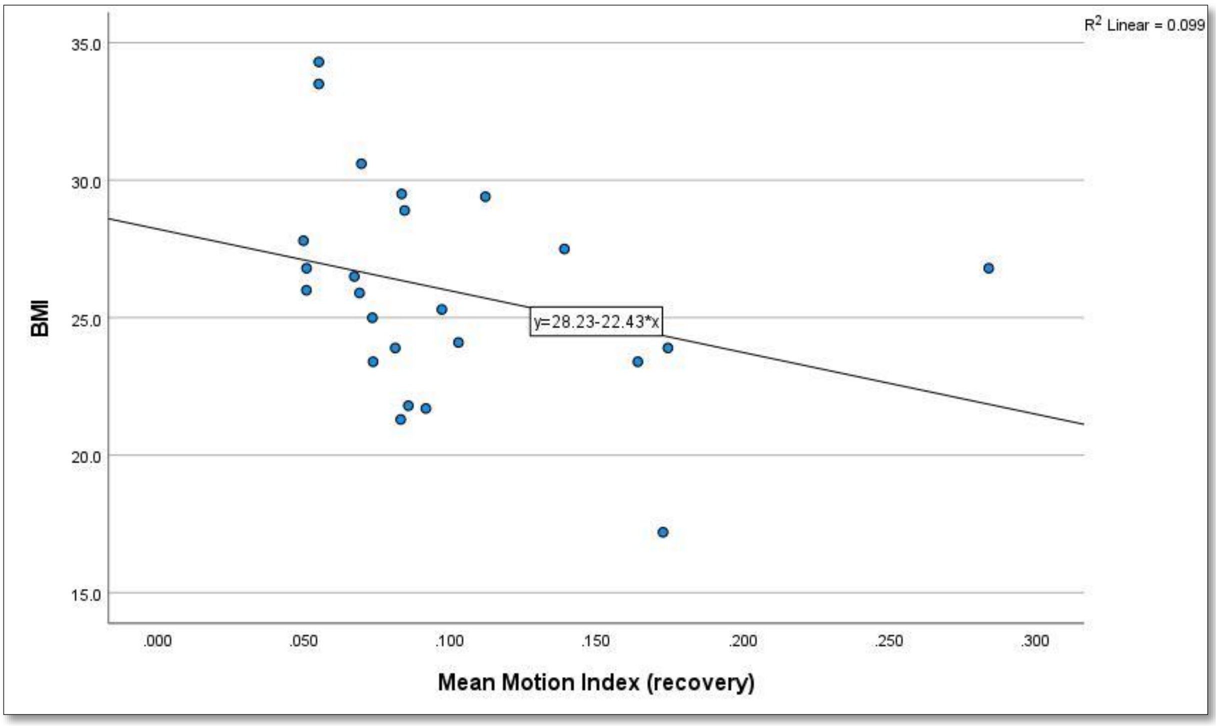


Figure 6. Scatter dot plot between mean Motion Index in ROI 3 (recovery) and Body Mass Index (linear regression $p < 0.05$).

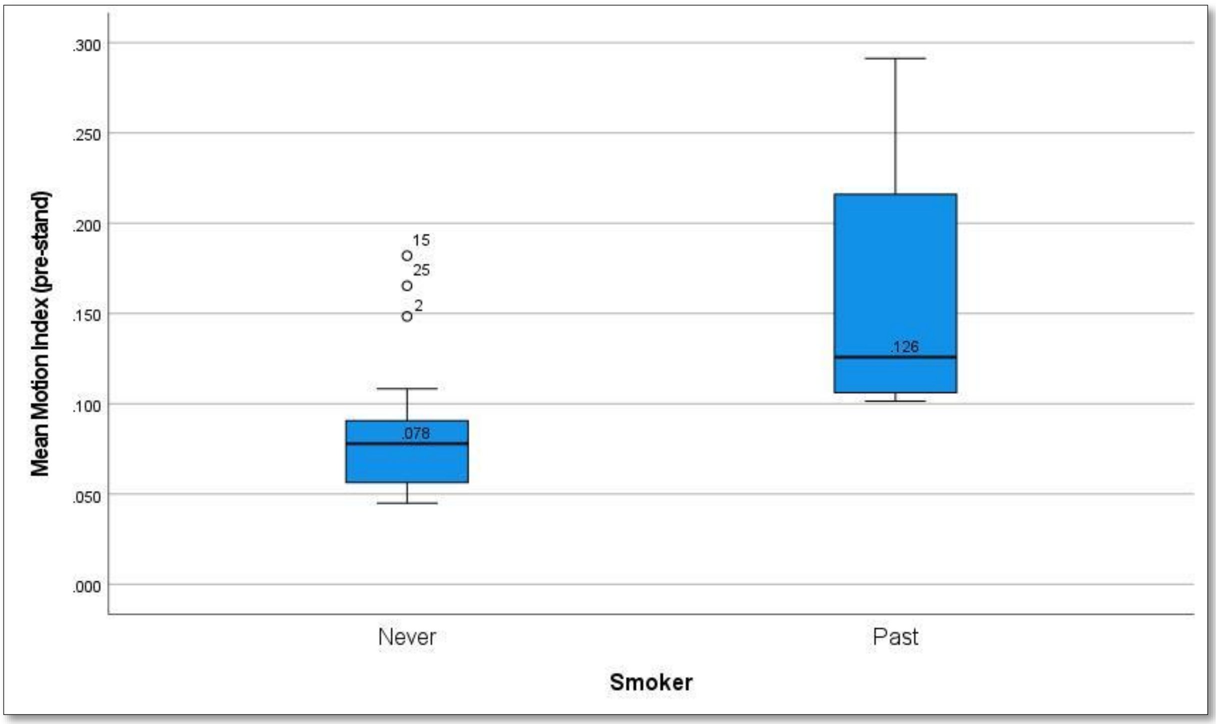


Figure 7. Grouped box-and-whisker plot of mean Motion Index in ROI 1 (pre-stand) versus smoking status (2-sided Mann-Whitney U test $p < 0.05$).

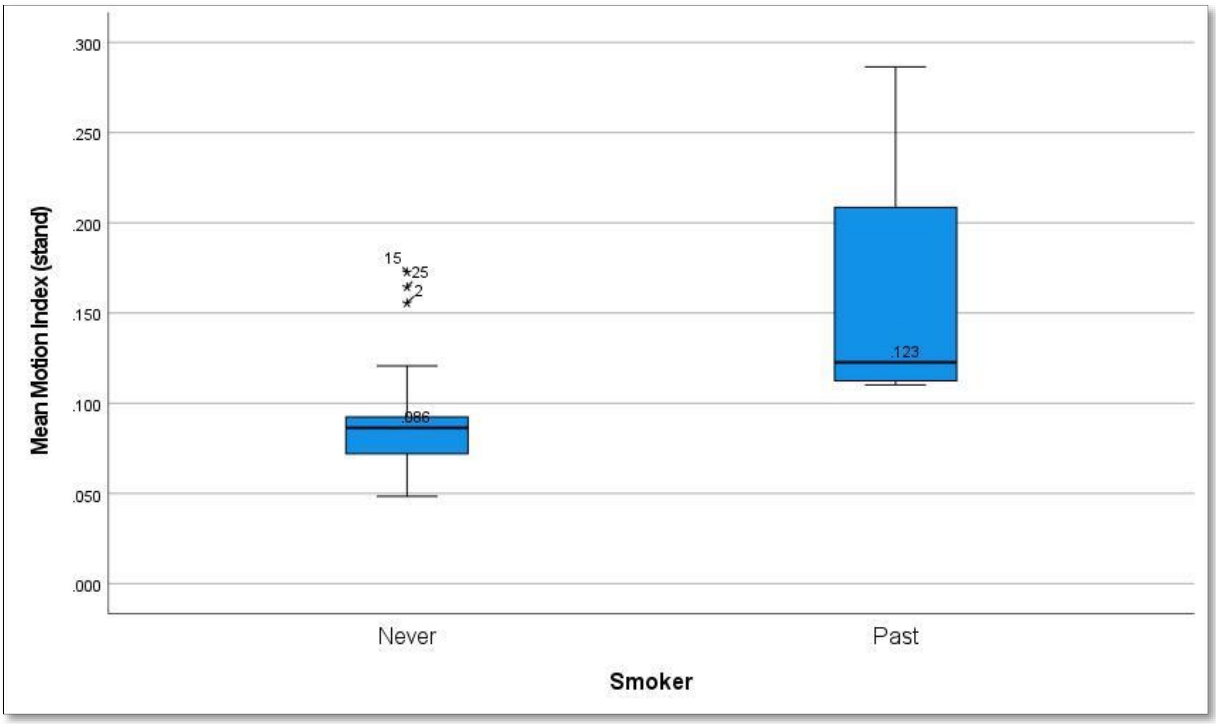


Figure 8. Grouped box-and-whisker plot of mean Motion Index in ROI 2 (stand) versus smoking status (2-sided Mann-Whitney U test $p < 0.05$).

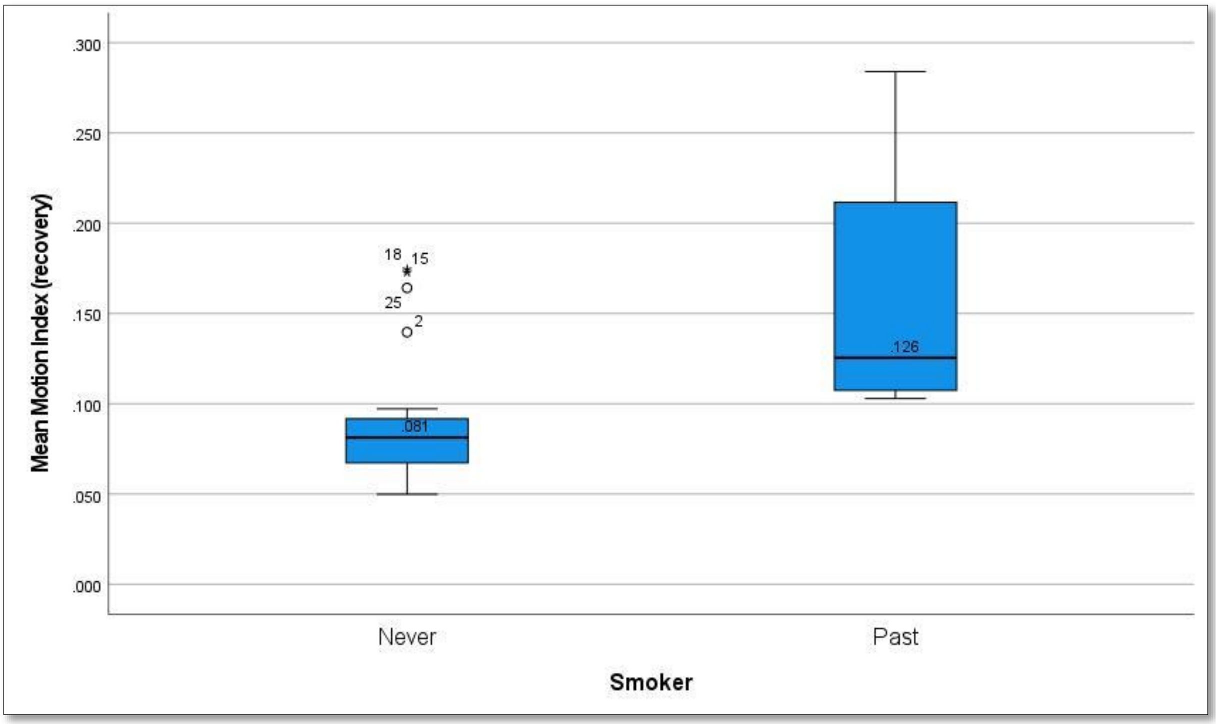


Figure 9. Grouped box-and-whisker plot of mean Motion Index in ROI 3 (post-stand) versus smoking status (2-sided Mann-Whitney U test $p < 0.05$).

Multivariate General Linear Models (MANOVAs) results were obtained as follows (Table 3):

Model 1 included age and Body Mass Index (BMI) as continuous covariates. This model did not reveal any statistically significant multivariate effects for either age (Wilks' Lambda = 0.948, $F(3,21) = 0.387$, $p = 0.763$) or BMI (Wilks' Lambda = 0.907, $F(3,21) = 0.722$, $p = 0.550$). Follow-up univariate tests showed that neither age nor BMI were independently associated with MI values in any ROI, and adjusted R^2 values were consistently low, indicating limited explanatory power.

Model 2 evaluated the effect of smoking history (never vs. ex-smoker), controlling for age. A significant multivariate effect of smoking status was observed (Wilks' Lambda = 0.658, $F(3,21) = 3.645$, $p = 0.029$, partial $\eta^2 = 0.342$). Univariate analyses revealed that ex-smokers had significantly higher MI values in both ROI 1 ($F(1,23) = 11.811$, $p = 0.002$, partial $\eta^2 = 0.339$) and ROI 3 ($F(1,23) = 9.995$, $p = 0.004$, partial $\eta^2 = 0.303$). No significant effect was found in ROI 2. These findings suggest that past smoking may influence postural stability during supine rest and recovery after standing, even when accounting for age.

Table 3. Summary of Multivariate General Linear Models (MANOVA) examining associations between participant characteristics and mean Motion Index (MI) values across three regions of interest (ROIs) during the Active Stand test.

Model	Independent Variable(s)	Multivariate Significance (Wilks' Lambda)	Significant Univariate Effects	Interpretation
Model 1	Age, BMI	Age: $F(3,21)=0.387$, $p=0.763$BMI: $F(3,21)=0.722$, $p=0.550$	None	No significant effect of age or BMI on postural MIs. Low explanatory power across all ROIs.
Model 2	Smoking status (ex vs. never), Age	Smoker: $F(3,21)=3.645$, $p=0.029$, $\eta^2=0.342$	ROI 1: $F=11.811$, $p=0.002$ ROI 3: $F=9.995$, $p=0.004$	Smoking history associated with higher postural instability at baseline and recovery, even after adjusting for age.

Results in Table 4 address the hypothesis that a positive Early Micromovement Index (EMI), indicated by the mean MI in ROI 2 (early post-stand phase) exceeding the mean MI in ROI 3 (recovery phase), may correlate with adverse clinical characteristics. However, among the 9 participants where EMI > 0 was observed, no statistically significant differences were found when compared to the 17 participants with EMI ≤ 0.

Table 4. Associations of a Positive Early Micromovement Index (EMI).

Characteristic	EMI > 0 N = 9	EMI ≤ 0 N = 17	p value
Mean Age, years (SD)	69.3 (3.9)	70.8 (6.1)	0.419
Female sex (%)	100%	75%	0.260
Median No. of medical conditions (IQR)	2 (4)	2 (1)	0.388
Median No. of medications (IQR)	2 (3)	2 (3)	0.890
Ex-smoker %	22.2%	12.5 %	0.602
Mean BMI, kg/m^2 (SD)	26.8 (4.6)	25.4 (3.4)	0.207
Median alcohol units per week (IQR)	8 (13)	3.3 (11.5)	0.207
Mean maximum grip strength, kg (SD)	21 (5.8)	25.6 (8.0)	0.251
Mean 5CST, seconds (SD)	11.1 (1.7)	10.7 (1.9)	0.558
Mean SHARE-FI score (SD)	-1.31 (0.791)	-0.535 (0.273)	0.419

SD: standard deviation; IQR: interquartile range; BMI: body mass index; 5CST: 5-Times Sit-to-Stand Test; SHARE-FI: SHARE Frailty Instrument for Primary Care.

To evaluate the statistical robustness of the comparisons between EMI positive and negative groups, post-hoc power calculations were conducted using MedCalc® version 23.2.6. Two comparisons were of interest: mean BMI and ex-smoker status. For the BMI comparison, the observed difference in means was 1.4 kg/m^2, with standard deviations of 4.6 and 3.4 in the respective groups. Using a significance level (α) of 0.05 and a target power of 80% (β = 0.20), the sample size required to detect this difference with sufficient statistical power was 111 participants in Group 1 and 210 in Group 2, totalling 321 participants. Given the actual sample size of 26, the analysis is underpowered for this effect size.

For the smoking status comparison, a chi-squared test for proportions was applied. The proportions of ex-smokers in the EMI > 0 and EMI ≤ 0 groups were 22.2% and 12.5%, respectively. To detect this difference with 80% power at a 0.05 significance level, the required sample size was 177 in Group 1 and 335 in Group 2, summing to 512 participants. Again, this highlights that the current study is significantly underpowered for detecting small-to-moderate differences in categorical predictors like smoking history.

3.10 Discussion

This first phase of the research explored the feasibility and utility of using accelerometry during the AS test to assess postural micromovements in older adults. By developing motion mean indices (MIs) across three defined regions of interest (ROIs) and introducing the Early Micromovement Index (EMI), the study sought to identify subtle postural changes that might reflect impaired neuromuscular or autonomic compensatory responses to orthostasis. A key and consistent finding was the association between past smoking and elevated postural motion, particularly in ROI 1 (supine rest) and ROI 3 (recovery phase). These associations persisted even after adjusting for age, and were detected through multivariate general linear modeling (MANOVA). This supports the hypothesis that smoking—though modifiable—may exert lasting effects on systems governing postural stability. Chronic exposure to tobacco has been implicated in endothelial dysfunction, altered sensory integration, and impaired neuromuscular control (Iki et al., 1994, Przybylski et al., 2013), all of which could plausibly contribute to increased sway before and after postural transitions. Furthermore, in individuals with smoking-related conditions such as chronic obstructive pulmonary disease (COPD), postural instability has been associated with diminished sensory processing and reduced muscular reserve (Janssens et al., 2014, Porto et al., 2015, Strandkvist et al., 2023), highlighting a possible physiological mechanism that underpins the observed associations in our sample.

The effects of past smoking were evident both during the supine resting phase (ROI 1) and the recovery phase after standing (ROI 3), but not during the early post-stand phase (ROI 2). This pattern suggests that smoking-related impairments in postural stability may persist even in non-challenging, resting conditions and re-emerge during the re-stabilisation phase following a postural transition. The absence of significant effects in the more acute upright phase (ROI 2) may reflect the complex interplay between compensatory neuromuscular responses and sensory integration during active adjustment. It is possible that a significant effect would have been detectable in ROI 2 with a larger sample size. These findings support the value of the AS test, when enhanced with accelerometry, as a sensitive approach for detecting subtle postural instability across varying physiological demands, including both static and dynamic conditions.

In contrast, while bivariate analyses suggested an inverse association between BMI and postural motion (i.e., lower BMI linked to higher MI), these findings were not retained in adjusted models. One interpretation is that lower BMI may increase the relative contribution of respiratory-induced movement to accelerometry signals, particularly given the anterior sensor placement. Research has shown that individuals with lower BMI tend to exhibit greater abdominal excursion during normal respiration (Kaneko and Horie, 2012, Du et al., 2024), potentially leading to overestimation of postural micromovement in this subgroup. This highlights the importance of considering anthropometric factors and sensor placement in the design and interpretation of accelerometry-based studies.

Methodologically, the anterior placement of the accelerometer was necessary to ensure comfort during the supine phase of the AS test. However, this placement may have increased the signal's sensitivity to respiration-related artefact. Although signal quality was rigorously assessed using spectrograms and low-pass filtering techniques, the potential for residual artefacts underscores the need for validation studies comparing anterior and posterior placement under controlled conditions.

The EMI was introduced as a novel composite metric designed to capture instability during the early upright phase relative to recovery. While exploratory in nature, the EMI provides a single-value summary of postural transition dynamics and may offer clinical interpretability for future research or screening tools. Although the present study found no statistically significant differences in clinical variables between EMI-positive and EMI-negative participants, the trend toward higher prevalence of smoking in the EMI-positive group is consistent with the broader findings of this phase. Importantly, post-hoc power analyses revealed that the study was significantly underpowered to detect small-to-moderate group differences in variables such as BMI and smoking history, requiring samples of over 300 and 500 participants respectively for adequate power. These findings reinforce the need for larger-scale studies to verify the EMI's predictive validity and refine its clinical application.

The lack of significant associations with age and sex may also reflect limitations in statistical power or sample variability, rather than a true absence of effect. Indeed, while previous studies have documented sex-based differences in balance control (e.g., due to lower muscle mass or hormonal differences affecting proprioception), these effects may be subtle and context-dependent (Kim et al., 2010). Similarly, the relatively narrow age range in this study may have limited the ability to detect age-related differences in postural micromovement.

Overall, this first pilot study demonstrates the feasibility of using a single waist-worn accelerometer to capture dynamic postural responses during the AS test. The simplicity of the equipment and the interpretability of derived metrics like MI and EMI suggest that this approach may be well-suited to both research and clinical environments. Given that the AS test mimics real-life transitions such as rising from bed, integrating accelerometry could enhance early detection of instability and frailty risk in older adults. Furthermore, the ability to detect subtle motion changes—potentially imperceptible to the naked eye—offers a promising avenue for proactive interventions aimed at fall prevention.

In conclusion, the results from this exploratory phase provide preliminary evidence that accelerometry-derived metrics during the AS test can reveal meaningful postural differences associated with modifiable risk factors such as smoking. While findings regarding the EMI require further validation, this metric shows promise as a dynamic marker of orthostatic postural control. Future studies should aim to replicate these results in larger, more diverse populations, evaluate test-retest reliability, and examine longitudinal associations with clinical outcomes such as falls, frailty progression, and quality of life.

4. Phase two: A pilot study of postural instability following an active stand procedure using force plate metrics

The Falls and Syncope Unit (FASU) at St. James's Hospital hosts a gait analysis laboratory equipped with a Codamotion system (https://codamotion.com/). Although not currently accredited by the Clinical Movement Analysis Society (CMAS) for diagnostic clinical use, the lab is regularly used for clinical research activities. With ethical approval granted through the FRAILMatics project, Phase Two of this research involved the MSc candidate collecting original data from patients attending routine FASU clinic appointments.

Following their clinical assessments, participants underwent an additional research evaluation in the FASU gait lab, where orthostatic postural motion was assessed using gold-standard force balance plates (FBPs). This phase of the study aimed to further develop my skills in collecting and processing primary FBP data in a clinical research environment, while also generating novel pilot data to complement the accelerometry-based findings from Phase One.

The specific aims of Phase Two were:

- To complement the method developed in Phase One by acquiring pilot data using the gold-standard force balance plate system.
- To conduct a dual-plate analysis by examining data from each plate separately, in order to explore whether meaningful differences or asymmetries could be detected between the left and right foot during the AS procedure.
- To explore associations between FBP data and clinical features of interest.

4.1 Ethics approval

Ethics approval for the study was obtained from the Tallaght University Hospital/St. James's Hospital Joint Research Ethics Committee (Project ID: 0221; initial approval date: 4 May 2021; amendment approval date: 2 August 2022). Additional approval was granted by the St. James's Hospital Research and Innovation Office (Reference: 6567; approval date: 26 July 2021). All participants provided written informed consent prior to taking part in the study. The study adhered to the ethical principles outlined in the World Medical Association Declaration of Helsinki for medical research involving human subjects (Duggan, 2024).

4.2 Participants' Recruitment and Research Set-Up

Patients aged 50 years or older attending the Falls and Syncope Unit (FASU) at St. James's Hospital for routine clinical appointments were invited to participate in this pilot study. Recruitment aimed to include 10 consecutive eligible participants (N = 10), with screening and inclusion occurring approximately once per week under the supervision of a consultant physician. Eligibility criteria included: recent completion of a standardised AS test as part of routine care, ability to provide informed consent for research participation (including the use of clinical data), and ability to safely transition from a supine to standing position with minimal assistance (i.e., one-person assist). Participants who met these criteria were provided with a Patient Information Leaflet (PIL) to explain the study, after which written informed consent was obtained.

Once consented, participants proceeded to the gait analysis laboratory, where they underwent a research assessment supervised by the attending doctor and led by the MSc candidate. The assessment focused on postural motion analysis using gold-standard FBPs over a three-minute standing period following the AS test. Relevant clinical characteristics were subsequently extracted from participants' medical records to support exploratory analysis. These included age, sex, smoking status (categorised as never or ex-smoker), and whether dizziness was reported during the clinical AS test.

Figure 10. Picture of the force balance plates (FBPs) test set-up.

4.3 Force Balance Plates' Data Processing

Force plate data were collected at a sampling rate of 400 Hz to capture fine-grained fluctuations in postural sway. The raw COP time series were pre-processed using custom

MATLAB scripts. Initial processing involved visual inspection to identify and exclude artefacts, followed by signal filtering to reduce high-frequency noise. A second-order Butterworth low-pass filter was applied, and an optimal cut-off frequency was determined by testing a range from 0.5 to 10 Hz. Based on this exploratory filtering, a 10 Hz cut-off was selected to retain relevant postural dynamics while suppressing extraneous noise.

4.4 Force Balance Plates' Data Analysis

Data analysis began by defining two key Regions of Interest (ROIs) to evaluate postural sway following the active stand. **ROI 2 covered the interval from 20 to 50 seconds post-stand**, representing early upright posture, while **ROI 3 spanned 60 to 90 seconds post-stand**, reflecting later upright balance. These intervals were selected to characterise potential changes in postural stability over time and to capture delayed effects such as those potentially associated with clinical characteristics.

Two metrics were selected from the literature to quantify sway from the force balance plate (FBP) data. The first was the **sway area (AREA-SW)**, which estimates the area enclosed by the COP trajectory over time. This metric was calculated by summing the area of triangles formed between successive points along the COP path and the mean COP location, as described by Prieto et al. (Prieto et al., 1996). This approach captures total displacement in mediolateral (X-axis), anteroposterior (Y-axis), and vertical force (Z-axis) planes, although variability along the Z-axis was expected to be minimal during quiet standing. The formula was implemented in MATLAB to compute sway area separately for each plate and ROI (equation 19):

$$Area - SW = \frac{1}{T}\sum_{n=1}^{N-1} |AP[n+1].ML[n] - AP[n].ML[n+1]| \qquad (19)$$

The second metric was the **95% prediction ellipse area (AREA-CE)**, derived using a MATLAB function developed by Schubert and Kirchner (Schubert and Kirchner, 2014). This metric estimates the area of an ellipse that is expected to enclose 95% of the COP points, offering a summary measure of variability in postural control. Ellipse area is commonly used to assess postural steadiness and provides an alternative view of balance fluctuations that is less sensitive to high-frequency noise than raw sway paths.

Both AREA-SW and AREA-CE were calculated for ROI 2 and ROI 3, and separately for each plate. These dual metrics were chosen to provide complementary insight into sway magnitude and dispersion, supporting a nuanced evaluation of postural dynamics across early and late post-stand periods, separately in each plate.

4.5. Core clinical hypotheses underlying Phase Two

Phase Two of the study was guided by three core clinically supported hypotheses:

First, it was hypothesised that postural sway, as measured by FBP metrics, would increase over time following the active stand—specifically, between the early (ROI 2: 20–50 seconds) and late (ROI 3: 60–90 seconds) upright phases—reflecting physiological fatigue or delayed autonomic adaptation (Mochizuki et al., 2006).

Second, based on findings from Phase One, it was anticipated that smoking history would be associated with greater sway and area of ellipse values, particularly during the recovery phase (ROI 3) (Pereira et al., 2001).

Third, the presence of orthostatic hypotension at 40 seconds post-stand (OH40) was hypothesised to correspond with elevated sway metrics in ROI 3, given its clinical association with impaired baroreflex function and postural instability, regardless of whether patients report symptoms of orthostatic intolerance or not (Claffey et al., 2022).

4.6. Statistical Analyses

Statistical analyses in Phase Two were conducted using IBM SPSS Statistics (Version 29). Given the small sample size (N = 10), a primarily exploratory approach was adopted. Descriptive statistics (mean, standard deviation, range) were used to summarise sway metrics—namely sway area (AREA-SW) and 95% prediction ellipse area (AREA-CE)—across the two postural regions of interest (ROI 2: 20–50s and ROI 3: 60–90s post-stand), and separately for each force plate.

To statistically evaluate (from an explorative perspective) within-subject changes in postural sway over time, repeated-measures Multivariate Analysis of Variance (MANOVA) was applied with ROI (2 vs. 3) as the within-subject factor. Each MANOVA model also included a relevant between-subject factor to test for associations with clinical features: smoking status (ex-smoker vs. never smoked) or presence of orthostatic hypotension at 40 seconds post-stand (OH40). This allowed for the detection of both main effects (e.g., overall changes in sway across ROIs) and interaction effects (e.g., whether changes in sway differed by smoking or OH40 status). Given the pilot nature of the study, alpha was set at 0.05 for significance testing, but p-values were interpreted cautiously due to low statistical power. Effect sizes (Partial Eta Squared) and observed power estimates were reported to provide additional context. Overall, data visualisations (i.e., simple or stratified line plots with 95% confidence intervals for means) were primarily used for interpretation and target statistical testing.

4.7. Results

4.6.1. Characteristics of the sample

The pilot sample consisted of 10 adults, ranging in age from 58 to 83 years, with a mean age of 73.5 years (SD = 8.5). The sample included 40% male (n = 4) and 60% female (n = 6) participants. In terms of smoking status, three participants (30%) were ex-smokers, while seven (70%) reported never having smoked. No current smokers were enrolled in the study. Regarding orthostatic blood pressure responses, three participants (30%) met the clinical criteria for OH40, regardless of symptom status (i.e., dizziness during the AS or not). The remaining seven participants (70%) did not exhibit OH40.

4.6.2. Force plate data processing

Figure 11 displays an example of filtered force plate data recorded during the full three-minute standing phase of the AS test. The plotted data correspond to the medio-lateral (ML) and antero-posterior (AP) directions of the COP trajectory as captured on Plate 1. The data have been processed using a second-order Butterworth low-pass filter with a 10 Hz cut-off frequency, as described in Section 4.3, to minimise high-frequency artefacts while preserving physiologically relevant sway patterns. Vertical markers indicate the start and end of ROI 2 (20–50 seconds) and ROI 3 (60–90 seconds), allowing visual inspection of postural sway behaviour across early and later upright intervals. As shown, the AP trajectory exhibits greater variability than the ML direction, consistent with typical sway dynamics in upright standing.

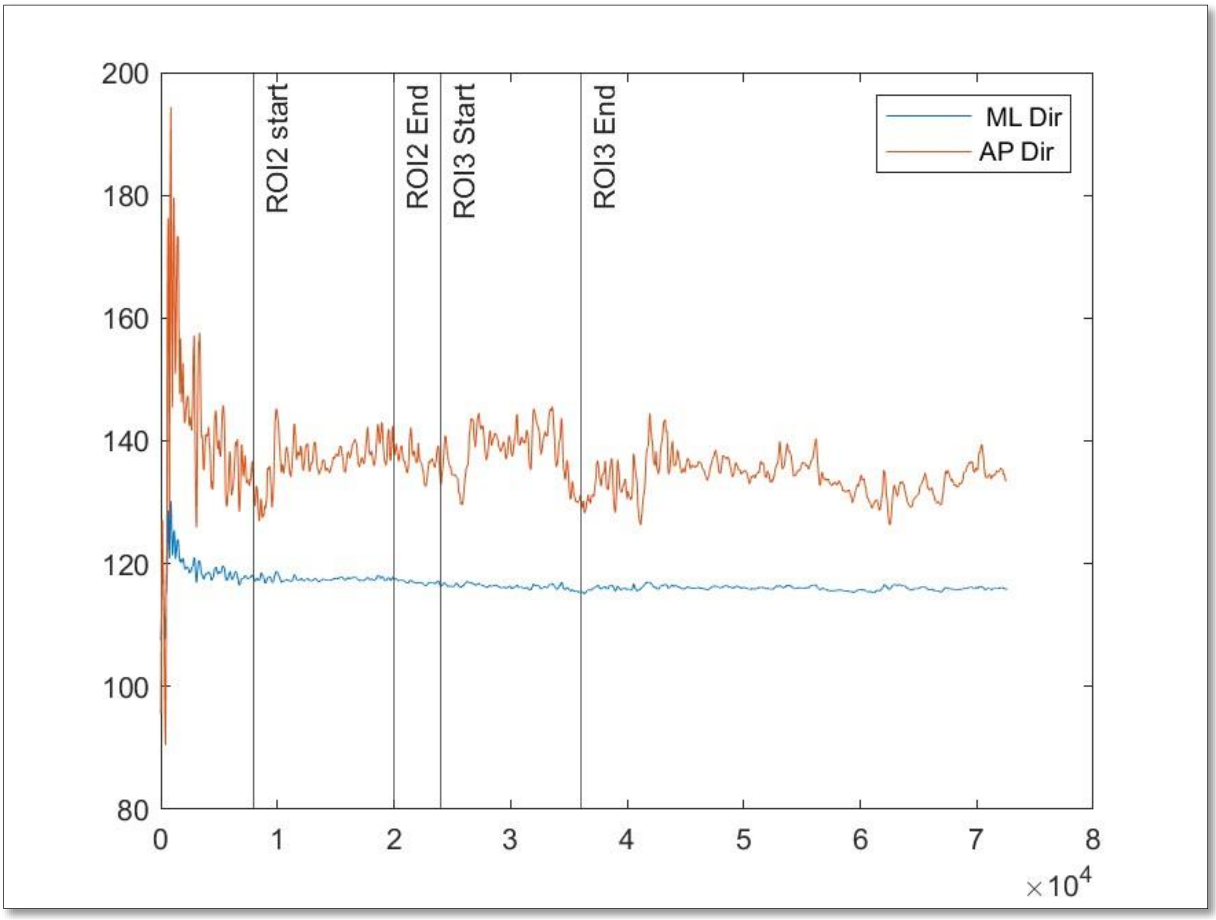


Figure 11. Filtered centre-of-pressure (COP) time series from Plate 1 in a participant during the full 3-minute standing phase of the Active Stand test, with ROI 2 (20–50 s) and ROI 3 (60–90 s) intervals annotated.

4.6.3. Force plate data descriptives

Tables 5 and 6, respectively, provide the descriptive statistics on the Area of Sway (mm^2/s) and Area of Ellipse (mm^2) for plates one and two for ROI2 and ROI3:

Table 5. Descriptive Statistics – Sway area (mm^2/s).

Sway Area (mm^2/s)	N	Minimum	Maximum	Mean
Plate one – ROI2	10	1.4	13.8	4.6 ± 3.6
Plate one – ROI3	10	1.1	75.4	15.5 ± 23.2
Plate two – ROI2	10	1.4	6.9	4.1 ± 1.8
Plate two – ROI3	10	0.7	45.7	10.8 ± 15.0

Table 6. Descriptive Statistics – Area of Ellipse (mm^2).

Area of Ellipse (mm^2)	N	Minimum	Maximum	Mean
Plate one – ROI2	10	17.1	173.5	67.1 ± 48.2
Plate one – ROI3	10	10.0	791.6	222.6 ± 274.5
Plate two – ROI2	10	11.2	96.6	55.4 ± 25.6
Plate two – ROI3	10	9.1	652.8	192.4 ± 256.8

4.6.2. Evaluation of the use of two force balance plates

One objective of this phase was to explore potential differences between the two force balance plates in the measurement of postural sway. This analysis was conducted on an exploratory pilot basis, using visual inspection of line plots with 95% confidence intervals. The intention was to assess whether any consistent discrepancies or asymmetries emerged between the measurements captured by each plate, particularly during the different standing phases defined as ROI2 and ROI3. To facilitate this, the average sway area and the mean area of the ellipse for each plate were graphically represented across the two ROIs. Figures 12 and 13 illustrate the mean sway area for Plate 1 and Plate 2, respectively, while Figures 14 and 15 show the corresponding plots for the area of the ellipse.

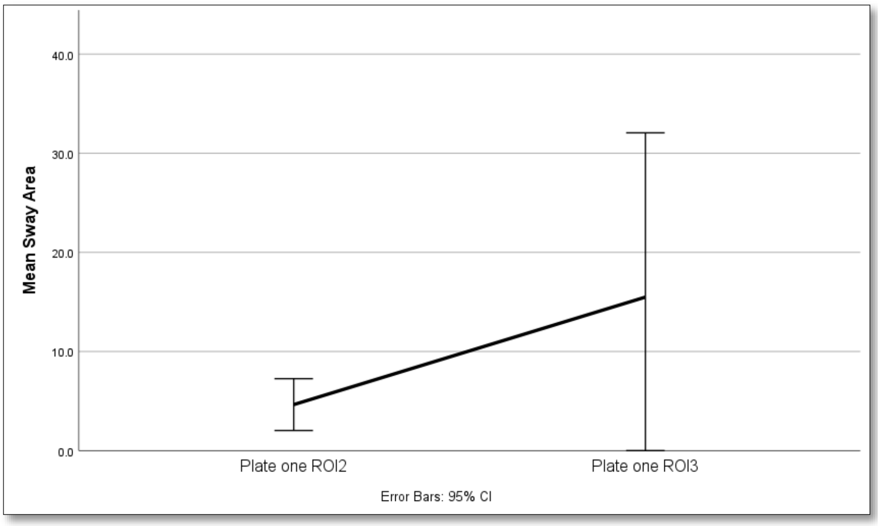


Figure 12. Comparative plot of mean sway area across ROI2 and ROI3 for Plate 1.

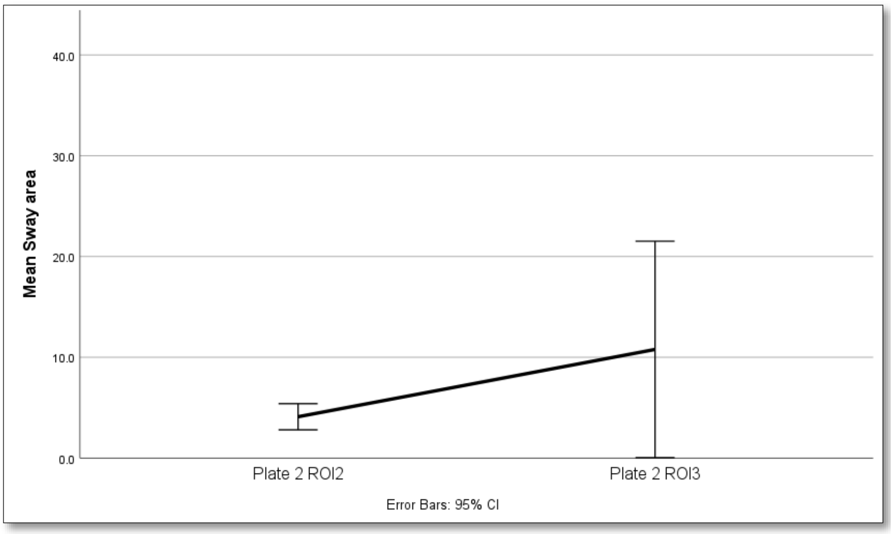


Figure 13. Comparative plot of mean sway area across ROI2 and ROI3 for Plate 2.

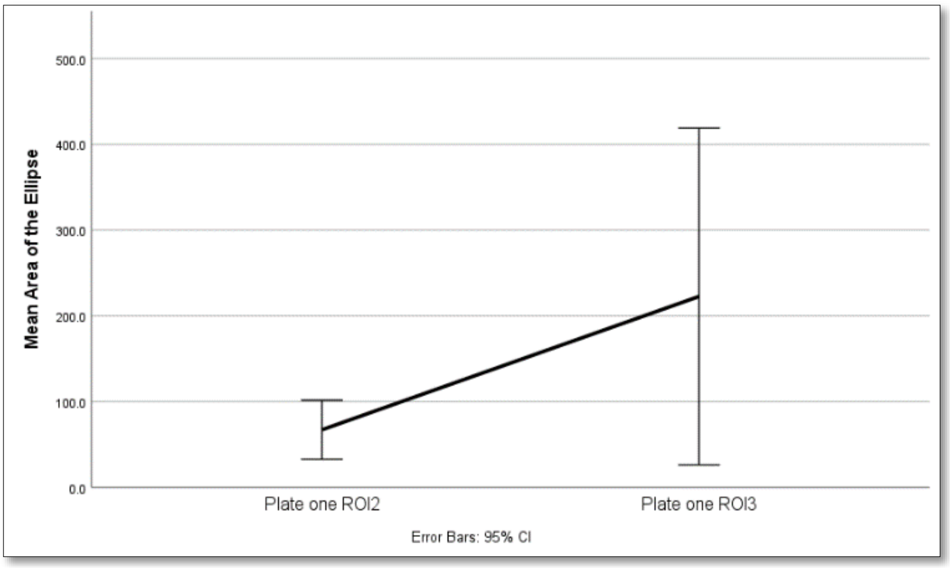


Figure 14. Comparative plot of mean area of ellipse across ROI2 and ROI3 for Plate 1.

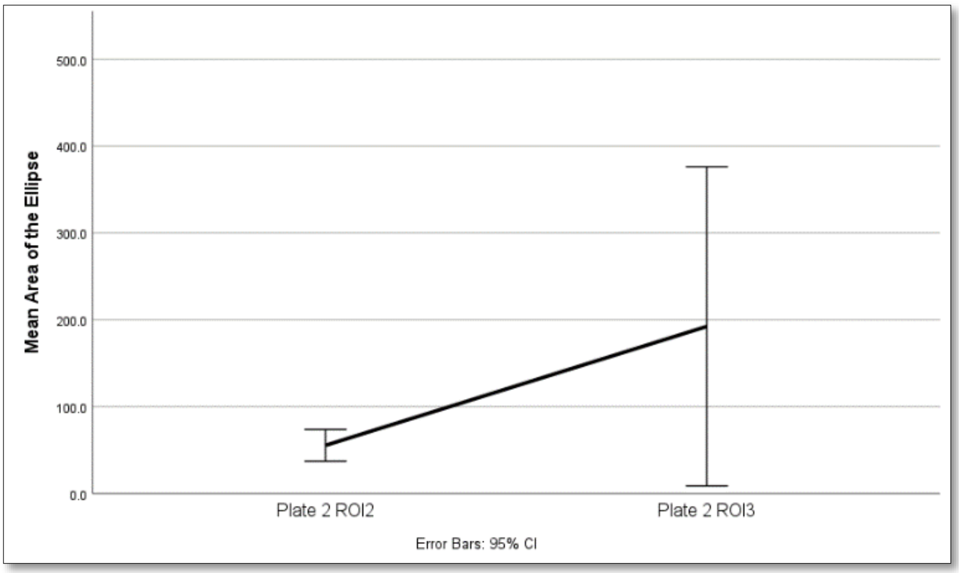


Figure 15. Comparative plot of mean area of ellipse across ROI2 and ROI3 for Plate 2.

The visual comparison reveals that postural sway tended to increase from ROI2 to ROI3 across both plates, consistent with a pattern of greater instability later in the standing period. However, Plate 2 demonstrated slightly lower variability compared to Plate 1, although overall the two plates seemed to produce comparable signals.

4.6.3. Associations between force plate features and clinical characteristics

To explore potential clinical correlates of postural sway, we examined associations between force plate metrics and two key clinical variables: smoking status and the presence of OH40 post-stand during the clinical AS. These variables were selected based on findings from Phase One and clinical hypotheses concerning autonomic regulation and balance control.

A series of repeated-measures Multivariate Analyses of Variance (MANOVA) were conducted to examine the influence of smoking status on postural sway metrics recorded using two FBPs during the Active Stand (AS) test. Sway Area and Area of Ellipse were analysed across two upright regions of interest: ROI 2 (20–50 seconds post-stand) and ROI 3 (60–90 seconds post-stand).

Across all four models, the main within-subjects effect of ROI showed trends toward significance (p-values ranging from 0.085 to 0.159), suggesting that postural sway changed over time following the orthostatic challenge. However, the interaction between ROI and smoking status was not statistically significant in any model (all p-values > 0.29), indicating no strong differential effect of smoking across the two standing phases.

Descriptive data suggested a tendency for greater sway and larger ellipse areas in smokers, particularly during ROI 3 (recovery), but these differences did not reach statistical significance. These findings imply that while there are indications of greater postural motion in smokers during recovery, the limited sample size may have constrained the power to detect meaningful effects.

To complement this analysis, I further explored the clinical hypothesis that participants may experience increased postural fatigue the longer they remain standing. Particular attention was directed toward the variable OH40, defined as the presence of clinically significant orthostatic hypotension measured at 40 seconds post-stand during the participants' preceding clinical AS. Individuals with OH40 often report sensations of dizziness and/or instability during sustained upright posture, which could plausibly result in reduced postural control during the later stages of the AS test—particularly within ROI 3 (60–90 seconds post-stand). In this pilot sample, three out of ten participants met the criteria for OH40. To explore potential differences, sway area and 95% ellipse area were visually stratified by OH40 status:

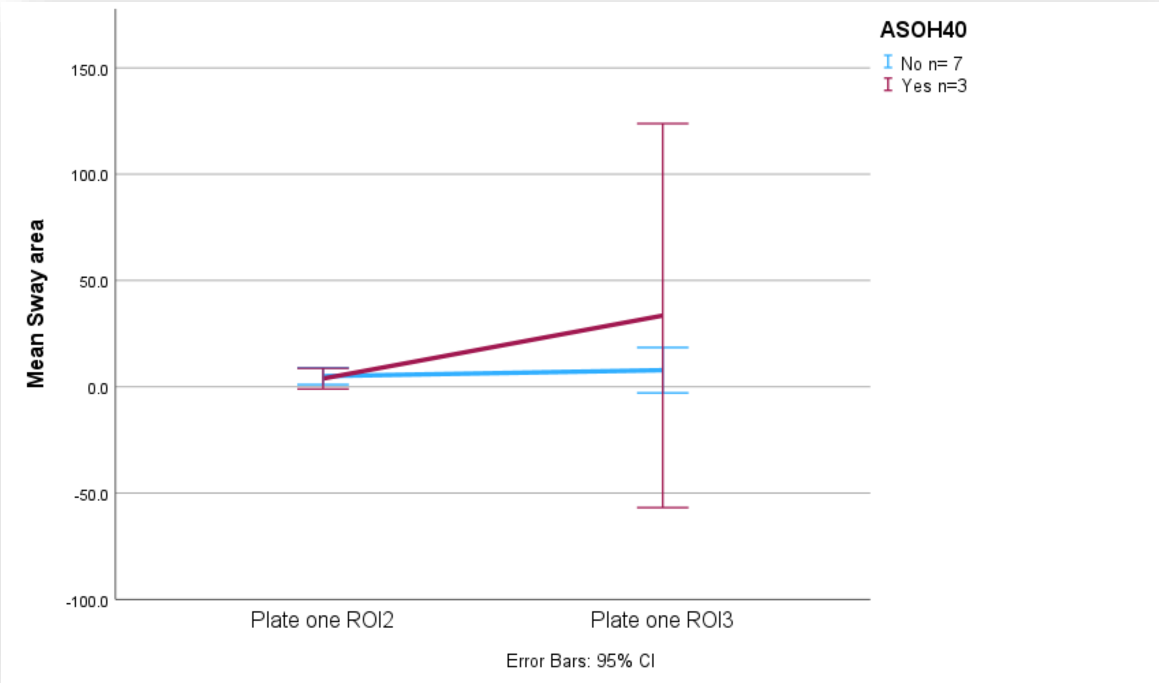


Figure 16. Stratified comparison of mean sway area between ROI2 and ROI3 on Plate 1 by OH40 status.

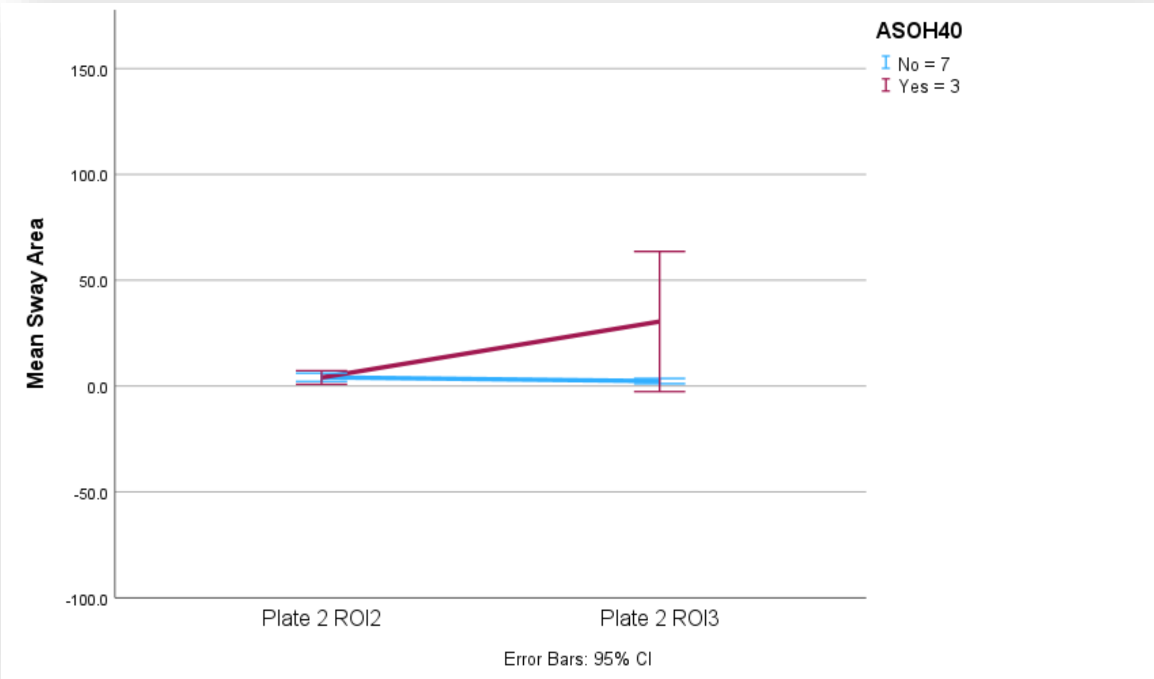


Figure 17. Stratified comparison of mean sway area between ROI2 and ROI3 on Plate 2 by OH40 status.

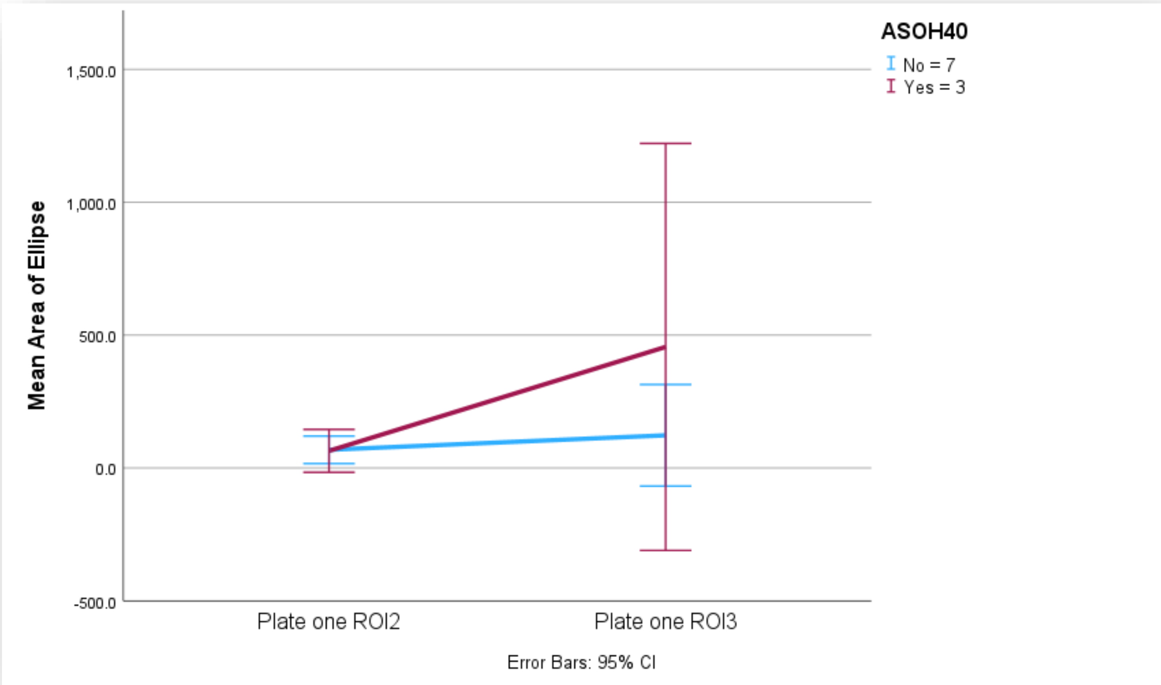


Figure 18. Stratified comparison of area of ellipse between ROI2 and ROI3 on Plate 1 by OH40 status.

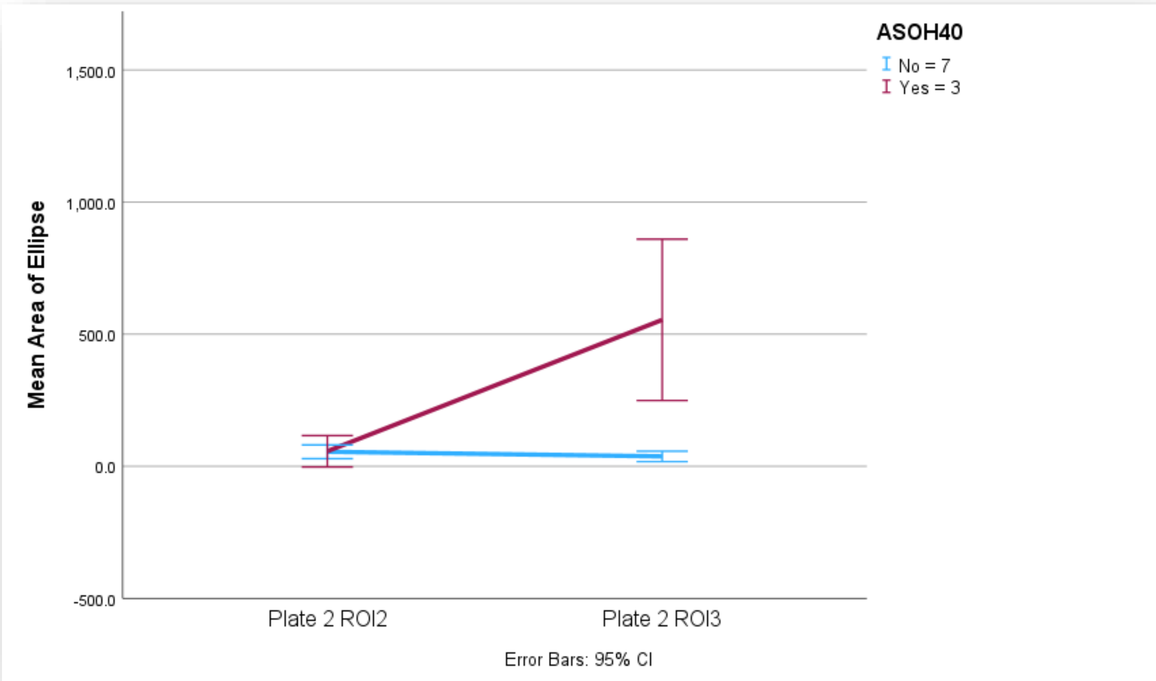


Figure 19. Stratified comparison of area of ellipse between ROI2 and ROI3 on Plate 2 by OH40 status.

Participants with OH40 (n = 3) exhibited a greater increase in mean sway area from ROI2 to ROI3 compared to those without OH40 (n = 7), and this was especially visible in Area of Ellipse Plate 2 analysis, where there was no overlap of confidence intervals in ROI3 between these 2 small groups.

This hypothesis was statistically examined using a repeated-measures General Linear Model (GLM) to assess the effect of OH40 status on postural sway, as measured by the area of the ellipse on force plate 2. The within-subjects factor was ROI, defined across two post-stand intervals: ROI2 (20–50 seconds) and ROI3 (60–90 seconds). Descriptive data showed comparable ellipse areas between groups during ROI2 (Mean ± SD: No OH40 = 54.8 ± 28.2 mm^2; OH40 = 56.8 ± 23.9 mm^2). However, substantial differences were observed in ROI3, where the OH40 group demonstrated markedly higher ellipse values (No OH40 = 37.3 ± 21.4 mm^2; OH40 = 554.1 ± 122.9 mm^2), suggesting increased postural sway during the later phase of standing.

Statistical analysis confirmed a significant main effect of ROI on ellipse area: $F(1,8) = 88.64$, $p < 0.001$, with a large effect size (Partial $\eta^2 = 0.917$), indicating that sway increased over time across participants. A significant interaction between ROI and OH40 status was also observed: $F(1,8) = 102.02$, $p < .001$ (Partial $\eta^2 = 0.927$), indicating that the change in ellipse area differed significantly between groups. Additionally, OH40 status was associated with a significant between-subjects effect on overall ellipse area: $F(1,8) = 132.42$, $p < 0.001$ (Partial $\eta^2 = .943$), reflecting consistently higher sway in the OH40 group. Parameter estimates showed no significant group difference in ROI2, but a pronounced disparity in ROI3, where ellipse area was significantly higher among OH40 participants (mean difference = –516.74 mm^2, $p < 0.001$). Furthermore, the 95% confidence intervals for ellipse area in ROI3 did not overlap between the two groups, reinforcing the potential clinical relevance of this finding.

While the small sample size (N = 10) limits the generalisability of these results, the magnitude and consistency of the observed effects suggest that ellipse area on force plate 2—particularly during later standing phases—may be a useful marker of OH-related postural instability. These findings also support the utility of analysing data from each force plate separately to capture subtle asymmetries or localised changes in postural control.

4.7 Discussion of Phase 2

This pilot study aimed to evaluate the feasibility and potential clinical relevance of using dual force balance plates (FBPs) to quantify postural sway during an AS test in older adults and to complement accelerometry-based findings reported in Phase One. While accelerometry in Phase One revealed increased postural motion among ex-smokers—especially during the recovery phase (ROI 3)—this pattern was not replicated using FBP-derived metrics.

This discrepancy likely reflects methodological differences. Accelerometers, particularly those positioned anteriorly at the waist, may be more sensitive to thoracoabdominal movement, including respiratory effort and upper-body compensatory adjustments. These signals could differ between ex-smokers and non-smokers due to smoking-related respiratory or neuromuscular impairments (Iki et al., 1994, Przybylski et al., 2013). In contrast, FBPs assess shifts in the COP via lower-limb forces and may not capture upper-body movements as effectively. Thus, FBP-based sway may reflect a narrower, though more stable, window into balance control.

The dual-plate setup enabled examination of potential asymmetries in postural control between limbs. Overall, sway and ellipse areas were similar across both plates, but Plate 2 showed slightly lower variability in ROI 3, particularly in ellipse area. Though not statistically significant and of uncertain clinical meaning, this could suggest subtle lateralisation of postural strategies, possibly influenced by limb dominance or individual biomechanics. Prior research has linked such asymmetries with fall risk in older adults, especially during dynamic tasks (Melzer et al., 2004). However, this trend requires validation in larger samples before drawing firm conclusions.

More prominently, this study identified substantial differences in postural sway between participants with and without OH40. Individuals with OH40 (n = 3) demonstrated much higher sway and ellipse areas in ROI 3 across both plates, especially on Plate 2. These findings are consistent with literature linking impaired autonomic compensation to instability during orthostasis (Finucane et al., 2019). In clinical terms, OH40 may signify vulnerability to falls during prolonged standing, possibly due to blunted baroreflex activity and reduced cerebral perfusion (Claffey et al., 2022). Notably, in ROI 3 of Plate 2, the 95% confidence intervals for OH40 and non-OH40 groups did not overlap, suggesting a potentially clinically meaningful distinction—albeit in a small sample.

The limited sample size (N = 10) is a key limitation, and the findings must be interpreted cautiously. The high variability observed, particularly in ROI 3, underscores the influence of individual factors such as fatigue, and perhaps medications and comorbidities. Nevertheless, the effect sizes observed in the OH40 comparisons were large, supporting the potential utility of FBP-derived measures—particularly ellipse area—as markers of orthostatic instability. Yet, it is essential to replicate this finding in independent, larger samples, for external validity of the result obtained in my study.

In conclusion, while FBP measures did not reproduce the smoking-related effects observed with accelerometry, they did reveal differences associated with OH40 during the recovery phase of the AS test. These findings support the feasibility of incorporating dual-plate FBP assessments into AS testing and suggest that such measures may help detect postural instability linked to autonomic dysfunction. Future research should explore the combined diagnostic potential of FBP and accelerometry in identifying subclinical instability and fall risk in older adults.

5. Conclusions of the thesis

In this thesis, I set out to explore the feasibility and potential clinical value of incorporating wearable accelerometry and force balance plate (FBP) technologies to assess postural stability during the Active Stand (AS) test in older adults. The project was undertaken as a pilot study and consisted of two phases, each targeting a distinct but complementary component of orthostatic assessment.

Phase One involved the analysis of accelerometer data from a waist-mounted sensor and provided early indications that such wearable technology could detect subtle postural motion patterns during upright standing. Notably, the analysis suggested that smoking history was associated with greater motion intensity, as detected by the anterior accelerometer, both at baseline (ROI 1) and during the recovery phase following the stand (ROI 3), indicating increased late orthostatic postural instability in ex-smokers compared to never-smokers. These preliminary findings point toward the possibility that anterior trunk accelerometry is sensitive to thoracoabdominal movement, potentially capturing respiratory-related compensatory mechanisms that differ in smokers. However, I fully acknowledge the exploratory nature of these results and the limitations due to the small sample size, which precludes generalisation at this stage.

In Phase Two, I piloted the use of dual force plates in the Falls and Syncope Unit (FASU) at St. James's Hospital. The aim was to complement the accelerometry data by capturing centre of pressure (COP) movements and exploring whether meaningful asymmetries or limb-specific contributions to balance could be observed. While force plate metrics did not replicate the smoking-related findings from Phase One—likely because they focus more on lower-limb dynamics—I found that participants with orthostatic hypotension at 40 seconds (OH40) exhibited notably larger sway and ellipse areas during the later standing phase (ROI 3). This was especially visible on Plate 2, where the difference in ellipse area between OH40 and non-OH40 participants was large, and confidence intervals did not overlap. To further examine this, I conducted a repeated-measures General Linear Model (GLM) analysis, which revealed statistically significant main effects of ROI and strong interactions with OH40 status, supported by large effect sizes. Although these statistical findings are promising, I am mindful of the small group sizes (N = 10, with only 3 OH40 cases), and I interpret the results with appropriate caution. The most important takeaway is not definitive proof of an effect, but rather the identification of key variables (e.g., ellipse area, late-phase sway) that merit investigation in larger, appropriately powered studies. Based on observed effect sizes and variability, future studies would likely require hundreds of participants to robustly test these associations and allow for stratification by other clinical variables such as frailty or medication use.

My analysis also explored asymmetry between the two force plates. While overall sway metrics were broadly comparable between plates, Plate 2 exhibited a trend toward reduced variability in ROI 3. Though not statistically significant, this may hint at asymmetries related to limb dominance or compensatory balance strategies—patterns that have been previously linked to fall risk in older adults. Again, the pilot nature of the study means such observations remain speculative but offer a useful direction for future work.

On a personal and professional level, this Master's programme allowed me to flourish in ways that a taught course may not have. Throughout this research-based MSc, I significantly enhanced my analytical and critical thinking skills. I learned to manage and interpret complex datasets, write custom MATLAB scripts, and perform advanced statistical analyses using SPSS. I also deepened my understanding of statistical theory and its application in clinical research. Perhaps most importantly, I had the opportunity to spend meaningful time in the falls clinic, where I engaged with both patients and clinical staff, gaining practical experience in a dynamic healthcare environment. This not only refined my research approach but also reinforced the real-world relevance of my project.

In sum, while the small sample size limits the generalisability of the results, this pilot research demonstrates the feasibility of incorporating both accelerometry and force plate technologies into clinical AS testing protocols. Each method offers unique insights into postural control—with accelerometry perhaps more sensitive to upper-body motion and FBPs offering precise quantification of COP shifts. Together, they may form a complementary toolkit for detecting subclinical instability in older adults, but this should be explored in future research. Moving forward, this work provides a foundation for larger studies that could help shape future falls prevention strategies and clinical assessment protocols in geriatric care.

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