Understanding the determinants of independent mobility in older adults

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

As aging occurs, safely maintaining an active lifestyle is critical for health and independence. Independent mobility is influenced by one's ability to perform three essential tasks of daily living: transitioning from a seated to standing posture, maintaining upright stance and walking. In spite of the apparent similarities in the predictive utility of these different tasks, there are few studies that have explored the specific relationship between these tasks that define independent mobility within individuals to determine if they reflect unique challenges to control. The thesis focused on two studies to advance understanding of the determinants of independent mobility in older adults. Study 1 explored the association between measures of standing, transitions and walking in 28 older adults. An important element was the assessment using portable low-cost measurement technology (Wii force boards and wearable accelerometers) so that testing could be done in the community. The results of this study revealed the potential importance of sit-to-stand performance as an independent measure of function in older adults. One important outcome was the need for a more detailed measurement of the sit-to-stand task, which is characterized by different phases that have unique control challenges. As a result, Study 2 was designed to evaluate different measurements of the sit-to-stand phases in order to provide a measurement tool that could be used in community and clinical testing. Ground reaction forces were found capable of identifying the different sit-to-stand phases and therefore afford the ability to quantify this behavior using portable technology. Identifying the underlying control mechanisms and relationships between these mechanisms allows clinicians to prescribe targeted and potentially more effective interventions focused on behavior specific control challenges.

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List of Abbreviations

5STS	5x Repeated Sit-to-Stand
6MWT	6 Minute Walk Test
ADL	Activities of Daily Living
AP	Anteroposterior
APA	Anticipatory Postural Adjustment
APR	Automatic Postural Response
BBS	Berg Balance Scale
BOS	Base of Support
CB&M	Community Balance and Mobility Scale
CI	Confidence Interval
CNS	Central Nervous System
COM	Centre-of-Mass
COP	Centre-of-Pressure
EC	Eyes Closed
EO	Eyes Open
ML	Medial-Lateral
mm	Millimeters
ms	Milliseconds
PASE	Physical Activity Scale for the Elderly
RMS	Root Mean Squared
ROM	Range of Motion
s	Seconds

SD	Standard Deviation
SMV	Sum Magnitude Vector
STS	Sit-to-Stand
TUG	Timed Up and GO
WBB	Wii Balance Board

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Chapter 1: Introduction

Background

As aging occurs, safely maintaining an active lifestyle is critical for our health and independence (13). To achieve a safe and active lifestyle, one must avoid falling, while also maintaining or improving their activity level. Decreasing fall risk and subsequent falls is essential because falls are a major source of injury (e.g. hip fractures) and cause of death in the elderly (10). Injuries resulting from falls also contribute to functional decline in the elderly population by causing loss of both mobility and independence (72). Falls are not only costly to an individual's health; they are also a substantial burden to Canadian healthcare, accounting for approximately $2 billion annually in healthcare spending (64). As community dwelling people age, falling becomes more prominent. More than 30% of individuals aged 65 and older fall, while 50% of people over 80 fall (50). In addition, 50% of elderly persons who fall do so repeatedly (73). Meanwhile, approximately 50% of older adults living in retirement homes fall at least once a year (61) and up to 40% fall more than once a year (70). Currently, more than 6 million individuals 65 years and older require long-term care to assist them with activities of daily living (ADLs) (35). These statistics become increasingly concerning with an aging population.

In spite of the age-related challenges to mobility, maximizing physical activity in older adults is vital for their health and wellbeing. Exercise improves overall health and physical fitness, cerebral function, lowers rates of mortality and is recognized as a major factor influencing independence through maintenance of mobility (13). Exercise has also been found to improve blood pressure and lipid profiles, reduce risk of coronary artery disease, while decreasing the risk of osteoporosis (2). However, some older adults may not be able to increase their activity level; therefore, preventing a decline in their current level of physical activity may help reduce their risk of the aforementioned comorbidities associated with a sedentary lifestyle. Therefore a focus on improving mobility can have wide ranging benefits that would be associated with increased activity.

The key element that reduces falls and maximizes activity is our capacity for independent mobility with stability control. Independent mobility can be operationally defined as the ability to stand, transition and walk without the use of a self-propelled assistive device (e.g. motorized wheelchair). Note that for the current definition the use of mechanical aids such as walkers, canes or assist poles is considered an element of independent mobility since the use of these mechanical devices requires the individual to generate the force that propels them during locomotion. It is also important to note that there are various types of transitions such as lying to sitting, bed to chair and standing to sitting. The current work will focus on the transition from a seated to standing position because it is one of the most biomechanically demanding tasks of daily life (59) and a barrier to independent mobility in older adults (1). The ability to perform combinations of these three essential tasks of daily living allows us to maintain our independence. Fundamental to the successful control and execution of each of these tasks is the control of balance, the ability to maintain control of the body without falling. This demand is highlighted by the need to generate the forces required to control the body's vertical position against the force of gravity when standing (82), to move the centre-of-mass (COM) when transitioning (e.g. sit-to-stand) and control the movement of the body (limbs, COM) during the dynamic task of walking. While each of the tasks demands a need for specific levels of strength and control, the challenges are quite unique and likely have very different demands on the nervous system control. Therefore, the objective of this thesis is to determine the relationship between performance on tests of independent mobility, standing balance, transitioning from a seated to standing posture and walking.

Chapter 2: Literature Review

2.1 Factors influencing independent mobility

Independent mobility can be influenced by both extrinsic and intrinsic factors. Extrinsic factors can be viewed as external elements that affect independent mobility. Fall prevention programs typically aim to reduce or remove extrinsic factors that affect independent mobility. Trip hazards such as uneven floors, throw rugs and inadequate support surfaces are removed from the homes of at risk individuals. Additional lighting is added to poorly lit corridors or rooms. Assistive devices, which increase the base of support (BOS) are prescribed to at risk individuals to decrease their risk of falling, making independent mobility safer. However, in some circumstances the mobility aid itself (e.g. 4 wheeled walker) can become a hazard if not prescribed and/or used properly. Medication can also affect independent mobility and balance control; therefore, it is essential that physicians are aware of the side effects of the medications they are prescribing (46).

Intrinsic factors can be defined as internal systems or properties required for independent mobility. Aging is associated with a degradation of various intrinsic factors that can affect independent mobility and balance control. For example, musculoskeletal impairments (e.g. restricted range of motion (ROM)) caused by damage to muscles, tendons, ligaments, joints and nerves are a common cause of physical disablement and make performing various ADLs difficult (33). Age related changes in cognition can also influence independent mobility. Studies have shown that activities once thought to be automatic (e.g. standing, walking) are in fact tightly linked to cognitive functioning (76). A decline in executive function, a set of cognitive skills necessary to plan, monitor, and execute a sequence of goal-directed complex actions (15) has been linked to decreased gait velocity, increased stride variability, increased falls and decreased performance on complex mobility tasks (68). Thus cognitive function, specifically executive function, plays an important role in gait control and independent mobility.

While there are a number of intrinsic factors that may potentially affect our capacity for independent mobility the current work is focused on balance control and strength. It is proposed that balance control (static and dynamic) and strength will have the greatest impact on our ability to stand, transition and walk, ergo our capacity for independent mobility.

2.2 Essential components of intrinsic control

Both static and dynamic balance control rely on sensory contributions to control postural orientation and stability. Postural control can be viewed as how the central nervous system (CNS) controls the body's position and limb orientation, whereas balance or stability control can be defined as the ability of the CNS to maintain equilibrium by keeping or returning the COM over the BOS.

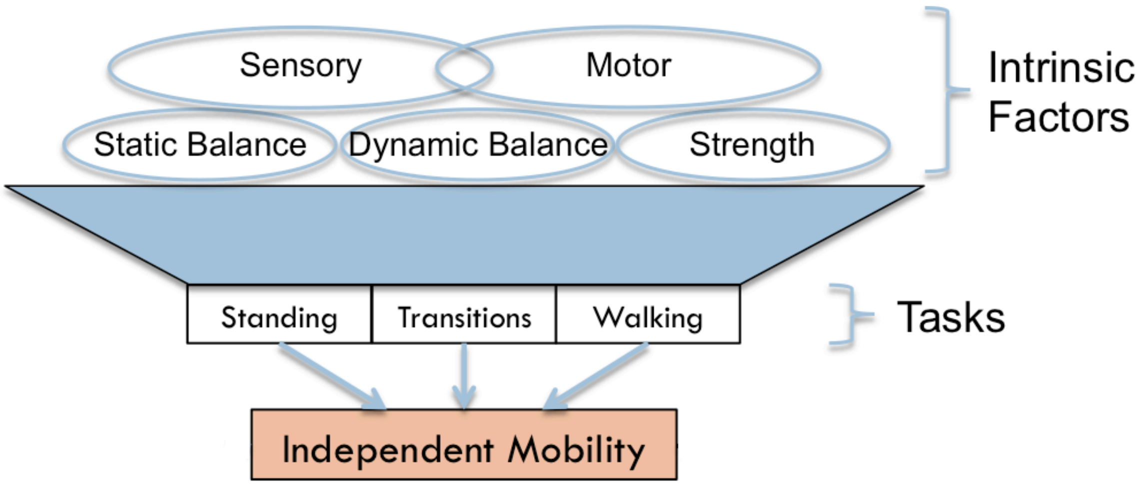


Figure 2-1: Depiction of the key intrinsic factors, balance control and strength, that effect performance on tasks of independent mobility.

2.2.1 Sensory/Motor

Maintaining balance requires integration of sensory information from the visual, vestibular and proprioceptive systems. These systems act through an adaptable hierarchy, where sensory inputs are differentially weighted and specific inputs are prioritized depending on the individual, task and the environment. The inability to reorganize sensory inputs and subsequently modify postural responses as a result of changing environmental conditions contributes to balance deficits (84).

Vision provides the information needed to effectively navigate through the world by contributing to a spatial map of the environment. This map allows for the rapid identification of any potential hazards and the ability to formulate an appropriate strategy to avoid or navigate the hazard (40). Vision also provides information on movement of the body with respect to the world, helping control upright posture (69). Impairments in visual acuity, contrast sensitivity, glare sensitivity, dark adaptation, accommodation and depth perception contribute to vision becoming progressively worse after the age of 50

(21). The inability to judge distances and perceive spatial relationships also affects the ability to identify hazards and move safely throughout the environment (69).

The vestibular system provides information regarding the position and movement of the head and assists in determining whether information is with respect to the surrounding or relative to the body. This information contributes to balance through corrective movements elicited by the vestibulo-ocular and vestibulospinal pathways. The vestibulo-ocular reflex helps maintain visual fixation during head movements by generating compensatory short latency eye rotations. The vestibulospinal reflex stabilizes the head and helps maintain upright stability by triggering muscle activation in the neck, trunk and extremities (26). Additionally, the vestibulocollic and cervicocollic reflexes stabilize the head by initiating neck responses (69). Vestibular dysfunction can cause symptoms from dizziness, orientation problems and postural disequilibrium to the distressing visual symptoms of vertigo and nystagmus during tasks that entail head movement (22). Vestibular hypofunction, typically characterized by postural instability and a broad-based staggering gait pattern with unsteady turns (71), impairs gait and posture in older adults and increases risk of falls (5).

Proprioception, perception of movement and spatial orientation, relies on the collection of sensory information from receptors in the muscles, tendons and joints that provide feedback regarding joint position sense, kinesthesia (the sense of movement) and touch to assist in balance control. Stretch sensitive muscle spindles found within the muscle belly detect changes in muscle length and contraction velocity. Golgi tendon organs located in muscle tendons are sensitive to changes in muscle tension occurring through active contraction or passive stretching. Tactile information provides sensory information regarding force distribution as well as detailed spatial and temporal information about contact pressures (69). Loss of peripheral sensation can result from diseases such as diabetes mellitus, alcohol misuse, vitamin B12 deficiency, chemotherapy and peripheral nerve damage (14). In addition, aging has been shown to decrease vibration perception (77) and touch thresholds (54) through a reduction in the number of Pacinian and Meissner's receptors. This reduces our ability to detect perturbations and moments of postural instability, and in turn affecting our capacity for independent mobility.

One essential function of visual, vestibular and proprioceptive information is to shape the actions of the motor systems. Converting this sensory information into motor outputs is achieved through sensorimotor transformations. Extrinsic information about the environment and intrinsic information about our body are converted into a sequence of muscle actions in order to successfully perform a movement or achieve a goal. Sensorimotor transformations are largely mediated through spinal pathways with very few synapses. Modifiability of these sensorimotor transformations may play an important role in balance control.

2.2.2 Balance control (anticipatory and reactive)

Balance control is composed of anticipatory (feedforward) and reactive (feedback) control. Feedforward control consists of anticipatory postural adjustments (APAs); postural changes that occur prior to a movement when one can anticipate the external perturbation (47). APAs result from an internal cue and act to minimize the impending perturbation and potential reliance on reactive control. Unloading the swing limb prior to foot-off in response to a perturbation is an example of an APA. The COM is pushed towards the stance limb by the centre-of-pressure (COP) initially moving towards the stepping limb to unload the limb prior to foot-off.

Reactive balance control consists of compensatory postural responses. Compensatory postural responses occur in response (feedback) to a stimulus (e.g. perturbation). These compensatory reactions are time dependent and much more rapid than volitional movements, occurring within approximately 100ms of perturbation onset (20). The amplitude of the reaction is scaled to the perturbation magnitude with the pattern of muscle activity being defined by the direction of the perturbation and the external environment (42). Compensatory postural responses are classified in two categories, fixed support and change-in-support strategies. In fixed support strategies, the BOS remains constant and an ankle or hip strategy is employed in response to the perturbation. Change-in-support strategies (e.g. taking a step or reaching for a hand rail) increase the BOS (45) and are typically used when the COM is beyond the BOS, as they are more effective in arresting COM movement (27). Reactive control is critical to correct for errors and limitations in anticipatory (feedforward) control. However, anticipatory and reactive control must act together as both are essential for maintaining stability. This is apparent when examining the role of somatosensory information from the lower limbs in the control of compensatory postural responses. Somatosensory information may be utilized for both sensory feedback and feedforward use, as prior experience such as that to scale the magnitude of the postural response (30).

Remaining independent and active requires anticipatory and reactive balance control in a variety of states, where the context of the movement can range from static standing to dynamic running. This creates a major challenge to our balance control system. Moments of postural instability challenge the balance control system, which in turn leads to reactive control errors such as postural sway during quiet standing and gait variability during ambulation.

2.2.3 Static balance control

The ability to stand independently on two feet is essential for remaining independent and active. Standing, once thought to be automatic, in fact relies on the static balance control system. Static balance consists of anticipatory and reactive control under static conditions, where the BOS remains constant (e.g. quiet standing). Spontaneous postural sway of the COP during quiet standing (static posturography) is the most common measure of static balance control. COP sway can be regarded as a measure of the CNS' ability to generate stabilizing responses in quiet standing, making COP sway a proxy measure of static balance control under static conditions. Therefore, an increase in COP sway (e.g. increased COP RMS sway) can be viewed as a degradation of static balance control resulting in a decreased ability to control balance reactions under static conditions. Inaccuracies in the reactive control response can arise from altered sensory input and/or imprecise motor output. Ankle proprioception has been proposed as the dominating sense in the adaptable hierarchy for standing balance control in the anteroposterior (AP) direction, while mechanoreceptors on the bottom of the feet are atop the hierarchy for medial-lateral (ML) stability control. This is likely a result of the proprioceptive threshold being lower than visual and vestibular for perceiving COP velocity (18). Older adults may have a decreased ability to perceive COP velocity and increased postural sway due to a reduction and/or decreased sensitivity in receptors and/or peripheral neuropathies, both of which are common in the elderly population.

Based on the inverted pendulum model, the COM and COP are directly related, where the difference in the horizontal acceleration of the COM and COP can be considered the error signal that the balance control system is sensing (83). Control of postural sway involves continuous muscle activity primarily from the leg and hip in response to integrated sensory information (19). However, as aging occurs visual, vestibular and proprioceptive degradation ensues hindering the ability to control the COM (13).

Static balance control, specifically the reactive control component, is also critical for maintaining stability following a transition from a seated to standing position. The time to restabilization, time from full hip extension until the magnitude of COP sway falls within the range of natural sway during quiet stance (1), can be viewed as the CNS' ability to stabilize the COM following a vertical transition. The time to restabilization is a measure of static balance control during static conditions following a dynamic movement, which may be important specifically in tasks that induce cerebrovascular challenges (e.g. sit-to-stand). Some individuals may be capable of independent standing with stability control or restabilizing after a transition yet struggle with walking. This occurs because static balance control is much simpler than dynamic balance control since the BOS remains constant. Comparatively, the BOS is constantly changing and moving during dynamic movements (57) making it more difficult for the CNS to maintain or return the COM over the BOS.

2.2.4 Dynamic balance control

The complexity of dynamic stability requires the input of higher brain centers to control movements generally perceived as automatic (e.g. walking). Locomotion often occurs in unfamiliar environments or environments with hazards, requiring gait to be continually modified to adapt to a given environment. In fact when walking, individuals are rarely “statically stable”, the COM is constantly outside the BOS. In order to accelerate the COM in a forward direction, the start of a fall forward must be voluntarily initiated to take the COM ahead of the BOS and only by safe placement of the swing foot is a fall avoided once every step (82). Inaccuracies in anticipatory and reactive balance control can lead to variability in gait patterns (e.g. increased step-to-step variability), indicating the CNS’ ability to generate stabilizing responses in locomotion is compromised, making step-to-step variability during gait a proxy measure of dynamic balance control under dynamic conditions.

Transitioning from a seated to standing position also requires dynamic stability control. The dynamic phase (previously reported in literature as movement phase (1)), time from seat off to full hip extension can be viewed as a marker of dynamic balance control. During this phase, the individual must anticipate where the COM will arrest, ensuring the end location falls within the BOS. Hughes (29) proposed two different standing strategies older adults may employ during the dynamic phase, momentum transfer strategy and stabilization strategy. The momentum transfer strategy primarily relies on flexing the trunk to generate momentum and a lesser contribution from knee musculature to stand, however this requires fine postural control to control the COM from overshooting the BOS in the AP direction. The stabilization strategy relies principally on knee musculature with minimal contribution from horizontal trunk momentum to stand, sacrificing efficiency to gain success and facilitate control by eliminating the dynamically unstable phase (where the BOS and COM do not coincide). Individuals with balance disorders often sacrifice efficiency to gain success and facilitate control in the dynamic phase by employing the stabilization strategy, eliminating the dynamically unstable phase (29). However, sacrificing efficiency requires a greater strength component in order to successfully stand.

2.2.5 Strength

While balance control is critical for independent mobility, strength also impacts the capacity for independent mobility. Strength is maintained at peak levels until the sixth decade, where accelerated loss occurs, with strength decreasing approximately 50% by the eighth decade (69). In order to avoid this potentially devastating loss of muscle mass and strength, older adults must remain physically active. Remaining physically active has been found to maintain higher levels of muscle mass and function (75), which allows individuals to safely perform ADLs and remain independent. Transitioning from a seated to standing posture is particularly difficult because it requires considerable strength to move our COM from the seated to vertical location (1). In fact, rising from a chair is one of the most biomechanically demanding tasks of daily life (59).

Strength also correlates with walking speed (6) (17) as strength is needed to propel the COM forward during locomotion. The propulsion phase of gait typically occurs at the ankle joint just prior to foot off. However, the strength requirement of gait or propulsion is highly dependent on the speed of locomotion, where increased gait speeds have higher strength demands. One exception may be extremely slow gait. In this case, individuals may initiate gait by moving the COM forward and beyond the BOS initiating a fall. Forward progression of the COM only occurs by passive movement of the swing foot in order to regain stability control. This process would be repeated for the entire duration of the bout of walking. Nevertheless, lower limb muscle weakness, specifically reduced ankle dorsiflexion (79), knee extension strength (39) and hip strength (60) correlate with an increased risk of falls. In summary, independent mobility is a complex interaction of the nervous system control of balance and the strength required to execute the necessary action.

2.3 Traditional assessments of independent mobility tasks

Developing a reliable, consistent and valid clinical assessment of fall risk and activity level monitoring is crucial for improving the health and wellbeing of the aging population, while simultaneously easing the economic burden to the healthcare system. These assessments could help clinicians reduce the subsequent frequency of falls and fall-related injuries, diagnose underlying disorders, gauge effectiveness of treatment and track recovery or age-related decline over time.

Clinicians employ a wide variety of traditional assessments to measure fall risk and balance control, the most common clinical assessments include: the Berg Balance Scale (BBS), Community Balance and Mobility Scale (CB&M) and the Timed Up and GO (TUG). Meanwhile, activity level is typically assessed through self-reported questionnaires such as the Physical Activity Scale for the Elderly (PASE). These tests are typically selected because they require little equipment (e.g. stopwatch), are inexpensive and can be easily administered. Traditional tests directly measure task performance (e.g. ability to walk) while indirectly measuring intrinsic factors (e.g. strength). It is essential to assess performance at the “task” level because performance at the “intrinsic factor” level may not translate into functional performance, as it is not known to what extent each of the intrinsic factors contributes to functional performance.

The BBS involves 14 subtests, which measure balance control during standing, sitting, transfers, reaching, leaning over, turning and taking a step. Each task is scored on a five-point scale (0-4) based on the quality of performance or the time required to complete the task, as measured by the clinician. The BBS is scored out of 56, where a score below 45 indicates the individual is impaired, with an increased risk for falls (7). These findings were further supported by Bogle et al (8) who concluded that older adults who scored higher than the cutoff score of 45 were less likely to fall than adults who scored below the cutoff score. They further established the BBS to have a sensitivity (ability to accurately identifying persons who should have a positive test result) of 53%. Other research has shown the BBS to be capable of helping understand functional deficit, discriminate among patients and serve as a screening tool (65). Regardless of the ability of the BBS to assess fall risk, it does not measure balance control during gait and it is well established that a large proportion of falls occur during walking (81). In addition, the BBS also has a ceiling effect, as it is insensitive to differences for individuals who have high levels of balance ability (65). As a result, the CB&M was developed to assess high functioning walkers with impairment.

The CB&M is a 13-item, 6-point scale that measures the performance of more challenging balance and mobility tasks requiring speed, precision, accuracy and sequencing of movement components representative of underlying motor skills necessary for function and participation within the community (31). Scores on the CB&M have been found to significantly correlate with walking velocity and step length, step width and step time (spatiotemporal measures of gait), as well as step length and step time variability (dynamic stability measures of gait) (31). The CB&M is a popular clinical test of balance control because it is a more challenging test of standing and walking, which is better suited to identify good versus poor walkers. It has been found to be reliable, consistent and valid, as determined for brain-injured individuals (28), however how these findings translate to other populations (e.g. elderly individuals in retirement home living) has yet to be determined, limiting its clinical applicability.

Currently the TUG, a test where individuals are observed and timed while they rise from an armchair, walk 3 meters, turn, walk back and sit down again is the most popular clinical test to measure basic functional mobility skills. The TUG's popularity is due to the fact that it has been proven to be a reliable and valid test for quantifying functional mobility, while TUG scores correlate well with log-transformed scores of the BBS and gait speed. TUG scores can also predict the patient's ability to go outside alone safely, fall risk (87% sensitivity) (66) and frailty (55).

Self reported questionnaires are the most common clinical method to assess physical activity in older adults. The PASE is a brief (5 minute) and easily scored survey designed specifically to assess physical activity in persons age 65 and older. The PASE score combines information based on leisure, household and occupational activity. PASE scores significantly correlate with grip strength, static balance, leg strength, resting heart rate, age and sickness impact profile. In addition, the PASE has a test-retest coefficient of 0.75, which exceeds those previously reported for other physical activity surveys (78). The PASE score may significantly correlate with various health outcomes, however its ability to accurately monitor activity level falls well below that of portable technology.

The ability to interpret the results of traditional clinical assessments (e.g. BBS) of independent mobility is limited because they do not measure each of these tasks separately, providing a distinct score for standing, transitions, and walking; an overall composite score is used to assess functional abilities. Without examining each of these tasks separately behavior specific control challenges cannot be identified, decreasing the ability to prescribe more effective individualized therapeutic interventions to decrease risk of falls and increase or maintain activity levels.

2.4 Assessing independent mobility in a research setting

Safe independent mobility is influenced by one's ability to perform three essential tasks to daily living: standing, transitioning from a seated to standing posture and walking. Therefore it should come as no surprise that tests of standing balance (standing for 30 seconds (s)) (43), transitions (5x sit-to-stand) (9), and walking (six-minute walk test) (23) can all identify fall risk in the elderly. An elderly individual may struggle with a specific task or a combination of the tasks. Thus, it is important to understand the source of the problem in order to prescribe individualized therapeutic interventions focusing on behavior specific control challenges to maximize capacity for safe independent mobility.

2.4.1 Standing

Static balance is generally assessed using the "gold standard" force plates, which measure ground reaction forces generated during quiet standing. Force plates are used to assess balance abilities because standing upright relies on static balance control, which is dependent on sensory input to control the COM. Errors in static balance control result in postural sway during independent standing. ML COP root mean squared (RMS) sway has been shown to predict future fall risk with moderate accuracy (closeness of a measurement of quantity to the quantity's true value) (67%) and high sensitivity (80%) (43). In addition, force plates can be used to measure the effect of vision on stabilization as calculated using the Romberg Quotient (30 second eyes COP RMS sway path/30 second eyes closed COP RMS sway path). Force plates can also accurately monitor stance load asymmetries. Thus, performance on tests of independent standing probe static balance control and can be used to reveal the continuum of balance control and identify those who are at an increased risk of falling.

2.4.2 Transitions

Understanding the unique determinants of one's ability to stand versus ability to stabilize is important to develop person-specific therapeutic techniques to enhance sit-to-stand (STS) ability (1). Force plates and kinematic video (e.g. Polhemus Motion Tracking) are used to feature extract events and assess performance during tests of transitions. The STS is a particularly interesting task because it consists of both static and dynamic balance components. Kinematic video is used to determine full hip extension, the point distinguishing the dynamic balance control component of the STS from the static balance control component. The dynamic balance portion of the STS, the dynamic phase (previously reported in literature as the movement phase (1)), is the time from seat off to full hip extension. Restabilization phase, the time from full hip extension until the magnitude of COP sway falls within the range of natural sway during quiet stance (1) requires static balance control. Force plates provide researchers with quantitative information regarding the CNS's ability to control the COM (indirectly) during a transition by measuring ground reaction forces throughout the STS.

Performance during the dynamic phase of the STS may be indicative of the strategy used, dynamic stability capacity and lower limb strength. Performance during the restabilization phase can potentially be used as a marker of age-related changes in dynamic and/or static balance control, reveal underlying balance impairments and potentially screen for cerebrovascular conditions (e.g. orthostatic hypotension).

Transitions are also commonly used to assess vascular control in a laboratory setting because it transpires rapidly over 1-5s. The STS requires active muscle contraction and engagement of “central command” for movement initiation, which leads to an immediate increase in heart rate as the subject contracts his/her muscles to initiate standing (51).

2.4.3 Walking

Traditionally, costly kinematic video (e.g. Vicon) or GAITrite systems are used to capture foot contact and foot off during locomotion in a research laboratory setting. From these variables a variety of spatiotemporal and dynamic stability measures of gait can be calculated (25). Increased stride-to-stride variability in stride length, speed and double support time are associated independently with falling. Of all gait measures, the single best predictor of falling is stride-to-stride variability in velocity. Using this variable, fallers and non-fallers were classified at 71% accuracy (93% sensitivity) (44). Stride time variability also correlates significantly with strength, functional status and mental health (24). Assessing dynamic postural control is essential because dynamic balance during locomotion is complex since it requires the control of the COM within a changing and moving BOS. Identifying common age-related changes that negatively impact locomotion (e.g. increased gait variability) is critical. The adoption of a more conservative basic gait pattern by older adults reduces the magnitude of accelerations experienced by the head and pelvis when walking, which is likely a compensatory strategy to maintain balance in the presence of age-related deficits in physiological function, particularly reduced lower limb strength (48). By identifying detrimental gait characteristics, appropriate therapeutic interventions can be implemented to improve gait and increase activity levels while decreasing risk of falling.

2.5 Assessing independent mobility in a clinical setting using portable technologies

The recent development of inexpensive portable technologies such as Wii Balance Boards (WBB) and wearable accelerometers offer clinicians a more sensitive and accurate tool for assessing balance control and monitoring activity levels than traditional measures (e.g. BBS, 6MWT, PASE). Portable technologies such as WBB and accelerometers offer a unique ability to quantify and assess these tasks with sensitivity approaching that of their "gold standard" comparators outside of a research laboratory setting for a fraction of the cost.

2.5.1 Standing

Recent work by Clark et al. (11) validated WBB against "gold standard" force plates and concluded the WBB is a valid tool for assessing standing balance. Both devices were found to exhibit good to excellent COP path length test-retest within-device (ICC = 0.66-0.94) and between-device (ICC = 0.77-0.89). As a result of its capabilities, the WBB can provide clinicians with a standing balance assessment tool suitable for the clinical setting (11). This allows for quantitative measures of standing balance control (e.g. COP sway during quiet standing, effect of vision on stabilization (Romberg Quotient), stance load asymmetries) to be captured in a clinical setting. However, the significantly more affordable WBB does have limitations. The inability of the WBB to measure shear forces is the most significant limitation of using the WBB instead of their "gold standard" comparator (force plate). Without shear forces COM displacement cannot be estimated, limiting their ability to assess the transition from a seated to standing posture.

2.5.2 Transitions

The time required to perform the 5 times sit-to-stand (5STS) test is a validated measure of lower extremity strength (12), capable of identifying balance impairment (80) and able to predict fall risk (9). While the 5STS may be a good measure of function and strength it may not be revealing of specific balance and orthostatic challenges. Since the 5STS requires a considerable amount of strength it is difficult to determine if the postural sway during the restabilization phase following the final stand is due to fatigue or a challenge to the balance control system. In addition, orthostatic tests such as the active stand typically require an individual to be supine or in a seated position for a prolonged period of time followed by a singular transition to a standing posture (67).

Potentially revealing phases (dynamic phase, restabilization phase) of the STS that may give insight into stability control are not currently evaluated during clinical examination. It is essential to independently measure performance during these phases to identify behavior specific control challenges in order to prescribe more targeted therapeutic interventions. Literature is also lacking studies that independently assess these two distinct phases of the STS. A study by Lindemann (38) used ground reaction forces, specifically the point when vertical force oscillates inside a corridor of 2.5% of body weight, to denote the transition from the dynamic to restabilization phase of the STS and concluded restabilization phase warranted further investigation. Akram and colleagues (1) used kinematic video to identify full hip extension, which was then used to distinguish the dynamic phase from the restabilization phase. However, the use of such "research" grade technology makes it difficult to use for clinical screening and assessments. Fortunately portable technologies (e.g. accelerometers, WBB) are available that may allow the assessment of the various components of the STS (32).

2.5.3 Walking

The development of portable and inexpensive accelerometers has afforded clinicians the ability to assess gait in a clinical setting (3) (23). Accelerometers are small, therefore walking is relatively unrestricted and direct measurement of 3D accelerations eliminates errors with differentiating displacement and velocity data. Accelerometers have also been found to provide insight into motor control during walking as well as age-related differences in dynamic balance control and gait patterns in people with movement disorders (34) (62) (37). Using accelerometers to assess gait offers clinicians an easy and inexpensive way to quantitatively assess dynamic balance control, risk of falling and monitor activity levels.

Walking is a rhythmic pattern with a distinguishable foot-off and foot-contact acceleration profile (56). The accelerometer's internal clock and data-logging format allows precise timing of these measures. Therefore, by attaching an accelerometer to the ankle over the lateral malleolus, accelerometers can capture every step throughout the day, duration and number of walking bouts, as well as the time of day the walking occurred. Not only can each bout of walking be measured, step-to-step variability for each bout can be calculated, providing insight regarding the quality of walking. This unobtrusive monitoring of daily walking can be used to assess levels of physical activity and in turn the risk of comorbidities associated with said lifestyle. Using accelerometers to monitor activity level in patients following neurological injury (e.g. stroke) can expose the characteristics and temporal qualities of ambulation (56), track recovery progress and effectiveness of therapeutic interventions.

2.6 Measuring key intrinsic factors in a clinical setting

By taking advantage of new inexpensive technology it is possible to bring assessments previously restricted to a laboratory setting into the clinic, providing a detailed assessment of the essential tasks that comprise independent mobility (e.g. standing, sit-to-stand, walking). The following describes tasks and measurements that portable technologies can offer clinicians to assess the various intrinsic factors of independent mobility in a clinical setting.

2.6.1 Static balance

Static balance is assessed using a 30 second independent standing balance test, where individuals are instructed to stand quietly with their arms at their sides and look straight ahead for 30 seconds. Individuals perform this test with their eyes open followed by a 30 second trial with their eyes closed. During each of these tests RMS of AP and ML COP sway (mm) is used as proxy measures of static stability control. While, the time to restabilization (s) following a single stand at a preferred pace is as a measure of static balance control.

2.6.2 Dynamic balance

Dynamic balance is measured using the 6-minute walk test (6MWT) and a single STS test. For the 6MWT, individuals are instructed to walk at a maximal pace in order to cover as much ground as possible during the allotted time. During this test step time variability (s) is used as a marker of dynamic balance control. The dynamic phase (s) of a single sit-to-stand is a proxy measure of dynamic stability control.

2.6.3 Strength

Strength is assessed using the 5STS test, a single STS and the 6MWT. In the 5STS, individuals are instructed to rise from a chair five times as fast as possible without using the arms of the chair. The time (s) to complete 5STS is used as a measure of lower body strength (12). The dynamic phase of a single STS can also be used as a proxy measure of strength. Gait velocity is strength dependent (6), therefore gait velocity (m/s) is used a proxy measure of strength during the 6MWT.

It is important to note that when measuring static and dynamic balance control it is difficult to distinguish between the anticipatory and reactive aspects of control. These tests of standing, transitions and walking are chosen to identify poor static or dynamic balance control. Further probing is needed for individuals who perform poorly on these tests to distinguish whether their anticipatory and/or reactive control is compromised

2.7 Rationale

In spite of the apparent similarities in the predictive utility of these different tasks there are few studies that have explored the specific relationship between tasks of independent mobility within individuals to determine if they reflect unique challenges to control. Woollacott (85) acknowledge that different control mechanisms are required during static balance (e.g. standing) than dynamic balance (e.g. walking), yet the relationship between control mechanisms still needs to be determined. Maki (44) found combining measures of static (RMS ML COP sway) and dynamic (stride-to-stride variability in velocity) balance improved the predictive accuracy of classifying fallers from non-fallers, however offered no explanation for this increase. Identifying the underlying control mechanisms and relationships between these mechanisms is important because age related decline in one task or mechanism might not translate to degradation in others. By identifying specific control challenges, clinicians would have the capability of prescribing more targeted rehabilitation therapies.

In order to assess the underlying control mechanisms that may affect independent mobility these behaviors must first assessed at the "task" level. Literature has deemed the WBB a suitable tool for assessing standing balance in a clinical setting (11) and found accelerometers to be capable of assessing gait in a clinical setting (3) yet literature is lacking studies examining the use of portable technologies to assess the transition from a seated to standing posture.

2.8 Research questions and objectives

This thesis consists of two studies designed to address the following research objectives:

Study 1: Relationship between performance on tests of independent mobility in older adults

Objective: To explore the relationship between performance on tests of standing balance, transitioning from a seated to standing posture and walking.

The results of the first study revealed the potential importance of sit-to-stand performance as an independent measure of function in older adults. In addition, it was clear that a simple time measure of the sit-to-stand was not sufficient. So the second study advanced the development of measurements to better assess the elements of the sit-to-stand task.

Study 2: Using portable technology measures to extract events during the sit-to-stand

Objective: To determine if ground reaction forces and body worn accelerometers are capable of identifying a marker to distinguish the two distinct phases, dynamic phase and restabilization phase, of the sit-to-stand.

The work from study 2 would help to determine if it will be possible to use portable technologies (used in study 1) to assess the STS. The ability to use such technology will allow quantitative assessments within clinical and community settings.

Measuring the specific relationship between tasks of independent mobility within individuals is essential to determine if they reflect unique challenges to control. By identifying specific control challenges, clinicians would have the capacity of prescribing more targeted intervention and rehabilitation therapies.

Chapter 3: Study 1: Relationship between performance on tests of independent mobility in older adults

3.1 Introduction

As aging occurs, safely maintaining an active lifestyle is critical for health and independence (13). Injuries resulting from falls contribute to functional decline in the elderly population by causing loss of both mobility and independence (72). Exercise improves overall health, physical fitness, cerebral function, lowers rates of mortality and is recognized as a major factor influencing independence through maintenance of mobility (13). Independent mobility is influenced by one's ability to perform three essential tasks of daily living: standing, transitioning from a seated to standing posture and walking. Therefore it should come as no surprise that tests of standing balance (standing for 30 seconds) (43), transitions (5x sit-to-stand) (9), and walking (six-minute walk test) (23) can all identify fall risk in the elderly. An elderly individual may struggle with a specific task or a combination of the tasks. Thus, it is important to understand the source of the problem in order to prescribe individualized therapeutic interventions focusing on behavior specific control challenges to maximize capacity for safe independent mobility.

Each of the three essential tasks pose specific control challenges to older adults. Transitioning from a seated to standing posture involves a vertical transition or stand, which is particularly difficult because it requires considerable strength and dynamic balance control to move the COM forward and vertically, while maintaining stability within a decreasing BOS (1). The body must also adapt to physiological changes induced by the vertical acceleration associated with rising from a seated position (52). Standing upright relies on static balance control, which is dependent on sensory input to control the

COM. However, as aging occurs visual, vestibular and somatosensory degradation ensues hindering the ability to control the COM (13) (36). It is important to note that static balance during standing is different than dynamic balance during walking (85). Dynamic balance during locomotion is more complex as it requires control of the COM within a changing and moving BOS (57). Additionally, locomotion often occurs in unfamiliar environments or environments with hazards, requiring gait to be continually modified to adapt to a given environment. Identifying common age related changes that negatively impact locomotion (e.g. increased gait variability, decreased gait velocity) is critical. By identifying markers of balance dyscontrol (e.g. increased COP sway, increased gait variability) and decreased strength (e.g. increased time to perform 5STS) specific therapeutic programs can be implemented to maintain or improve capacity for independent mobility and decrease subsequent risk of falling.

The ability to interpret the results of traditional clinical assessments (e.g. Berg Balance Scale, Community Balance & Mobility Scale) of independent mobility is limited because they do not measure each of these tasks separately to provide a distinct score for standing, transitions, and walking; an overall composite score is used to assess functional abilities. Without examining each of these tasks separately behavior specific control challenges cannot be identified, decreasing the ability to prescribe more focused therapeutic interventions to decrease risk of falls and increase or maintain activity levels.

Currently the TUG, a test where individuals are observed and timed while they rise from an armchair, walk 3 meters, turn, walk back and sit down again is the most popular clinical test to measure a combination of all three of these tasks. The TUG's popularity is due to the fact that it has been proven to be a reliable and valid test for quantifying functional mobility, while TUG scores correlate well with log-transformed scores of the BBS and gait speed. TUG scores can also predict the patient's ability to go outside alone safely, fall risk (87% sensitivity) (66) and frailty (55). However, the TUG does not look at each of these tasks independently, therefore cannot identify behavior specific control challenges.

Standing balance, transitioning from a seated to standing posture and walking are also being individually assessed yet summarized using a composite score in a clinical setting. Common clinical tests include the BBS and the CB&M. The BBS involves 14 subtests, which measure balance control during standing, sitting, transfers, reaching, leaning over, turning and taking a step. Each task is scored on a five-point scale (0-4) based on the quality of performance or the time required to complete the task, as measured by the clinician. The BBS is scored out of 56, where a score below 45 indicates the individual is impaired, with an increased risk for falls (7). The CB&M is a 13-item, 6-point scale that measures performance of more challenging balance and mobility tasks that require speed, precision, accuracy and sequencing of movement components representative of underlying motor skills necessary for function and participation within the community (31). Scores on the CB&M have been found to significantly correlate with walking velocity and step length, step width and step time (spatiotemporal measures of gait) as well as step length and step time variability (dynamic stability measures of gait) (31). The 5STS test is capable of predicting fall risk (9), identifying balance impairment (80) and being a measure of lower limb strength (12).

Research settings have individually examined performance during these tasks and revealed that static balance measures during quiet standing such as ML-RMS COP sway

(43) and dynamic balance measures during gait tests (step-to-step variability) (23) are capable of predicting fall risk and recurrent fallers as this variability may be a marker of a general decline in motor and balance control (44).

In spite of the apparent similarities in the predictive utility of these different tasks there are few studies that have explored the specific relationship between these tasks of independent mobility within individuals to determine if they reflect unique control challenges. Woollacott (85) concluded different control mechanisms are required during static balance (e.g. standing) than dynamic balance (e.g. walking), yet further work was need to explore the relationship between mechanisms. Maki (44) found combining measures of static (ML-RMS COP sway) and dynamic (stride-to-stride variability in velocity) balance improved the predictive accuracy of classifying fallers from non-fallers, however did not discuss possible explanation for this increase. Identifying the underlying control mechanisms and relationships between these mechanisms is important because age related decline in one task or mechanism may not translate to decline in others. By identifying specific control challenges, clinicians would have the capability of prescribing more focused interventions. The objective of this study was to determine the relationship between performance on tests of standing balance, transitioning from a seated to standing posture and walking. It was hypothesized:

1) Independent of task, balance control performance measures will be related as reflected by a positive correlation between COP sway (RMS sway), time to restabilization (s) and step time variability (s) during walking.

2) Independent of task, performance on tests of strength will be related as reflected by a negative correlation between 5STS time (s) and dynamic phase (s) with gait velocity (m/s).

3) Performance on tests of balance control will not be related to measures of strength. This will be revealed by an absence of correlation between measures of balance control (COP sway (RMS sway), time to restabilization (s) or step time variability (s)) and measures of strength (5STS time (s), dynamic phase (s) or gait velocity (m/s)).

3.2 Methods

3.2.1 Subjects

Subjects were recruited from the Schlegel Functional Fitness Assessment performed over three different locations. Older adults who were unable to stand independently for 30 seconds (eyes open, eyes closed), unable to stand without the use of the arms of the chair or required an assistive device to perform the six-minute walk test were excluded from the study. In total 28 subjects were recruited who met the criteria. Characteristics of the individuals tested are highlighted in Table 3.1.

Table 3.1: Demographics of residents assessed using the Schlegel Functional Fitness Assessment.

Demographics	
# of residents assessed	28
Sex	33% male; 67% female
Age	Range: 72-96 years Mean: 85.2 ± 5 years

All subjects signed an information and consent form approved by the University Office of Research Ethics.

3.2.2 Experimental approach

Dependent measures of AP and ML COP RMS sway (mm), dynamic phase (s), time to restabilization (s), 5STS duration (s), step time variability (s) and gait velocity (m/s) were recorded using a cross-sectional observational study design.

Table 3.2: Tests of independent mobility and the associated proxy measures of balance control (static, dynamic) and strength.

	Measure of intrinsic factor		
Task	**Static balance**	**Dynamic balance**	**Strength**
30s standing balance	AP & ML COP RMS sway (mm)	-	-
STS	AP & ML time to restabilization (s)	Dynamic phase (s)	Dynamic phase (s)
5STS	-	-	Time (s)
6MWT	-	Step time variability (s)	Gait velocity (m/s)

3.2.3 Procedures

Acceleration: Acceleration of the lower limbs was collected via tri-axial accelerometers (Gulf Coast Data Concepts, Waveland, MS) placed over the right and left lateral malleoli. Accelerometer data were sampled at 40Hz. Lower limb accelerometer data were dual pass filtered using a 2nd order low-pass 20Hz Butterworth to determine gait characteristics (e.g. step time variability) during the 6MWT.

Kinetics: Measures of standing balance were collected using two Nintendo Wii Balance Boards (Nintendo, Kyoto, Japan) sampling at 100HZ and dual pass filtered using a 4th order Butterworth filter with a low-pass cut-off frequency of 10Hz. WBB data were 4th order dual pass filtered with a low-pass cut-off 6Hz Butterworth filter during the sit-to-stand.

Timing and distance: A tablet with a customized Labview program was used for data acquisition and entry. Timing for the 5STS test and distance walked during the 6MWT were recorded using an electronic stopwatch developed using Labview. Distance walked during the 6MWT (used for the calculation of gait velocity) was measured using an executive measuring wheel and entered into the customized Labview program.

3.2.4 Tasks

Residents performed the following four tasks in the order they are listed. Residents performed these tests of independent mobility as part of the Schlegel Functional Fitness Assessment; the order of the tasks was selected to make the assessment as efficient as possible, with the most fatiguing tasks (5STS, 6MWT) at the end. In general, one trial of each task was performed, however if the data were contaminated (e.g. the resident used their hands to stand in the STS) another trial was collected and used for analysis. For tests of standing balance and transitions residents were instructed to place their feet in a comfortable position as long as one foot remained on each Wii Board. This foot position was not normalized across residents and could change between task conditions.

Standing balance: residents were instructed to stand quietly with their arms at their sides while looking straight ahead for 30 seconds. This test was performed in an eyes open condition followed by an eyes closed condition (58).

Sit-to-stand: residents were instructed to stand from a standard chair (height= 45cm, seat depth= 39cm, backrest= 40cm high) with their arms across their chest and remain standing quietly looking straight ahead for 10 seconds.

5 times sit-to-stand: residents were asked to rise from a standard chair five times as fast as possible with their arms across their chest (41). Timing began on the "go" command and ceased when residents sat after the fifth stand-up (63).

Six-minute walk test: residents were instructed to walk at maximal pace, covering as much ground as possible during the allotted time (6 minutes) (74).

3.2.5 Data analysis

A customized Labview program was used to analyze the aforementioned dependent measures. COP measurements for the standing balance tests were parameterized in terms of mean location (43); specifically, RMS displacement relative to the mean location in both the AP and ML axes (mm).

Performance on tests of transitions was quantified using: dynamic phase (s) – time from Fz exceeding a 99% confidence interval (CI) of quiet sitting Fz (seat off) until Fz broke a 99.99% CI based on 100% body weight, a surrogate measure of full hip extension (4); time to restabilization (s) – full hip extension (denoted using Fz) until the magnitude of the COP sway fell within the range of natural sway during quiet stance, calculated from a 95% CI of COP sway (this variable was calculated for both the anteroposterior and mediolateral axes) (4); and 5STS duration (s) - time to stand up and sit down five times as quickly as possible.

Dependent variables on tests of walking included: mean gait velocity (m/s) - dividing the total distance walked by the duration of the walk time (24) and step time variability (s) – variability in time between corresponding successive points of heel contact of the opposite foot.

3.2.6 Statistical analysis

Normality of data was assessed to ensure the data meets the assumption of linearity. No violations of diagnostic plots were found. Correlational analyses, specifically Pearson product moment correlation coefficients were performed between performance on tests of the following standing balance, transitions and walking measures;

1. Correlation analysis between COP sway (RMS sway), time to restabilization (s) following a single stand and step time variability (s).
2. Correlation analysis of 5STS time (s) and dynamic phase (s) during a single stand with gait velocity (m/s).
3. Correlation analysis of COP sway (RMS sway), time to restabilization (s) following a single stand and step time variability (s) with 5STS time (s), dynamic phase (s) following a single stand and gait velocity (m/s).

A level of α =0.05 was used to denote statistical significance.

3.3 Results

Overall, balance measures within a task (e.g. ML COP sway versus AP COP sway) appear to be related, however there appears to be no relationship between measures of balance control compared across the tasks (e.g. quiet standing versus walking). Similarly, measures of strength comparing related tasks are associated (e.g. time to complete 5STS versus dynamic phase duration of STS), however, strength measures across tasks (e.g. 5STS versus walking velocity) were not related. Measures of balance control and strength were only related during the sit-to-stand task. Dependent measures are summarized in summarized in Table 3.3.

Table 3.3: Mean, standard deviation and range of dependent measures on tests of standing, transitions and walking.

Measure	Mean	SD	Range
5STS time (s)	18.3	5.8	10.7-33.2
Dynamic phase (s)	1.4	0.3	1-2.6
ML time to restabilization (s)	7.5	4.5	3.2-21.7
AP time to restabilization (s)	7.7	4.4	2.7-17.4
EO AP-sway (RMS)	5.5	2.3	2.7-13.1
EO ML-sway (RMS)	2.9	1.8	0.8-7.5
EC AP-sway (RMS)	7.2	4.0	3.5-24.2
EC ML-sway (RMS)	3.6	2.6	0.4-12.0
6MWT gait velocity (m/s)	1.1	0.2	0.8-1.4
6MWT step time variability (s)	0.03	0.01	0.01-0.07

With respect to the association between task performance on tests of balance control (Table 3.4), there was a strong statistically significant correlation between performance on test of quiet standing, eyes open (EO) and eyes closed (EC) standing balance in both the ML ($r(28)$, $=0.82$, $p<0.0001$) and AP ($r(28)$, $=0.75$, $p<0.0001$) planes. A strong statistically significant correlation was found between EC AP-RMS sway and EC ML-RMS sway, ($r(28) =0.73$, $p=<0.0001$). A moderate statistically significant correlation was also found between EO AP-RMS sway and EO ML-RMS sway, ($r(28) =0.67$, $p=<0.0001$). Statistically significant correlations were also found between measures of balance control during the sit-to-stand. ML restabilization phase duration was found to have a strong statistically significant correlation with AP restabilization phase duration ($r(26) =0.85$, $p=<0.0001$) and a moderate statistically significant correlation with duration of the dynamic phase ($r(24) =0.60$, $p=0.002$). Similar to ML restabilization phase, AP restabilization duration had a moderate statistically significant correlation with dynamic phase duration ($r(24) =0.50$, $p=0.01$).

Table 3.4: Pearson product moment correlation coefficients for sample variables representing tests of balance control: 30 seconds eyes open standing balance, a single sit-to-stand and 6MWT. Shaded cells denote statistically significant correlations (p<0.05).

		1	2	3	4	5
1	EO ML-sway (RMS)	*				
2	EO AP-sway (RMS)	.67	*			
3	ML restabilization phase (s)	-.07	.11	*		
4	AP restabilization phase (s)	.05	.26	.85	*	
5	Dynamic phase (s)	.10	.25	.60	.50	*
6	Step time variability (s)	.07	.19	-.09	-.03	.02

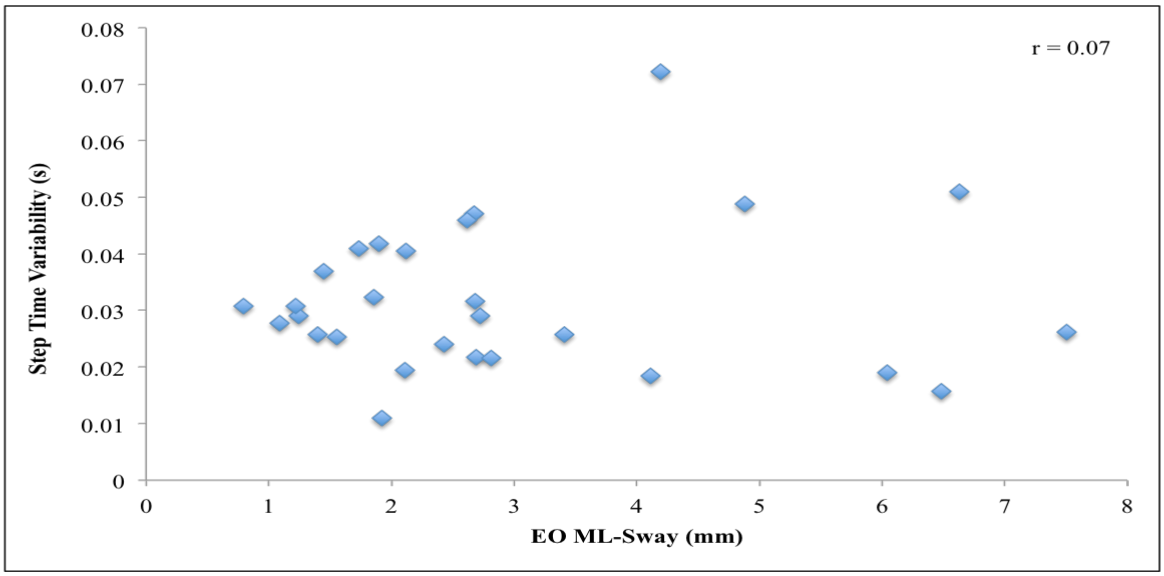


Figure 3-1: No statistically significant correlation (r(28) =0.07, p=0.74) was found between balance measures during walking (step time variability) and standing (EO ML-sway).

Balance measures during quiet standing were not associated with balance measures during transitions and walking. There was no statistically significant association between EO ML-RMS sway and ML restabilization phase (r(26) =-0.07, p=0.74), AP restabilization phase (r(26) =0.05, p=0.80), dynamic phase (r(24) =0.10, p=0.64) or 6MWT step time variability (r(28) =0.07, p=0.74) (Figure 3-1). There was also no statistically significant correlation between performance on tests of EO standing in the AP plane and ML restabilization phase (r(26) =0.11, p=0.59), AP restabilization phase (r(26) =0.26, p=0.20) (Figure 3-2), dynamic phase (r(24) =0.25, p=0.25) or 6MWT step time variability (r(28) =0.19, p=0.33).

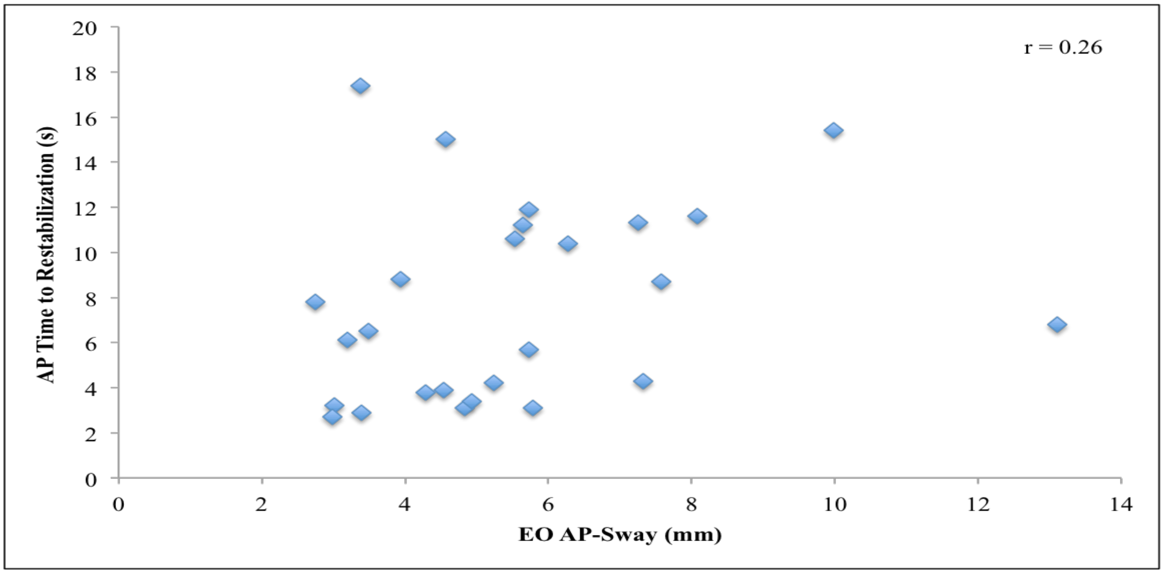


Figure 3-2: No statistically significant relationship (r(26) =0.26, p=0.20) between balance measures during transitions (AP time to restabilization) and standing (EO AP-sway).

Measures of balance control during transitions, ML restabilization (r(26), = -0.09, p=0.66), AP restabilization (r(26), = -0.03, p=0.87) and dynamic phase (r(24), = 0.02, p=0.94) (Figure 3-3) were also independent of balance measures during walking, 6MWT step time variability.

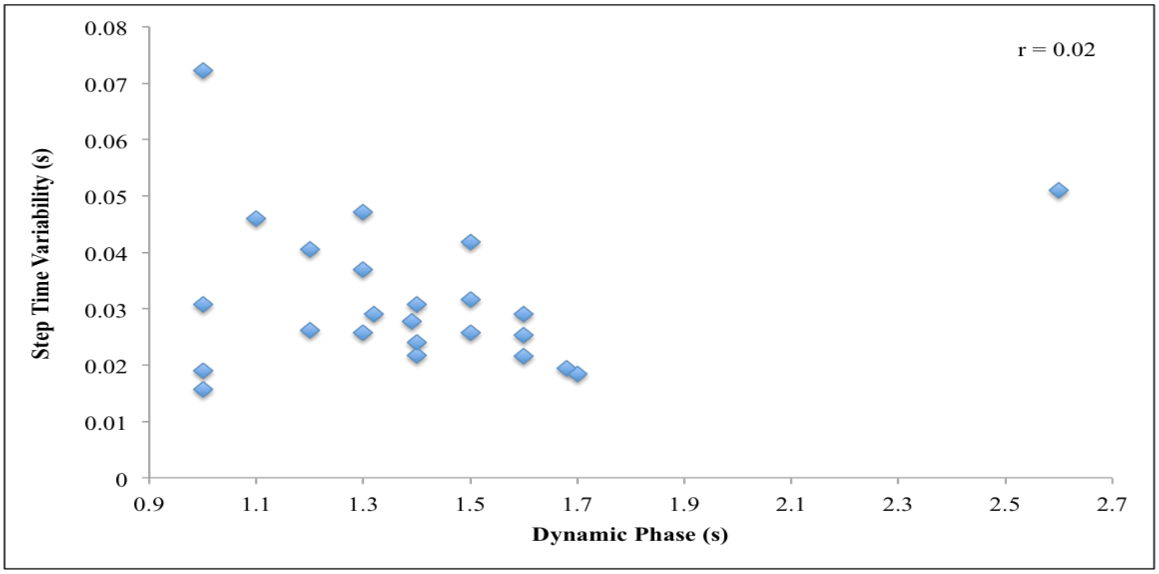


Figure 3-3: No statistically significant correlation (r(24), = 0.02, p=0.94) between dynamic balance measures during walking (step time variability) and transitions (dynamic phase).

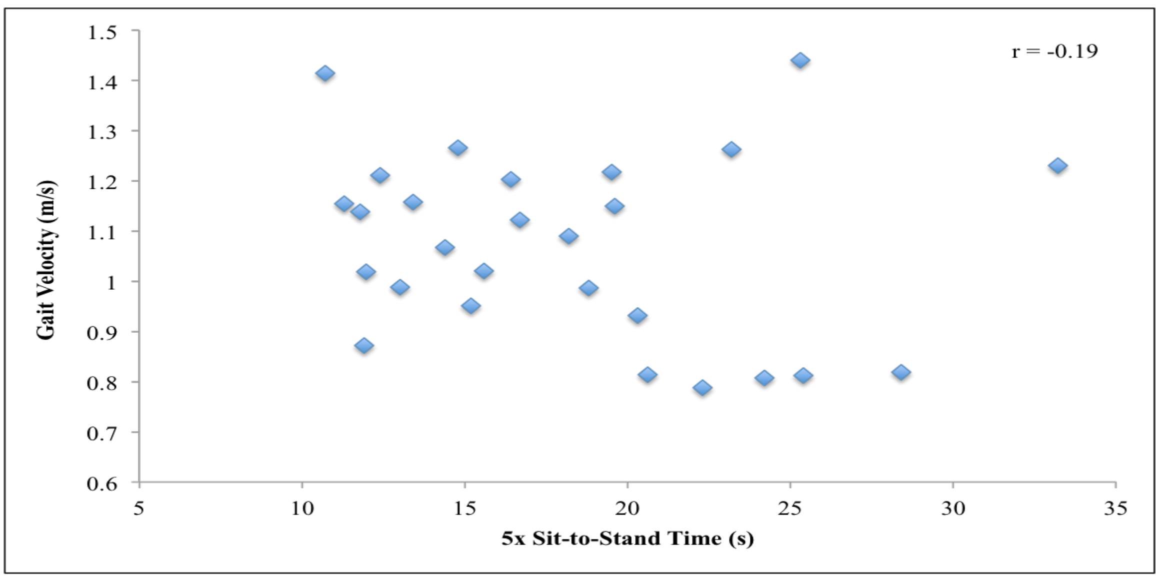


Figure 3-4: Strength measures during walking (gait velocity) were independent of strength measures during transitions (5STS time) (r(27), = -0.19, p=0.35).

Proxy measures of strength during transitions are related, however these strength measures during transitions are independent of a strength measure during walking. A moderate statistically significant correlation was found between 5STS time and dynamic phase (r(24), = 0.50, p=0.01). No other significant correlations were found between measures of strength, 5STS time and 6MWT gait velocity (r(27), = -0.19, p=0.35) or dynamic phase and 6MWT gait velocity (r(23), = 0.25, p=0.25).

The association between measures of balance control and strength are summarized in Table 3.5. Statistically significant correlations between measures of balance control and strength were only found on tests of transitions. AP restabilization was found to have a strong statistically significant correlation with 5STS time (r(26), =0.73, p<0.0001) and moderate significant correlation with dynamic phase duration (r(24), =0.50, p=0.01). Dynamic phase also had a moderate statistically significant correlation with 5STS time (r(24), =0.50, p=0.01). With the exception of transitions, no other measures of balance control and strength were related. Measures of standing balance control, highlighted by EO ML-RMS sway, were independent of strength measures. No statistically significant correlation was found between EO ML-RMS sway and dynamic phase duration (r(24) =0.10, p=0.64), 5STS time (r(28) =0.10, p=0.62) or 6MWT gait velocity (r(27) =-0.01, p=0.97). Measures of balance control during transitions, AP restabilization phase (r(25) =0.04, p=0.85) and dynamic phase (r(23) =0.25, p=0.25), were not associated with 6MWT gait velocity. 6MWT step time variability, an index of dynamic balance control during walking, was also independent of strength measures during transitions, dynamic phase (r(24) =0.02, p=0.94) and 5STS time (r(28) =0.14, p=0.47), as well as strength measures during walking, 6MWT gait velocity (r(27) =-0.08, p=0.68).

Table 3.5: Pearson product moment correlation coefficients highlighting the relationship between performance on tests of balance control and strength: standing balance, transitions and walking. Shaded cells denote statistically significant correlations ($p<0.05$).

		1	2	3	4	5	6	7	8	9
1	5STS time (s)	*								
2	Dynamic phase (s)	.50	*							
3	ML time to restabilization (s)	.58	.60	*						
4	AP time to restabilization (s)	.73	.50	.85	*					
5	EO AP-sway (RMS)	.26	.25	.11	.26	*				
6	EO ML-sway (RMS)	.10	.10	-.07	.05	.67	*			
7	EC AP-sway (RMS)	.28	.07	.18	.27	.75	.57	*		
8	EC ML-sway (RMS)	-.03	-.09	-.04	.02	.58	.82	.73	*	
9	6MWT gait velocity (m/s)	-.19	.25	.17	.04	.23	-.01	.11	.05	*
10	6MWT step time variability (s)	.14	.02	-.09	-.03	.19	.07	.08	.003	-.08

3.4 Discussion

Measures of balance control and strength in older adults were assessed at the task level using quantitative measurement techniques. The findings revealed the independence of standing, transition and walking tasks, yet highlighted the link of measures within specific tasks. The independence of balance control measures across tests of standing (EO, EC), transitions (dynamic phase, restabilization phase) and walking (step time variability) did not support the hypothesis that independent of task, measures of balance control would be related. The independence of strength measures on tests of transitions (5STS, sit-to-stand) and walking (6MWT) did not support the hypothesis that strength performance measures would be related. With the exception of the association between strength and balance measures within tests of transitions, the findings support the hypothesis that balance control and strength performance measures would not be related.

The link of balance measures within a task may be indicative of task-specific control mechanisms. With respect to the relationship of balance control measures during quiet standing, the statistically significant correlation between tasks of standing balance (EO, EC) in both the ML and AP planes is likely related to the reliance on the static balance control system. Degradation of static balance control occurs through loss of sensory elements, ability to integrate information and issue motor commands, and a loss of musculoskeletal function (36), resulting in an increase in postural sway in both the ML and AP axes. The statistically significant correlations between balance measures (dynamic phase and restabilization phase) during transitions may be associated with different possible mechanisms, specifically: 1) the natural link between stability control in the two sequential phases, 2) the common dependence of reactive control for the two phases or 3) the common influence of movement strategy (confidence/slowing). With respect to the first option, the dynamic phase of the STS requires dynamic stability to control the AP overshoot, while the restabilization phase requires static stability control to restabilize the COM following the transition (1). Consequently, a decreased ability to restabilize the COM following the transition resulting in an increased time to restabilization may be associated with an inability to control the COM AP overshoot during the dynamic phase, as seen by a longer dynamic phase. An alternative explanation is that both phases depend on reactive balance control as a key aspect; an altered ability to react rapidly or with the appropriate responses (as noted in the previous section) may reduce stability in both phases. Finally, it is possible that the changes in strategy (e.g. moving rapidly or slowing) will alter the challenges to stability control (29) and that variation in movement strategy across individuals leads to this association.

The independence of balance control measures across tests of standing, transitions, and walking does not support the hypothesis that measures of balance control should be related. The independence of balance measures during standing and walking may reflect unique stability control challenges during these tasks. Static stability governs balance control during quiet standing where the COM is maintained within a constant BOS. However, during gait, dynamic stability is required to maintain balance where the COM is constantly outside of an ever-changing BOS. These findings support previous literature (85) that concluded static balance during standing is different than dynamic balance during walking; based on Winter's work (82), which found ankle muscle activity sufficient to maintain static balance during quiet standing yet insufficient to maintain balance of the whole body during walking.

The independence of static balance measures, COP sway and restabilization phase, during standing and transitions, may be due to several possible reasons related to task differences in sensory or vascular challenges. With respect to the sensory challenges, standing balance relies on plantar cutaneous mechanoreceptors and ankle proprioception for stability (53) (54). Conversely, transitioning from a seated to standing posture increases COM motion and in turn challenges sensory contributions, specifically visual contrast sensitivity, lower limb proprioception and tactile sensitivity have all been found to play an important role in sit-to-stand performance (41). With regards to the vascular challenge, in quiet standing the body has had time to adapt to any vascular changes that may have occurred; therefore, the postural sway exhibited is a result of an inability of the CNS to generate stabilizing responses during quiet standing. Conversely, transitioning from a seated to standing posture occurs prior to the restabilization phase. This vertical transition creates a cerebrovascular challenge through a gravitationally mediated redistribution of the blood volume, which may affect the time to restabilization. Therefore, the restabilization phase may not solely be a measure of static balance control, but a complex measure of neuromuscular and vascular control explaining the dissociation between the two variables.

The independence of COP sway and the dynamic phase may be a result of different control challenges or strength demands. The BOS remains constant with relatively little COM motion during quiet standing, relying on ankle musculature to keep the COM within the BOS (82). However it is unlikely ankle musculature alone is sufficient to address the sit-to-stand task-related challenges: 1) move the COM forward, 2) vertically raise the COM from the seated to standing location and 3) make the transition from a large and stable BOS in sitting to a considerably smaller BOS in standing (1). Quiet standing has a relatively low strength demand, to maintain the body's vertical position against the force of gravity when standing (82) and generate stabilizing responses. Transitioning from a seated to standing posture requires considerable strength from the knee extensors to move our COM from the seated to vertical location (1).

The lack of association between balance measures during transitions and walking can be attributed to the distinct biomechanical challenges. Although the dynamic phase and step time variability may be considered proxy measures of dynamic balance control, the dynamic phase during sit-to-stand has an increased strength component (compared to walking) required to move the COM vertically during the transition, making it importantly dependent on strength as well as balance control (1). Walking consists of four basic subtasks: 1) generation of continuous movement to progress towards a goal, 2) maintenance of stability during progression, 3) adaptability to meet any changes in the environment, and 4) initiation and termination of locomotor movements. The first task requires strength to propel the body; however, the second and third tasks require a complex interaction of locomotor and balance abilities to maintain upright posture and properly modify the ongoing locomotor behavior to suit the environment (49). In addition, it is difficult to determine the extent of strength and balance control contributions to dynamic phase performance, possibly explaining the dissociation between dynamic phase and step time variability.

The independence of strength measures across tasks of independent mobility does not support the hypothesis that measures of strength should be related. The common importance of strength on tests of transitions was revealed by the moderate statistically significant correlation between 5STS time and dynamic phase. The increased knee extensor (quadriceps) demands required to move the COM from the seated to vertical location in transitions (41) may explain why the residents who stood faster in the single STS, shorter dynamic phase, were able to complete the 5STS faster. However using the composite measure of 5STS performance (time) makes it is difficult to detect the different phases of the 5STS and to determine whether strength or balance control was the limiting factor when performing the test. Gait velocity also requires strength to propel the COM forward during locomotion with the propulsion phase of gait typically occurring at the ankle joint via ankle plantar flexors (soleus, gastrocnemius) just prior to foot off (82). Thus, the independence of strength measures on tests of transitions and walking may be related to the different muscle groups being challenged in the specific tasks. As was the case for the composite measure of the 5STS, using the 5STS time, gait velocity during the 6-minute walk is not likely a simple measure of strength. Instead factors such as health status (e.g. aerobic fitness, cognitive status, angina, arthritis) height, age, gender and medication can all affect gait velocity (16), making it difficult to interpret the limiting factor of gait velocity during the 6MWT.

The independence of balance and strength measures across tests of standing, transitions and walking, support the hypothesis that measures of balance control and strength should not be related. While there were associations within tasks that had common balance or strength demands, as noted above, there was little association between the balance and strength tasks. The only measure of balance control that linked to strength performance was the association between a single STS and 5STS time, which may be indicative of the important challenge to AP stabilization during the standing phase of STS tasks (1). This also speaks to the limitation of using the composite measure of 5STS performance (time). There was no association of gait velocity and balance (quiet standing, STS, gait variability), while it supports the hypotheses, one may argue that the gait velocity should also be impacted by challenges to balance control. Individuals with balance control challenges often increase double-support time, decreasing time spent balancing on one leg, and/or reduce stride length, minimizing the forward excursion of the COM beyond the BOS of the stance foot, both of which increase stability. However, these stabilizing gait adaptations result in slower gait speeds. Decreased gait velocity also benefits those with balance control challenges during walking by allowing more time to react to obstacles and potential perturbations (44). Additionally, the tasks rely on different intrinsic factors, standing balance relies primarily on static balance control, measured via COP sway, transitions rely on a complex interaction of static balance and vascular control measured through the restabilization phase and step time variability is a measure of dynamic balance control, whereas gait velocity is strength dependent (6).

The independence of strength measures during transitions and balance measures during standing and walking is also likely attributed to the reliance on different intrinsic factors. As previously stated, performance on test of transitions is strength dependent while walking requires a complex interaction of locomotor and balance abilities to maintain an upright posture (49), while quiet standing relies on ankle musculature to generate stabilizing responses.

It is important to note the lack of association between tasks of independent mobility may be due inadequate power. Also, since only residents who were able to stand without the use of the arms of the chair and who did not require a walking aid were recruited, there may be a lack of variability in the sample population to determine if these tasks are truly independent. In this regard, the average 5STS time (18.3s) and walking velocity (1.1m/s) portrays a high functioning group of older adults in spite of their average age (mean: 85 years). Regardless of the time required to complete the 5STS, the ability to perform the task without the use of the arms of the chair is indicative of a decreased risk of falling, ADL disability and independent ADL disability (86). However, Buatois (9) concluded individuals who took longer than 15s to complete the 5STS (without use of their hands) had a 74% greater risk of recurrent falls than those who took less time. With respect to the average gait velocity, Hausdorff (23) found no significant difference in gait velocity between fallers (1.13±0.42m/s) and non-fallers (1.13±0.39m/s) in community-dwelling older adults. Therefore, although the population tested may have been high functioning for their age they still may be experiencing balance and/or strength challenges and at an increased risk for falls. Using the arms of the chair changes the force profile during standing, making it difficult to distinguish the different phases of the sit-to-stand (38). Therefore, the ability of ground reaction forces to measure sit-to-stand events in an impaired population who require the arms of the chair to stand may be limited. In regards to the correlation of measures across tasks in a more impaired population, there may be an association between measures of balance control across tasks. It is likely individuals with balance control challenges during “relatively simple” quiet standing (BOS constant, control COM) are also experiencing control challenges during “complex” walking, as they must now control both the COM and BOS. Consequently in a more impaired population, an increased COP sway during quiet standing may be associated with an increase in step time variability during walking.

Each of the three essential tasks of independent mobility pose specific control challenges to older adults. Thus, it is important to understand the source of the problem in order to prescribe individualized therapeutic interventions focusing on behavior specific control challenges to maximize capacity for safe independent mobility. The findings of this study revealed the independence of standing, transition and walking tasks. Furthermore, this study found measures of these tasks that may have exposed the independent components (e.g. strength, balance) and revealed the importance of measures that highlight the unique control challenges between standing, transitions and walking.

3.5 Conclusions

This work revealed the independence of standing, transitions and walking tasks, supporting the need for behavior specific performance scores to replace less informative composite scores to better inform clinical intervention and assess capacity for independent mobility. The independence of balance measures across tasks and strength measures across tasks revealed the complexity of each task, indicating different control mechanisms may govern task performance. The within task relationships between measures may advocate for only a single test of each task (standing, transitions, walking) being required to assess capacity for independent mobility.

Chapter 4: Study 2: Using portable technology measures to extract events during the sit-to-stand

4.1 Introduction

The ability to transition from a seated position to a standing posture is essential for independent mobility. While the task itself seems "simple" the underlying neuromuscular and vascular control is quite remarkable. In fact rising from a chair is one of the most biomechanically demanding functional tasks (59). Older adults may find this task difficult because it requires strength and balance control to move the COM forward and vertically, while maintaining stability within a decreasing BOS (1). Additionally, the body must also adapt to vascular challenges induced by the associated vertical acceleration of the body (52).

The 5x sit-to-stand test has become a clinical staple for assessing an individual's ability to transition from a seated to standing posture. In addition to the 5STS being a functional measure of one's ability to transition, this inexpensive and easy to administer test has been proven to be a validated clinical tool shown to measure lower extremity strength (12), identify balance impairment (80) and predict fall risk (9). However the 5STS test does not provide details about specific challenges to balance control, strength or cerebral autoregulation.

Focusing on a single sit-to-stand and then stand still (STS) task can provide more information about the different phases of movement. The STS consists of both static and dynamic balance components making it a particularly interesting task (1). It is essential to independently measure performance during these distinct phases to identify behavior specific control challenges in order to prescribe more effective therapeutic interventions to improve ability to transition. For example, reveal individuals who have difficulty with strength versus those with balance control or to reveal the potential underlying challenges to balance control such as orthostatic hypotension. Two potentially revealing phases, dynamic phase and restabilization phase, that may give insight into balance, strength and cerebrovascular determinants are not routinely evaluated during clinical examination largely due to the absence of clinical measurement tools. The dynamic balance portion of the STS, the dynamic phase, is the time from seat off until the hip joint has reached full extension. The restabilization phase, time from full hip extension until the magnitude of COP sway falls within the range of natural sway during quiet stance (1), requires static balance control. Additionally, it has been proposed that by denoting the onset of the restabilization phase, stability control at the end of the STS may be used as a prognostic marker for subclinical cerebrovascular and neurological control challenges (1).

Literature is lacking studies that independently assess these two distinct phases of the STS. A study by Lindemann (38) used ground reaction forces to denote the transition from the dynamic to restabilization phase of the STS and concluded the restabilization phase warranted further investigation. Akram and colleagues (1) used kinematic video to identify full hip extension, which was then used to distinguish the dynamic phase from the restabilization phase. However, the use of such "research" grade technology makes it difficult to use for clinical screening and assessments. Fortunately portable technologies (e.g. accelerometers, Wii Balance Boards) are available that may allow the assessment of the various components of the STS (32). The focus of the current work is to develop and evaluate the use of low-cost portable technologies as tools to assess performance during transitions such as the sit-to-stand. The measurement tools that have been used to assess different aspects of sit-to-stand control are wearable accelerometers (32) and force plates

(38). The current work compares the use of these two classes of instruments to reveal the specific phases (dynamic, static) during STS behavior. Older adults stand using a variety of strategies (29) and at different speeds (41) making it essential to validate the ability of portables technologies to denote the transition from the dynamic to restabilization phase across a range of standing conditions and speeds. The first objective of this study was to assess the ability of tri-axial accelerometers and/or Wii Balance Boards to distinguish the two distinct phases, dynamic phase and restabilization phase, of the STS across a variety of speeds and standing strategies. The second objective was to determine which portable technology measure should be used in a clinical setting to identify the transition point from the end of the dynamic phase to the start of the restabilization phase. The gold standard measurement for this point is based on kinematic measurement of full hip extension, which is impractical in the conventional clinical or community setting. For the purpose of this study a surrogate of full hip extension, the point at which the trunk stops moving vertically, was used to identify the dynamic from the restabilization phase. This point will be referred to as full hip extension to remain consistent with previous literature (1). It is specifically important to distinguish these two phases in order to better determine the strength (dynamic phase) and balance (restabilization phase) determinants of STS performance. Overall it is anticipated that force plate and accelerometry will be able to provide a reliable indication of the transition between the dynamic and static phases of standing reflected by the gold-standard kinematic measure of full hip extension. More specifically it was hypothesized that when attempting to measure the transition point from the end of the dynamic phase to the start of the restabilization phase:

1. Independent of speed and standing condition, there will be no significant difference in the time to zero vertical acceleration (accelerometry) or the time at which vertical force reaches full body weight (portable force plate) as compared to the time to full hip extension (kinematic gold-standard of end of dynamic phase).
2. Independent of speed and standing condition, the estimate of timing, compared to time of full hip extension, will be more precise (smaller mean difference) and less variable (lower standard deviation (SD)), when acceleration is measured from the head as opposed to the sternum and vertical force.

Using measures that could be assessed using low-cost portable technologies to feature extract events during the STS would allow clinicians the opportunity to quantitatively assess behavior specific control challenges during the transition from a seated to standing posture, a measurement that was previously restricted to a laboratory setting. Furthermore, stability control at the end of the STS may be used as a prognostic marker for subclinical cerebrovascular and neurological control challenges (1).

4.2 Methods

4.2.1 Subjects

Ten healthy adults (4 females, 6 males, average age = 27.8, range: 23-31 years) were recruited. Due to the possible influence of movement speed and body position on the characteristics of STS performance the current study assessed the ability of portable technologies to feature extract sit-to-stand events across specific STS conditions in healthy young adults. All subjects signed an information and consent form approved by the university Office of Research Ethics.

4.2.2 Experimental approach

The gold standard measure of the end of the dynamic phases is denoted by the occurrence of full hip extension. However, a surrogate measure of the time of full hip extension, peak trunk verticality, was calculated using kinematic motion tracking. The use of accelerometry (head, sternum) and vertical force was used to estimate the surrogate of time of full hip extension using a cross-sectional test-retest observational study design.

A secondary analysis was also performed to examine the ability of portable technologies to denote seat off (start of the dynamic phase). The dependent measure of seat off (seconds) was calculated using kinematic motion tracking (head, sternum) acceleration and vertical force using the same cross-sectional test-retest observational study design. See Table 4.2 for a detailed description of dependent variables.

4.2.3 Procedures

Sit-to-stand: Subjects were instructed to stand from a chair (height= 45cm, seat depth= 39cm, backrest= 40cm high) with their arms across their chest and remain standing quietly looking straight ahead for 10 seconds on the experimenter's "go" signal. Participants performed 10 trials of each condition with the order of the conditions being randomized.

Table 4.1: List of sit-to-stand conditions and instructions for trials performed.

STS condition	Starting position and task instructions
Preferred	Sit normally in the chair and stand at a self-selected *preferred* speed – preferred foot position
Slow	Sit normally in the chair and stand at a self-selected *slow* speed – preferred foot position
Backrest	Sit in the chair with your back against the backrest and stand at a self-selected *preferred* speed – preferred foot position
Edge	Sit with your ischial tuberosities on the edge of the chair and stand at a self-selected *preferred* speed – preferred foot position
Narrow	Sit normally in the chair and stand at a self-selected *preferred* speed – foot width narrow

Foot placement for the conditions was normalized within individuals by having subjects place their feet in a position of their choice prior to the initial practice trials. This position was marked on the ground and subjects used this to ensure constant foot position across trials. In one task condition (*narrow*), the foot position was challenged. For these trials all subjects stood with their feet two inches apart. With respect to the speed of standing, individuals were instructed to stand at self-selected speeds. Therefore, the absolute speed at which individuals stood within the various task conditions (e.g. *slow, preferred)* may have varied between subjects.

Kinematic displacement: Vertical trunk displacement, used as a surrogate measure of full hip extension, was recorded using a Polhemus Liberty motion capture system (Colchester, VT) sampling at 120Hz with the source cube placed 3 feet in front of the subject. Trunk location was captured via a Polhemus marker secured to a belt at the level of the navel. Kinematic data were filtered using a dual pass 4^{th} order Butterworth filter with a 5Hz low-pass cut-off frequency to smooth digitizing noise (29).

Acceleration: Acceleration of the head was collected via a tri-axial MTw accelerometer (Xsens, Enschede, Netherlands) placed superior to the left ear. Sternum acceleration was captured using a tri-axial MTw accelerometer secured to a vest inferior to the suprasternal notch. Accelerometer data were sampled at 60Hz. Head and sternum accelerometer data were dual pass filtered using a 2^{nd} order low-pass 6Hz Butterworth filter (32).

Kinetics: Force plate data were collected using AMTI force plates (Watertown, MA) sampling at 240Hz. Force plate data were filtered using a 4^{th} order dual pass Butterworth filter with a 6Hz low-pass cut-off frequency. Note that in the current study, as it was conducted in the lab setting, research grade force plates were used. These results are generalizable to testing (such as conducted in study 1) that use portable technology such as the Wii force boards.

4.2.4 Data analysis

A 100ms 3.3V square wave pulse was sent from a customized Labview collection program to the Polhemus motion capture system and accelerometers to temporally synchronize the data. A customized Labview program was used to analyze the aforementioned dependent variables. See Table 4.2 for a detailed description of dependent variable calculations. The measure of seat off was denoted using kinematics (point where vertical trunk location broke a CI greater than 99.99% of the original vertical location), vertical sternum acceleration (local minimum prior to peak vertical acceleration), a sum magnitude vector (SMV) of sternum acceleration (point where sternum acceleration SMV broke a 99% CI of quiet standing) and Fz (peak Fz and point where Fz exceeded a positive 99% CI of quiet sitting Fz). A SMV measures the accelerations in all axes (AP, ML, vertical), thereby quantifying the overall acceleration of a segment. A sternum acceleration SMV was used in addition to the vertical channel acceleration as the orientation of the accelerometer changes throughout the stand, consequently not measuring true vertical acceleration. The transition from the end of the dynamic phase to the start of the restabilization phase was identified using kinematics (point where vertical trunk location broke a 95% CI of final vertical location), vertical acceleration of the head and sternum (point where vertical channel acceleration broke a 95% CI of quiet standing acceleration), a SMV of head and sternum acceleration (point where acceleration SMV broke a 99% CI of quiet standing) and Fz (point where Fz returned to a CI greater than 99.99% of quiet standing Fz). The various confidence interval widths were chosen to accommodate for the difference in the variability of baseline measures (e.g. vertical force versus vertical channel acceleration) to ensure the absolute value of the CI relatively consistent across measurement techniques.

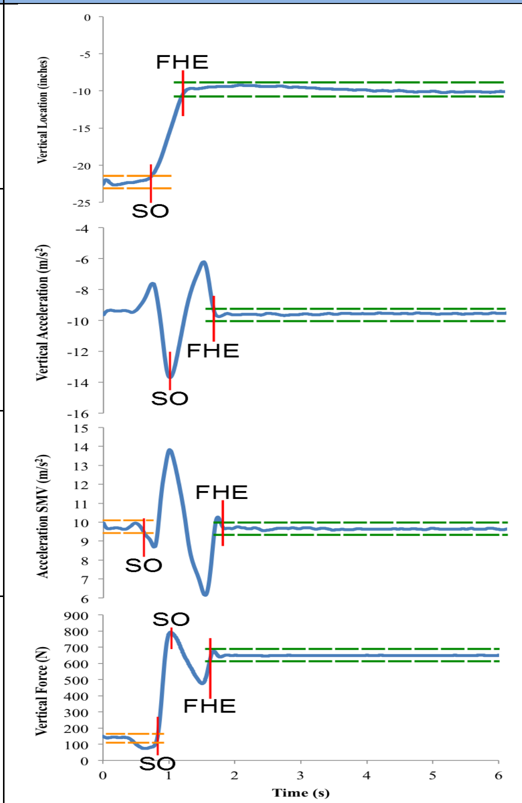
4.2.5 Statistical analysis

Normality of data was analyzed to ensure the data meets the assumption of linearity. No violations of diagnostic plots were found. Difference scores between time to full hip extension measured from vertical kinematic displacement, acceleration and vertical force were calculated for each of the standing conditions (*slow, preferred, narrow, backrest, edge*) and each dependent measure of full hip extension.

A two-way repeated measures ANOVA (α= 0.05) was used to determine if estimates of the end of dynamic phase measured using accelerometers and vertical force were significantly different than the gold standard (full hip extension denoted via vertical kinematic displacement).

A two-way repeated measures ANOVA (α= 0.05) was also used to determine if the estimate of timing, compared to kinematic identification of seat off, was more accurate (smaller mean difference) and less variable (lower standard deviation), when acceleration was measured from the head as opposed to the sternum and vertical force.

Table 4.2: Detailed description of study two phase points that indicate the start and end of the dynamic phase (seat off and full hip extension).

Marker	Seat-off	Full hip extension	
Kinematics Polhemus trunk location	Point where the trunk started to move vertically - Vertical trunk location broke a 5 SD CI of original vertical location	Point where the trunk was no longer moving vertically - Vertical trunk location broke a 2 SD CI of final vertical location	
Accelerometer (vertical acceleration channel of the head & sternum)	Point where the head/sternum accelerometer started to accelerate in the vertical direction -Local minimum prior to peak vertical acceleration	Point where there was 0 vertical head/sternal acceleration following seat off - Vertical acceleration broke a 3 SD CI of quiet standing acceleration	
Accelerometer (sum magnitude vector of head & sternum)	Point where the sternum started to accelerate - Sternum acceleration SMV broke a 3 SD CI of quiet sitting	Point where there was 0 head/sternal acceleration following seat off - Acceleration SMV broke a 3 SD CI of quiet standing	
Force plate (Fz)	Peak Fz Point where Fz broke quiet sitting threshold -Fz exceeded a positive 3 SD CI of quiet sitting Fz	Point where Fz returned to a quiet standing value following peak Fz -Fz returned to a 6 SDCI of quiet standing Fz	

4.3 Results

Two-way repeated measures ANOVA revealed a significant effect of condition ($F[4,36] = 37.93$, $p < 0.0001$) and measure ($F[5,45] = 20.14$, $p < 0.0001$) on time to peak verticality of the trunk (surrogate measure of full hip extension) with significant interaction effects ($F[20,180] = 6.85$, $p < 0.0001$). A two-way repeated measures ANOVA also revealed a significant effect of condition $F[4,36] = 18.31$, $p < 0.0001$) and measure ($F[4,36] = 19.06$, $p < 0.0001$) on time to seat off with significant interaction effects ($F[16,144] = 4.51$, $p < 0.0001$).

4.3.1 Identifying the transition from the dynamic to restabilization phase

As expected, there was an effect of sit-to-stand task condition ($F[4,36] = 37.93$, $p < 0.001$) on time to full hip extension. The *slow* condition had a statistically significant higher ($p < 0.0001$) time to full hip extension (400(SD=118)ms) than the *preferred* (262(SD=76)ms), *narrow* (246(SD=61)ms), *backrest* (268(SD=81)ms) *and edge* (221(SD=57)ms) conditions. The *backrest* condition also had a significantly higher ($p = 0.04$) time to full hip extension than the *edge* condition. There was also an effect of measure ($F[5,45] = 20.14$, $p < 0.0001$) on time to full hip extension. Portable technology measures (vertical channel acceleration, sum magnitude vector, vertical force) had a significantly higher ($p < 0.0001$) time to peak trunk verticality (proxy measure of full hip extension) than the "gold standard" kinematic displacement.

Head accelerometer: Overall the timing of the estimate of full extension from a head worn accelerometer (Figure 4-1) was approximately 430ms after the end of the dynamic phase determined from kinematics and was similar in all task conditions.

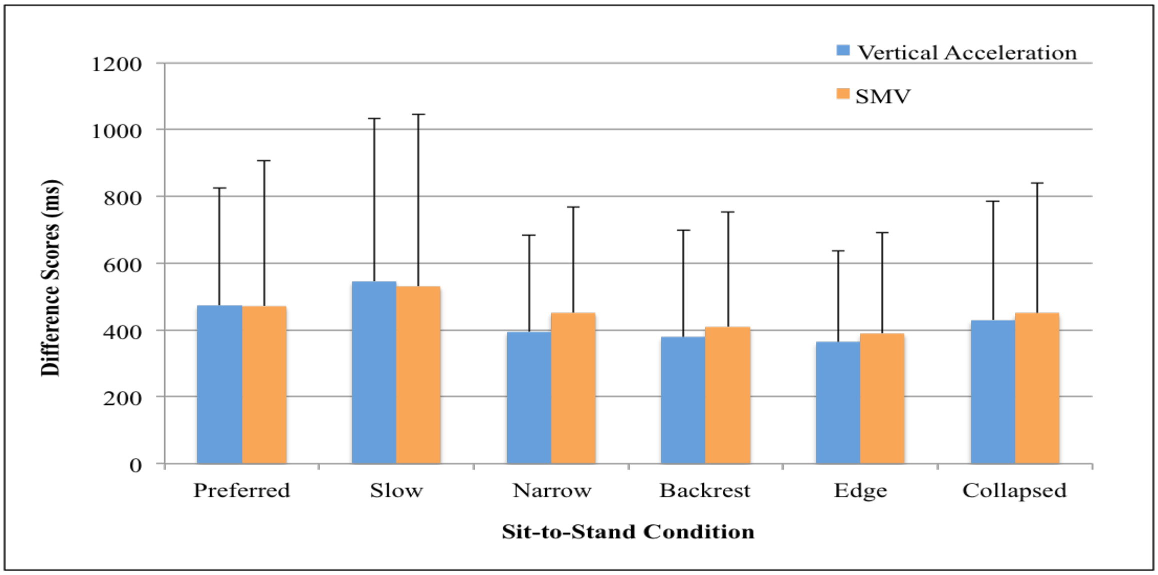


Figure 4-1: Mean difference scores with standard deviations (bars) between full hip extension identified using kinematic displacement and acceleration of the head across all conditions. No significant differences (p>0.05) in difference scores were found between vertical head acceleration and a SMV of head acceleration to denote full hip extension between conditions.

There was a statistically significant difference when using head accelerometry (vertical channel, SMV) to denote the transition from the dynamic to restabilization phase compared to vertical kinematic displacement ($p<0.0001$) across all task conditions. At the *preferred* speed, the point at which the head is no longer accelerating vertically occurred 474(SD=351)ms after full hip extension identified using vertical kinematic displacement. The SMV of head acceleration denoted the point where acceleration broke a 99% CI of quiet standing acceleration 472(SD=434)ms after full hip extension identified using vertical kinematic displacement. For the self-selected *slow* speed, vertical channel acceleration of the head identified the end of the dynamic phase 546(SD=486)ms after full hip extension identified using vertical kinematic displacement. Head acceleration SMV denoted the transition from the dynamic to restabilization phase 531(SD=515)ms after kinematic motion tracking. In the *narrow* condition, the point at which the head is no longer accelerating vertically occurred 395(SD=290)ms after full hip extension identified using the vertical kinematic displacement. The SMV of head acceleration identified the end of the dynamic phase 452(SD=315)ms after kinematic motion tracking. For the *backrest* condition, the vertical channel of head acceleration denoted full hip extension 379(SD=320)ms after the kinematic marker of full hip extension. The estimate of full hip extension using the SMV of head acceleration occurred 410(SD=342)ms the point identified using vertical kinematic displacement. In the *edge* condition, the point at which the head is no longer accelerating vertically occurred 364(SD=272)ms after full hip extension. Using the SMV of head acceleration, the point where acceleration broke the CI of quiet standing acceleration occurred 390(SD=301)ms after full hip extension identified using vertical kinematic displacement. When collapsed across all tasks, the vertical channel of head acceleration denoted the transition from the dynamic to restabilization phase 430(SD=356)ms after kinematic motion tracking. The head acceleration SMV estimate of full hip extension occurred 450(SD=390)ms after the kinematic estimate.

Sternum accelerometer: The timing of the estimate of the transition from the dynamic to restabilization phase from a sternum worn accelerometer (Figure 4-2) was approximately 360ms after the end of the dynamic phase determined from kinematics and was similar in all task conditions.

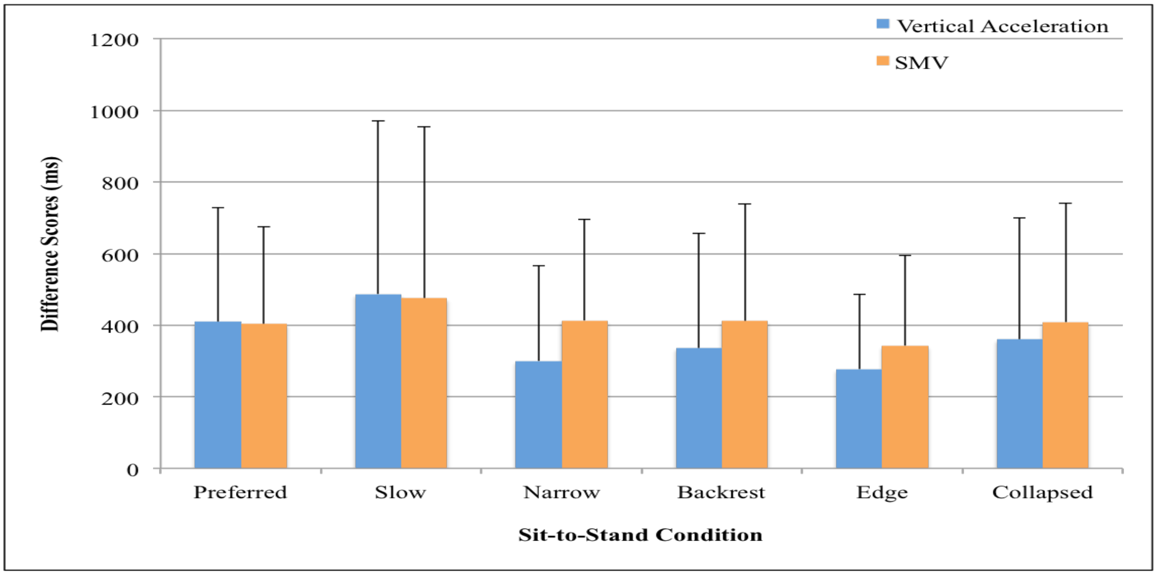


Figure 4-2: Mean difference scores with standard deviations (bars) between full hip extension identified using kinematic displacement and acceleration of the sternum across all conditions. No significant differences ($p>0.05$) were found when using vertical sternum acceleration or a SMV of sternum acceleration to denote full hip extension between conditions.

There was also a statistically significant difference when using sternum accelerometry (vertical channel, SMV) to denote the transition from the dynamic to restabilization phase compared to vertical kinematic displacement ($p<0.0001$) across all task conditions. At the *preferred* speed, the point at which the sternum is no longer accelerating vertically occurred 409(SD=319)ms after full hip extension identified using vertical kinematic displacement. The estimate of full hip extension using a SMV of sternum acceleration occurred 403(SD=271)ms after full hip extension identified using vertical kinematic displacement. For the self-selected *slow* speed, the vertical channel of sternum acceleration identified the transition from the dynamic to restabilization phase 487(SD=483)ms after vertical kinematic displacement. Using the SMV of sternum acceleration, full hip extension was denoted 475(SD=479)ms after the marker using vertical kinematic displacement. In the *narrow* condition, full hip extension identified using the vertical channel of sternum acceleration occurred 300(SD=266)ms after full hip extension denoted using vertical kinematic displacement. The SMV of sternum acceleration identified the end of the dynamic phase 413(SD=282)ms after vertical kinematic displacement. For the *backrest* condition, the estimate of the onset of the restabilization phase denoted using the vertical channel sternum acceleration occurred 336(SD=321)ms after full hip extension identified using vertical kinematic motion tracking. The SMV of sternum acceleration estimated the point of full hip extension 412(SD=326)ms after full hip extension identified using vertical kinematic displacement. In the *edge* condition, the transition from the dynamic to restabilization phase denoted using the vertical channel sternum acceleration occurred 277(SD=208)ms after the same marker measured via vertical kinematic displacement. Using the SMV of sternum acceleration, full hip extension was identified 343(SD=252)ms after vertical kinematic displacement. When collapsed across all speeds, the point at which the sternum is no longer accelerating vertically occurred 361(SD=338)ms after full hip extension identified using vertical kinematic displacement. The SMV of sternum acceleration denoted the end of the dynamic phase 409(SD=333)ms after vertical kinematic displacement.

Vertical force: The timing of the estimate of the transition from the dynamic to restabilization phase from vertical force was statistically similar ($p=1.0$) to that of the "gold standard" kinematic displacement in the *slow* condition. Overall, vertical force (Figure 4-3) identified the transition from the end of the dynamic phase to the onset of restabilization phase 255ms after the marker identified using vertical kinematic displacement.

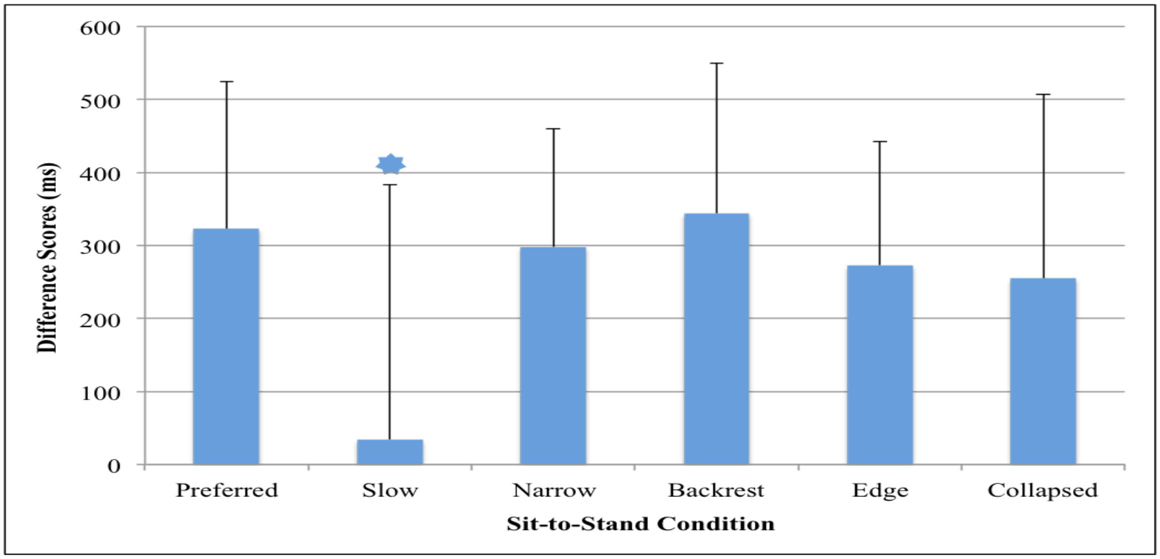


Figure 4-3: Mean difference scores with standard deviations (bars) between full hip extension identified using kinematic displacement and vertical force across all conditions. * denotes no statistically significant difference ($p>0.05$) when denoting full hip extension using Fz compared to kinematic displacement.

In the *slow* condition, there was no statistically significant difference ($p=1.0$) when using vertical force to denote full hip extension compared to vertical kinematic displacement. At the *slow* speed, the estimate of full hip extension using Fz occurred 34(SD=349)ms after vertical kinematic displacement. However, there was a statistically significant difference ($p<0.0001$) when using vertical force to denote full hip extension compared to vertical kinematic displacement during the *preferred, narrow, backrest and edge* conditions. At the *preferred* speed, the point where Fz returned to a CI greater than 99.99% of quiet standing Fz occurred 323(SD=201)ms after full hip extension identified using vertical kinematic displacement. For the *narrow* condition, the transition from the dynamic to restabilization phase identified using Fz occurred 298(SD=162)ms after full hip extension identified using vertical kinematic displacement. For the *backrest* condition, Fz denoted full hip extension 344(SD=206)ms after vertical kinematic displacement. In the *edge* condition, Fz identified the onset of restabilization phase

273(SD=169)ms after vertical kinematic displacement. When collapsed across all conditions, Fz denoted the transition from the dynamic to restabilization phase 255(SD=252)ms after the point identified using kinematic motion tracking.

To address the second hypothesis and compare the estimates of timing from portable technologies to vertical kinematic displacement (independent of task), the effect of measure on time to full hip extension was examined. Two-way repeated measures ANOVA reveled a statistically significant difference when denoting full hip extension using vertical force compared to the vertical channel of head acceleration (p=0.03) and a SMV of head acceleration (p=0.01) (Figure 4-4).

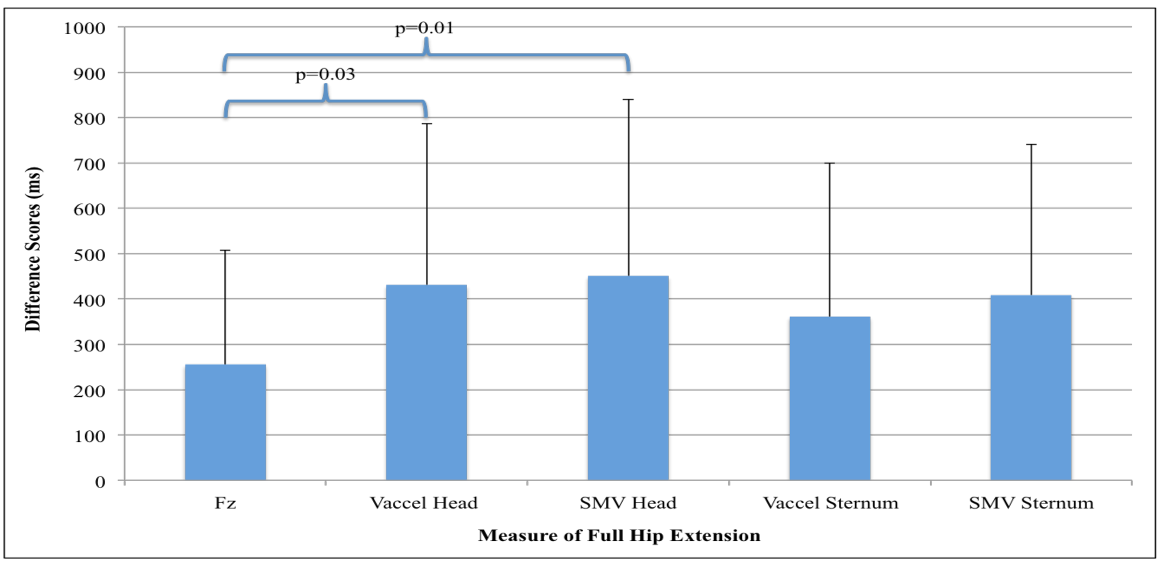


Figure 4-4: Mean difference scores with standard deviations (bars) between full hip extension identified using kinematic displacement compared to force plate and accelerometry measures when collapsed across all sit-to-stand conditions.

Difference scores between the “gold standard” kinematic motion tracking and portable technologies were significantly lower for Fz 255(SD=252)ms than vertical channel of head acceleration 430(SD=356)ms (p=0.03) and a SMV of head acceleration 450(SD=390)ms (p=0.01). Although not statistically significant, there was a trend (p=0.07) for lower mean difference using Fz compared to a SMV of sternum acceleration 409(SD=333)ms.

4.3.2 Within subject differences when using Fz to denote full hip extension

To examine the ability of Fz to denote full hip extension within subjects (Figure 4-5), within subject difference scores were compared to the group mean 323(SD=201)ms at the self-selected *preferred* speed. With the exception of subject 1 (mean difference = 714ms), mean difference scores remained relatively constant, mean range =190-347ms. In addition, Fz had similar variability when denoting full hip extension within all subjects, standard deviation range = 87-233ms.

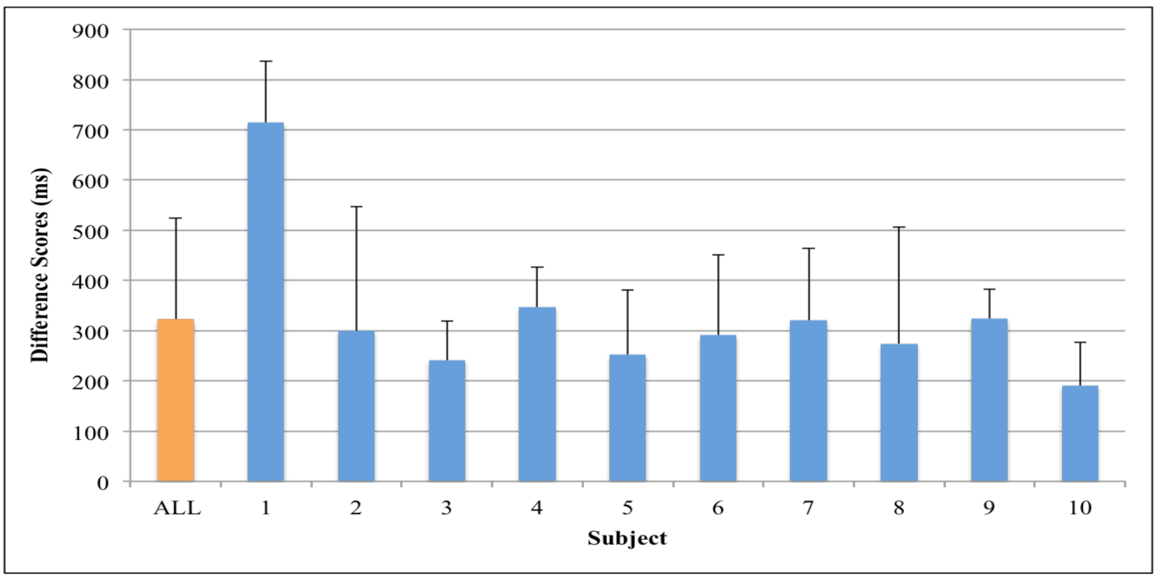


Figure 4-5: Within subject mean difference scores with standard deviations (bars) between full hip extension identified using kinematic displacement and Fz for the self-selected *preferred* speed.

4.3.3 Denoting the onset of the dynamic phase

Two-way repeated measures ANOVA revealed an effect of sit-to-stand task condition ($F[4,36] = 18.31$, $p< 0.0001$) on time to seat off. The *slow* condition had a significantly higher ($p<0.0001$) time to seat off (208(SD=111)ms) than the *preferred*

(148(SD=68)ms)*, narrow* (133(SD=56)ms)*, backrest* (152(SD=76)ms) *and edge* (120(SD=50)ms) conditions*.* In addition, there was also an effect of measure (F[4,36] = 19.06, p<0.0001) on time to full hip extension. Portable technology measures (vertical channel acceleration, sum magnitude vector, vertical force) had significantly higher (p<0.0001) times to seat off than the "gold standard" kinematic displacement.

Kinetics: Two-way repeated measures ANOVA revealed no statistically significant difference when using a baseline measure of vertical force (the point where Fz exceeded the 99% CI of quiet sitting Fz) to identify the onset of the dynamic phase compared to vertical kinematic displacement in the *slow* 121(SD=518)ms (p=0.96) and *edge* 50(SD=383)ms (p=1.0) conditions. There was, however, a statistically significant difference in the *preferred* 331(SD=232)ms (p<0.005), *narrow* 269(SD=223)ms (p=0.03) and *backrest* 475(SD=341)ms (p<0.0001) conditions.

Acceleration: Two-way repeated measures ANOVA found no statistically significant difference in the *preferred* 160(SD=491)ms (p=0.85), *narrow* 149(SD=416)ms (p=0.88)*, backrest* 174(SD=633)ms (p=0.49) *and edge* 142(SD=374)ms (p=0.69) conditions when denoting seat off as the point where the sternum acceleration SMV broke a 99% CI of quiet sitting compared to vertical kinematic displacement. However, a statistically difference (p<0.0001) was found when denoting seat off using this point during the *slow* condition 370(SD=557)ms.

A statistically significant difference (p<0.0001) was also found in all conditions when using the local minimum of the sternum vertical acceleration channel to denote seat off compared to kinematic displacement.

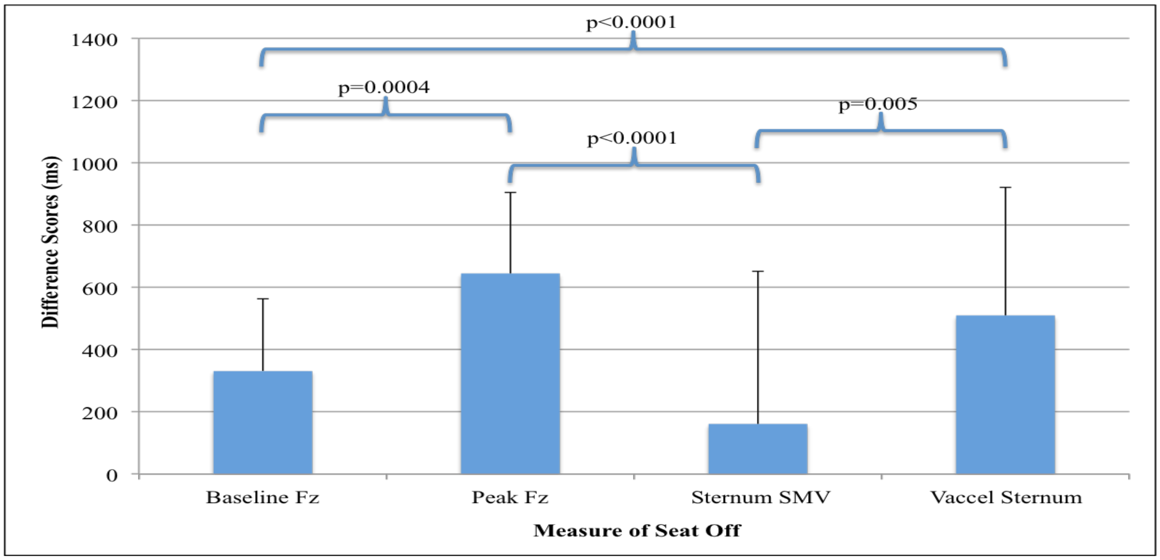


Figure 4-6: Mean difference scores with standard deviations (bars) between seat off identified using kinematic displacement compared to force plate measures and acceleration when collapsed across all sit-to-stand conditions.

There was no statistically significant difference (p=0.95) in the difference scores when measuring seat off using vertical force or a SMV of sternum acceleration. It is important to note the estimate of seat off recorded via vertical forces appeared to have a lower standard deviation (across subjects and task conditions) compared to estimates from accelerometers.

4.4 Discussion

In order to determine the future use of portable technologies to assess components of the sit-to-stand, the current study independently compared the use of accelerometry and force plate measures to estimate the start and end of the dynamic phase as reflected by the "gold standard" measured using kinematics. In contrast to the hypotheses, there were statistically significant differences between the timing of the gold standard (full hip extension) and the estimates from accelerometry and force plates. There was, however, a statistically significant difference when denoting full hip extension using vertical force compared to head accelerometer measures, with vertical force having a statistically lower difference score. In addition to having a lower mean difference score, using Fz to denote full hip extension appeared to be less variable than accelerometer measures of full hip extension for each condition and when collapsed across all conditions. The following discussion addresses: 1) the possible reasons for the measured differences between the gold standard and the estimates tested in this study, 2) the justification for the proposed use of force plates as a method of estimating the timing of the sit-to-stand phases, and 3) the potential clinical importance of these measures.

The presence of statistically significant differences between the gold standard and both ground reaction forces and accelerometry data could have resulted for several reasons. In all cases, the measures of acceleration and ground reaction forces occurred after the point identified using kinematic motion tracking. The possible reasons for the measured delay between the surrogate measure of full hip extension (peak verticality of trunk) and the force and acceleration estimates could arise from the fact that this was a comparison of displacement versus acceleration or kinetics. Given that the timing of the acceleration and force events tend to lead the kinematic timing measures, it is likely that other factors are associated with the person specific systematic differences between the events denoted using force and acceleration estimates compared to the timing of full hip extension. In contrast, the specific measures that were used to reflect sit-to-stand events (seat off, full hip extension) were chosen because these measures have previously been established in literature (1). Other measures may exist that are capable of quantifying and differentiating the balance and strength components of the sit-to-stand.

The error in the ability of vertical force to denote the transition from the dynamic to restabilization phase could be attributed to the definition of the “gold standard” full hip extension or the speed of standing. The current study used peak verticality of the trunk, a surrogate measure of full hip extension, to identify the distinct phases of the sit-to-stand. Therefore, the previously reported measure of full hip extension (1) was not actually measured in this study. The speed at which individuals stood and the associated degree of Fz overshoot may also affect the ability of portable technology measures to identify the transition from the dynamic to restabilization phase. Nevertheless, this Fz overshoot can be viewed as an indication of the COM acceleration, where a decreased Fz overshoot, lower COM acceleration, may be indicative of a standing strategy that facilitates control throughout the dynamic phase.

It is clear the source for the error was not task specific as the differences were evident regardless of the sit-to-stand movement characteristics. The variations of tasks were used to simulate the strategies and challenges that may be experienced or performed across young and older adults. For example, the *backrest* condition was used to facilitate the momentum transfer strategy, where the trunk flexes forward, changing the orientation of the accelerometer, resulting in the vertical channel of the tri-axial accelerometer having a decreased vertical component and an increased AP component. Whereas the *edge* condition was implemented to facilitate the stabilization strategy, where individuals stand straight up with little trunk flexion keeping the vertical axis of the accelerometer measuring acceleration in the vertical direction. This issue was also addressed in post processing using a vector approach when analyzing the accelerometer data. A sum magnitude vector was calculated, where the magnitude of accelerations in all planes were used to identify the transition from the dynamic to restabilization phase. Nevertheless, this measurement technique did not decrease the mean difference score when identifying a proxy measure of full hip extension using accelerometry compared to motion tracking.

In spite of the differences in timing when denoting the transition from the dynamic to restabilization phase, the current results led to the view that measures from ground reaction forces may be a suitable method to extract STS measures indicative of balance and strength and might help future work reveal underlying challenges (e.g. sensory, vascular). This is based on statistically lower mean difference scores and qualitatively less variability in the *slow* condition when denoting full hip extension using Fz compared to accelerometer measures. Although statistically significant differences were found between force plate measures and vertical kinematic displacement to denote full hip extension, these differences are likely not clinically significant. These differences are unlikely clinically significant as it is doubtful that an error of 225ms will affect the time to restabilization so vastly that it will be not capable of screening for individuals with restabilization challenges. Akram (1) found healthy older adults required more time (9 seconds) to restabilize following a single sit-to-stand than healthy young adults required (7 seconds). Thus, 225ms is well within the clinically meaningful difference between healthy young and older adults, however, the difference in time to restabilization between healthy older adults and older adults with control challenges still needs to be determined. It is important to note, the link between restabilization phase and cerebrovascular challenges still need to be explored. However, the ability of vertical force to identify full hip extension in a clinical setting for a fraction of the cost of research grade equipment makes Wii force boards a potential tool for clinical screening and assessments.

Secondary analysis also revealed vertical force to be the most suitable measure of seat off. No statistically significant difference was found in difference scores when measuring seat off using vertical force or sternum SMV. In spite of no statistically significant differences between vertical force and accelerometer measures of seat off, Fz had appeared to have less variability making it the recommended measure to identify the onset of the dynamic phase.

As previously mentioned, other measures may exist to denote the transition from the dynamic to restabilization phase. This issue may also be addressed in future studies by the use of more advanced sensor technologies with the ability to detect orientation (e.g. use of a gyroscope). If the accelerometer contains a gyroscope, the orientation is known; therefore, a baseline measure during quiet sitting can be calculated and vertical acceleration in each plane can be calculated using the formula $A_{vert=}\ A_{plane} Cos\ \theta$, where θ is the change in orientation from baseline. Unfortunately, in the current study we were limited to the more clinically feasible sensors without a gyroscope. In addition, there may be an approach that relies on multiple sensor modalities/technologies to reflect such phase points. Combining measures of ground reaction force and acceleration may offer a more precise measure (smaller mean difference) when identifying full hip extension. Overlaying the sternum acceleration and ground reaction force profiles may provide additional insight into COM movement, which is used by the “gold standard” kinematic motion tracking systems to denote the transition from the dynamic to restabilization phase.

Without a measure of kinetics (e.g. WBB) in the seat pan of the chair it is difficult to precisely measure the onset of dynamic phase, seat off. The difference score of vertical force to denote seat off may be attributed to the width of the quiet sitting confidence interval chosen. The confidence interval chosen (3 SD) may have been too large, therefore increasing the load required to break the band leading to the delayed identification of seat off. Thus, the difference score of a 2 SD confidence of quiet sitting Fz, which would decrease the load required and in turn lead to an earlier identification of seat off, should be examined to denote the onset of the dynamic phase.

These findings are clinically relevant as all tasks of independent mobility can now be assessed in a clinical setting using portable technologies. This work provides insight into the ability of ground reaction forces (which can be measured using WBB) to feature extract events during the STS, specifically the ability to quantify and differentiate the balance and strength components of the task. Having the capability to characterize the balance and strength components during the sit-to-stand will address the inability of composite measures (e.g. 5STS time) to reveal behviour specific control challenges during tests of transitions. Performance during the dynamic phase of the STS may be indicative of the strategy used, dynamic stability capacity and lower limb strength. Performance during the restabilization phase can potentially be used as a marker of age-related changes in dynamic and/or static balance control, reveal underlying balance impairments and possibly screen for cerebrovascular conditions (1).

4.5 Conclusions

The current study revealed the ability to rely on ground reaction forces to approach the phases of sit-to-stand and therefore the ability to quantify this behavior using portable technology. As a result, the three core tasks (walking, standing and sit-to-stand) can be adequately measured using existing low cost and portable technology that can be used in clinical and community settings. With respect to sit-to-stand behavior, the ability to denote the transition between dynamic and static (restabilization phase) is likely important to be able to use quantitative measures of this task performance as a more sensitive marker of specific control challenges (e.g. vascular, sensory, strength). These measurements will provide insight into the complex relationship between the sit-to-stand and the other tasks of independent mobility (standing, walking) to determine if they reflect unique challenges to control.

Chapter 5: General discussion

5.1 General discussion

Independent mobility is influenced by one's ability to perform three essential tasks of daily living: maintaining upright stance, transitioning from a seated to standing posture and walking. An elderly individual may struggle with a specific task or a combination of the tasks since each task poses both unique and commonly shared control challenges. It is important to understand the source of potential problems to mobility and balance in order to prescribe individualized therapeutic interventions that focus on behavior specific control challenges to maximize capacity for safe independent mobility.

The findings from study 2 provided insight into the ability to use kinetic and accelerometry measures to feature extract events during the STS, specifically the ability to quantify and differentiate the balance and strength components. Since vertical force was found suitable to assess the various phases of the sit-to-stand, it is now possible to use Wii boards to quantify and distinguish the strength and balance control components of the sit-to-stand, as well as assess standing balance. These measures can be complemented with wearable accelerometry to provide insight into the complex relationship between the STS and the other tasks of independent mobility (standing, walking) to determine if they reflect unique challenges to control. Combining vertical force measures with accelerometry may offer other measurement techniques to characterize COM movement, thus affording the ability to explore the relationship between the COM and COP in a clinical setting.

The findings from study 1 revealed the independence of standing, transition and walking task performance, reinforcing the importance of assessing behavior specific performance scores in contrast to the traditional composite scores (e.g. TUG score), which do not identify behavior specific control challenges. The within task relationships between measures may advocate for only a single test of each task being required to assess independent mobility. In addition, the findings from study 1 identified measures of these tasks that may have exposed the independent strength and balance components and revealed measures that highlight the unique control challenges between tasks of independent mobility.

Developing and understanding the relationship between measures of specific balance or strength control challenges affords clinicians the ability to prescribe targeted and potentially more effective interventions. By focusing treatment on behavior specific control challenges, an individual's fall risk can be decreased, while improving or maintaining their current activity level, ergo maximizing their capacity for safe independent mobility.

5.2 Future direction

Subsequent key steps in the line of research should focus on: 1) the ability of the restabilization phase to detect cerebrovascular control challenges (e.g. orthostatic hypotension), 2) the generalizability of findings to a broader range of individuals, and 3) the ability of these tools to predict fall risk and mobility.

Individuals with known cerebrovascular control challenges (e.g. orthostatic hypotension) should be tested to determine the ability of the restabilization phase to detect cerebrovascular control challenges. A comparison between the time to restabilization in a control group of healthy older adults and older adults suffering from cerebrovascular control challenges can then be made to determine if regulation of brain blood flow affects performance during the restabilization phase.

The residents recruited for this thesis were relatively high functioning older adults; therefore the generalizability of these findings to a broader range of individuals still needs to be determined. A greater number of impaired older adults need to be recruited to determine the relationship between balance and strength measures across and within the tasks of independent mobility. However, as previously mentioned using the arms of the chair changes the force profile during standing, making it difficult to distinguish the different phases of the sit-to-stand (38). Consequently, this may require the use of new technologies (e.g. sensors with gyroscopes) to help in sit-to-stand phase identification.

The predictive utility of these tools to assess fall risk and mobility needs to be addressed. Fortunately, Schlegel Villages keep a detailed log of all falls hence a prospective cohort study can be performed using residents who have completed the Schlegel Functional Fitness Assessment to determine which performance measures best predict fall risk. Additionally, an ambulatory monitoring study is currently being conducted on Schlegel Functional Fitness Assessment participants, therefore a prospective cohort study can also be used to determine which performance measures best predict mobility levels in older adults living in congregate care.

5.3 Limitations

The current work was limited by: 1) the sensors used to characterize the various phases of the sit-to-stand in study 2, 2) the population tested in study 1, and 3) sample size. The inability to measure true vertical acceleration from a head-mounted and sternum-mounted accelerometer made it difficult to dissociate the vertical and AP acceleration during the dynamic phase of the sit-to-stand, resulting in a decreased ability to estimate when the COM stopped moving vertically, a surrogate measure of full hip extension. However, as highlighted in study 2, an accelerometer containing a gyroscope can quantify orientation; therefore, a baseline measure during quiet sitting can be calculated and vertical acceleration in each plane can be calculated using the formula $A_{vert} = A_{plane} \cos\theta$, where θ is the change in orientation from baseline. Unfortunately, in the current study we were limited to sensors without a gyroscope. As previously mentioned, study 1 participants were relatively high functioning older adults. To truly determine if different control mechanisms govern performance on tests of standing, transitions and walking, lower functioning older adults should be recruited, which would also make these results more generalizable. This work was also limited by the sample size of both studies. More participants should be recruited to increase the power of these studies to confirm ground reaction forces can characterize the distinct phases of the sit-to-stand and determine if the control mechanisms governing performance on tests of standing, walking, and transitions are truly independent.

5.4 Conclusions

All tasks of independent mobility and underlying control challenges can now be assessed in a clinical setting, as ground reaction forces which can be measured using Wii Balance Boards were found capable of extracting measures indicative of balance and strength and help future work reveal underlying challenges (e.g. sensory or vascular). The independence of standing, transition and walking tasks supports the need for behavior specific performance scores to replace traditional less informative composite scores to better inform clinical intervention and assess capacity for independent mobility. The lack of association of measures across tasks of independent mobility revealed the complexity of each task and indicates different control mechanisms govern task performance.

References

1. Akram, SB, and McIlroy, WE. Challenging horizontal movement of the body during sit-to-stand: impact on stability in the young and elderly. *J. Mot. Behav.* 43: 147-153, 2011.
2. Allen, J, and Morelli, V. Aging and Exercise. *Clin. Geriatr. Med.* 27: 661-671, 2011.
3. Auvinet, B, Berrut, G, Touzard, C, Moutel, L, Collet, N, Chaleil, D, and Barrey, E. Reference data for normal subjects obtained with an accelerometric device. *Gait Posture* 16: 124-134, 2002.
4. Badiuk, B, Van Ootenghem, K, Hughson, R, and McIlroy, W. Using accelerometers to feature extract events during the sit-to-stand: A validation stufy. *(in progress) - no journal* .
5. Baloh, RW, Enrietto, J, Jacobson, KM, and Lin, A. Age-related changes in vestibular function: a longitudinal study. *Ann. N. Y. Acad. Sci.* 942: 210-219, 2001.
6. Bassey, EJ, Bendall, MJ, and Pearson, M. Muscle strength in the triceps surae and objectively measured customary walking activity in men and women over 65 years of age. *Clin. Sci. (Lond)* 74: 85-89, 1988.
7. Berg, KO, Wood-Dauphinee, SL, Williams, JI, and Maki, B. Measuring balance in the elderly: validation of an instrument. *Can. J. Public Health* 83 Suppl 2: S7-11, 1992.
8. Bogle Thorbahn, LD, and Newton, RA. Use of the Berg Balance Test to predict falls in elderly persons. *Phys. Ther.* 76: 576-83; discussion 584-5, 1996.
9. Buatois, S, Miljkovic, D, Manckoundia, P, Gueguen, R, Miget, P, Vancon, G, Perrin, P, and Benetos, A. Five times sit to stand test is a predictor of recurrent falls in healthy community-living subjects aged 65 and older. *J. Am. Geriatr. Soc.* 56: 1575-1577, 2008.
10. Carter, ND, Kannus, P, and Khan, KM. Exercise in the prevention of falls in older people: a systematic literature review examining the rationale and the evidence. *Sports Med.* 31: 427-438, 2001.
11. Clark, RA, Bryant, AL, Pua, Y, McCrory, P, Bennell, K, and Hunt, M. Validity and reliability of the Nintendo Wii Balance Board for assessment of standing balance. *Gait Posture* 31: 307-310, 2010.
12. Csuka, M, and McCarty, DJ. Simple method for measurement of lower extremity muscle strength. *Am. J. Med.* 78: 77-81, 1985.
13. Daley, MJ, and Spinks, WL. Exercise, mobility and aging. *Sports Med.* 29: 1-12, 2000.
14. Deshpande, N, and Patla, AE. Visual-vestibular interaction during goal directed locomotion: effects of aging and blurring vision. *Exp. Brain Res.* 176: 43-53, 2007.
15. Deutsch Lezak, M, Howieson, D, Bigler, E, and Tranel, D. Neuropsychological Assessment. New York, New York; Oxford Univerversity Press, 200.
16. Enright, PL. The six-minute walk test. *Respir. Care* 48: 783-785, 2003.

17. Fiatarone, MA, Marks, EC, Ryan, ND, Meredith, CN, Lipsitz, LA, and Evans, WJ. High-intensity strength training in nonagenarians. Effects on skeletal muscle. *JAMA* 263: 3029-3034, 1990.
18. Fitzpatrick, R, and McCloskey, DI. Proprioceptive, visual and vestibular thresholds for the perception of sway during standing in humans. *J. Physiol.* 478 (Pt 1): 173-186, 1994.
19. Fitzpatrick, R, Rogers, DK, and McCloskey, DI. Stable human standing with lower-limb muscle afferents providing the only sensory input. *J. Physiol. (Lond.)* 480: 395-403, 1994.
20. Gage, WH, Zabjek, KF, Hill, SW, and McIlroy, WE. Parallels in control of voluntary and perturbation-evoked reach-to-grasp movements: EMG and kinematics. *Exp. Brain Res.* 181: 627-637, 2007.
21. Gittings, NS, and Fozard, JL. Age related changes in visual acuity. *Exp. Gerontol.* 21: 423-433, 1986.
22. Halmagyi, GM, and Cremer, PD. Assessment and treatment of dizziness. *J. Neurol. Neurosurg. Psychiatry.* 68: 129-134, 2000.
23. Hausdorff, JM, Edelberg, HK, Mitchell, SL, Goldberger, AL, and Wei, JY. Increased gait unsteadiness in community-dwelling elderly fallers. *Arch. Phys. Med. Rehabil.* 78: 278-283, 1997.
24. Hausdorff, JM, Rios, DA, and Edelberg, HK. Gait variability and fall risk in community-living older adults: a 1-year prospective study. *Arch. Phys. Med. Rehabil.* 82: 1050-1056, 2001.
25. Hausdorff, JM. Gait variability: methods, modeling and meaning. *J. Neuroeng Rehabil.* 2: 19, 2005.
26. Highstein, SM. How does the vestibular part of the inner ear work. In: Anonymous New York, New York: Oxford University Press, 1996.
27. Horak, FB, and Nashner, LM. Central programming of postural movements: adaptation to altered support-surface configurations. *J. Neurophysiol.* 55: 1369-1381, 1986.
28. Howe, JA, Inness, EL, Venturini, A, Williams, JI, and Verrier, MC. The Community Balance and Mobility Scale--a balance measure for individuals with traumatic brain injury. *Clin. Rehabil.* 20: 885-895, 2006.
29. Hughes, MA, and Schenkman, ML. Chair rise strategy in the functionally impaired elderly. *J. Rehabil. Res. Dev.* 33: 409-412, 1996.
30. Inglis, JT, Horak, FB, Shupert, CL, and Jones-Rycewicz, C. The importance of somatosensory information in triggering and scaling automatic postural responses in humans. *Experimental Brain Research* 101: 159-164, 1994.
31. Inness, EL, Howe, JA, Niechwiej-Szwedo, E, Jaglal, SB, McIlroy, WE, and Verrier, MC. Measuring Balance and Mobility after Traumatic Brain Injury: Validation of the Community Balance and Mobility Scale (CB&M). *Physiother. Can.* 63: 199-208, 2011.
32. Janssen, WG, Kulcu, DG, Horemans, HL, Stam, HJ, and Bussmann, JB. Sensitivity of accelerometry to assess balance control during sit-to-stand movement. *IEEE Trans. Neural Syst. Rehabil. Eng.* 16: 479-484, 2008.
33. Jette, AM, Branch, LG, and Berlin, J. Musculoskeletal impairments and physical disablement among the aged. *J. Gerontol.* 45: M203-8, 1990.

34. Kavanagh, JJ, and Menz, HB. Accelerometry: a technique for quantifying movement patterns during walking. *Gait Posture* 28: 1-15, 2008.
35. Kaye, HS, Harrington, C, and LaPlante, MP. Long-term care: who gets it, who provides it, who pays, and how much? *Health. Aff. (Millwood)* 29: 11-21, 2010.
36. Konrad, HR, Girardi, M, and Helfert, R. Balance and aging. *Laryngoscope* 109: 1454-1460, 1999.
37. Lee, J-, Cho, S-, Lee, J-, Lee, K-, and Yang, H-. Wearable accelerometer system for measuring the temporal parameters of gait. In: Proceedings of Annual International Conference of the IEEE Engineering in Medicine and Biology - Proceedings. Anonymous , 2007. pp. 483-6.
38. Lindemann, U, Muche, R, Stuber, M, Zijlstra, W, Hauer, K, and Becker, C. Coordination of strength exertion during the chair-rise movement in very old people. *J. Gerontol. A Biol. Sci. Med. Sci.* 62: 636-640, 2007.
39. Lord, SR, Ward, JA, Williams, P, and Anstey, KJ. Physiological factors associated with falls in older community-dwelling women. *J. Am. Geriatr. Soc.* 42: 1110-1117, 1994.
40. Lord, SR, and Dayhew, J. Visual risk factors for falls in older people. *J. Am. Geriatr. Soc.* 49: 508-515, 2001.
41. Lord, SR, Murray, SM, Chapman, K, Munro, B, and Tiedemann, A. Sit-to-stand performance depends on sensation, speed, balance, and psychological status in addition to strength in older people. *J. Gerontol. A Biol. Sci. Med. Sci.* 57: M539-43, 2002.
42. Macpherson, J, Rushmer, D, and Dunbar, D. Postural responses in the cat to unexpected rotations of the supporting surface: evidence for a centrally generated synergic organization. *Exp. Brain Res.* 62: 152-160, 1986.
43. Maki, BE, Holliday, PJ, and Topper, AK. A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *J. Gerontol.* 49: M72-84, 1994.
44. Maki, BE. Gait changes in older adults: predictors of falls or indicators of fear. *J. Am. Geriatr. Soc.* 45: 313-320, 1997.
45. Maki, BE, and McIlroy, WE. The role of limb movements in maintaining upright stance: The 'change- in-support' strategy. *Phys. Ther.* 77: 488-507, 1997.
46. Maki, BE, and McIlroy, WE. Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. *Age Ageing* 35 Suppl 2: ii12-ii18, 2006.
47. Massion, J. Movement, posture and equilibrium: Interaction and coordination. *Prog. Neurobiol.* 38: 35-56, 1992.
48. Menz, HB, Lord, SR, and Fitzpatrick, RC. Age-related differences in walking stability. *Age Ageing* 32: 137-142, 2003.
49. Mori, S, Sakamoto, T, Ohta, Y, Takakusaki, K, and Matsuyama, K. Site-specific postural and locomotor changes evoked in awake, freely moving intact cats by stimulating the brainstem. *Brain Res.* 505: 66-74, 1989.
50. O'Loughlin, JL, Robitaille, Y, Boivin, J, and Suissa, S. Incidence of and Risk Factors for Falls and Injurious Falls among the Community-dwelling Elderly. *Am. J. Epidemiol.* 137: 342-354, 1993.

51. Olufsen, MS, Tran, HT, Ottesen, JT, Lipsitz, LA, Novak, V, Benim, R, Joyner, S, Tweedy, E, Vogl, C, and Dixon, A. Modeling baroreflex regulation of heart rate during orthostatic stress. *American Journal of Physiology - Regulatory Integrative and Comparative Physiology* 291: R1355-R1368, 2006.
52. Olufsen, MS, Alston, AV, Tran, HT, Ottesen, JT, and Novak, V. Modeling heart rate regulation - Part I: Sit-to-stand versus head-up tilt. *Cardiovascular Engineering* 8: 73-87, 2008.
53. Perry, SD, McIlroy, WE, and Maki, BE. The role of plantar cutaneous mechanoreceptors in the control of compensatory stepping reactions evoked by unpredictable, multi-directional perturbation. *Brain Res.* 877: 401-406, 2000.
54. Perry, SD. Evaluation of age-related plantar-surface insensitivity and onset age of advanced insensitivity in older adults using vibratory and touch sensation tests. *Neurosci. Lett.* 392: 62-67, 2006.
55. Podsiadlo, D, and Richardson, S. The timed "Up & Go": a test of basic functional mobility for frail elderly persons. *J. Am. Geriatr. Soc.* 39: 142-148, 1991.
56. Prajapati, SK, Gage, WH, Brooks, D, Black, SE, and McIlroy, WE. A novel approach to ambulatory monitoring: investigation into the quantity and control of everyday walking in patients with subacute stroke. *Neurorehabil. Neural Repair* 25: 6-14, 2011.
57. Priest, AW, Salamon, KB, and Hollman, JH. Age-related differences in dual task walking: A cross sectional study. *Journal of NeuroEngineering and Rehabilitation* 5, 2008.
58. Prieto, TE, Myklebust, JB, Hoffmann, RG, Lovett, EG, and Myklebust, BM. Measures of postural steadiness: Differences between healthy young and elderly adults. *IEEE Transactions on Biomedical Engineering* 43: 956-966, 1996.
59. Riley, PO, Schenkman, ML, Mann, RW, and Hodge, WA. Mechanics of a constrained chair-rise. *J. Biomech.* 24: 77-85, 1991.
60. Robbins, AS, Rubenstein, LZ, Josephson, KR, Schulman, BL, Osterweil, D, and Fine, G. Predictors of falls among elderly people. Results of two population-based studies. *Arch. Intern. Med.* 149: 1628-1633, 1989.
61. Rubenstein, LZ, Josephson, KR, and Robbins, AS. Falls in the nursing home. *Ann. Intern. Med.* 121: 442-451, 1994.
62. Saremi, K, Marehbian, J, Yan, X, Regnaux, J-, Elashoff, R, Bussel, B, and Dobkin, BH. Reliability and validity of bilateral thigh and foot accelerometry measures of walking in healthy and hemiparetic subjects. *Neurorehabil. Neural Repair* 20: 297-305, 2006.
63. Schlicht, J, Camaione, DN, and Owen, SV. Effect of intense strength training on standing balance, walking speed, and sit-to-stand performance in older adults. *Journals of Gerontology - Series A Biological Sciences and Medical Sciences* 56: M281-M286, 2001.
64. Scott, V, Wagar, B, Sum, A, Metcalfe, S, and Wagar, L. A public health approach to fall prevention among older persons in Canada. *Clin. Geriatr. Med.* 26: 705-718, 2010.
65. Shumway-Cook, A, Baldwin, M, Polissar, NL, and Gruber, W. Predicting the probability for falls in community-dwelling older adults. *Phys. Ther.* 77: 812-819, 1997.

66. Shumway-Cook, A, Brauer, S, and Woollacott, M. Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Phys. Ther.* 80: 896-903, 2000.
67. Sorond, FA, Serrador, JM, Jones, RN, Shaffer, ML, and Lipsitz, LA. The Sit-to-Stand Technique for the Measurement of Dynamic Cerebral Autoregulation. *Ultrasound in Medicine and Biology* 35: 21-29, 2009.
68. Springer, S, Giladi, N, Peretz, C, Yogev, G, Simon, ES, and Hausdorff, JM. Dual-tasking effects on gait variability: the role of aging, falls, and executive function. *Mov. Disord.* 21: 950-957, 2006.
69. Sturnieks, DL, St George, R, and Lord, SR. Balance disorders in the elderly. *Neurophysiol. Clin.* 38: 467-478, 2008.
70. Thapa, PB, Gideon, P, Fought, RL, and Ray, WA. Psychotropic Drugs and Risk of Recurrent Falls in Ambulatory Nursing Home Residents. *American Journal of Epidemiology* 142: 202-211, 1995.
71. Tian, JR, Crane, BT, Wiest, G, and Demer, JL. Effect of aging on the human initial interaural linear vestibulo-ocular reflex. *Exp. Brain Res.* 145: 142-149, 2002.
72. Tinetti, ME, and Williams, CS. The effect of falls and fall injuries on functioning in community-dwelling older persons. *J. Gerontol. A Biol. Sci. Med. Sci.* 53: M112-9, 1998.
73. Tinetti, ME, and Speechley, M. Prevention of Falls among the Elderly. *N. Engl. J. Med.* 320: 1055-1059, 1989.
74. Troosters, T, Gosselink, R, and Decramer, M. Six minute walking distance in healthy elderly subjects. *Eur. Respir. J.* 14: 270-274, 1999.
75. Tseng, BS, Marsh, DR, Hamilton, MT, and Booth, FW. Strength and aerobic training attenuate muscle wasting and improve resistance to the development of disability with aging. *J. Gerontol. A Biol. Sci. Med. Sci.* 50 Spec No: 113-119, 1995.
76. van Iersel, MB, Kessels, RP, Bloem, BR, Verbeek, AL, and Olde Rikkert, MG. Executive functions are associated with gait and balance in community-living elderly people. *J. Gerontol. A Biol. Sci. Med. Sci.* 63: 1344-1349, 2008.
77. Verrillo, RT, Bolanowski, SJ, and Gescheider, GA. Effect of aging on the subjective magnitude of vibration. *Somatosens. Mot. Res.* 19: 238-244, 2002.
78. Washburn, RA, Smith, KW, Jette, AM, and Janney, CA. The Physical Activity Scale for the Elderly (PASE): development and evaluation. *J. Clin. Epidemiol.* 46: 153-162, 1993.
79. Whipple, RH, Wolfson, LI, and Amerman, PM. The relationship of knee and ankle weakness to falls in nursing home residents: an isokinetic study. *J. Am. Geriatr. Soc.* 35: 13-20, 1987.
80. Whitney, SL, Wrisley, DM, Marchetti, GF, Gee, MA, Redfern, MS, and Furman, JM. Clinical measurement of sit-to-stand performance in people with balance disorders: validity of data for the Five-Times-Sit-to-Stand Test. *Phys. Ther.* 85: 1034-1045, 2005.
81. Wild, D, Nayak, US, and Isaacs, B. Description, classification and prevention of falls in old people at home. *Rheumatol. Rehabil.* 20: 153-159, 1981.
82. Winter, DA. A.B.C. (Anatomy, Biomechanics, Control) of Balance during Standing and Walking. Waterloo, Ontario; Waterloo Biomechanics, 1995.

83. Winter, DA, Patla, AE, Prince, F, Ishac, M, and Gielo-Perczak, K. Stiffness control of balance in quiet standing. *J. Neurophysiol.* 80: 1211-1221, 1998.
84. Woollacott, MH, Shumway-Cook, A, and Nashner, LM. Aging and posture control: Changes in sensory organization and muscular coordination. *International Journal of Aging and Human Development* 23: 97-114,1986.
85. Woollacott, MH, and Tang, P-. Balance control during walking in the older adult: Research and its implications. *Phys. Ther.* 77: 646-660, 1997.
86. Zhang, F, Ferrucci, L, Culham, E, Metter, EJ, Guralnik, J, and Deshpande, N. Performance on five times sit-to-stand task as a predictor of subsequent falls and disability in older persons. *J. Aging Health* 25: 478-492, 2013.