Instrumenting the Clinical Test of Sensory Interaction and Balance:

A Comparison of Different Measures of Balance

by

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Supervisory Committee

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Abstract

BACKGROUND: Balance is integral to an individuals' ability to stand and walk, often deteriorates with age and has been identified as one of the major risk factors for fall risk. The Clinical Test of Sensory Interaction and Balance (CTSIB) has been shown as a valid tool to characterize postural sway during quiet standing to serve as a measure of balance control and this measurement can be reflected as either Centre of Pressure (COP) and Centre of Mass (COM) in different kinematics forms including acceleration, velocity and displacement. In addition, within the kinematics of acceleration, velocity and displacement, there is an array of metrics, such as maximal sway, sway mean, sway standard deviation and area of an ellipse that can be calculated to quantify postural sway. It is currently not clear which of these potential measures and metrics of postural sway are best suited to differentiate balance performance between the CTSIB trials.

RESEARCH QUESTION: Which metrics and measures of postural sway are best suited to differentiate balance performance between the CTSIB trials?

METHODS: Thirty-nine community-dwelling older adults' (70+) completed the six trials of CTSIB. We determined the capacity of displacement and velocity of COP [COP_{FPd} and COP_{FPv}, respectively] measured by a force plate and velocity and acceleration of COM [$COM_{IMUa,}$ and $COM_{IMUv,}$ respectively] measured with inertial measurement unit (IMU), as well as displacement of COM [COM_{Pend}]) measured with a Pen-tail to differentiated the balance performance between trials of CTSIB by using repeated measures analysis of variance conducted across all metrics and the six trials of CTSIB separately for each measure for main effects for trial and significant interactions between trials and metrics. Post hoc analysis was conducted on significant

interactions ($p < 0.05$) for main effects and interactions using Tukey's Honest Significant Difference test. Effect sizes for comparisons between significant trials were calculated using Hedge's G.

RESULTS: All ANOVA's for each kinematic measure yielded significant main effects for trial, metric and trial by metric interactions. Out of 135 comparisons, the number of significant differences between CTSIB trials was seen in COP_{FPd} (56), followed by COM_{IMUv} (54), COP_{FPv} (48), and COM_{IMUa} (28). For COM_{Pend} 6 out of a possible 45 significant difference between CTSIB trials.

DISCUSSION: This study is the first to compare different measures and metrics across trials of the CTSIB to characterize unique features of postural sway. Importantly, all the measures were capable of detecting differences between trials of the CTSIB. Further, the differences between Trials 1-5 and 1-6, and AoE metrics, were sensitive to more differences between trials across all measures. Finally, AP and ML metrics had the largest effect size across trials.

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LIST OF ABBREVIATIONS

AoB_B - Area of Ellipse Bounded

AoE_E - Area of Ellipse Eigenvalue

AP - anteroposterior

BOS - Base of Support

BBS - Berg Balance Scale

CB&M - Community Balance and Mobility and scale

CNS - Central Nervous System

COG - Centre of Gravity

COM - Centre of Mass

COP - Centre of Pressure

CTSIB - Clinical Test of Sensory Interaction (Organization) and Balance

CTSIB T1-6 - Clinical Test of Sensory Interaction and Balance trials 1 - 6

FAB - Fullerton Advanced Balance Scale

FP - Force Plate

COP FP_d - Force Plate in the kinematic measure of displacement (position)

COP FP_v - Force Plate in the kinematic measure of velocity

FRT - Functional Reach Test

GRF - Ground reaction forces

IMU - Inertial Measurement Unit

COM IMU_a - Interial Measurement Unit in the kinematic measure of acceleration

COM IMU_v - Interial Measurement Unit in the kinematic measure of velocity

mCTSIB - modified Clinical Test of Sensory Interaction (Organization) and Balance

mCTSIB T1-4 - modified Test of Sensory Interaction and Balance trials 1 - 4

Max_{AP} - Maximal excursion in the anteroposterior direction

Max_{ML} - Maximal excursion in the mediolateral direction

$Mean_{AP}$ - sway mean in the anteroposterior direction

$Mean_{ML}$ - sway mean in the mediolateral direction

ML - mediolateral

Older Adults - Adults aged 70 or more years

COM Pen_d - Pen-tail device in the kinematic measure of displacement (position)

SD_{AP} - Standard deviation in the anteroposterior direction

SD_{ML} - Standard deviation in the mediolateral direction

SPPB - Short Physical Performance Battery

$Sway_{AP}$ - Anteroposterior Sway

$Sway_{ML}$ - Mediolateral Sway

TSP - Total Sway Path

TUG - Timed Up-and-Go

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Dedication

I want to dedicate this work to my family. You have always been there to support me, and I would not have been able to complete this without you.

Chapter 1: Introduction

1.1 General Statement of the Problem

Balance is integral to an individuals' ability to stand and walk, and it often deteriorates with age impairing gait function and leading to reduced mobility, independence and quality of life (Gill et al. 2001; Pizzigalli et al. 2016). Decreasing control of balance has been identified as one of the major risk factors for falling, and further research could lead to improved screening of fall risk (Piirtola and Era 2006; Pizzigalli et al. 2016). The Clinical Test of Sensory Interaction and Balance (CTSIB) was originally designed to assess the sensory systems (proprioception, vision, and vestibular) that contribute to balance by using six distinct quiet standing trials that include eyes open, eyes closed and a visual conflict dome to alter the visual stimulus on both firm and foam surfaces (Shumway-Cook and Horak 1986). Originally, to quantify balance, research indicated three techniques for clinicians: 1) subjective assessment of postural sway based on a numeric ranking system (e.g., 1 = minimal sway, 2 = mild sway, 3 = moderate sway, 4 = fall), 2) use of a stopwatch to record the amount of time the patient maintains balance (maximum = 30 seconds), and 3) use of Pen-tail to record displacement of postural sway (Cohen, Blatchly, and Gombash 1993; Ricci et al. 2009; Shumway-Cook and Horak 1986). These methods afforded limited accuracy but were clinically viable. More recently, researchers have instrumented the CTSIB using either a force plate (FP) or an inertial measurement unit (IMU) to measure postural sway, which has mitigated some of the measurement limitations (Swanenburg et al. 2010). However, each device provides different outputs; FP's provide outputs of Centre of Pressure (COP) kinematics and IMU's provide outputs of Centre of Mass (COM) kinematics. It is currently not clear what measures of postural sway best represent the balance function of older adults.

A force plate measures ground reaction forces generated by an individual standing on the platform and from this measurement centre of pressure (COP) is calculated (Howcroft et al. 2017). An IMU placed in the lumbar region measures acceleration and rotational motion of trunk movements using accelerometers and gyroscope sensors, respectively, to the approximate centre of mass (COM) sway, assuming that peripheral limbs do not change position (Allum et al. 2001). Though estimates of postural sway can be derived from both COP and COM measures, they are not synonymous and provide unique, but related, information about kinematics relevant to postural stability. Further COP and COM can both be provided in different kinematic domains (e.g. displacement, velocity, acceleration). For example, FP data can be used to calculate COP displacement (FP_d), and its first derivative yields COP velocity (FP_v), while the IMU signal can be used to calculate COM acceleration (IMU_a) and its first integral yields COM velocity (IMU_v). It is currently unclear which kinematic domain of COP or COM is the most sensitive in detecting changes in postural stability.

In addition, to the three kinematic domains, there is an array of metrics within each domain, such as maximal sway, sway mean, sway standard deviation and area of an ellipse that can be calculated to quantify postural sway. Different metrics have proven valuable to detect changes in tasks and populations. For example, COP path length, COP velocity and sway in medial-lateral (ML) and anterior-posterior (AP) are variables that can distinguish older adult fallers from non-fallers (Pizzigalli et al. 2016). It is currently not clear which metric is best able to capture key features of postural sway that are most relevant to postural stability.

1.2 Significance

It is not currently known which device in which kinematic domain using which metric would be able to detect the key features of postural sway between trials of CTSIB. Different metrics from different kinematics domains and devices may characterize unique elements of postural sway during CTSIB. We aim to determine which kinematics (acceleration, velocity and displacement) of which measure (CoP or CoM) is best suited to differentiate balance performance between CTSIB trials. This will enhance the capacity of the CTSIB's to detect changes in postural stability over time as well as identify individuals with deficits in postural control, including older adults at risk of falls.

1.3 Research Questions

This project sought to determine which combination of the metrics across the three kinematic subdomains measured by IMU and FP best differentiates postural performance between CTSIB trials.

First, we compared the typical outputs for each device FP displacement (COP FP_d), IMU acceleration (COM IMU_a), and Pen-tail displacement (COM Pen_d) across the relevant metrics. Second, we compared FP velocity (COP FP_v) and IMU velocity (COM IMU_v).

1.4 Limitations

The sample size of 39 older adults limits the power of the study, but there is support for this sample size in previous older adult research (Chaikeeree et al. 2015; Criter and Honaker 2016; Filar-Mierzwa et al. 2017; High et al. 2018; Lee 2017). The sample was selected from community-dwelling, older adults living in Victoria, British Columbia, Canada, and, therefore, should not be considered representative of the general Canadian population. The study was further limited because it only examined FP and IMU devices and should not be regarded as an exhaustive study of the possible tools that measure COP and COM during CTSIB. Other possible devices include, but are not limited to, motion capture systems, infrared imaging systems, the Wii Balance Board, insole pressure sensors systems, and mobile phone assessments. The type of high mass IMU sensor that was used limited the sensitivity of the device to small amplitudes when measuring trunk accelerations. Finally, the Visual Conflict Dome employed in this study did not always wholly obstruct the participant's peripheral vision.

Chapter 2: Literature Review

2.1 Significance of Falls

Currently, in North America, unintentional injuries are the third leading cause of mortality, of which falls are the largest contributor (Heron 2019). In Canada, 20-30%, and worldwide 35-40%, of older adults experience a fall each year (Smartrisk 2009). Falls among older people are a public health concern because of their frequency and adverse consequences in terms of morbidity, mortality, and quality of life, as well as their impact on health system services, including cost (Peel 2011). The Frailty and Injuries Co-operative Studies of Intervention Techniques define a fall as “unintentionally coming to rest on the ground, floor or other lower-level” (Masud and Morris 2001; Peel 2011; Public Health Agency of Canada 2014). In the most recent systematic review of international fall-related costs, researchers found that costs ranged between 0.85 percent 1.5 percent of the total health care expenditure of countries such as Canada, the United States, Australia, and Europe (Heinrich et al. 2010). The costs ranged between 0.07 percent to 0.20 percent of the Gross Domestic Product of the countries mentioned above (Peel 2011). These costs ranged from $3,476 to $10,749 (USD) per injurious fall and $26,483(USD) per fall that required hospitalization (Davis et al. 2010). From 2011 to 2016, the population of older adults in Canada had increased from 5% to 17%, and by 2025 the population of older adults was expected to increase to 21% (Statistics Canada 2017). The combination of the cost of falls and the growing population of older adults heralds the importance of further research trying to reduce falls in older adult populations.

2.2 Increased Fall Risk in Older Adults

Research has shown that increased fall risk is an aspect of normal ageing, and this is due to multiple risk factors that interact with one another (Ambrose, Paul, and Hausdorff 2013; Masud and Morris 2001; Organization 2007; Rubenstein and Josephson 2002). Due to the complex interplay between these risk factors, further research is needed to determine which factors are key contributors to falls. The Canadian Public Health Agency has classified these risk factors into four major categories: biological or medical, behavioural, environmental, and socio-economical (Public Health Agency of Canada, 2014). Within the biological and medical group of risk factors, impaired control of balance and gait is a crucial factor that can lead to instability and falls. In particular, age-related changes to the systems that control balance, such as sensory (somatosensory, visual, and vestibular), musculoskeletal and associated neural control systems, can lead to impaired ability to maintain an upright stance or to react to a sudden loss of balance. Decreasing control of balance has been identified as one of the major risk factors for falling, and further research could lead to improved screening of fall risk (Piirtola and Era 2006; Pizzigalli et al. 2016).

2.3 Quiet Standing and Balance

During quiet standing, the human body maintains a static upright position through posture and balance. Posture is an angular measure from vertical that describes the orientation of any body segment relative to the gravitational vector (Winter 1995). Balance describes the dynamics of body posture to maintain stability and prevent falling, and it can be both dynamic and static. Static balance describes the ability to keep the centre of gravity (COG) within the base of support (BOS) during quiet standing (Gage et al. 2004; Horak and Nashner 1986; Pollock et al. 2000; Winter et al. 1996). Dynamic balance describes the ability to prevent falling while the body is in motion where the COG is outside the BOS; examples include running and gait. The exception to this is the double support phase during gait (Winter 1995).

The COG of a body is the point at which all the linear forces and torques are balanced or in equilibrium. The COG is the point in which a body would balance without a tendency to rotate, and it is the point in which the weight of the body may be said to act (Hamilton and Luttgens 2012). The COG is often used interchangeably with the COM, but in regards to the COM, the COG is only one axis representing the force vector of gravity (Winter 2009). The COM is the point location of the body that is the net sum of the three-dimensional space vectors, which includes COG, and comprises and accounts for the distribution of all the body segments (Pollock et al. 2000; Winter 1995, 2009). Centre of Pressure (COP) is the point of application of the ground reaction force vector, which represents the sum of the vertical force and sheer force vectors on a force platform (Winter 1995, 2009). The trajectory of the COP is independent of the COM but acts as a neuromuscular response to the imbalances of the body's COM (Winter 2009).

The greater the displacement of the COM or the more substantial the external force that can be applied before an individual becomes unbalanced, the better the stability of the individual, which is known as postural stability (Pollock et al. 2000). During quiet standing, an individual's' postural stability can be quantified by measuring postural sway, which is the horizontal movement of the COG. Recordings of the COP and COM are used to calculate measures that quantify postural stability (Robertson et al. 2014).

The central nervous system (CNS) controls balance by receiving and processing appropriate sensory input and responding to it by sending motor commands through the somatic nervous system to skeletal muscles (Bugnariu and Fung 2007). Three main sensory systems send information to the CNS. They are the somatosensory, visual, and vestibular systems, and they work together in a complex interplay to send this information to the CNS. The visual system primarily senses external environmental visual cues to trigger, plan and execute adjustments to maintain an upright posture (Hay et al. 1996; Jeka, Allison, and Kiemel 2010; Pyykko, Jantti, and Aalto 1990; Saftari and Kwon 2018). Primarily, the visual system will use vertical and horizontal lines as visual input that assist in maintaining an upright posture. Similarly, the vestibular system acts to sense linear and angular accelerations to give information about the position of the head relative to gravity which assists to maintain upright posture (Allen et al. 2016; Khan and Chang 2013; Liston et al. 2014; Lobel et al. 1998; Park et al. 2001; Tang, Lopez, and Baloh 2001). The somatosensory system uses receptors in muscles, joints, skin, and other tissue spread throughout the body to sense things like the chemical environment, temperature, and physical forces. Of interest, mechanoreceptors sense position and rate of change of position (i.e. velocity) to facilitate postural adjustments of the body (Judge et al. 1995; Robbins, Waked, and Mcclaran 1995; Winter 1995). Of particular importance is the sensory input from the cervical spine to provide a reference for the vestibular system input about the position of the head and the lower extremities. The visual, vestibular, and somatosensory systems work in concert to perceive and respond to environmental stimuli to maintain balance.

2.4 Balance and Normal Aging

During the natural ageing process, the integration of sensory information decreases and reliance on visual information increases. One study removed visual stimulus during a test of quiet standing balance on a firm surface (CTSIB trials 1 & 2) and found an approximate doubling of sway velocity (Pyykko, Jantti, and Aalto 1990). Hay et al. found when comparing older adults to younger adults, postural stability in older adults was more affected by the removal of visual stimulus. In contrast, the alteration of proprioceptive information affected the two cohorts similarly (Hay et al. 1996). The study suggested that older adults rely more on visual stimulus than a proprioceptive stimulus to maintain balance. Another study compared younger adults, older adults, and fall prone older adults and found that older adults relied on visual stimulus more than younger adults and fall prone older adults tended to weigh visual stimulus higher than healthy older adults (Jeka, Allison, and Kiemel 2010). The findings of the three studies show the importance of visual stimulus in the maintenance of balance for older adults as they age. These findings demonstrate the need for any test that assesses balance or fall risk in older adults to include protocols that analyze vision. An example of this is trials 2 and 5 of the CTSIB.

Older adults display a reduced ability to adapt to diminished sensory information or inaccurate sensory inputs while maintaining balance. One study examined the influence of cutaneous sensory receptors. It showed a significant difference in postural control regulation between older adults with and without the sole of the plantar foot anesthetized (Do, Bussel, and Breniere 1990). Further research showed the diminished ability to respond to these incorrect sensory stimuli, which suggested an inability to increase the weighting of vestibular inputs (Hay et al. 1996). When assessing balance and fall risk in older adults, these studies demonstrate the importance of evaluating proprioceptive ability in the plantar foot region and the integration of vestibular inputs for maintaining balance.

2.5 Clinical Balance Assessments

Clinicians use targeted assessments specifically developed to identify salient characteristics in the maintenance of balance to discriminate fallers from non-fallers (Lusardi et al. 2017). These assessments can range from simple tests of muscular strength and endurance to tests of standing balance, to analyses of dynamic balance during gait, to comprehensive test batteries, which include elements of all four. The Five-times Sit-to-Stand (Teo, Mong, and Ng 2013), Five-Step Test (Murphy et al. 2003), and 30-second Sit-to-Stand (Jones, Rikli, and Beam 1999) are all representative of simple muscular strength and endurance tests which have been successfully used to identify older adult fallers. Examples of assessments of standing balance include the Functional Reach Test (FRT) which assess the individual's ability to maintain static balance while reaching (Duncan et al. 1990) to the Clinical Test of Sensory Interaction and Balance (CTSIB) (Shumway-Cook and Horak 1986) which assess the changes in standing balance when vision, proprioception, and vestibular systems are challenged. The Timed Up-and-Go (TUG) is commonly used by clinicians, due to its ease of use, to assess gait speed as an indicator of fall risk (Nordin et al. 2008). Comprehensive tests such as the Short Physical Performance Battery (SPPB), the Community Mobility and Balance scale (CB&M) (Balasubramanian 2015), Fullerton Advanced Balance Scale (FAB) (Rose, Lucchese, and Wiersma 2006), Berg Balance Scale (BBS) (Berg et al. 1992) utilize a battery of tasks to evaluate muscular strength and endurance, standing balance, and dynamic balance during gait to assess fall risk (Guralnik et al. 1994).

2.6 Clinical Test of Sensory Interaction (Organization) and Balance

Shumway-Cook and Horak developed the CTSIB to test patients with neurological problems using six different intersensory conditions (trials) that either eliminated or altered visual and somatosensory and, in particular, proprioceptive input meant to challenge patients postural control systems. Proprioceptive input was changed by having participants stand on foam in some trials. In trials 1 through 3, subjects stood on a firm surface; in trials 4 through 6, subjects stood on a foam surface. For trials 1 and 4, subjects stood with their eyes open; for trials 2 and 5, subjects stood with their eyes blindfolded; for trials 3 and 6, subjects wore a visual conflict dome to disrupt their vision and give them an incorrect visual stimulus. The dome was constructed using a paper lantern and a dental headband with eight vertical lines spaced 2.5 centimetres apart and an “X” at the approximate position of the patient's straight-ahead gaze. They did not normalize foot position across the participants, but normalized it within each participant, saying that each participant’s foot position should be “similar in each condition” (Shumway-Cook and Horak 1986).

Shumway-Cook and Horak concluded that for the treatment of patients with neurological disorders, it was essential to know which sensory input the individual was dependent on for sway orientation and how well the patient could adapt to reliance on the proprioceptive, visual, and vestibular sensory systems in a situation of intersensory conflict. To test this, they implemented trials 1 through 6. To quantify sway, they indicated three techniques for clinicians: 1) subjective assessment based on a numeric ranking system (e.g. 1 = minimal sway, 2 = mild sway, 3 = moderate sway, 4 = fall), 2) uses of a stopwatch to record the amount of time the patient maintained balance (maximum = 30 seconds), 3) use of a grid and a plumb line to record body displacement. Shumway-Cook and Horak found that most healthy adults could easily maintain their balance under all six conditions, while individuals with marked increases in sway or falls under conditions of sensory conflict (trials 3-6) indicated sensory interaction problems. For example, a fall during condition 3 (visual-conflict dome on a firm surface) suggested an abnormal reliance on vision to maintain balance. Shumway-Cook and Horak found that for most subjects, the amount of anterior-posterior sway increased over the six conditions, and the greatest sway was during trials 5 and 6 (Shumway-Cook and Horak 1986). They planned CTSIB as an adjunct tool to help clinicians assess balance in a clinical setting, but due to limitations, the test has not seen major use in clinical settings. These limitations include, but are not limited to, the bias of practitioners during clinical assessment, impracticality of measurement tools in the clinical setting (a grid and plumb line, and later a Pen-tail), and limitations with measurement variables (absolute displacement).

Cohen, Blatchly, and Gombash (1993) furthered research using the CTSIB to establish norms of performance for older adults. They created norms by dividing participants into three groups: one aged 25 to 44 years, the second aged 45 to 64 years, and the third aged 65 to 84 years. They found that for trials 5 and 6, group 3 had time scores that were significantly shorter than the other two groups. They also concluded that a score of 20 seconds on trials 4 – 6 with feet together was within normal limits for older adults. A drawback they found was that the CTSIB did not specify the exact nature of a subject balance problem. Still, it's simple and less expensive nature made it a valid addition to clinical balance assessments (Cohen, Blatchly, and Gombash 1993). However, the test still did not see widespread adoption clinically.

In 2009, Ricci et al. examined the difference between older adults who have never fallen, who have only fall once, and who have fallen on recurrent occasions. They found that older adults who have fallen on recurrent occasions have more cases where their balance was not maintained for 30 seconds during trial 5 (eyes closed on foam) compared to older adults who had fallen once. Furthermore, when comparing recurrent fallers to older adults who have not fallen, they concluded that recurrent fallers were more dependent on somatosensory and visual systems than older adults who had only fallen once or had not fallen at all (Ricci et al. 2009). Ricci’s findings indicated that sensorial interaction varied according to fall history.

In 2016, Criter & Honaker compared the sensitivity and specificity of the four trials of the modified CTSIB (mCTSIB). The mCTSIB followed all the same protocol as CTSIB, except mCTSIB did not include trials with altered visual stimuli (CTSIB trials 3 & 6). The mCTSIB tested eyes open standing on a firm surface (CTSIB/mCTSIB Trial 1), eyes open standing on a foam surface (CTSIB Trial 4/ mCTSIB Trial 3), eyes closed standing on a firm surface (CTSIB/ mCTSIB Trial 2), and eyes closed standing on a foam surface (CTSIB Trial 5/ mCTSIB Trial 4). Criter & Honaker wanted to assess Shumway-cook & Horak's original rating system, which was commonly used in the clinical setting. Participants were visually rated and scored by the following: 0 equalled little or no sway, 1 equalled mild sway, 2 equalled moderate sway, and 3 equalled the inability to perform the task or a fall. They found that the sensitivity and specificity had an inverse relationship. The highest sensitivity was for mCTSIB Trial 4 with a visual rating of 1 (91.7%), which has the lowest specificity (11.1%), whereas, mCTSIB Trial 3 with a visual rating of 3 had the highest specificity (55.6%) and the lowest sensitivity (50.0%) (Criter and Honaker 2016). It should be noted that only measures of total trial duration were used to calculate the validity of researchers' visually based scores. No research has examined whether sensitivity and specificity would be improved with the introduction of technologies that can measure postural sway during standing balance.

Originally, Shumway-cook and Horak designed the CTSIB to measure postural sway by using grids and a plumb line, and later, Cohen, Blatchly & Gombash updated the measurement tool to use a Pen-tail instead. Both systems were cutting edge for their time, but they had limitations that prevented their clinical application (Cohen, Blatchly, and Gombash 1993; Shumway-Cook and Horak 1986). The three primary limitations were measurement inaccuracy due to human or practitioner error, impracticality of measurement tools in the clinical setting (a grid and plumb line, and later a Pen-tail), and limitations with measurement variables (absolute displacement). Later, researchers found a major drawback of the CTSIB was that it did not specify the exact nature of a subject's balance problem (Cohen, Blatchly, and Gombash 1993).

Recently, different technologies like FPs and IMUs have made advancements that increased their ability to quantify postural sway accurately, which increases their clinical viability (Gill et al. 2001; Johansson et al. 2017; Swanenburg et al. 2010). Dawson et al. found that the FP demonstrated a strong test-retest reliability (ICC [3,1] = 0.75), during mCTSIB (Dawson et al. 2018). Both the FP and IMU reduced human error by having devices measure COM and COP and computers calculate measures of postural sway. Further, both the FP and IMU calculated absolute displacement as well as measures of velocity and acceleration, which improved the characterization of sway. These new technologies have the potential to reduce the major limitations, including human and practitioner error, the impracticality of measurement protocol in a clinical setting, and the limited measurement variables. Currently, FP’s and IMU’s have not been compared to determine which device is better characterizing postural sway during the CTSIB; refer to Appendix 1.

Further, there is a need to standardize the protocol for CTSIB as discrepancies that exist between different protocols could influence the outcomes. Some studies perform the mCTSIB, which tests trials standing on foam and a firm surface while the participant's eyes are open and close. Other research includes two extra trials where a third visual condition was included, where the participant’s eyes are open, while an object obstructs their vision (giving incorrect visual information). The type of object used to obstruct each participant's vision ranges from a Japanese-style lantern to goggles to the most common, a flip-down shield (Cohen, Blatchly, and Gombash 1993; Ricci et al. 2009; Shumway-Cook and Horak 1986). The lantern wraps entirely around the participant's head, whereas the shield only covers the face. This inconsistency could allow a participant to use their peripheral vision, which means their vision isn't wholly disrupted and would invalidate the test. More recently, some researchers are using head-mounted displays to disrupt participants' vision with a more complex visual stimulus (Alahmari et al. 2014). For trials, with eyes-closed, it is most common to have participants close their eyes when instructed by a practitioner. Some studies, however, used a blind-fold to obstruct the vision of the participant (Amor-Dorado et al. 2008).

Further, standardization of duration for the trials of CTSIB is required. Trial durations range from 10 seconds (Rahal et al. 2015) to no time limit except when the participant loses balance (Chung, Kim, and Yang 2016), but the most common duration was 30 seconds (Dawson et al. 2018; High et al. 2018; Nair et al. 2018). The number of attempts a participant gets varied as well, from one to three for each trial (Amor-Dorado et al. 2008; Moalla et al. 2008; Vaz et al. 2013). To account for the element of learning, researchers should standardize the number of attempts for a participant. To add variability in study results, researchers have not standardized the order of trials, which can vary from sequential, one to four or six, to random; refer to Appendix 1.

Footwear and foot position also needs to be standardized in the testing protocol. Whitney and Wrisley looked at the influence of footwear and foot position and found that for trials of CTSIB with eyes-open and eyes-closed on a firm surface and foam, there was no difference between the type of footwear and the foot position (Whitney and Wrisley 2004; Wrisley and Whitney 2004). Their research used a stopwatch to time each trial and averaged the results. This conclusion should be further researched using devices and metrics that can further characterize postural sway. In 2015, Chaikeeree et al. examined the interaction of foam types and age. They found sway in the mediolateral direction was significantly larger for older adults than younger adults as the density of foam decreased (Chaikeeree et al. 2015). They used an IMU (acceleration-based OPAL system) to characterize sway indicting more research should be done examining variability introduced at the foot by foam type, foot position, and footwear.

The unstandardized nature of CTSIB protocols creates variability in results affecting their interpretation and ability to make comparisons between results. The standardization of the test protocol and the implementation of new technology could improve the characterization of postural sway during CTSIB and increase the clinical utility of the test.

2.7 Technologies that Measure Balance: Force Plates

Force Plates (FP) measure three-dimensional force vectors that comprise a vertical component plus two shear components acting along the force plate's surface (Winter 2009). The shear forces are usually resolved into anterior-posterior and medial-lateral directions. Commonly, four triaxial transducers located at the corners of the plate are used to calculate COP, which is derived from the forces and location of the transducers relative to the other transducers (Winter 2009). A force plate uses either piezo-electric crystals, strain gauges, or capacitors to convert the forces to electrical signals (Robertson et al. 2014). It is essential to select a force plate with the capacity and sensitivity (measurement range and resolution) to accommodate for the ground reaction forces and level of precision that are expected for the experiment. There is a trade-off when doubling the capacity; the sensitivity is halved. Static balance applications generate tiny forces and involve little visible movement. Vertical loads seldom exceed body weight. The horizontal forces are minimal, and therefore, it is best to choose the lowest capacity and the highest sensitivity of a force plate for these experiments (Robertson et al. 2014; Winter 2009).

Originally, Kapten et al. proposed the standardization of force plate, which they called posturography and defined as the "recording of body movements of subjects in standing position" (Kapteyn et al. 1983). They outlined the general procedures to be used when recording postural sway with a FP that included the designations of body displacement (Anterior, Posterior, Left, and Right), and they defined a stabilogram as body movement as a function of time. They suggested that the time of recording should begin 10 seconds before the time of analysis to eliminate transient phenomenon, and scale units should be in Newtons-metre-second (Kapteyn et al. 1983). In 2012, Scoppa et al. reviewed the general procedure outlined by Kapteyn et al., and while keeping the general procedures, they updated two critical elements. They concluded that the acquisition interval should not be less than 25 seconds, and the sampling frequency should be at least 50 Hz (Scoppa et al. 2012). They concluded that with these standards, it would ensure technical performances for accuracy of better than 0.1 mm, precision better than 0.05 mm, resolution higher than 0.05 mm, and linearity better than 90% over the whole range of measurement parameters (Scoppa et al. 2012).

In 1996, Prieto et al. evaluated the relative sensitivity of FP measures of postural sway related to age. Under eyes-closed and eyes-open conditions, they compared twenty healthy young adults (21-35 years) and twenty healthy older adults (65-77 years) for a variety of time and frequency measures. They found that COP measures of mean, velocity, and elliptic area could characterize differences between young and older adults during both conditions (Prieto et al. 1996). The impact of vision was also demonstrated by Choy, Brauer, and Nitz who looked at four-hundred and fifty-three women aged between 20 and 80 years and found that for each decade when the participant's eyes were closed in a single-leg stance, there was a correlated decline in postural stability (Choy, Brauer, and Nitz 2003). These findings were further supported by a study between young and older adults when testing the age differences for conditions with eyes-open and eyes-closed where they found significant differences in area, length, and mean velocity of COP (Benjuya, Melzer, and Kaplanski 2004). Later, Vieira, Oliveira, & Nadal examined the age-related changes in quiet standing FP measures of fifty-seven participants divided into young (19-29 years), middle-aged (38-51 years), and older adults (65-73 years). They found no differences between the COP time series and mean velocity, which contradicted the previous studies mentioned above. They found that only age-related differences for the frequency band that encompassed 80% of the area under the COP power density spectrum could detect differences between young and old cohorts (Vieira, Oliveira, and Nadal 2009). This contradiction showed the need for further study of the measures of postural sway.

Recently, with the improvement of new technologies, researchers have been expanding on traditional FP measures. One such study investigated both traditional standing balance measures and dynamic measures. They used principal component analysis to quantify the contributions of each measure. They found that measures of sample entropy, central tendency measures, variability, length, and area in both AP and ML directions were complementary for the characterization of postural sway in older women (Tallon et al. 2013). Another study divided a group of community-dwelling older adults (70+ years) into quantiles looking at COP total sway path length with eyes-closed and eyes-open. They hoped to develop cut-off scores for prospective fallers, single-faller, multi-faller and non-faller classifications. They looked at measures of range in AP and ML COP motion, AP and ML COP root-mean-square distance from the mean, and vector sum magnitude of COP velocity (Howcroft et al. 2017). The improvements in technology have increased the number of options to characterize postural sway (balance). Still, no studies have directly compared the measurement options to determine which are best to characterize sway during CTSIB. Further, no studies have looked at technology's abilities to characterizer sway of an individual when different sensory systems are perturbed. Studies have focused on finding differences between measurement tools and the cohort of participants; see Appendix 1.

A FP is considered the gold standard for measures of COP, but the device still has some sources of error. These sources of error should be understood to reduce the amount of signal variability from a device. A FP measures GRF and calculates COP measurements, but they can not describe the location of peak pressure under the foot. A source of signal variability is the constant component of gravity that can be minimized by subtracting the signal mean and correcting for the offset. To reduce variability, measurement standards were established in collection and analysis of FP data that included recording 10 seconds before the time of examination to eliminate transient phenomenon, and the use of a standard set of units for the reporting of FP signal outputs (Kapteyn et al. 1983). Later, researchers concluded that the acquisition interval should not be less than 25 seconds, the sampling frequency should be at least 50 Hz and the general procedure outlined by Kapten should be maintained (Scoppa et al. 2012). A low-pass double Butterworth filter can be used to remove signal artifacts not associated with postural sway and control for phase shift. It is essential to choose a sampling frequency that is two times the frequency expected in the activity the participant is performing, which follows the Nyquist Sampling Theorem (Robertson et al. 2014; Winter 2009). It has been shown that standard COP characteristics are strongly dependent on individual experimental design and are susceptible to distortions such as the noise of signal digitalization, which often makes the results from different laboratories incomparable and unreliable (Błaszczyk 2016).

2.8 Technologies that Measure Balance: Inertial Measurement Units

Inertial measurement units (IMU), use accelerometers that measure the change in acceleration forces caused by gravity. They use gyroscopes to measure angular velocity and sometimes a magnetometer to measure changes in the magnetic field (Robertson et al. 2014). To quantify quiet standing, IMUs are generally placed in the lumbar region of a participant's torso to approximate sway of the COM (trunk sway); unfortunately, no exact location has been established. Instead, placement of the sensor ranges from the third lumbar vertebrae to the first sacral vertebrae, with the majority of researchers using the third vertebrae (Borg et al. 2007; Ozinga and Alberts 2014; Turcot et al. 2009; Yokoyama et al. 2002). Early research modelled quiet stand with frequencies ranging from 0 to 55 Hertz (Borg et al. 2007; Yokoyama et al. 2002), but the improvement of sensors has allowed for the recording of frequencies up to 200 Hertz (Mancini et al. 2012; Ozinga and Alberts 2014).

In 1998, Moe-Nilssen found that an IMU with a single axis accelerometer placed in the lumbar region was a reliable measure of acceleration of the trunk during walking and standing (Moe-Nilssen 1998). In 2015, Chaikeeree et al. used an acceleration-based IMU system to examine the interaction of age and foam type during the mCTSIB. They quantified postural sway by root-mean-square of COM acceleration in the ML and AP direction between young (21.6 ± 3.3 years) and older (53.2 ± 4.9 years) adults. Four types of foam were examined, and as the density of the foam increased, the firmness decreased, and the sway excursions of both groups increased. The results showed that root-mean-square in the ML direction in older subjects was larger than the younger subjects ($p \leq 0.001$), particularly on the foam with the highest density and lowest firmness ($p = 0.001$) (Chaikeeree et al. 2015). This research demonstrated the importance of the type of foam selected and, more importantly, IMU's were a good tool to measure the differences in postural sway between young and old adults.

In 2001, two studies examined the ability of angular-velocity transducers to calculate trunk sway (COM) measures of postural stability during clinical balance tests. Both studies examined angular velocity (pitch and roll), angular range (pitch and roll), and trial duration for two-legged stance, one-leg stance, semi-stance and gait tasks. The tasks were performed with eyes-open, eyes-closed, standing on a firm surface, and standing on a foam surface for a two-legged stance (Allum et al. 2001; Gill et al. 2001). The data analysis consisted of measuring the range of pitch and roll angular displacement and velocity. The range was measured in two ways. The first method used the peak-to-peak extent of the values in the roll and pitch directions following the removal of the first second and the last two seconds for each trial. The second method involved binning all values together and creating a histogram for each measure examined. From these histograms, 5% and 95% limits were calculated, and the extent of these limits was assigned to a 90% range value (Gill et al. 2001). Gill's research was able to find differences when comparing older adults with the middle-aged and younger adults for roll angle (ML range), pitch angle (AP range), roll velocity, and pitch velocity during trials on a firm surface with eyes-open (CTSIB Trial 1) and eyes-closed (CTSIB Trial 2). During CTSIB Trial 4, where the participant was standing on a foam surface with eyes-open, differences were only found for pitch angle between cohorts. For CTSIB Trial 5, when the participant's eyes were closed on a foam surface, differences were found for pitch angle and velocity (Gill et al. 2001).

An IMU has several possible sources of error that need to be understood to use the device properly. Gravity can be a source of signal variability that can be minimized by subtracting the signal mean and correcting for the offset. Another source of error is the acceleration caused by visceral movement due to the respiration of the participant, which can be controlled by locating the IMU in the lower lumbar region (Robertson et al. 2014; Winter 2009). When an IMU is used to record postural sway, the electrical signal can create artifacts due to skin movement, incorrect digitization, electrical interference, and moving wires, so it is crucial to select the correct filters to remove these artifacts (Robertson et al. 2014).

2.9 Measures Postural Sway

Recently, technologies that quantify postural sway by measures of COM or COP have been improved. The ability to measure and/or calculate new metrics of COM and COP in several kinematic variables with increased accuracy is now possible. Both improvements in the portability of FPs and wearable technologies, like IMUs, have improved to a level where they are capable of research-grade data with high sample rates and high signal to noise ratios (Moe-Nilssen 1998). These improvements allow for the integration of signals to calculate linear and angular kinematics with minimal measurement error (Masani et al. 2014). Linear kinematics of postural sway, such as velocity and displacement, have been used to characterize the differences between cohorts of young and older adults using the CTSIB (Alahmari et al. 2014; Burke, Franca, Meneses, Cardoso, Pereira, et al. 2010; Desai et al. 2010; High et al. 2018; Rombaut et al. 2011). In 2008, Swanenburg et al. concluded that COP FP measures of the maximal (ML & AP directions) and root-mean-square amplitude (ML & AP directions) and AoE_B were reliable at differentiating older adult fallers from non-fallers (Swanenburg et al. 2008). Later, Melzer et al. used a retrospective analysis of COP displacement, measured with a FP, in older adult fallers and non-fallers where they found that older adult fallers had higher sway amplitudes in the ML direction, but not in the AP direction. They concluded that ML sway amplitude seen in narrow stance conditions was a good measure of balance control to identify older adult fallers (Melzer, Kurz, and Oddsson 2010). Some studies have also explored the utility of metrics in different kinematic domains. Still, no studies have compared these metrics between kinematic domains during the conditions of the CTSIB. See Appendix 1 for comparisons.

Masani et al. compared FP measures of COP displacement with a laser sensor measure of COM displacement and later converted both devices signals to the kinematic domains of velocity and acceleration. For each kinematic domain, only mean measures in the AP direction were considered during quiet standing with eyes open. Comparing these kinematic measures revealed that the COP velocity was correlated with the COM velocity, but more highly correlated with the COM acceleration (Masani et al. 2014). Further research needs to be conducted to explore if these conclusions are valid for measures in the ML directions for older adult populations.

Masani's findings suggest that COM acceleration could be used in place of COP velocity, which has previously been shown to be the most sensitive metric for detecting changes in balance due to ageing (Era et al. 2006; Masani et al. 2007; Prieto et al. 1996). This general understanding of the value COP velocity has led many researchers to rely on this kinematic metric exclusively for studies involving the CTSIB (Bulat et al. 2007; Burke, Franca, Meneses, Cardoso, Pereira, et al. 2010; Hill et al. 2010; Rahal et al. 2015; Ramirez, Nogal, and Casado 2008). While it is valuable to include COP velocity as a metric when measuring postural sway, Masani's findings showed the need for continued research comparing kinematics of displacement, velocity and acceleration for measures in both the ML and AP direction, and AoE for both COP and COM kinematics. Until the results of studies comparing measures of postural sway in displacement, velocity, and acceleration have been thoroughly explored, researchers need to be cautious when limiting the measures employed. As an example, past research has shown that older adult fallers had higher sway amplitudes in COM acceleration in the ML direction (Chaikeeree et al. 2015). Without examining COM acceleration, this characteristic of postural sway would be missed. Researchers should be careful not to leave out valuable postural sway characteristics that may improve the identification of features of balance associated with ageing and help predict fall risk.

Further, in the past five years, several studies have used only a timing device, such as a stopwatch, during the conditions of CTSIB to examine different cohorts and compare CTSIB to other tests of balance for their ability to assess fall risk; see Appendix 1 (Ahmed et al. 2016; Amor-Dorado et al. 2017; Criter and Honaker 2016; Dannenbaum et al. 2016; Horn et al. 2015; Jirikowic et al. 2016; Moghadam et al. 2015; Nair et al. 2018). Recently, technologies such as FPs and IMUs have made advancements that increase their ability to quantify postural sway accurately and make them more clinically viable (Gill et al. 2001; Johansson et al. 2017; Swanenburg et al. 2010). Using only measures of time limits, the characterization of postural sway and provides only a gross representation of balance ability. Both the FP and IMU reduce human error by having automated timing measures, and they increase the richness of measurement by calculating kinematic measures, such as absolute displacement, velocity, and acceleration throughout the entire trial as well as within each individual sway. These improvements could lead to the improved utility of CTSIB in clinical settings and an improved tool for fall risk assessment and prediction.

Mulavara et al. compared the Sensory Organization Test (SOT) to the CTSIB for measures of linear acceleration (AP and ML directions) and tri-axial angular velocity. They found that the CTSIB was more likely than the SOT test to indicate balance deficits between healthy adults and adults with vestibular disorders (Mulavara et al. 2013). Macedo et al. looked at measures of displacement for COP AoE_B and mean velocity of trunk sway. They used these measures to compare older adults with vestibular disorders with healthy age-matched controls and found that both measures differentiated the cohorts during trials 5 of the CTSIB (Macedo et al. 2015).

Howcroft et al. examined static posturography to predict elderly fall risk. They hoped to develop appropriate outcome measure cut-off scores for prospective fallers, single-faller, multi-faller and non-faller classifications. They looked at measures of range in AP and ML COP motion, AP and ML COP root-mean-squared distance from the mean, and vector sum magnitude of COP velocity. They were able to achieve an accuracy of 84.9% using measures of Eyes-closed AP velocity, Eyes-closed vector sum magnitude velocity, root-mean-square AP velocity, and vector sum magnitude velocity with a cut-off score of 0.541 to screen older people at risk of multiple falls (Howcroft et al. 2017). High et al. examined measures of displacement (COP total path length, COP AoE_B, COP mean velocity, and alpha values in ML & AP directions) and found no differences between healthy older adults and older adults at high risk of falls (High et al. 2018). The conflicting results of these studies demonstrate the importance of future research to compare measures in different kinematic domains with each other to understand better what measures best characterize postural sway during CTSIB. Additionally, replication studies are warranted where agreement is not observed for the utility of postural sway metrics to classify cohorts correctly.

Chapter 3 Manuscript

3.1 Introduction

Postural stability is integral to an individuals' ability to stand and walk and often deteriorates with age, impairing gait function and leading to reduced mobility, independence and quality of life (Gill et al. 2001; Pizzigalli et al. 2016). Having the means to measure and characterize postural stability accurately allows this critical function to be monitored and could be used to screen for fall risk as well as guide timely clinical interventions. The Clinical Test of Sensory Interaction and Balance (CTSIB) was originally designed to assess the sensory systems (proprioception, vision, and vestibular) that contribute to postural control by using six distinct quiet standing trials that include eyes open, eyes closed and a visual conflict dome to alter the visual stimulus on both firm and foam surfaces (Shumway-Cook and Horak 1986). Originally, to quantify postural sway, clinicians and researchers employed three techniques: 1) subjective assessment based on a numeric ranking system; 2) use of a stopwatch to record the amount of time the patient maintains balance; 3) use of Pen-tail to record body displacement (Cohen, Blatchly, and Gombash 1993; Ricci et al. 2009; Shumway-Cook and Horak 1986). More recently, researchers have instrumented the CTSIB using either a force plate (FP) or an inertial measurement unit (IMU) to measure postural sway, which has mitigated some of the measurement limitations (Swanenburg et al. 2010). However, a FP provides outputs of centre of pressure (CoP) kinematics while an IMU worn at the lumbar spine provides outputs of centre of mass (COM) kinematics, and it is currently not clear which postural sway characteristics are most indicative in detecting changes in postural stability.

Though estimates of postural sway can be derived from both CoP and CoM measures, they are not synonymous and provide unique, but related information about kinematics relevant to postural stability. Further CoP and CoM can both be provided in different kinematics (e.g. displacement, velocity, acceleration). For example, FP data can be used to calculate CoP displacement (COP_{FPd}), and its first derivative yields COP velocity (COP_{FPv}), while the IMU can be used to calculate CoM acceleration (COM_{IMUa}) and its first integral yields CoM velocity (COM_{IMUv}). Additionally, COM can be measured using a simple Pen-tail to record body displacement ($COM_{PENd)}$. It is currently unclear which CoP or CoM measures are most sensitive in detecting changes in postural stability.

In addition, within the kinematics of acceleration, velocity and displacement, there is an array of metrics, such as maximal sway, sway mean, sway standard deviation and area of an ellipse that can be calculated to quantify postural sway. Different metrics have proven valuable to detect changes in tasks and populations. For example, CoP path length, CoP velocity and sway in medial-lateral (ML) and anterior-posterior (AP) are variables that can distinguish older adult fallers from non-fallers (Pizzigalli et al. 2016). It is currently not clear which kinematic metric is best able to capture key features of postural sway that are most relevant to characterize postural stability.

We aim to determine which kinematics (acceleration, velocity and displacement) of which measure (CoP or CoM) is best suited to differentiate balance performance between CTSIB trials. This will enhance the capacity of the CTSIB to detect changes in postural stability over time as well as identify individuals with deficits in postural control, including older adults at risk of falls.

3.2 Methods

3.2.1 Participants

Thirty-nine community-dwelling older adults' (70+ years old) with no history of neurological, metabolic, cardiovascular, or musculoskeletal disorders volunteered to participate in this study. Participants were screened with the Physical Activity Readiness Questionnaire (PAR-Q), and physician approval was obtained if participants reported 'yes' to any question. Participant characteristics are outlined in Table 1. All participants participated with written informed consent, and the study was conducted in accordance University of Victoria Human Research Ethics Board.

Table 1: Participant Characteristics

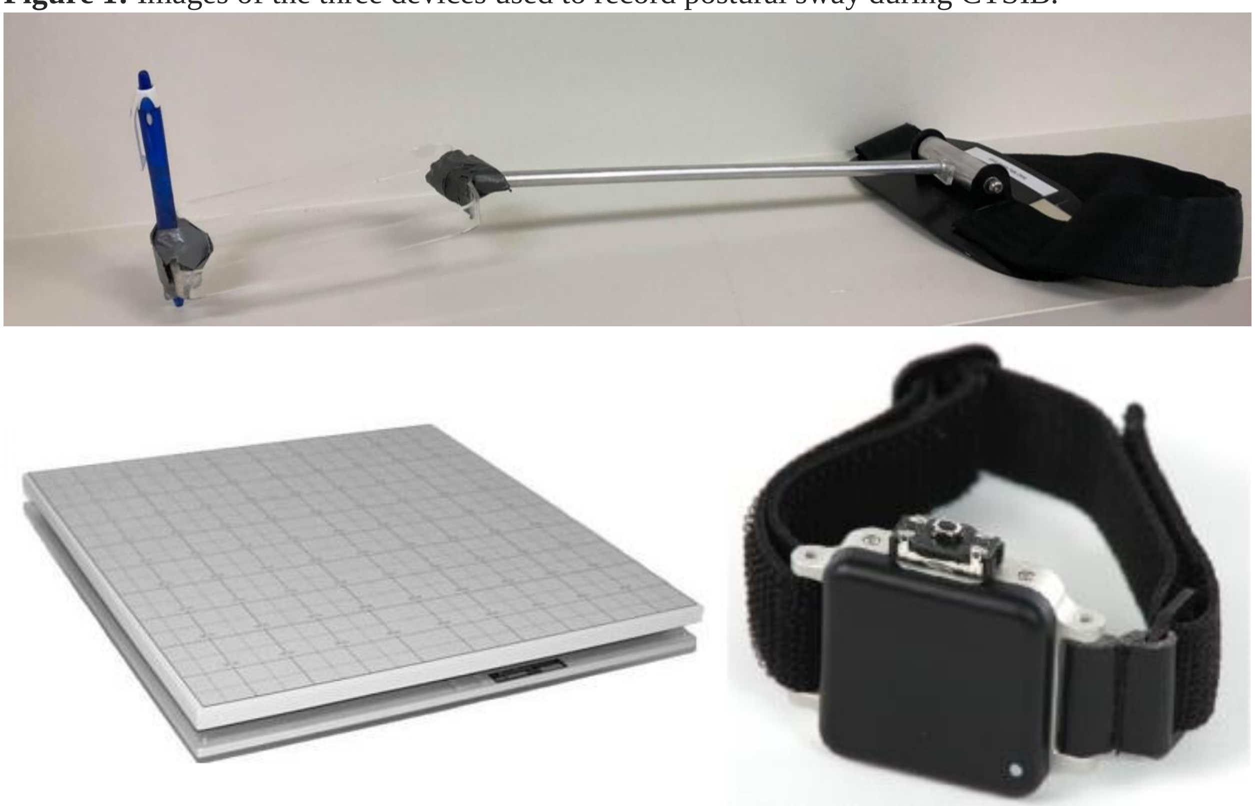
		Age (yrs)		Height (cm)		Weight (kg)		MMSE (#)	
	n	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*
Female	24	76	*3.3*	162.7	*7.9*	70.6	*15.8*	28.7	*1.6*
Male	15	76.80	*3.7*	175.2	*6.1*	81.7	*20.4*	28.5	*1.3*
Total	39	76.3	*3.4*	167.5	*9.4*	74.9	*18.3*	28.6	*1.5*

Abbreviations: n = number of sample, yrs = years, cm = centametres, kg = kilograms, MMSE = Mini Mental State Examination, # = MMSE score, SD = standard deviation

3.2.2 Protocol

The CTSIB protocol was employed as outlined in Shumway-Cook & Horak (1986), where participants stood without shoes on a firm surface, feet together, arms crossed with hands-on opposite shoulders for six trials total. The first three trials respectively comprised eyes open, eyes closed and viewing a visual conflict dome that included a face shield with a visual target centred in the middle of the participants' field of vision. These three conditions were repeated while the participant stood on a medium-density foam rubber mat (15 cm thick) for a total of 6 trials. During all trials, three devices were used to record direct measures of postural sway. An AMTI AccuSway system comprising a portable force platform and AMTI's Balance Clinic software (Advanced Mechanical Technology, Inc., Watertown, MA) measures forces (Fx, Fy, Fz) sampled at a 200 Hz and then calculates CoP displacement. An IMU system comprising of a tri-axial accelerometer and a mechanical gyroscope sampled at 200 Hz (+/- 4g's), was placed at the 5th Lumbar spinous process to measure acceleration and angular velocity of trunk movements and then calculate trunk (approximating COM movement) acceleration. A Pen-tail consisting of a belt apparatus holding a wand (52 cm) and a pen that recorded trunk movements on a sheet of paper and was used to calculate trunk displacement. See Figure 1 for the measurement apparatus.

Figure 1: Images of the three devices used to record postural sway during CTSIB.



Note: Top = pen-tail device, bottom left = force plate, bottom right = inertial measurement unit.

3.2.3 Data Analysis

Custom-written Labview software was used for all FP and IMU data signal processing and analysis. FP data were low pass filtered at 20 Hz using a zero-phase 2nd order Butterworth filter. IMU data were high pass filtered 0.4 Hz using a 2nd order Elliptic filter and a lowpass Cascade filter (0.4 Hz). Nine metrics defined in Table 2 (e.g. Medial-lateral mean excursion) were calculated across the six trials of CTSIB for each COP or COM measures (COP_{FDd}, COP_{FDv}, COM_{IMUa}, COM_{IMUv}). Three metrics were calculated for COM_{PENd} (i.e. Sway and Area of Elipse Bounded). The data for each metric for each measure was normalized to the mean of their respective trial 1 to standardize values for analyses. Repeated measures analysis of variance (Statistica software vs. 13, TIBCO Software, Palo Alto, USA) was conducted across the nine metrics (3 for COM_{PENd}) and the six trials of CTSIB separately for each measure for main effects for trial and significant interactions between trials and metrics. Post hoc analysis was conducted on significant interactions ($p < 0.05$) for main effects and interactions using Tukey's Honest Significant Difference test. Effect sizes for comparisons between significant trials were calculated using Hedge's G.

Table 2: Definitions of each metric of postural sway calculated during the CTSIB.

Metric	Definition
$Sway_{ML}$/ $Sway_{AP}$*	The max excursion (i.e. range) in ML (left-right) and AP(front-back) directions, calculated in radii.
AoE_B*	Product of half max excursion in AP and ML directions times pi.
AoE_E	Calculated from a covariate matrix using the orthogonal first and second eigenvalues. Comparable to a 95% confidence interval of AoE_B.
$Mean_{ML}$/ $Mean_{AP}$	Mean of the individual max excursions in the ML/ AP directions.
SD_{ML}/ SD_{AP}	The standard deviation of the individual max excursions in the ML/ AP directions.
TSP	Sum of the distance covered.

*Metrics measured by the Pen

Abbreviations: ML = medio-lateral, AP = anteroposterior, COM = Centre of Mass, COP = Centre of Pressure, FP = Force Plate, IMU = Inertial Measurement Unit, Pen = Pen-tail device, AoE_B = area of ellipse bounded, AoE_E = area of ellipse eigenvalue, SD = standard deviation, TSP = total sway path.

3.3 Results

All ANOVA's for each kinematic measure yielded significant main effects for trial, metric and trial by metric interactions. Post hoc analysis for the trial main effects showed no differences between trials 2-3, 3-4 and 5-6 for COP_{FPd}, trials 1-4, 2-3, 3-4 and 5-6 for COP_{FPv}, trials 2-3, 2-4, 3-4 and 5-6 for COM_{IMUa}, trials 2-3 and 5-6 for COM_{IMUv} and trials 2-3, 2-4, 3-4 and 5-6 for COM_{Pend}. Post hoc results from trial by metric interactions are summarized in Table 3.

Out of 135 comparisons, the number of significant differences between CTSIB trials was seen in COP_{FPd} (56), followed by COM_{IMUv} (54), COP_{FPv} (48), and COM_{IMUa} (28). For COM_{Pend} 6 out of a possible 45 significant difference between CTSIB trials.

When considering all measures and metrics, there were the most differences observed between trials 1x5 (32/39) and 1x6 (32/39); this was followed by trials 4x5 (28/39) and trials 4x6 (19/39).

Within COM_{IMUv} and COP_{FPv} significant differences were found for all 9 metrics for trials 1x5, 1x6, 4x5. Further for trials 1x5, 1x6, 4x5 for both COM_{IMUv} and COP_{FPv} the effect sizes were all large across the 9 metrics with the following ranges: COP_{FPv} 1x5 - 1.0-1.6; 1x6 – 1.0-1.5: 4x5- 0.8-1.1; COM_{IMUv} 1x5 - 0.8-1.4; 1x6- 0.8-1.3; 4x5- 0.7-1.0. As shown in Table 3.

From the nine metrics, AoB_B has the most differences (49/75) between CTSIB trials across all five measures. However, the largest or second-largest effect sizes were seen in Sway AP and, in particular, between trials 1x5 and 1x6 across all the comparisons. The amplitude of AP Sway across the CTSIB trials for the five different measures (COMIMUa, COMIMUv, COPFPd, COPFPv, COMPENd) is displayed in Figure 2. The amplitude of Area of Ellipse Eigen Values across kinematic measures (COM_{IMUa}, COM_{IMUv}, COP_{FPd}, COP_{FPv}) when across CTSIB trials are displayed in Figure 3.

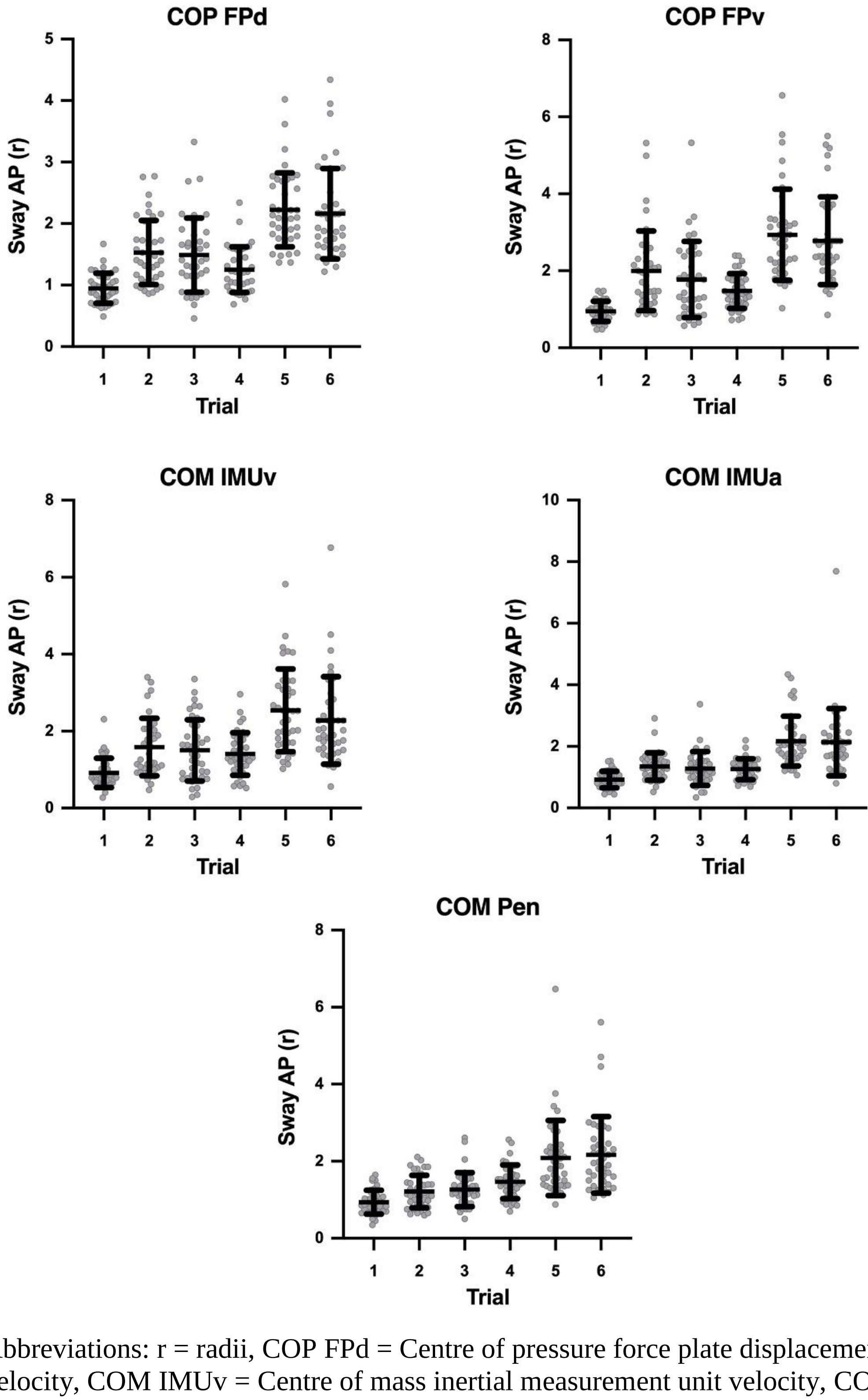
Table 3: Effect Size and significant findings of kinematic measures (COM_{IMUa}, COM_{IMUv}, COP_{FPd}, COP_{FPv}, COM_{PENd}) when comparing trials of CTSIB.

	Trials	$Sway_{ML}$	$Sway_{AP}$	AoE_B	AoE_E	$Mean_{ML}$	SD_{ML}	$Mean_{AP}$	SD_{AP}	TSP		$Sway_{ML}$	$Sway_{AP}$	AoE_B	AoE_E	$Mean_{ML}$	SD_{ML}	$Mean_{AP}$	SD_{AP}	TSP		$Sway_{ML}$	$Sway_{AP}$	AoE_B
COM IMU a	1x2										COM IMU v			0.8***	0.8***	0.9*					COM Pend			
	1x3													0.8***	0.8***									
	1x4																							
	1x5	0.8**	1.4***	0.7***	0.9***				1.4***			1.0***	1.4***	0.8***	1.0***	1.2***	1.1***	1.2***	1.3***	1.4***				0.8***
	1x6	0.9***	1.0***	0.6***	0.7***				1.3***			1.2***	1.1***	0.8***	0.9***	1.0**	1.2***	1.1***	1.3***	1.3***				0.8***
	2x3																							
	2x4													0.5**										
	2x5			0.5***	0.7***				0.9**				0.7**	0.5***	0.6***			0.7***	0.7**					0.6***
	2x6			0.5***	0.6***				0.9*					0.4***	0.4***									0.6***
	3x4																							
	3x5			0.6***	0.7***				0.9**				0.7***	0.5***	0.6***			0.7***	0.7**					0.6***
	3x6			0.5***	0.6***				0.8*					0.4***	0.5***									0.6***
	4x5		1.0*	0.6***	0.7***				0.9**			0.7***	0.9***	0.7***	0.8***	1.0***	0.9**	0.9***	0.8***	1.0**				
	4x6			0.5***	0.5***								0.7**	0.6***	0.7***			0.7*	0.7**					
	5x6													0.1**	0.2**									
COP FP d	1x2			0.9***	0.8***			0.9*	0.9*		COP FP v			0.7***	0.7***									
	1x3			0.9***	0.8***			0.7*						0.9***	0.9**									
	1x4			0.8***	0.8***																			
	1x5	1.6***	1.9***	1.3***	1.1***		0.7***	1.4***	1.4***			1.4***	1.6***	1.0***	1.0***	1.3***	1.5***	1.4***	1.5***	1.4***				
	1x6	1.6***	1.5***	1.1***	1.1***		1.2***	1.2***	1.2***			1.3***	1.5***	1.0***	1.0***	1.4***	1.4***	1.4***	1.4***	1.4***				
	2x3													0.3**										
	2x4													0.5***	0.5***									
	2x5	0.8**	0.8**	0.7***	0.7***			0.7***	0.7**					0.5**	0.5**									
	2x6	0.9**	0.7*	0.7***	0.7***									0.4**	0.4**									
	3x4																							
	3x5	0.9**	0.8**	0.8***	0.7***			0.7***	0.7**					0.7***	0.6***									
	3x6	0.9**	0.7*	0.7***	0.7***		0.7*							0.7***	0.6***									
	4x5	0.4***	0.4***	0.6***	0.6***			0.5***	0.4***			1.0**	1.1**	0.8***	0.8***	1.0***	1.1**	1.0**	1.1*	1.0***				
	4x6	1.1***	1.1***	0.8***	0.8***			0.9***	0.9***			0.9**		0.8***	0.8***	1.0**	1.0*			1.0**				
	5x6																							

Effect size: 0.2 ≥ small < 0.5 ≥ medium < **0.8 ≥ large**. Significant findings: <0.05 = *, <0.01 = **, <0.001 = ***

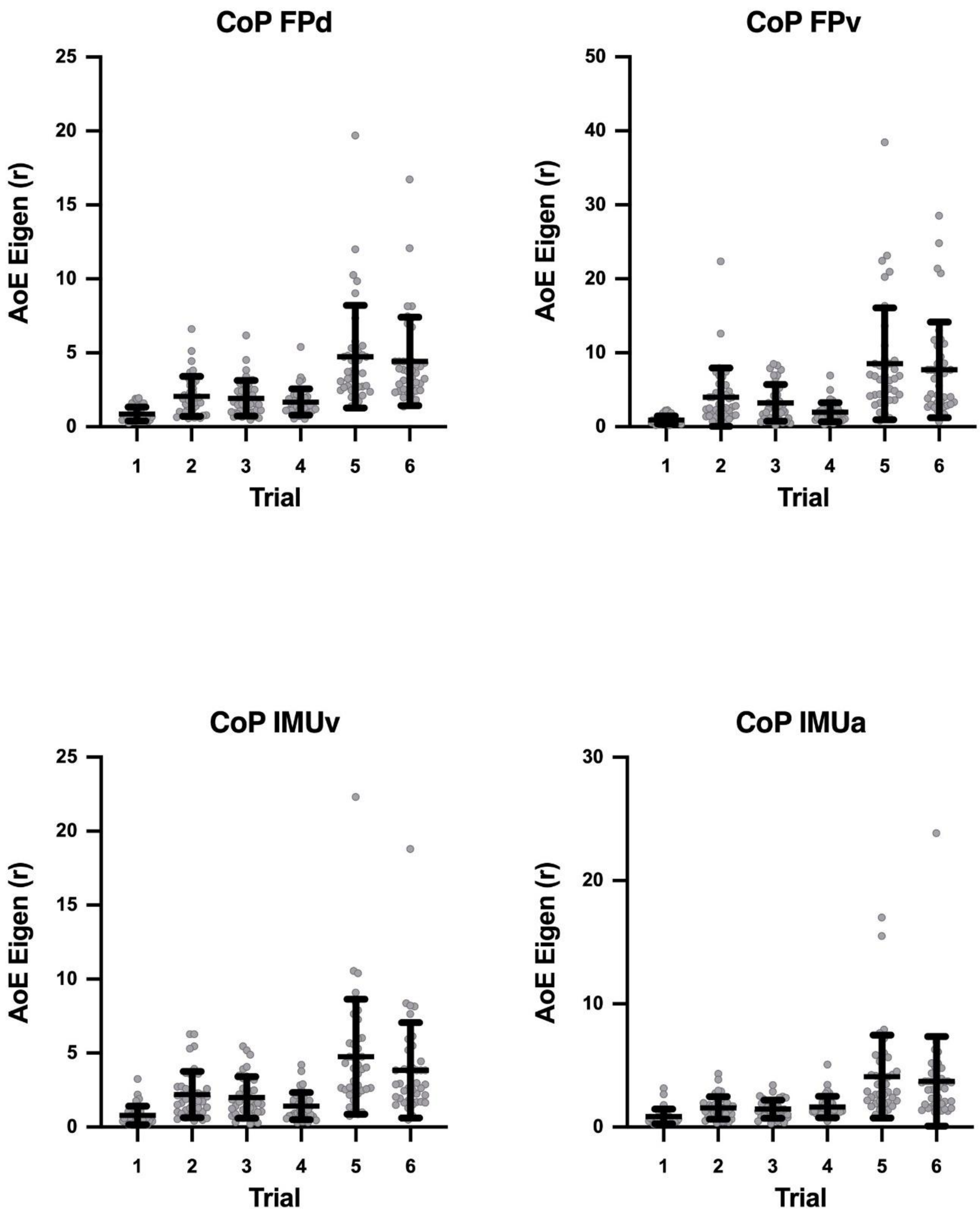
Abbreviations: COM_{IMUa} = Centre of mass inertial measurement unit acceleration, COM_{IMUv}, = COM IMU velocity, COP_{FPd} = Centre of Pressure force plate displacement, COP_{FPv} = COP FP velocity, COM_{PENd} = COM Pen-tail device displacement, ML = medio-lateral, AP = anteroposterior, AoE_B = area of ellipse bounded, AoE_E = area of ellipse eigenvalue, SD = standard deviation, TSP = total sway path.

Figure 2: Amplitude of Sway in the Anteroposterior direction across kinematic measures (COM_{IMUa}, COM_{IMUv}, COP_{FPd}, COP_{FPv}, COM_{PENd}) when comparing trials of CTSIB.



Abbreviations: r = radii, COP FPd = Centre of pressure force plate displacement, COP FPv = velocity, COM IMUv = Centre of mass inertial measurement unit velocity, COM IMUa, = acceleration, COM Pen = COM Pen-tail device displacement, AP = anteroposterior.

Figure 3: Area of Ellipse Eigen Values across kinematic measures (COM_{IMUa}, COM_{IMUv}, COP_{FPd}, COP_{FPv}) when comparing trials of CTSIB.



Abbreviations: r = radii, COP FPd = Centre of pressure force plate displacement, COP FPv = velocity, COM IMUv = Centre of mass inertial measurement unit velocity, COM IMUa, = acceleration, AoE Eigen = Area of Elipse Eigen Value.

3.4 Discussion

This study is the first to compare different measures and metrics across trials of the CTSIB and determine which measures and metrics are best suited to differentiate postural performance. Overall, we have four major findings. First, both COP and COM measures were capable of detecting differences between trials of the CTSIB. Second, a greater number of differences were seen between certain trials, primarily trials 1-5 and 1-6, by more measures and metrics. Next, there were a greater number of differences in AoE metrics (AoE_B and AoE_E) between trials across all measures. Finally, AP and ML metrics had the largest effect sizes across trials and specifically between trials 1-5 and 1-6 comparisons.

3.4.1 Utility of COP and COM Measures across all CTSIB trials and metrics.

Previous research examining instrumented measures of postural sway during CTSIB have commonly used either COM or COP displacement or velocity (Bulat et al. 2007; Burke, Franca, Meneses, Cardoso, and Marques 2010; Darlington et al. 2000; Hill et al. 2010; Rahal et al. 2015; Ramirez, Nogal, and Casado 2008; Rombaut et al. 2011; Shumway-Cook and Horak 1986; Whitney, Poole, and Cass 1998). COP velocity had previously been shown to be the most sensitive metric when detecting changes in balance due to ageing (Era et al. 2006; Masani et al. 2007; Prieto et al. 1996). While it is valuable to include COP velocity as a measure, many researchers have relied only on this measure for studies that involved the CTSIB (Bulat et al. 2007; Burke, Franca, Meneses, Cardoso, Pereira, et al. 2010; Hill et al. 2010; Rahal et al. 2015; Ramirez, Nogal, and Casado 2008). Our findings suggest the general utility of multiple measures to differentiate between CTSIB trials across all metrics, COP_{FPd} (56/134), COM_{IMUv} (54/134), and COP_{FPv} (48/134) all had similar significant differences between trials. Notably, Masani et al. also compared different measures of COP and COM in the AP direction during quiet standing with eyes open. They found that COP velocity was correlated with COM velocity, but more highly correlated with COM acceleration (Masani et al. 2014). Masani's findings suggest that COM acceleration could be used in place of COP velocity. Interestingly in our study, COM_{IMUa} did not have as many significant differences (28/135) as COP velocity. However, this could be due to the different conditions compared in our study to that by Masani et al. When considering the utility of measures to detect differences in postural sway conditions, it is also essential to appreciate that COM and COP activity represent unique aspects of balance and maybe differently impacted by balance strategies, populations and sensory challenges. For example, COM and COP might uniquely capture the relative use of ankle versus hip strategy as a balance control mechanism (Masani et al. 2014). While our study was limited to comparing trials of the CTSIB in healthy older adults, our results demonstrate the importance of not limiting studies of balance control to any single measure or metric.

3.4.2 Different CTSIB trials across all measures and metrics.

In this study, differences between certain trials (1-5, 1-6) were more susceptible to observed changes by more measures and metrics. Consistent with research examining older adults with vestibular disorders, the greatest differences were found when comparing trials 1 to 5 and 6 (Macedo et al. 2015), which suggests these trials have a strong capacity to differentiate balance performance between trials across the five measures and nine metrics. If time constraints were a factor in administering the CTSIB test, completing trials 1, 5 and 6 would provide a viable means to assess balance performance as well as determine the influence of proprioception and visual inputs on balance performance. In trial 1, the participants' balance performance utilizes inputs from typical visual and proprioceptive inputs. In contrast, trial 5 removes visual inputs by closing the eyes and further challenges the balance system by altering the proprioceptive inputs by having the participant stand on foam instead of a firm surface. Therefore, the participant must rely more heavily on the vestibular system in trial 5 when compared to trial 1. Trial 6 provides altered visual input by changing the environmental visual cues related to vertical and horizontal lines. This may be relevant in some visual dysfunctions common in older adults where visible environmental landmarks may be obscured, such as macular degeneration, glaucoma or cataracts. CTSIB Trial 5 is emerging as the preferred trial at characterizing differences between young and older adult cohorts (Boulgarides et al. 2003; Burke, Franca, Meneses, Cardoso, and Marques 2010; Darlington et al. 2000; Hagedorn and Holm 2010; Murray et al. 2005). Recently, Horn and colleagues' found that altered visual inputs are not different from eyes closed trials when comparing young and older adult cohorts (Horn et al. 2015). Based on these findings, along with ease of clinical implementation, many studies have removed trial 3 and 6, and the modified CTSIB (mCTSIB) has emerged. Research investigating the utility of the mCTSIB also found the most differences between trials 1 and 5 (Boulgarides et al. 2003; Burke, Franca, Meneses, Cardoso, and Marques 2010; Hagedorn and Holm 2010). While this conclusion is sound when comparing cohorts, more research needs to be conducted looking at differences within-subjects (i.e. comparisons of differences in performance between trials 1 and 5), and with subjects with visual or vestibular disorders before trials 3 and 6 are abandoned. Our study, along with Macedo's found trials 5 and 6, detected the most differences across all metrics. The utility of trial 6 is also supported by research that found trial 6 to be useful for discriminating older adults with vestibular disorders from healthy older adults (Cohen, Blatchly, and Gombash 1993). Consistent with Golder, Earl and Mallery's research, the second-highest number of differences was found between trials 1 and 6, showing the importance of including trial 6 (Golder, Earl, and Mallery 2012).

If more time is available for testing, trial 4 could be added to trial 1, 5 and 6. The current results suggest that the participant's balance performance in trials standing on foam (trials 4) was significantly affected by removing (trial 5) or altering (trial 6) the visual inputs with from trials. The addition of trial 4 to 1, 5 and 6 would allow clinicians and researchers to differentiate between the influence of altered visual inputs versus proprioceptive inputs. In the current work, several differences were found between all trials, each adding its own characterization of postural sway. Examining different metrics across all the trials of CTSIB or a reduced version with trials 1, 5 and 6 provides a characterization of balance that could become a valuable tool for clinicians assessing populations with challenges to maintain their balance, like older adults with vestibular disorders.

3.4.3 Utility of Different Metrics to determine differences across all CTSIB trials and measures.

In this study, the metrics of AoE_B (49/75) and AoE_E (41/60) exhibited the most significant differences between trials across all measures. Therefore, if prioritizing metrics to be calculated, AoE_B and AoE_E appear to be a robust choice. These results are consistent with Swanenburg et al., who found AoE_B COP displacement was able to differentiate older adult fallers from non-fallers (Swanenburg et al. 2008). As well as, Macedo et al. who found that COP displacement AoE_B were different between older adults with vestibular disorders and healthy age-matched controls during trials 5 of the CTSIB (Macedo et al. 2015). However, these results are also in contrast to High et al., who found no differences between healthy older adults and older adults at high risk of falls using COP AoE_B (High et al. 2018). As AoE is a metric that is impacted by movement in both AP and ML directions, it may be more sensitive to overall change during common balance control strategies occurring at the ankle and the hip captured in COP and COM in AP and ML directions (Horak and Nashner 1986; Pyykko, Jantti, and Aalto 1990). In fact, among the nine metrics compared in this study, AoE_B and AoE_E encompass the greatest number of characteristics of balance performance. Therefore, in a population of subjects (e.g. older adults) who may respond differently in each direction, AoE_B or AoE_E may best characterize changes in their overall state of balance. However, given these metrics are a product of multiple directions (AP and ML) of activity, they may not be able to discriminate or be uniquely sensitive to specific changes in control strategies seen in different sensory conditions or differences between population groups.

3.4.4 Effect size of Different Metrics across all CTSIB trial and measures.

In the present study, AP, followed by ML metrics, had the highest effect size for specific trial comparisons. $Sway_{AP}$ had either the largest or second-largest effect size in trials 1-5 and 1-6 across all the four measures (COP_{FPd}, COP_{FPv}, COM_{IMUv}, and COM_{IMUa}). Swanenburg et al. concluded that COP measures of the maximal (ML & AP directions) and root-mean-square amplitude (ML & AP directions), as well as AoE_B, were reliable at differentiating older adult fallers from non-fallers (Swanenburg et al. 2008). Later, Melzer et al. used a retrospective analysis of COP displacement measured with a FP in older adult fallers and non-fallers. They found that older adult fallers had higher sway amplitudes in the ML direction, but not in the AP direction. They concluded that ML sway amplitude seen in narrow stance conditions was a good measure of balance control to identify older adult fallers (Melzer, Kurz, and Oddsson 2010). These findings suggest the clinical utility of AP and ML metrics in the analysis of characteristics of balance to identify fall risk.

3.4.5 Utility of Different technologies to measure balance performance.

Recently, technologies that quantify postural sway by measures of COM or COP have been improved. The ability to measure and/or calculate new metrics of COM and COP in several kinematic variables with increased accuracy is now possible. Improvements in wearable technologies, like IMUs, as well as, the portability of FPs, has resulted in tools that provide research-grade data including high sample rates and high signal to noise ratios (Moe-Nilssen 1998). These improvements allow for the integration of signals to calculate linear and angular kinematics with minimal measurement error (Masani et al. 2014). Added to the technologies improved technical characteristics, they are continually being developed, which makes them more accessible for clinicians in terms of their cost and ease of use.

3.5 Summary

These findings suggest that CTSIB could have improved clinical utility when instrumented with technologies like FPs and IMUs. Both the COP_{FPd} (56/135) and COM_{IMUv} (54/135) were able to improve on the traditional technique, COM_{Pend} (6/45), which suggests these new measures were able to characterize unique features of postural sway. Improvements in these new technologies have made it possible to calculate more kinematic measures. This improved characterization of postural sway and balance could add a useful tool for clinicians, but more research needs to be conducted comparing kinematic measures and specific metrics. A limitation of this study was that only static balance challenges were assessed during quiet standing where the COM was within the base of support; however, our results demonstrate the importance of different measures and metrics provide, in characterizing balance. trial 5 is emerging as the preferred trial to characterize differences between older adults to other cohorts (e.g. young or older adults with vestibular disorders). However, before trials 3 and 6 are abandoned, more research needs to be conducted looking at differences within-subjects (i.e. comparisons of differences in performance between trials 1 and 5) as well as further comparisons to those with visual or vestibular disorders. For older adult populations who may respond differently in ML and AP direction measures of AoE, which take into account these directions, captures more differences between trials of the CTSIB. In contrast, measures in the AP and ML directions capture larger differences (i.e. effect sizes) between trials of CTSIB.

In closing, instrumented CTSIB with either a FP or an IMU should undergo further research to explore their utility to measure balance performance during sensory-based balance tests in the clinical setting. Further research should explore the utility of trial 3 and 6 and their inclusion in the test battery of CTSIB.

3.6 Limitations

The sample size, though relatively small (39 community-dwelling older adults), robustly detected significant differences given the large effect sizes suggesting this sample size was adequate. Further, the sample size was typical in other older adult research (Chaikeeree et al. 2015; Criter and Honaker 2016; Filar-Mierzwa et al. 2017; High et al. 2018; Lee 2017). The sample was selected from community-dwelling, older adults living in Victoria, British Columbia, Canada, and, therefore, should not be considered representative of the general Canadian population. The study only examined FP and IMU devices and, therefore, should not be regarded as an exhaustive study of the possible tools that measure COP and COM during CTSIB. Other possible devices include, but are not limited to, motion capture systems, infrared imaging systems, the Wii Balance Board, insole pressure sensors systems, and mobile phone assessments.

Though there is the potential that fatigue may influence performance across the test battery, it is improbable given that the total test time is only three minutes. Further, the participants were community dwellers with active lifestyles, and their typical activity levels were much greater than the test demands. In addition, though randomizing trials would address the potential effects of fatigue, it would make the test cumbersome to implement potentially involving the removal and addition of equipment for each trial, which would also significantly influence the total test time. Further, for comparisons to the existing literature, where the trial order is not typically randomized, it was prudent to maintain the consistent order of tests. Lastly, participants are over halfway through the battery at Trial 4, and if there were an effect of fatigue, we would have expected to see more differences between Trial 1x4. However, as noted, there are very few differences between these two trials despite the higher level of difficulty of Trial 4 over Trial 1, suggesting that fatigue was not an issue at this point in the testing protocol. Moving forward, we will incorporate a 7th Trial, which is a repeat of the baseline Trial 1, for comparison to ensure fatigue is not a confounding influence.

Though there were multiple comparisons in the current analysis, repeated measures ANOVA analysis, with a Post Hoc Tukey's, should control for Type I Error. However, to be more conservative, we could apply Dunn-Sidak correction to our p-value. Based on the current results, it is not likely that there is concern about Type I error for several reasons. First, all the results were as expected with larger excursions in postural sway seen consistently across all measures as the balance difficulty of the trials increased. The effect sizes and detection of significant differences also matched our expected outcomes suggesting the results were not due to chance and Type I error. If Type I error were influencing the results, we would see inconsistent, spurious significant differences, which was not the case.

3.7 Future Directions

This research provides an excellent understanding and baseline for informing the choice of measures and metrics that would be best suited for future research exploring prospective identification of several clinical populations such as those at risk of falls or onset of Parkinson's disease. The specificity of this detection could be enhanced by monitoring change in the CTSIB balance measures over time, as well as looking at individual differences as noted by the different measures in postural sway between Trial 1 (baseline) compared to the other trials. Further, a cognitive load during the balance trials could be incorporated into the test battery. In addition, new technologies such as a virtual headset in place of a visual conflict dome, pressure sensor insoles, and mobile phone assessment should undergo further research to explore their clinical utility for practitioners. Finally, research should explore the utility of Trial 3 and 6 and their inclusion in the test battery of CTSIB.

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Appendix 1

Table A: Literature review in PubMed of research using CTSIB and mCTSIB to measure postural sway and balance ordered by year.

(Author)	CTSIB Trials (Trial Duration)	Comparison	Metrics	Device
(Shumway-Cook and Horak 1986)	CTSIB (30sec)	Baseline (Trial 1)	Trial Duration, ML Sway, AP Sway, AoE	Stop watch, grid and plumb line
(Cohen, Blatchly, and Gombash 1993)	CTSIB (30sec)	Cohort	Duration	Timing Device
(Di Fabio and Seay 1997)	CTSIB (30 sec)	Tests	Duration	Timing Device
(Bernhardt et al. 1998)	CTSIB (30 sec) 1 attempt -feet 10cm apart	Tests	Duration	Timing Device
(El-Kashlan et al. 1998)	CTSIB (30sec) 3 attempts for each trial	Cross-sectional comparison of Tests, Pretest-Posttest	Cumlative Mean Duration	Force Plate (Balance Master)
(Whitney, Poole, and Cass 1998)	CTSIB	Tests	CoM position, Duration	Timing Device, Postural Grid
(Darlington et al. 2000)	CTSIB (30sec)	Cohort (insulin-dependent vs non)	Postural sway (mVolts)	Force Plate (Strain gauge)
(Allum et al. 2001)	mCTSIB (20sec)	Cohorts (unilateral vesitbular loss, acute peripheral vestibular dificit with canal paresis, & Normal subjects age matched)	Duration, peak-to-peak excursions in roll and pitch directions for angular displacement and velocity	IMU (2 digitally-based angular-velocity transducers)

(Nicholson et al. 2002)	CTSIB(25.5s, 2 attempts of each trial, choice of foot position)	Cohorts	ML/ AP sway, Trial duration	Force Plate
(Boulgarides et al. 2003)	mCTSIB (Trial 5 predictive)	Cohort; Regression (Dependent Variable History Falls)	mean CoP sway speed (deg/sec), ave CoP position	Force Plate (Balance Master 6.1)
(Sjostrom et al. 2003)	mCTSIB (20 sec)	Tests	Angular velocity deviation (roll/ML & pitch/AP)	IMU (2 angular velocity transduces) - L2-L3
(Whitney and Wrisley 2004)	mCTSIB (30sec) 2 attempts (shoes on/off)	Test, With & without shoes	Duration	Timing Device
(Wrisley and Whitney 2004)	mCTSIB (30sec) 2 attempts (feet together, feet apart)	Test, Feet together & Apart	Duration	Timing Device
(Murray et al. 2005)	CTSIB (30sec)	Cohort (Faller vs NF)	Duration	Timing Device
(Lebib et al. 2006)	mCTSIB (10 sec) 3 attempts	Cohort	mean position of CoP, mean velocity CoP	Force Plate (Neurocom Balance Master)
(Bulat et al. 2007)	mCTSIB, 30 sec	Cohort	mean CoG sway velocity	Force Plate (Nurocom Balance Master)
(Madureira et al. 2007)	CTSIB (30sec)	Cohort, Pretest-posttest	Duration (mean of group for each trial)	Timing Device
(Amor-Dorado et al. 2008)	CTSIB (30sec) 3 attempts for each trial	Cohort	Duriation (mean of 3 attempts)	Timing Device
(Moalla et al. 2008)	mCTSIB (10sec) 3 attempts	Cohort	position of CoP, velocity (AP & ML) mean	Force Plate (Balance Master Neurocom
(Ramirez, Nogal, and Casado 2008)	mCTSIB	Tests	Speed of displacement	Force Plate (Neurocom Balance Master)

(Hill et al. 2010)	mCTSIB (10sec)	Corhort	ave sway velocity (degrees/sec)	Force Plate (Neurocom Balance Master)
(Ricci et al. 2009)	CTSIB (30sec)	Cohort	Duration	Timing Device
(Burke, Franca, Meneses, Cardoso, and Marques 2010)	mCTSIB	Cohort	CoP mean velocity (degs/sec)	Force Plate (Nurocom Balance Master)
(Desai et al. 2010)	mCTSIB T4 & T5 (20sec)	Cohort (Faller vs NF)	postural sway (peak-to-peak AP-ML CoP excursions: AoE, ML Sway, AP Sway, Sway path length)	Flexible pressure mat
(Hagedorn and Holm 2010)	mCTSIB	Cohort, Preintervention-postintervention	Duration	Timing Device
(Rombaut et al. 2011)	mCTSIB (30sec)	Cohort	mean sway velocity (cm/sec), ML Sway, AP Sway, 95% ellipse sway area	Force Plate (AccuGait strain portable FP)
(Suttanon et al. 2011)	mCTSIB (10sec)	Reliability (ICC),	Duration (composite of 4 trials)	Force Plate (Neurocom)
(Szturm et al. 2011)	mCTSIB	Cohort	Duration	Timing Device
(Golder, Earl, and Mallery 2012)	CTSIB (30sec)	Trials (Time only)	Duration	Timing Device
(Mulavara et al. 2013)	mCTSIB (30 sec)	Tests, Cohort	Angular motion, Acceleration (AP & ML) Roll angular velocity, pitch angular velocity, Yaw angular velocity - all degrees	IMU (Xsens) mid-thoracic level

(Vaz et al. 2013)	CTSIB (30sec) one try at each condition	Pretest-posttest comparison	Duration	Timing Device
(Zur et al. 2013)	CTSIB (30sec)	Cohort	Duration	Force Plate
(Alahmari et al. 2014)	CTSIB (20sec) with head mounted display	Cohort	CoP area & velocity (95% confidence ellipse, max AP & ML excusion)	Force Plate (Balance Rehabilitation Unit)
(Macedo et al. 2015)	CTSIB (30sec)	Compare Trials	Duration; CoP disp (elliptical area with 95% distribution), Velocity of Body sway (total distance/ 60s)	Balance Rehab Unit (BRU)
(Rahal et al. 2015)	mCTSIB (10sec)	Cohort	Ave Sway Velocity	Force Plate (NueroCom Balance Master)
(Acar et al. 2015)	mCTSIB (time not reported)	Cohort	Mean CoG Sway Velocty (Degrees/sec)	Force Plate (Neurocom Balance Master)
(Chaikeeree et al. 2015)	mCTSIB (30 sec) each trial repeated 3 times	Cohort	RMS of CoM Accel(ML & AP)	IMU (OPAL inertial sensor measuring CoM acceleration at L5 at 50Hz)
(Horn et al. 2015)	mCTSIB/CTSIB (30sec)	Corhort (5 percentiles)	Trial Duration (ave of 3 trials)	Timing Device
(Lefaivre and Almeida 2015)	mCTSIB (20sec)	Pretest-posttest comparison	Average CoP deviations (AP & ML) of 4 trials	Force Plate (Biodex Balance System SD)
(Moghadam et al. 2015)	mCTSIB (30sec)	Cohort	Trial Duration	Timing Device

(Ahmed et al. 2016)	mCTSIB (30sec)	Cohort	Duration	Timing Device
(Amor-Dorado et al. 2017)	CTSIB (30 sec)	Cohort, Test	Duration with T1 used as Baseline	Timing Device
(Chung, Kim, and Yang 2016)	CTSIB (total time can maintain standing position)	Tests	Acceleration (Root mean square and Signal Vector Magnitude in AP & ML directions)	IMU (Trigno tri-axial acceleromter)
(Criter and Honaker 2016)	mCTSIB (20sec)	Cohort; Linear Regression (History of Falls dependent variable)	Trial Duration; Amount of sway (0 = little to no sway, 1 = mild sway, 2 = moderate sway, 3 = inability to perform task)	Timing Device
(Dannenbaum et al. 2016)	CTSIB (30 sec)	Cohort	Duration	Timing Device
(Eyvazov et al. 2016)	CTSIB T1	Longitudinal, Tests	Median Interquartile range of Postural Measurements	Force Plate (950-460 BioSway system)
(Jirikowic et al. 2016)	CTSIB (30sec)	Cohort, Longitudinal.	Duration (Mean of all trials combined)	Timing Device
(Collado-Mateo et al. 2017)	mCTSIB (30sec)	Cohort, Longitudinal	Mean Sway Indices?	Force Plate (Biodex Stability System)
(Dawson et al. 2018)	mCTSIB (30sec)	Tests construct validity	Overall sway index (standard deveation of sway)	Force Plate (Biodex Stability System)
(Filar-Mierzwa et al. 2017)	mCTSIB	Pretest-posttest comparison	Average, median, min, max, Lower Quartile, Top Quatile	Force Plate (BioSway)

(Lee 2017)	CTSIB (20sec)	Cohort	mean of 10 sec median sway acceleration (m/s2), normalized to baseline of each condition	Smartphone 100Hz (Apple iPhone 4S with SPARKvue software)
(Nair et al. 2018)	mCTSIB (30 sec)	Cohorts	Duration	Timing Device
(High et al. 2018)	mCTSIB (30sec) with and without vibrotactile feedback	Cohort	CoP total path length, mean velocity, 95% elliptical sway area, alpha values (AP & ML).	Force Plate (AccuGait 600Hz)
(Moran et al. 2019)	mCTSIB (20sec) - feet shoulder width apart	Cohort	Sway Index Score (average sway from CoP), Composite test score	Force Plate (Biodex Balance System SD)

Note: The search terms used: CTSIB, mCTSIB, Balance, Stability, and Postural Sway.