Reliability and Validity of an Accelerometer-based Balance Assessment for Fall Risk Screening

DISSERTATION

Presented in Partial Fulfillment of the Requirements for the Degree Doctor of Philosophy in the Graduate School of The Ohio State University

By

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2013

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Abstract

Postural sway occurs spontaneously during static stance as a result of the unstable properties of the body. In balance dysfunction, often a consequence of the normal aging process, the amplitude and frequency of this sway may exceed the limits of stability and contribute to a fall. Therefore, it is important to develop reliable and valid tools to assess postural sway and fall risk. When compared with force plate center of pressure (COP) measurements (one of the most commonly used postural sway assessment tools), accelerometers have been as good as, or better at discriminating postural sway differences between populations with and without balance deficits and between tasks requiring different levels of balance control.

A wireless tri-axial accelerometer has recently come on the market that is smaller, less expensive, and more sensitive than those previously utilized in postural control studies. However, no reliability or validity studies on this device exist. Therefore, we aimed to assess the new accelerometer’s 1) test-retest reliability, 2) ability to discriminate between tasks with differing sensory conditions (i.e., visual and somatosensory), 3) ability to discriminate between individuals who are fallers versus non-fallers, and 4) and its concurrent validity with other fall risk screening tools in older adult populations.
Twenty healthy and independent living older adults (mean age 81±4 years; 8 male and 12 female) participated in the reliability and balance task discrimination study. An accelerometer was taped to the lower back of the participant with a gait belt positioned just below. Participants completed three 30-second trials of the four classic Romberg conditions (standing on a firm or compliant surface with eyes open or closed) in random order. Following a 20 minute rest, participants underwent a second block of testing for a total of 24 trials. Raw data was collected at approximately 250 Hz and low-pass filtered at 55Hz. Data were transformed to correct for pelvic tilt and adjust for low frequency. Following processing of raw data, root mean squared (RMS), with the unit gravity (g), for the anterior-posterior (A-P) and mediolateral (M-L) acceleration data was calculated. Intraclass Correlation Coefficients (ICC) were used to test for reliability, while repeated measures analysis of variance (ANOVA) models were used to test for a main effect of balance condition on the A-P and M-L acceleration RMS.

To validate the accelerometer-based balance assessment (ABA) as a fall risk screening tool, we recruited 95 healthy older adults (mean age 86±6 years) residing in five different independent living facilities. Participants reported the number of falls they had in the previous six months prior to testing. Participants then completed, in random order, the Berg Balance Scale (BBS), Timed up-and-go (TUG), Activities-specific Balance Confidence (ABC) scale, and the ABA. The ABA consisted of the same four Romberg conditions used in the reliability assessment with and without a cognitive task (counting backward by 3’s) for a total of eight balance conditions. The best attempt of two successful trials (the one with the lowest RMS values) for each condition was
included in the data analysis. The fall risk discriminative power of various combinations of the TUG, BBS, ABC, and/or the ABA was evaluated using logistic regression and chi-square analyses. Spearman’s rank correlation coefficients were used to investigate the relationship between RMS, BBS, and the ABC scale measures. Pearson’s product-moment correlation was used to determine the relationship between RMS and TUG scores.

ICCs were all good to excellent with values ranging from 0.736 to 0.972 for trial-to-trial and from 0.760 to 0.954 for block-to-block. There was a significant stepwise increase in A-P and M-L acceleration RMS from conditions 1 to 4. With respect to A-P sway, the mean acceleration RMS increases from conditions 1 to 2, 2 to 3, and 3 to 4 were 0.0031±0.0008g, 0.0035±0.0011g, and 0.0075±0.0022g, respectively. Similarly, the M-L mean acceleration RMS increased significantly from conditions 1 to 2, 2 to 3, and 3 to 4 (0.0025±0.0007g, 0.0068±0.0010g, and 0.0084±0.0018g, respectively). A-P and M-L acceleration RMS were statistically similar when standing on a firm surface. However, M-L acceleration RMS significantly exceeded A-P for condition 3 ($F(1, 19) = 4.61, p = 0.045$) and condition 4 ($F(1, 19) = 6.68, p < 0.018$).

For the prediction of fall risk, all prediction models with the ABA, alone or in combination with the TUG, BBS, and ABC scale, had AUC’s ≥ 0.80. Models without the ABA had AUC’s in the range of 0.45 to 0.75. In general, most of the prediction models were better at correctly identifying non-fallers than fallers, indicated by substantially larger specificity than sensitivity values. Only for the comparison of multiple fallers to non-fallers did a model, the combination of the TUG, BBS, and ABC
scale with the ABA, exhibit both good sensitivity (85.0) and specificity (82.5). No clinical test on its own or in combination exhibited good sensitivity (all ≤ 61.1).

Surprisingly, the associations between the ABA and the TUG, BBS, and ABC scale were not only insignificant but extremely weak with correlation coefficients of 0.11, -0.04, and -0.02 for the TUG, BBS, and ABC scale, respectively. As expected, the TUG, BBS, and ABC scale were all significantly correlated with each other and in the proper direction.

In summary, the accelerometer exhibited good to excellent trial-to-trial and block-to-block reliability. The accelerometer was also able to discriminate between visual and supporting surface conditions and acceleration axes. Though not significantly correlated with the clinical tests, the ABA alone was better than the TUG, BBS, and/or ABC at discriminating fallers from non-fallers.
Acknowledgments

Professionally, I would to thank Dr. Devor for his guidance during my growth as a scholar and the undeniable influence he has had on my development as a teacher. The research presented here would not exist without the many contributions of a great number of people. Drs. Kloos and Kegelmeyer, thank you for helping me bridge the gap between my knowledge of exercise science and a useful and practical extension of that knowledge in my work with older adults. Panos Koutakis and Nick Hanson, thank you for helping me turn a noisy accelerometer signal into convenient and useable information. Finally, in no particular order of contribution, the following students were integral to the success of this study as they interacted with participants, collected data, and in some cases even contributed to the study design and data interpretation: Brenna Congeni, Jessica Dicke, Jessica Fortney, Deanne Gauch, and Josh Stone.

On a personal level, my family has been unbelievably supportive and patient during the last 10 consecutive years of higher education. Some colleagues have commented that it must be challenging to succeed in an academic environment with a wife and children. For me it has been just the opposite. Thank you Karyn, Kayleigh, and Brie for providing me with a much needed distraction and adding perspective. It is not in spite of you, but because of you that I have achieved this level of success.
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Saunders, N.W. Myth Busters: Eating fat makes you fat. Faculty Staff & Fitness Newsletter College of PAES, The Ohio State University, Columbus, OH.
Fields of Study

Major Field: Physical Activity and Educational Services
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Chapter 1: Introduction

Postural sway occurs spontaneously during static stance as a result of the unstable properties of the body [1]. In balance dysfunction, often a consequence of the normal aging process, the amplitude and frequency of this sway may exceed the limits of stability and contribute to a fall. One out of every three older adults (those ≥ 65 years of age) fall each year [2]. Twenty to 30% of falls result in moderate to severe injuries, and older adults are hospitalized for these fall-related injuries five times more often than any other cause [2]. In addition to increased dependency and decreased quality of life, fall-related injuries are expected to result in annual medical costs of $55 billion by the year 2020 [2].

While falls may result from extrinsic and intrinsic factors, the relative contribution of each changes with age [3]; with aging there is a decrease in extrinsic causes due to less participation in risky behaviors, but there is also a concomitant increase in intrinsic causes due to normal age-related functional decline. Part of the normal aging process involves decrements in peripheral receptors and nerve function [4-11], cognition and attention [12-16], and muscle strength and power [17-21], all of which are highly involved in the regulation of upright posture. Not surprisingly, balance abnormality/impairment has been identified as a major risk factor for future falls[22].
Therefore, it is important to develop reliable and valid tools to assess postural sway and fall risk.

In recent years, accelerometers have shown promise in their ability to properly assess postural sway. Most researchers have placed a single accelerometer at the small of the back near lumbar vertebrae 3, the approximate height of the center of mass (COM). Accelerometers, depending on the model, have the ability to approximate linear COM accelerations in all three planes. In addition to good reliability [23, 24], accelerometers have consistently demonstrated the ability to discriminate between distinct populations [23, 25-27], and balance tasks [23, 25-30]. In fact, Mayagoitia et al. [28] found that, for four out of five balance conditions, an accelerometer was better than a force plate at discriminating between balance conditions.

A wireless tri-axial accelerometer has recently come on the market that is smaller, less expensive, and more sensitive than those previously utilized in postural control studies. However, no reliability or validity studies on this device exist. Therefore, we aimed to assess the new accelerometer’s 1) test-retest reliability, 2) ability to discriminate between tasks with differing sensory conditions (i.e., visual and somatosensory), 3) ability to discriminate between individuals who are fallers versus nonfallers, and 4) and its concurrent validity with other fall risk screening tools in older adult populations.
Chapter 2: Review of Literature

CDC Fall Statistics

One out of every three older adults (those ≥ 65 years of age) fall each year [2]. Twenty to 30% of falls result in moderate to severe injuries, and older adults are hospitalized for these fall-related injuries five times more often than any other cause [2]. In addition to increased dependency and decreased quality of life, fall-related injuries are expected to result in annual medical costs of $55 billion by the year 2020 [2].

Fall Risk Outcome Measures

Hallmarks of the normal aging process include decreases in muscle strength and power, bone density, neural conduction velocity, visual acuity, and proprioception, each of which may contribute to an increased risk of falling. With pathology, such as vestibular dysfunction, diabetic neuropathy, Parkinson’s disease, Huntington’s disease, etc., the consequences of aging become much more severe and the risk of falling increases dramatically. As bone health deteriorates the risk of sustaining a serious injury from a fall increases as well. The causes of falls in the older adult population are many and include both internal and environmental factors. While environmental factors like weather, home hazards, and activity level are important and have received attention, a
tremendous emphasis has been placed on developing instruments to identify internal risk factors for falls.

*Comprehensive Balance and Mobility Screens*

*Performance-Oriented Assessment of Mobility*

The Performance-Oriented Mobility Assessment (POMA) was developed by Tinetti [31] to evaluate fall risk with little equipment or expertise required. The test includes several static and dynamic balance tasks along with a comprehensive gait assessment. Subjects are judged somewhat subjectively on their ability to perform each task and categorized as normal (2), adaptive (1), or abnormal (0). Retrospectively, the POMA has displayed the ability to discriminate between fallers and non-fallers [31]. In fact, fall risk increased with the number of balance and gait abnormalities.

In a prospective study of falling in older adults, Tinetti et al. [22] administered a comprehensive baseline assessment including orthostatic changes in blood pressure, vision, hearing, reflexes, position sense, muscle function, and the POMA. Subjects were then followed for one year with falls reported weekly. The risk of falls was associated with the number of total risk factors. However, while statistically significant, compared with other risk factors, balance and gait abnormalities, determined via the POMA, exhibited the weakest association with falls. Tinetti [31] suggested that the POMA may have traded sensitivity for simplicity. In a different 1-year prospective study of falls, Hausdorff et al. [32] found that the POMA was unable to discriminate fallers from non-fallers. Therefore, while the POMA may have use in identifying balance and gait
abnormalities and directing the design of appropriate interventions, it does not appear to
capture the functional characteristics that predispose a person to future falls.

Berg Balance Scale

The Berg Balance Scale (BBS) is a 14 item test battery developed by Berg et al. [33] to measure static and dynamic balance, sit-to-stand and stand-to-sit transitions, and
functional ability to perform simple mobility tasks. Each item is scored 0-4 with four
representing the highest level of function. The highest overall possible score is 56. The
BBS is one of the most frequently used instruments to assess fall risk and has been
considered the gold standard [34]. However, while its inter-rater reliability and test-retest
reliability have been consistently high, the validity of the BBS as a fall prediction tool has
delivered vastly varied results.14

Some studies have reported that the BBS was capable of retrospectively
discriminating fallers from non-fallers [35, 36], while others have found the instrument
insensitive to individuals with a history of falls [37, 38]. In general, the specificity of the
BBS far outweighs the sensitivity, making it a useful tool for ruling out high risk but not
necessarily at properly categorizing individuals as high risk [39]. Brauer et al. [40] found
the BBS to have a ceiling effect for higher functioning individuals, which may explain
the findings of Santos et al. [36], that the sensitivity and specificity were both high for
sedentary subjects but low for active subjects.

Prospective studies, too, have delivered mixed results. Brauer et al. [40] reported
that the BBS was unable to prospectively predict falls. Muir et al. [41], on the other
hand, found that the predictive ability of the BBS was dependent on the definition of a
fall and how the BBS scores were categorized. For example, the BBS only moderately predicted multiple falls and was unable to significantly predict single and/or injurious falls. Many studies have used a cutoff score of 45 to categorize fall risk. Muir et al. suggested that it is inappropriate to dichotomize the BBS evidenced by the fact that 40% of the subjects with a score greater than 45 experienced a fall during the follow-up period, while 40% with a score less than 45 did not fall. Instead, it was suggested that because risk is gradient the BBS should therefore be analyzed in a gradient manner.

Nuels et al. [38] attempted to establish an optimal cutoff score for the BBS by calculating sensitivities and specificities from the results of previous studies. Unfortunately, an optimal cutoff could not be established because of too much diversity in study population and study design. It was determined that the BBS on its own was unable to predict falls in older adults, and that it should only be used to supplement other tests to screen for fall risk.

Other than assessing fall risk, the BBS has also been used to assess change in functional status over time, possibly following a period of treatment. However, while the relative test-retest reliability, as determined by intra-class correlation coefficients, has been reported to be high, the absolute test-to-test variability for an individual may be Romero et al. [42] used the standard error of measurement to calculate the minimal detectable change (MDC) in the BBS for cognitively intact older adults ≥65 years. The MDC represents the absolute change in score necessary to have 95% confidence that there is change in the functional status of the individual. For the BBS, the MDC was
calculated to be 6.5 units. Therefore, any test-to-test variation in score less than 6.5 can be considered measurement error.

*Timed Up-and-Go*

The 3-meter Timed Up-and-Go (TUG) is a simple, objective, and relatively quick assessment of lower body strength, agility, and basic mobility skills. Subjects begin in a seated position, rise from the chair, walk as quickly as possible to a marker three meters in front of the chair, turn around the marker, and return to a seated position. The singular outcome measure is total time in seconds to complete the task. The TUG has high inter-rater and test-retest reliability [39].

Several studies argue against the use of TUG for fall prediction. Herman et al. [37] and Weiss et al. [43] reported the inability of the TUG to retrospectively discriminate fallers from non-fallers. The prospective fall studies of Hausdorff et al. [32] and Nordin et al. [44] reported no significant predictive ability of the TUG. Similar to the BBS, due to consistently high specificity but variable sensitivity, the TUG may be best suited to rule out rather than rule in high risk [44]. Also like the BBS, no consistent cutoffs have been established, though a MDC of 3.4 seconds has been suggested for Parkinson’s patients [45].

Weiss et al. [43] argued that simply the total time to completion of the TUG does not fully capture differences that may exist between fallers and non-fallers. While the traditional TUG was unable to discriminate fallers from non-fallers, several accelerometry derived measures, acquired during the TUG, were sensitive to group differences. Specifically, fallers tended to be more restricted and cautious during the two
transfers, as indicated by lower AP acceleration amplitudes. The benefit of supplementing the TUG with accelerometry is that each phase of the task is objectively and automatically evaluated. The drawback to the use of this equipment, or any other equipment like it, is that it is less accessible, more expensive, and requires at least some expertise to interpret the data. In addition, no normative data or prospective data on fall risk exist for these measures.

*Roger’s Modular Obstacle Course*

The obstacle course developed by Rogers et al. [46] was designed to assess fall risk characteristics so that therapies could specifically target deficiencies. The course has nine stations that challenge different aspects of balance and mobility similar to some everyday challenges. Subjects are scored objectively on the time to complete each task as well as the entire course. The quality of movement and postural control for each task are also qualitatively assessed. No reliability or validity information was offered. The tradeoff for this type of assessment is likely to be increased sensitivity for decreased convenience. The obstacle course requires a good deal of equipment and space and would therefore be impractical for many clinical and community settings.

*Static Balance*

Static balance tasks assess the ability of subjects to maintain their COM over their BOS. Typical outcome measures include pass/fail, time to failure, or the amount of postural sway during a fixed balance duration.

*Center of Pressure*
It is common practice to assess postural stability during quiet stance using force plates. As the COM shifts along the ML and AP axes, force plates detect and record the corresponding changes in the center of foot pressure (CoP). While not directly related, the CoP exhibits similar patterns of movement of the COM as long as the body behaves as a rigid inverted pendulum pivoting only about the ankle (in quiet stance this is assumed to be the case). Therefore, CoP metrics have provided an objective measure of postural stability. However, CoP has proven to be ineffective at prospectively predicting falls [40], having displayed poor sensitivity (29%) and high specificity (88%). One major problem with CoP data is that the final results are heavily influenced by the type and severity of filtering, and there is currently little consistency in filtering methods.

**Balance Duration**

Hausdorff et al. [32] had community living older adults (≥70 years old) perform three different balance tasks, single leg stance with eyes open and eyes closed, and tandem stance with eyes open. The outcome measure was balance duration. None of the three durations were associated with future falls.

**Romberg**

In healthy individuals, postural stability is maintained via contributions of the visual, vestibular, and somatosensory systems. Various versions of the Romberg test systematically eliminate, or make less reliable, the aforementioned sensory systems. This is typically accomplished by having subjects balance on a firm surface or compliant foam with their eyes open or closed. Sometimes a secondary cognitive task is added to evaluate the attentional demands of a Romberg condition. Subjects with vestibular loss
perform poorly and often fall when balancing on foam with their eyes closed. The Romberg test has also demonstrated the ability to retrospectively discriminate fallers from non-fallers [47]. No literature was found evaluating the sensitivity of the Romberg test for fall prediction.

Dynamic Balance

Functional Reach

Duncan et al. [48] developed the functional reach (FR) to test the limits of stability (LOS) of subjects during anterior volitional lean and was meant to be a measure of lower extremity strength and function. LOS is a concept based on the notion that upright static balance is possible only while the center of mass (COM) remains within the base of support (BOS); in this case the BOS would be approximately the length of the foot. However, how close the COM can get to the support boundary before taking a step is also a function of strength and confidence. As originally designed, the subject stands with their feet at shoulder width and their right shoulder flexed anteriorly to 90 degrees. They then reach forward as far as possible without taking a step. The absolute distance, measured by a stationary yardstick, is recorded for each trial.

Duncan had subjects (age 70-87 years; other demographic data unknown) perform the FR while standing on a force plate to characterize movement of the CoP. FR was significantly correlated with AP CoP deflections. Additionally, inter-rater and test-retest reliability were high. Finally, FR was observed to be inversely associated with age. The following limitations were acknowledged: the reach distance is influenced by body
height; it may not be able to be used in many populations such as those with dementia, spinal deformities, and upper extremity impairment; and it only measures AP stability.

Weiner et al. [49] applied the FR to older adults along with the following other physical performance measures: mobility skills, physical and instrumental activities of daily living, 10-foot walking speed, single leg balance, and tandem walking. FR was shown to be moderately to highly correlated with all other measures. In addition, FR was able to correctly classify frail older adults. In contrast to the apparent ability of the FR to discriminate individuals based on frailty, Brauer et al. [40] and Hausdorff et al. [32] found that FR was unable to prospectively predict falls. Brauer also noted a ceiling effect with the FR in better functioning subjects.

It has been suggested that FR does not really measure LOS, but instead is primarily a function of trunk rotation [50]. In a comprehensive evaluation of the FR, Jonsson had subjects perform the FR on a force plate while simultaneously collecting kinematic data via motion capture and ankle muscle response amplitudes and timing via electromyographic (EMG) recordings. Unlike the results of Duncan et al. [48], FR was not correlated with CoP. There, too, was no correlation between FR and ankle movement or ankle EMG. There was, however, a high association between FR and degree of trunk rotation, thereby indicating that the FR was not in fact evaluating LOS. This may explain the poor fall prediction sensitivity identified by Brauer et al. and Hausdorff et al.

Demura and Yamada [51] compared a modified version of the FR to that developed by Duncan. In the modified version subjects push an elastic stick. The change in length of the elastic stick is the outcome of interest. Japanese male and female older
adults (mean age = 70 years) performed the traditional and elastic stick FR on a force plate with ankle EMG. CoP excursions and EMG amplitudes were significantly greater for the elastic stick FR, which indicates a greater demand on the lower extremities and greater challenge to the LOS. Another proposed modification to the FR involves reaching for an object and/or standing on an unstable surface. It has been shown that FR distance improves in the presence of an external object and decreases when standing on foam or another unstable support [52]. Though no associations between these modified versions of FR and fall risk exist, they do show promise.

Maximal Lean

In yet another somewhat modified FR, Brauer et al. [40] investigated maximal volitional lean on a force plate. Older adult females (mean age = 70) were asked to lean as far forward, backward, and laterally as they could without taking a step. The maximal CoP displacement in each of the four directions was analyzed. Unlike the traditional FR, this method assessed lean in all four directions and did not involve reaching with an upper extremity. Once again, though, the CoP displacement data was not significantly associated with future falls, likely due to deviation from a pure ankle strategy.

Reaction Time Step Task

The reaction time step task implemented by Brauer et al. [40] involved older adult females (mean age = 70) standing on a force plate with a force-sensitive box on the floor in front of them. Upon seeing a warning and response light the subjects would step as quickly as they could (right or left depending on the warning) on the box. The force plate measured weight transfer from side to side. EMG measured lower extremity muscle
activation times. Finally, the force-sensitive box measured the latency between the onset of the response light and the foot touching the top of the box. The general finding of this prospective falls study was that fallers were slower to respond than non-fallers.

Maximum step length

Maximum step length requires that a subject step with one foot as far from the starting position as possible (without moving the stationary foot) and return to the starting position without destabilizing. This is performed anteriorly, laterally, and posteriorly for left and right legs. The distance between feet is measured for each direction. The test would appear to be heavily influenced by confidence and lower extremity strength/power. This test has shown high test-retest reliability; at least moderate correlations with other balance, gait, and mobility tests; and the ability to predict future multiple falls [53]. No sensitivity or specificity values were reported.

Confidence and Knowledge

Activities-Specific Balance Confidence Scale

One of the most widely used balance confidence questionnaires is the Activities-Specific Balance Confidence (ABC) scale. The survey asks the subject to express their confidence (as a percentage) in performing many activities of daily living (ADL). The ABC scale has exhibited the ability to reliably discriminate groups based on mobility status [54], fall history [35], future near falls [55], and a host of other subject characteristics including gender, age, education level, and health status [55]. It may also be sensitive to a change in functional status over time [55]. The advantage of the ABC scale over the related Falls Efficacy Scale of Tinetti is that it inquires about a greater
number of specific ADLs. The downside to being so specific with ADL descriptions is that it increases the likelihood that a subject will not have experienced that scenario. However, evaluation of cases where this has happened has shown that the overall score is relatively unaffected [54].

*Fall Risk Behaviors and Perceptions Scale*

The FRBPS, developed and tested by Yuen and Carter [56], inquires about 1) the frequency of occurrence of specifically listed risky behaviors, and 2) the likelihood of older adults falling because of a specifically listed internal and external hazards. The exploratory study by Yuen yielded the following results: compared with those less than 75 years of age, those older than 75 participated less in risky behaviors and were more aware of fall risk factors; compared with women, men more often participated in risky behaviors; and, compared with less mobile older adults, more mobility increased the likelihood of participating in risky behaviors.

*Gait*

*Normal Gait Parameters*

Typical outcome measures for gait assessment have included stride length, comfortable gait speed, fast gait speed, and walking distance. These measures are fairly easy to obtain and require minimal equipment (stopwatch and measuring tape). Inter-rater and test-retest reliability are consistently high. Comfortable and fast gait speeds are highly correlated with the BBS and TUG, and are sensitive predictors of self-reported physical function [39]. Of these measures, only gait speed has demonstrated good predictive ability [57]. In a prospective falls study in well-functioning older adults (mean
age = 74.2 years), Cesari et al. [57] determined that a cutoff gait speed of 1 m/s best discriminated future fallers from non-fallers. Fall risk was much greater in those with gait speeds less than 1 m/s. In addition fall risk increased in a gradient manner with decreasing gait speeds.

Gait variability

In many physiological systems variability is considered healthy and reactive. In gait, however, variability in stride length, stride time, and swing time are sensitive measures of fall risk. In the prospective study of Hausdorff et al. [32], older adults (≥ 70 years) walked at a normal pace for 6 minutes while wearing force sensitive insoles. Stride time variability and swing time variability were greater in those that fell. Variability was also greater in those with multiple falls than those with only one fall. The force sensitive insoles are a convenient tool for many gait parameters due to tremendous portability, which permits the assessment of gait under more natural and realistic conditions. They do not directly provide information about stride length or gait speed, though means of both could be calculated with a stopwatch and measuring tape.

Verghese et al. [58] also conducted a prospective falls study. Gait was assessed at baseline using a GaitRite system. The GaitRite is a short force sensitive walkway that permits the evaluation of temporal and special properties of gait. While gait speed, stride length, and cadence exhibited no significant association with future falls, stride length and stride time variability were greater in future fallers. The GaitRite appears to be very user friendly with all outcome measures automatically calculated and reported. It is also expensive and may not fully characterize gait with such a short walkway.
Dynamic Gait Index

The Dynamic Gait Index (DGI) is an instrument used to assess gait characteristics under normal conditions, variable speeds, obstacle avoidance, stair climbing, and with head movements. It correlates moderately with the BBS [34, 37], and with the TUG and ABC [37]. Both Whitney (subjects had impaired vestibular function; 14-88 years old) and Herman (subjects were community living older adults; 70-90 years old) reported that the DGI was able to discriminate fallers from non-fallers based on self-reported fall history. However, no prospective evidence was offered and no threshold could be established that produced good sensitivity and specificity [37]. Two separate studies reported the MDC as 2.9 units for Parkinson’s patients [45] and older adults in general [42].

Of the studies reviewed here no one assessment has consistently demonstrated the ability to predict falls in older adults. This is likely the result of the association between test performance and falls varying by population. The causes of falls are vast. Therefore, tests which assess many different aspects of postural control and the ability to perform ADLs are preferred.

Fall Risk Factors

Prospective Studies of Falls

For the sake of convenience and brevity, most investigations of fall risk have based their results off of self-report fall history. Subject characteristics and test performance are compared between those with and without a history of falls. The
problem with a retrospective design is that, while associations can be made, no causal relationship between current functional status and previous falls can be established. For example, it is often reported that the overall score for the BBS is lower for those with a history of falls. However, there is no way of knowing if the lower level of functioning preceded the fall, or if the fall led to a change in functional status or movement strategy.

Therefore, the determination of fall risk factors and the magnitude of their contribution to falling are best achieved through a prospective study design. Most prospective studies collect baseline data and then follow subjects for an extended period of time. Due to poor fall recall, it is customary to perform weekly or monthly callbacks to inquire about any change in status. In addition to reporting falls, information is collected regarding the circumstances of the fall, any change in medication or use of assistive devices. In this way associations can be more confidently made between baseline characteristics and performance and future falls.

Tinetti et al. [22] found that the risk of future falls was most greatly associated with bedroom hazards, cognitive impairment, a history of falls at baseline, medication use, and balance and gait abnormalities. Fall risk was also strongly associated with number of risk factors of any kind. Most falls occurred at home.

Berg et al. [33] found that older adults (mean age = 83.5) scoring < 45 on the BBS were 2.7 times more likely to fall in the next year than those scoring ≥ 45. It was also noted that those using an assistive device scored significantly lower on the BBS than those who ambulated unassisted.
Speechley and Tinetti [59] classified subjects at baseline based on “frailty” and “vigor.” Frail subjects were those possessing several frail attributes such as balance and gait abnormalities, infrequent physical activity, and lower extremity disabilities. Subjects were considered vigorous if they were under the age of 80, cognitively intact, and frequently engaged in physical activity. There was also a “transitional” group that was somewhere between frail and vigorous. Falls were reported in all three groups, though the circumstances of the falls for these groups were very different. Vigorous individuals mostly fell away from home during displacing activities or as a result of a variety of environmental factors. Frail individuals almost exclusively fell at home. While frail individuals were more likely to fall, vigorous individuals were more likely to sustain an injury. The greatest proportion of fallers were those classified as transitional, and, as would be expected, the circumstances of their falls were somewhere between those of the other two groups.

Lowery et al. [60] studied the prevalence of environmental hazards in the homes of patients with dementia and their association with falls. Subjects were followed prospectively for three months. Subjects self-reported home hazards via a standardized checklist. Subjects were also classified as living at home or in a care environment. Falls were not significantly associated with place of residence, any single environmental hazard, or group of environmental hazards. The sample size may have been too small and the study duration too short for significant findings. Also, there was no cognitively intact control group to evaluate the possibility of dementia as a risk factor.
Cesari et al. [57] conducted an enormous 5-year prospective falls study with over 3000 subjects. All subjects underwent a gait assessment where they were asked to walk at a normal pace over a very short distance. At the conclusion of the study the association between gait speed and falls was analyzed. The group with gait speeds less than 1 m/s were more likely to fall, had more lower extremity limitations, more hospitalizations, and a greater number of deaths. Therefore, slower gait speed seems to be an indicator of other underlying pathology or functional decline.

Like Cesari et al., Brach et al. [61] also investigated the association between gait parameters and future falls. It has been speculated that qualities of gait other than gait speed may be independently associated with falls. Because of the strong correlation between stride length, stance time, and gait speed, stride length and stance time did not provide additional information about fall risk or mobility disability. Variability measures of gait, however, were significantly associated with falls and mobility disabilities, even after controlling for a host of other commonly reported risk factors and potential confounders.

According to Faulkner et al. [62] the strongest fall risk factors for elderly women are a history of falls and any previous use of anti-epileptic drugs. Fear of falling, poor health, use of antidepressants, and surprisingly a faster gait speed were considered moderate risk factors. It was suggested that faster gait speeds may be associated with less time to react to a disturbance. In agreement with the finding of Tinetti et al. [22], the number of baseline risk factors was positively and strongly associated with future falls.
Mertz [63] found that women (age = 20-87 years) fell more and suffered more fractures than men. Additionally, men and women with lower cardiorespiratory fitness (determined via a submaximal stress test) fell more, after controlling for many other factors. Compared with younger adults, older adults were more likely to fall while walking and more likely to sustain a fracture.

*Fall Risk Reviews*

Nickens [3] reported that falls result from both extrinsic and intrinsic factors, but that the relative contribution of each changes with age. With aging there is a decrease in extrinsic causes due to less participation in risky behaviors. There is concomitant increase in intrinsic causes due to the normal functional decline. It was suggested that there is no common time of day, season, or location among fallers. Women were observed to fall more than men.

Cumming [64] reviewed the association between medication use and falls in older adults. In general, falls were seen to increase with increasing medication use. However, not all medications were associated with falls. For example, most cardiovascular medications were not significantly associated with falls. On the other hand, psychotropics generally doubled the risk of falls.

Shaw [65] reviewed the association between cognitive impairment and falls. Those diagnosed with cognitive impairment (scoring less than 24 on the Mini Mental State Examination) were twice as likely to fall and three times more likely to sustain a fracture. Furthermore, those with cognitive impairment were less likely to make a good functional
recovery following a fall. Gait, balance, and cardiovascular abnormalities were the best predictors of falls in this population.

Tinetti and Kumar [66] found that the strongest risk factors were previous falls; strength, gait, and balance abnormalities/impairments; and specific medications. Understandably, fall risk increased with an increasing number of risk factors.

Cefalu [16] reviewed the association between aging and falls. With increasing age there were suggested to be increasing attentional requirements allocated to balance to prevent falls. The brain exhibits a decrease in processing speed which corresponds to a diminished reaction time in general. Cognitive decline primarily manifested itself in executive function deficits, not memory. There is physical decline as well which is associated with falls. Perhaps most important is the decrease in lower extremity power which may be associated with a decrease in the probability of recovering from a misstep. Possibly as a compensatory strategy, fallers tended to exhibit slower gait speeds, shorter stride length, and a longer stance phase.

Central Nervous System Control of Balance and Gait

Postural control in stance and during locomotion is achieved through the integration of information from higher brain centers, spinal cord interneuronal networks, and peripheral afferents. The degree to which each of these systems contribute to posture depend largely on the complexity of the task.
In the simplest form of standing posture, static balance, peripheral afferents and the spinal cord play the primary role in healthy individuals [67]. The body is assumed to behave as a single-link inverted pendulum pivoting about the ankle. Muscle spindles, anatomically arranged in parallel with extrafusal fibers, provide information on muscle length as well as velocity of stretch and relaxation. Golgi tendon organs, anatomically arranged in series with skeletal muscle, provide information about whole-muscle contractile force. Finally, a variety of receptors in the joints and soles of the feet provide information about joint position and center of pressure. Afferents from these sensors arrive at the appropriate level of the spinal cord where they synapse directly or indirectly with motor neurons. The overall effect is to appropriately control the duration and amplitude of skeletal muscle contraction about the ankle, thus maintaining upright posture.

The spinal cord is capable of far more than moderating static balance. For a century it has been known that interneuronal circuits in the spinal cord are able to generate and maintain straight line walking on a flat surface [68]. This was generally demonstrated in cats with a transected spinal cord permitting no supraspinal influence. Under these conditions, cats were still able to produce unobstructed locomotion. These interneuronal spinal cord circuits have been termed central pattern generators (CPGs). CPGs function by coordinated excitation and inhibition of ipsilateral and contralateral flexors and extensors at the appropriate phases of the gait cycle. Furthermore, CPG activity may be adaptive. Spinalized cats walking on a treadmill exhibit a rapid paw
shake after stepping on a piece of sticky tape. What is so extraordinary is that the shaking response is superimposed upon the gait cycle demonstrating that it had been integrated into the existing CPG command. While the spinal cord demonstrates impressive independent function, it is important to note that spinalized cats move at a slower speed and exhibit lateral instability. This in and of itself suggests the presence of other forms of regulation in healthy individuals.

*Motor Cortex*

Lawrence and Kuypers [69] investigated the contribution of supraspinal structures on movement. They monitored the behavior of monkeys before and following lesions to pyramidal tracts, unilaterally and bilaterally. With bilateral loss, the animals exhibited difficulty releasing food from the hand but had no trouble releasing their grip while climbing. Those with asymmetric loss showed a preference for the contralateral limb, and used that limb with unusual agility and speed. They concluded that the corticospinal pathways provide for speed and agility.

Twenty years later, with advancements retrograde tracers and electrical stimulation/recording, Armstrong [68] reported a more involved role of the motor cortex. Investigations of cats walking on treadmills with a variety of obstacles and walking on ladders demonstrated that the corticospinal tract modulates CPG cycling, especially for cases where a specific foot placement is necessary, or where gait must be voluntarily modified to avoid an obstacle. In support of this functional hypothesis, Beloozerova and Sirota [70] found that cats were unable to step over barriers or walk across ladders following the inactivation of the motor cortex via tetrodotoxin. In a different experiment,
it was observed that walking across a flat floor continued in darkness, but stopped immediately when lights were turned off while walking across a ladder. Beloozerova and Sirota agreed that the motor cortex must at least partially control the accuracy of movement.

The pyramidal tract neuron (PTN) recordings of Drew et al. [71] provide yet another functional role of the motor cortex. Groups of PTNs were found to be temporally associated with flexor/extensor EMG activity during skillful changes in gait. It was found that the descending commands were integrated at the level of the spinal cord and superimposed upon the existing CPG program.

**Brainstem**

There are additional neural structures capable of modifying the existing spinal cord patterns, and many of them reside in various levels of the brainstem. Lawrence and Kuypers [72], again using a monkey model, made lesions in the medulla, pons, and red nucleus. The descending pathways from the brainstem were found to be grouped into ventromedial and lateral systems based on their terminal position in the grey matter of the spinal cord. It was observed that the ventromedial system was largely involved with the maintenance of erect posture, integrated movements of the body and limbs, and directing the course of progression while walking. The lateral system, on the other hand, provided for independent use of the extremity, especially the hand.

Orlovsky [73], using a more direct approach in cats, pioneered investigations into the role of the brainstem during locomotion. Recordings indicated that the Vestibulospinal tract, originating at Deiter’s nucleus, was primarily active in recruiting
extensors prior to the termination of the swing phase and on into the stance phase (a finding of Degtyarenko et al. [74] as well). The Rubrospinal tract, originating at the red nucleus, was most active in recruiting flexors at the start of and during the swing phase. These descending pathways were implicated in modifying the magnitude, but not the timing, of these muscular responses. One of the more interesting findings from this study was that stepping could be induced by excitation of these descending pathways.

Supporting the work of Orlovsky, Rho et al. [75] found short stimulation trains to the red nucleus gave rise to increased flexor activity during the swing phase. Long trains led to an increase in amplitude and duration with no change in gait cycle duration. Long trains to the motor cortex, though, led to a resetting of the step cycle. Recordings taken from the Rubrospinal tract and motor cortex during voluntary gait modifications suggests they are both involved in general gait adjustments, especially just before the contralateral limb would step over an obstacle [76]. The red nucleus was shown to additionally contribute to inter and intra-limb coordination.

Two locomotor regions have been discovered in the brainstem, the subthalamic locomotor region (SLR) and the mesencephalic locomotor region (MLR) [68]. While stimulation of either region initiates locomotion, the MLR is the dominant of the two. The MLR receives input from the basal ganglia and sensorimotor cortex. It was suggested that the MLR may mediate locomotor drives associated with these higher brain centers. The MLR gives rise to reticulospinal pathways that are now known to energize/enhance CPGs and modify force development. In fact, Mori et al. [77] were able to induce galloping and jumping in cats via stimulation of MLR. Stimulation of SLR in
these same cats induced a much more natural walk with searching behavior. Stimulation of dorsal and ventral tegmental fields in the pons suppressed and initiated walking respectively. Taken together, these experiments suggested a locomotor control hierarchy going from rostral to caudal in the brainstem.

**Cerebellum**

The classical view of the function of the cerebellum is that it serves as an error detector between intended movement and the movement that is actually occurring [68]. This hypothesis makes sense as the cerebellum receives information as to the state of the environment, the state of the body, and intended movements [74]. Adaptive learning can then take place based on the perceived error. One of the best known examples of this phenomenon is the air puff with tone classical conditioning experiments in rabbits [78]. When an air puff to a rabbit’s eye is temporally associated with a tone, the cerebellum is highly involved in adaptive learning. Given that the puff and tone occur within a narrow latency period, the cerebellum can modify the motor program controlling blinking for subsequent stimuli.

In addition to error detection and adaptive learning, the cerebellum has been implicated, somewhat controversially, in the rhythmic nature of spinal cord networks in intact animals [68]. Reticulospinal and Rubrospinal pathways are known to be phasically active during the normal gait cycle. Removal of the cerebellum abolishes this rhythmic firing completely. Recordings demonstrated that Purkinje cells exert an inhibitory influence on the interpositus nucleus. The interpositus neurons phasically excite
Rubrospinal neurons. Therefore, when Purkinje fibers are turned off the Rubrospinal tract is disinhibited.

The cerebellum may also serve to initiate and terminate movement [79]. Monkeys were trained to hold a rod with three different wrist positions. In some trials the monkeys were expected to pull the rod. In others the monkeys were expected to hold the rod in position while it was pulled away. Recordings from the cerebellum and motor cortex unveiled the activation sequence for each of these scenarios. When the monkeys initiated the movement, the sequence of activity was dentate, motor cortex, interpositus, followed by muscle contraction. In reaction to the rod being pulled, the sequence was reversed. The results indicated that the dentate and motor cortex were responsible for the postural hold and start of the movement while the interpositus was related to the postural hold and termination of movement. DeLong and Strick [80] demonstrated, though, that cerebellar activity was absent during slow movements, thereby indicating that the contribution of the cerebellum may be activity dependent.

**Basal Ganglia**

To this point systems and their functions have been identified independently. In reality these systems are highly integrated. It has been shown that movement may be stimulated from a multitude of neural structures including a simple reflex arc. Therefore, there must be a control center that determines which movements should be enhanced and which should be suppressed. Mink [81] offered a good example of why a control center would be necessary. If a person is standing on a movable platform and the platform tilts forward, a sequence of muscle activity in the legs and trunk maintains upright posture. If,
while sitting, the person voluntarily plantar flexes the ankle, like stepping on the gas, activation of the aforementioned postural mechanisms would resist the desired movement.

The basal ganglia is the most likely candidate for largely solving this problem of competing motor programs [82]. The striatum (caudate and putamen, collectively) receives excitatory input from nearly every cortical area along with thalamic nuclei. The pallidum (globus pallidus pars interna, substantia nigra pars reticulate, and ventral pallidum, collectively) sends inhibitory output to the MLR and areas of the brainstem involved in postural control. At rest, the pallidum is tonically active, thereby inhibiting undesired movements. When a movement is desired a rather significant excitatory input to the striatum is necessary. This is because the medium spiny neurons that make up the majority of striatal neurons have a high threshold of activation. Striatal activity inhibits the pallidum, which in turn disinhibits the thalamus, MLR, and brainstem areas involved in postural control. The subthalamic nucleus can rapidly terminate movement via excitation of the pallidum (this may be further enhanced by cortical excitation of the subthalamic nucleus).

Integration of movement and postural control

Because the human body in upright posture is inherently unstable, with the center of mass located 2/3 the height of an individual above the ground, postural adjustments are necessary for the simplest tasks (static stance) to complex movements like locomotion and reaching. For voluntary movements there must be feedforward activation of postural muscles [83]. One way this feed forward activation is achieved is through corticospinal
collaterals synapsing with pontine and medullary reticulospinal pathways. In this way, postural adjustments may precede the primary movement intended by cortical stimulation. This coordination between posture and movement is in many cases disrupted in Parkinson’s patients, making a case for the involvement of the basal ganglia in this coordination as well [83]. The reticulospinal system also receives input from the cerebellum and limbs [84]. Taken together, the reticulospinal system, in conjunction with its many inputs, seems to be the dominant means by which postural adjustments precede voluntary movements.

Postural Control during Perturbations and Altered Senses

For static balance, postural control represents the ability to maintain the center of mass (COM) over the base of support (BOS) [85]. Humans are inherently unstable with 2/3 of their body weight distributed 2/3 their height above the ground. Additionally, humans in bipedal stance have a relatively small BOS. In the anterior-posterior (AP) direction the BOS is simply the length of the foot. The medio-lateral (ML) BOS is a function of stance width and therefore fluctuates. Limit of stability (LOS), unlike BOS, is dependent on strength, tissue properties, and confidence.

Static Unperturbed Stance

Equilibrium in healthy individuals is maintained by the contributions of the somatosensory, vestibular, visual, and motor systems [86]. The degree to which each of these systems contributes to postural control depends on the challenge presented. For
static unperturbed stance with feet at shoulder width, upright posture is maintained primarily through the rapid compensatory reactions mediated by somatosensory information [86]. The body is expected to behave as an inverted pendulum pivoting only about the ankle in ML and AP directions [87]. ML sway is limited due to the stability of a wide stance, and AP sway is partially restricted by the intrinsic stiffness of the ankle joint.

Muscle spindles, anatomically arranged in parallel with extrafusal fibers, provide information on muscle length as well as velocity of stretch and relaxation. Golgi tendon organs, anatomically arranged in series with skeletal muscle, provide information about whole-muscle contractile force. Finally, a variety of receptors in the joints and soles of the feet provide information about joint position and center of pressure. Afferents from these sensors arrive at the appropriate level of the spinal cord where they synapse directly or indirectly with motor neurons. The overall effect is to appropriately control the duration and amplitude of skeletal muscle contraction about the ankle, thus maintaining upright posture [67]. This somatosensory information does not have to come from the lower extremities, however, to influence lower extremity musculature. Zhang et al. [88] showed that the presence of very light contact between the index finger and a stationary object was sufficient to influence postural strategy selection (indicated by a decrease in the variability of trunk and leg segments).

**Locomotion**

Postural challenges to locomotion are quite different than those of static stance [85] for the following reasons: 1) the COM and BOS are continually moving, 2) the
COM is not maintained within the BOS, and 3) there are two single-limb support periods that take up to 80% of the gait cycle. Visual and vestibular responses become much more important during locomotion, especially when gait must be skillfully altered. During voluntary gait modifications to visually triggered stimuli (obstacles), there is evidence that collaterals from the motor cortex with instructions on how to alter gait patterns and limb trajectory arrive first at the pontomedullary reticular formation in the brainstem [67]. There is a wealth of evidence to suggest the reticular formation is involved with the integration of posture and movement. The reticular formation also receives inputs from the cerebellum and limbs. With all of the sensory inputs and a copy of the motor program from the motor cortex, the reticulospinal pathway can deliver instructions in a feedforward manner to the appropriate supporting musculature (axial in many cases) in preparation for the intended movement. While a proactive model of obstacle avoidance relies primarily on vision and attention, a reactive model to a perturbation relies heavily on vestibular and somatosensory input [85]. The majority of responses are a combination of polysynaptic spinal reflexes and direct vestibular input to postural muscles.

*Postural Responses to Standing Perturbations*

Platform perturbations in combination with EMG have been widely used to study postural control recovery strategies. These perturbations are typically either horizontal translations or rotations about a ML or AP axis. Nashner and Berthoz [87] studied the responses of subjects to posterior support translation in combination with different visual stimuli. Subjects stood on a movable cart inside a movable visual surround (both the cart and the surround were able to move in the AP direction only). In the normal condition,
the cart moves and the surround is stationary. In this case the visual stimulus is reliable. In the stabilized condition, the surround move in an equal and opposite direction to the cart. Vision is incongruent with vestibular and somatosensory cues for this condition. As a result, the same support translation elicited an inappropriate response (sway increased and gastroc EMG amplitude decreased). In yet a third condition, enhanced, the surround moved in the same direction as the cart. Sway, in this case, was reduced as was gastroc EMG amplitude. One of the more interesting findings from this study was a rapid habituation exhibited with multiple trials under all visual conditions. This indicated a rapid reweighting of the sensory systems from that which was learned to be less reliable (vision) to more reliable information (somatosensory and vestibular). Others have altered vision during surface perturbations by simply using eyes open (EO) and eyes closed (EC) trials [89]. Ledin and Odkvist [89] noted that the absence of vision led to a decreased time to maximum displacement of the center of pressure (CoP) despite there being no significant change in sway amplitude. It was suggested that the EC condition caused a more rapid correction.

Horak and Nashner [90] too studied the responses of AP translations. Rather than manipulating visual conditions, support surfaces of various lengths were employed. On a normal support surface, translations resulted primarily in a distal to proximal sequence of leg muscle activation, characteristic of the ankle strategy. When a short surface was used the activation pattern was reversed, which is characteristic of a hip strategy. Finally, when an intermediate length support surface was used a combination of the two strategies was observed. It was suggested that postural strategies occur on a continuum resulting
from different combinations of ankle and hip strategies. Motor learning did occur in these trials as EMG amplitude and timing were modified with practice. Of interest, when moving from one condition to the next the strategy that was employed previously was initially used for the new condition until adaptation occurred. The response latencies exhibited in this study suggests that the motor learning was a function of the automatic postural control system.

Goodworth and Peterka [91] studied subjects exposed to frontal plane tilting of a support surface. The tilting was automated and was set to occur at different frequencies. Subject stance width was also adjusted from trial to trial. At frequencies < 1 Hz an ankle strategy was observed, whereas frequencies > 1 Hz elicited a hip strategy. The most interesting finding was that of a non-linear response strategy, especially for the narrower BOS conditions (sensitivity was reduced with increasing stimulus amplitudes). This indicates sensory reweighting similar to that observed by Nashner and Berthoz.

*Postural Responses to Altered Senses*

As previously discussed, in cases where the same unreliable information is repeatedly delivered, a reweighting of the sensory systems commonly occurs. This is clinically relevant and useful information as most physiological systems decline with age. Furthermore, there are many pathological conditions which may diminish or remove a sensory system (i.e. peripheral neuropathy, macular degeneration, vestibular loss, etc.). It is therefore of great interest how these systems are organized and weighted.
When researchers or clinicians are interested in identifying the effect of a deficiency many will utilize a sensory organization test (SOT). Conditions 2–4 alter either visual or somatosensory information, leaving two of the three senses unchanged. Conditions 5 and 6 alter both vision and somatosensory information, making the vestibular system the only reliable input. Sway-referencing refers to the support surface and/or the visual surround rotating proportionally to match spontaneous AP sway of a subject.

Horak et al. [92] had three distinct groups of subjects (bilateral labyrinth deficiency, somatosensory loss, and healthy controls) perform all six SOT conditions. The somatosensory loss was artificially induced via hypoxic anesthesia of the feet and ankles. Results indicated no difference between the somatosensory loss group and the control group for any of the six conditions. This suggests that the somatosensory system was not necessary, even in visual sway-referenced conditions, as long as vestibular information was present and reliable. Vestibular loss subjects, too, performed equally to controls for the first four conditions. However, for conditions 5 and 6, where all three systems were either absent or unreliable, subjects went into immediate free fall.

Others have compared those with bilateral vestibular loss to healthy controls utilizing visually induced sway in fixed and sway-referenced conditions [93]. The visual stimulus consisted of sinusoidal surround rotations delivered at a variety of amplitudes. The general response was increasing AP sway with increasing stimulus amplitude. In fact, at lower stimulus amplitudes there were no between-group differences. This suggests that vestibular subjects were not more reliant on visual information than
controls. At higher amplitudes, visually induced sway in normal subjects reached a saturation point. This saturation effect was absent in those with vestibular loss indicating that the saturation in controls was due to an increased reliance on vestibular information. In the sway-referenced condition, both groups swayed to the same extent indicating vestibular loss subjects had not become more reliant on somatosensory information. While these experiments did not show vestibular loss subjects to have an increased reliance on vision for standing balance, an increased visual reliance has been observed in vestibular loss subjects walking from a fixed platform onto a moving sled (like a moving walkway at an airport) [94]. Once again, the weight that each system is given appears to be task specific.

In the absence of subjects with a particular sensory deficiency, many researchers have made attempts to simulate deficiencies via artificial sensory deprivation. One tool that has been utilized to diminish somatosensory information is vibration to the gastrocnemius muscle [95]. Again, under normal conditions upright posture is primarily mediated with somatosensory information. Vibration diminished this process and induced AP sway in healthy subjects. The additional removal of vision led to increased sway but quicker reactions. Vestibular deficiencies have been induced via galvanic stimulation to the vestibular nerve [96]. This resulted in increased ML and AP sway (AP > ML).

In a normal healthy person, the somatosensory, vestibular, and visual systems contribute to upright posture during stance and locomotion. Much of this information is redundant, evidenced by the ability of normal individuals to maintain equilibrium in the
face of misleading information from two of the three systems. With increasing challenges (perturbations, walking, sensory conflicts) the nervous system adapts to potentially misleading information by placing more weight on the information that appears to be consistently reliable. In many cases these adaptations occur rather quickly. An inability to adapt might indicate an underlying pathology.

Attention and Postural Control

The Neurophysiological Basis of Dual Task Interference

Along with a number of other physiological processes, postural control declines with age, and there is evidence that the resultant increase in attentional requirements for postural tasks are a contributing factor to falls in older adults [97]. While the relationship between attention and postural stability may seem obvious, the mechanism by which a cognitive task leads to a change in postural stability is largely unknown. One of the leading theories is dual task interference (DTI). According to the Cortical Field Hypothesis, if two tasks require the same cortical area, they cannot be performed simultaneously [98]. Therefore, in a dual task scenario there would be increases in reaction time and/or error for one or both tasks.

A number of investigators have utilized brain imaging technology to better understand the mechanisms behind DTI [15, 98-100]. Hayashi et al. [99] and Kazui et al. [100] each had their young healthy adult subjects perform three different tasks, serial subtraction, recitation of single-digit multiplication tables, and simple counting. The
serial subtraction was intended to act as a true calculation, while the recitation of multiplication tables was expected to involve no more than rote arithmetic memory. Simple counting served as a control exercise to account for the lingual articulation component of both tasks of interest. Both studies contributed similar results despite having used different imaging technology (Hayashi et al. utilized PET scans and Kazui et al. utilized fMRIs). Serial subtraction and recitation of the multiplication tables primarily led to an increase in the activity of the premotor cortex and the supplementary motor cortex. Hayashi et al. additionally found increased activity in the basal ganglia for the multiplication task. The three brain regions identified are highly involved in maintenance of posture, planning and anticipation of movement, and initiation/suppression of movement. It seems plausible in light of these findings that the aforementioned cognitive tasks would interfere with static and dynamic balance, if simultaneously attempted.

Herath et al. [98] looked at brain activity in response to somatosensory and visual stimuli. Subjects were expected to press a button with their middle finger as soon as they perceived a poke from a blunt stylus. Another button was to be pressed by their index finger in response to a green LED stimulus. These stimuli were offered in rapid succession with short and long inter-stimulus intervals. Three important observations were made. First, the latency and error rate for the second stimulus only increased for the short inter-stimulus intervals (an argument for competition of resources). Second, the supplementary motor area, basal ganglia, and parts of the thalamus were activated in response to both somatosensory and visual stimuli. Finally, the dual task condition led to the activation of cortical regions not utilized for either task independently. It was
suggested that when two concurrent tasks interfere the brain compensates by enlisting the services of other brain regions.

Importantly, the aforementioned imaging studies have only been performed on young healthy adults and it would be wrong to generalize these findings across all age groups. In fact, Van Impe et al. [15] showed the effect of dual tasking is vastly different between young and older adults. All subjects were required to draw circles with and without the secondary cognitive task of serial addition. Their outcome measure of interest was dual task cost, a measure of the percent change in performance from the single task condition to the dual task condition. It was overwhelmingly clear that brain activation was much higher and more expansive in older adults for both conditions. If the Cortical Field Hypothesis (the same brain structure cannot process information for multiple tasks simultaneously) holds, compared with their younger counterparts, older adults may be much more susceptible to DTI because of the greater number of structures utilized at a given time. What may be most alarming about this is that the older adults in this study were relatively young with a mean age of 68. It would not be surprising to find an even greater disparity between young adults and those in their 80s and 90s.

Task Prioritization

When two tasks are performed simultaneously, there are likely to be increases in reaction time and/or error for one or both tasks [98]. Whether cognition or postural control is prioritized has been the topic of much debate.

Schaefer et al. [101] studied 9, 11, and 23 year olds balancing on a wobble board with and without a memory task. While the young adults exhibited poorer balance and
memory performance during the dual task conditions, children actually improved their postural stability at the expense of the memory task. The difference between these age groups appeared to be a result of task prioritization rather than level of interference. Another possible explanation is that adults may have been more embarrassed about making memory mistakes than losing their balance. There have been other reports of balance performance improving during dual task conditions. Melzer et al. [102] observed that older adults standing with their feet together decreased their mediolateral sway and elliptical area with the addition of a cognitive task. EMG recordings taken from the ankle plantar and dorsiflexors indicated a greater degree of co-contraction during the dual task condition. Like the children in Schaefer’s study, these older adults appear to have prioritized postural stability over the cognitive task during the more challenging balance tasks.

There are conflicting reports as well. Huxhold et al. [103] found that both young adults and older adults decreased body sway when assigned a simple cognitive task. However, as the task difficulty increased so did the sway for older adults only. Huxhold et al. hypothesized that under normal standing conditions postural control is a relatively automated process and that the subjects were consciously trying to control their sway. In the dual task condition, their focus was diverted, thereby permitting the automated process to occur unimpeded. Donker et al. [104] contributed similar findings and thoughts. For this study, healthy young adults stood quietly with their eyes closed, with or without a cognitive task. The cognitive task improved postural stability for these subjects. Donker et al. argued that the eyes closed condition led to a greater internal
focus, and that the internal focus might disrupt automated postural control. The cognitive task, on the other hand, led to a greater external focus permitting automaticity of postural control.

Shumway-Cook et al. [97] expected there to be a hierarchy of task priority when presented a balance challenge with a cognitive task, such that postural stability would be favored over the cognitive measure. Instead, postural stability in these older adults deteriorated with no change in cognitive performance. It was suggested that the balance challenge was not difficult enough to express that hierarchy.

Unlike the static balance tasks offered by Shumway-Cook et al., Van Lersel et al. [105] tested the effects of cognitive tasks on gait and posture parameters. The healthy older adult subjects were asked to perform straight line walking at different speeds with and without a cognitive task (serial subtraction or animal naming). Both cognitive tasks resulted in an increase in stride length variability, stride time variability, and ML trunk displacement. However, there was no significant decline in cognitive performance for serial subtraction or animal naming. This would suggest that even during challenging tasks priority is often given to cognition over posture, a finding that may at least partially explain the relatively high proportion of falls that occur during gait in older adults.

The Effects of Aging on Postural Stability and Cognitive Performance during Dual Task Conditions

As previously discussed, there are many cognitive tasks that utilize the same brain regions involved with postural control, planning and anticipation of movement, and initiation/termination of movement. Older adults were also shown to exhibit more
expansive brain activation. It is therefore likely that the effects of DTI are age-dependent with older adults exhibiting the greatest response.

Barin et al. [106] compared young and middle-aged adults. All subjects performed the sensory organization test (SOT) with and without serial subtraction. Both groups performed equally well for all six SOT conditions with and without the cognitive task. It was concluded that the older group was not old enough to express the known attentional consequences of aging.

Maylor et al. [13] studied two groups with the same numerical age disparity as Barin et al., but instead compared middle-aged adults to older adults. ML and AP postural sway was measured in single task and dual task conditions; several different cognitive tasks were evaluated. Compared with middle-aged adults, older adults swayed more during single and dual task conditions and performed more poorly on all cognitive tasks.

This result of increased sway and decreased cognitive performance with age has been replicated in a number of other studies comparing young and older adults [14, 97, 102, 103]. Brown et al. [12] similarly found an age-dependent response to platform perturbations with and without serial subtraction. While all subjects responded to the addition of the cognitive task by taking larger and more frequent steps, the step length and frequency of the older adults were greater than those of the younger adults. Cognitive performance was also significantly lower for the older adults, but only with respect to speed of counting, not accuracy. Prado et al. [14] also observed a slower counting speed but no change in accuracy with age.
Dual Task Interference and Non-linear Dynamics of Postural Sway

There have been many reports of standard center of pressure (CoP) measures, such as path length (PL), the length of the path the CoP travels over the duration of a given trial, and root mean squared (RMS), the square root of the arithmetic mean of the squares of the original values, being insensitive to functionally different subject groups (i.e. gymnasts vs. non-gymnasts, dancers vs. non-dancers, figure skaters vs. non-skaters, young vs. older adults …). In these same groups postural control has been found to be qualitatively rather than quantitatively different. For example, compared with younger adults, older adults are known to exhibit a more regular and less complex CoP signal. Complexity here refers to the degree of irregularity of a time series over multiple timescales and is quantified by multiscale entropy (MSE).

Kang et al. [107, 108] hypothesized a similar relationship would be found with increasing frailty. Subjects clinically classified as non-frail, pre-frail, and frail stood quietly with a wide base on a force plate, with and without serial subtraction. Both standard (RMS and PL) and dynamic (complexity) measures of CoP were calculated. RMS, PL, and complexity of the CoP time series were able to discriminate frail individuals from pre-frail or non-frail individuals (frail subjects swayed more and more regularly than pre-frail or non-frail subjects). However, only complexity was sensitive enough to discriminate pre-frail from non-frail subjects (complexity decreased with increasing level of frailty). Serial subtraction elicited increased sway and more regular sway in all subjects. It was also noted that those with the worst cognitive performance exhibited the greatest change in postural dynamics.
There is sufficient evidence to support the hypothesis of DTI, though the precise mechanism remains elusive. Which task a subject subconsciously or consciously chooses to prioritize is likely influenced by the difficulty of the postural challenge, the type of cognitive task, functional abilities, level of confidence, and a host of other variables. While somewhat inconclusive because of the wide variety of study populations and research protocols, there does appear to be a non-linear age-dependent effect of DTI such that older adults sway more than younger and middle-aged adults. There also appears to be a non-linear change in cognitive performance with age, with respect to counting speed, but not accuracy, though this relationship may not extend to other cognitive tasks. Dual tasking has been shown to influence the amount and variability/complexity of postural sway.

Anatomical and Physiological Adaptations during the Aging Process

The aging process is characterized by impaired function in nearly every physiological and anatomical system. Of interest here are the adaptations which challenge the ability to maintain upright posture during activities of daily living (ADL). These ADL include things like shopping, cleaning, cooking, walking up and down stairs, and taking a bath or shower. Impaired postural control increases the risk of experiencing a fall and typically leads to dependency.
**Muscle Morphology and Strength**

One of the most obvious and influential deleterious adaptations to aging is a loss of strength, but more importantly power. Relative to dynamic activities, quiet standing is not very demanding. Therefore, relatively little strength or power is required. However, strength and power may be quite necessary in response to a perturbation during stance or gait. In a comparison of young and older men responding to decelerating perturbations on a treadmill, Granacher et al. [17] observed that older men not only exhibited lower maximum isometric force, but a decreased rate of force development. Older men also had a lower magnitude of tibialis anterior reflex activity. It has been shown that individuals with a history of falls have weaker ankle dorsiflexors (tibialis anterior), compared with non-fallers [109]. Therefore, decrements in maximum isometric force and rate of force development may contribute to the incidence of falls in older adults.

The mechanisms behind the loss of strength and power with age have been well studied. Lexell et al. [20] studied, post mortem, previously healthy individuals from 15 to 80 years old. A slice of the vastus lateralis from each subject was extracted for further analysis. Muscle area, fiber density, fiber number, and fiber type proportion were determined. A number of differences between age groups were reported. First, fibers from older muscles were more loosely packed, compared with their younger counterparts. This was accompanied by a decrease in fiber density (i.e. the percent of muscle area represented by muscle fibers decreased with age). Second, older muscles exhibited greater fiber size heterogeneity. The average fiber diameter decreased with age. However, this was completely a result of the atrophy of type II fibers, as type I fiber
diameter did not differ by age. Finally, in addition to fiber atrophy, Lexell et al. observed a 40% decrease in the number of fibers from the age of 20 to 80. Lexell and Downham [19] later showed that hypoplasia occurs in type I and type II fibers to the same extent. Therefore, the proportion of type II to type I fibers remains unchanged with age.

Narici et al. [21] examined the difference in anatomical cross-sectional areas (CSA) of the medial gastrocnemius. The subjects were young and older healthy men matched for daily caloric expenditure. The anatomical CSA was determined via computerized tomography scans, while fascicle length and pennation angle were determined via ultrasound. Compared with younger men, older men had smaller anatomical CSAs (combined atrophy and hypoplasia) accompanied by shorter fascicle lengths (decrease in the number of sarcomeres in series) and smaller pennation angles. While not tested, it was suggested that these anatomical changes would influence length-tension, force-velocity, and power-velocity relationships.

In a review by Kostek and Delmonico [18], several other important adaptations were pointed out. First, the mosaic distribution of fiber types in young adults moves to a more grouped distribution in older adults. It is thought that this occurs from denervation followed by reinnervation via axonal sprouting. Second, the level of reactive oxygen species (ROS) rises with age. ROS are known to degrade myofibrillar proteins, impair the function of existing proteins, and lead to an uncoupling between ATPase activity and force production. Finally, aging is characterized by an increase in intramuscular triglycerides (IMTG). This may be partially due to the differentiation of satellite cells into adipocytes. The increased concentration of IMTGs has been independently
associated with impaired muscle function. In summary, these studies demonstrate an anatomical explanation for the observed muscular strength and power deficit seen in older adults.

*Peripheral Receptors and Nerve Function*

While strength and power may be necessary to respond appropriately to a disturbance, the somatosensory system, especially in the lower extremities, is highly involved with detecting the magnitude and direction of a perturbation. Muscle spindles, composed of intrafusal muscle fibers and anatomically arranged in parallel with extrafusal muscle fibers, provide information about muscle length and velocity of contraction. When the whole muscle is lengthened, as would be the case for the tibialis anterior during a forward platform tilt, muscle spindles send an afferent signal to the spinal cord. This afferent signal synapses with an $\alpha$-motor neuron, with projections back to the same muscle, causing the muscle to shorten. The muscle spindle itself must contract and shorten along with the extrafusal fibers to maintain the gain of the system.

Swash and Fox [7] studied distal, proximal, and axial muscle spindles from post-mortem individuals (newborn to 81 years of age). The outcome measures of interest were the number of spindles, number of intrafusal fibers in each spindle, intrafusal fiber diameter, capsular thickness and diameter, and pattern of innervation. Both the capsular thickness and variability of capsular thickness increased with age. It was also shown that aging produced a pattern of denervation and atrophy of the muscle spindles. The consequences of these anatomical changes would be a decreased sensitivity and
magnitude of response. This may partially account for the increased tibialis anterior response latency to platform translations in older adults [109].

Sense of vibration, particularly in the sole of the foot, plays an important role in the postural control system. Receptors such as Pacinian Corpuscles are known to respond to vibration frequencies in the range of 25-600 Hz, though these vibrotactile thresholds may increase with age. Verrillo [9] had subjects from a broad spectrum of ages try to detect vibrations of different frequencies with the palm of their hand. While there were no gender differences, Pacinian Corpuscle sensitivity was shown to decrease with age.

In a much more elaborate study design, Wells et al. [11] tested the vibrotactile sensitivities at 55 different locations on the sole of the foot in young and older adults. Results first demonstrated that there were, in general, three areas of sensitivity for both age groups: the ball and medial arch, the lateral border and the heel, and the toes. While these regions were unaffected by age, the thresholds increased by 250-700% from young to old. It was proposed that the decrease in sensitivity may negatively influence balance and gait. Indeed, Kristinsdottir et al. [110] found that subjects with the lowest vibration perception also exhibited the greatest high frequency sway variability and absolute sway. While not directly attributed to vibrotactile thresholds, older adults are known to exhibit slower gait and shorter stride length [111].

Mechanoreceptors, found in ligaments and skin, convey information about joint position (proprioception) and touch, respectively. In fact, the number of mechanoreceptors has been shown to be directly proportional to the level of proprioception. It has therefore been of interest what happens to the number and function
of these receptors with age. Morisawa [6] studied the acromioclavicular ligament receptors from shoulder surgery patients. It was reported that the total number of mechanoreceptors and free nerve endings decreased with age. Similar findings have come from the study of anterior cruciate ligaments (ACL) of rabbits. Aydog et al. [4] discovered that aging brought about a decrease in the number of ACL mechanoreceptors. In addition, many of the remaining ACL receptors in the older rabbits were morphologically changed in a way that was assumed to diminish their function. Iwasaki et al. [5] reported an age-dependent decrease in the CSA and density of mechanoreceptors, specifically Meissner’s Corpuscles, in the index finger skin.

Because of its major contribution to the maintenance of upright posture, position sense of the ankle has been heavily scrutinized. The typical experimental setup has been described elsewhere [10]. Briefly, the experimental knee of the subject is isolated to limit movement. The ankle is braced into a mechanical device similar to a Biodex. Vision of the ankle is obstructed by an opaque screen. The ankle is then passively plantarflexed and dorsiflexed. The computer screen identifies three target ankle angles. When the subject feels the ankle is at the target angle they are instructed to release an electronic contact device held between their thumb and index finger. This results in a break in the circuit. This protocol is performed with and without vibration of the tibialis anterior. The error between the true ankle position and the perceived ankle position has been shown to increase with age [10, 112].

In addition to changes in CSA, number, and sensitivity of peripheral receptors, there is also impairment of stimulus transmission via peripheral nerves with age. In a
review by Verdu et al. [8], a number of significant findings were highlighted. First, the number of both myelinated and unmyelinated nerve fibers decreases with age. Second, myelin sheaths were found to deteriorate with age, possibly as a result of a decrease in the expression of myelin proteins. Third, older adults exhibit decrements in conduction velocity, sensory discrimination, and automatic responses. Finally, axonal regeneration and reinnervation following injury tend to be delayed and diminished with age.

Aging and Postural Sway

What are the consequences of all of these adaptations with respect to postural control during stance and gait?

Peterka and Black [113] utilized the SOT in 7-81 year olds. A u-shaped curve was observed with respect to the amount of postural sway. Both children and older adults exhibited equilibrium deficiencies, with young and middle aged adults demonstrating the least sway. Children seemed to be especially reliant on and sensitive to somatosensory cues. In a similar study, Speers et al. [114] compared the performances delivered by young and older adults during the SOT. Older adults swayed more than young adults for all conditions; this was especially noticeable in the vision altered conditions. It was concluded that the degradation of the somatosensory inputs with age led to a reweighting of the sensory systems with an increased reliance on vision [114]. It was also noted that the reweighting of sensory systems seemed to be part of the healthy aging process.

Older adults are primarily concerned with maintaining independence, and independence is often lost after falling. There were many anatomical, and consequently physiological, adaptations discussed here that contribute to postural instability. These
adaptations included decreased muscle strength and power, increased reaction time,
decreased receptor sensitivity and peripheral nerve conductivity, and a reweighting of the
sensory systems to increase reliance on vision. Whether or not these processes may be
attenuated or reversed is beyond the scope of this discussion.

Force plates vs. accelerometers regarding their ability to assess postural sway

Postural sway occurs spontaneously during static stance as a result of the unstable
properties of the body [1]. In balance dysfunction, the amplitude and frequency of this
sway may exceed the limits of stability and contribute to a fall. Therefore, it is important
to better understand the mechanisms of postural control for maintenance of upright
posture. Postural control assessments aim to measure deviations in the body center of
mass (COM) or center of gravity (COG). However, COM is difficult to measure directly,
so many investigators estimate its change in position over time by measuring changes in
the center of applied pressure (COP) on a force plate, or by measuring accelerations of
the COM with an accelerometer. In order to properly interpret the results from studies
involving force plates and accelerometers, it is important to fully understand the
relationship between the COP, COM, and COG. In a review, Winter [115] defined all
three. COM was defined as the weighted average of the COM of each body segment in
3-D space. The COG represents the vertical projection of the COM onto the ground (or
force plate). Finally, the COP is the weighted average of all the pressures in contact with
the ground (or force plate). Importantly, while COP and COM are related, COP is completely independent of the COM.

**Validation of Force Plates for Analysis of Postural Control**

Force plates measure the three orthogonal components of force (vertical and two horizontal forces) and their respective moments. These components are collectively used to calculate COP in the sagittal and frontal planes. At any instant the COP is a single point with the coordinates \((X_{\text{net}}, Y_{\text{net}})\). Once the COP coordinates are determined a variety of variables may be calculated which correspond to different aspects of postural control.

Murray et al. [116] studied the relationship between the COP and the COG during static balance by measuring both simultaneously. The COP was calculated from force plate output signal, while the COG was determined via photographic inspection. While the COG exhibited a smooth curve throughout the trials, the COP constantly fluctuated in front of and behind the COG. It was explained that no purposeful fore-aft movement may occur when the COP is equal to the COG. However, the application of the COP behind the GOG would result in a forward sway, while a backward sway would be the result of the COP applied in front of the COG. The COP fluctuations were suggested to represent automatic muscular contractions about the ankle to maintain the COG within the base of support (BOS); for the case of static balance, the BOS is simply a rectangle with sides equal to the length of the foot and stance width. Indeed, Winter [115] concluded that for static balance the average COP should equal the average COG, though the path of the COP would be much greater than the COG.
Hasan et al. [117] also simultaneously measured COP and COG, this time utilizing a force plate and motion capture, respectively. Thirty-one anthropometric measurements were taken to calculate the COG and create an 11-segment rigid body model. Subjects performed three different static balance tasks: two feet with their eyes open, two feet with their eyes closed, and single leg stance with their eyes open. In agreement with Winter [115], the amplitude of the COP was greater than the COG for all conditions, while the offset between the mean COP and COG was negligible. In addition, the COP was highly correlated with the COG for all three static conditions. However, with increasing task difficulty, COG excursions increased less than did COP excursions. This was suggested to indicate an intact and responsive postural control system.

The key assumption underlying the relationship between the COP and COG is that the body is assumed to behave as a rigid structure rotating only at the ankle in the sagittal and frontal planes like an inverted pendulum [118]. Karlsson and Persson [1] studied whether or not the body truly behaves as a rigid inverted pendulum. For this purpose, motion capture analysis of human subject sway was compared to a theoretically constructed ideal model. In this theoretical model, the ankle represents the average position of the COP over a given time period of data collection, and it is the only joint about which movement may occur. Subjects performed four tasks: normal quiet standing with their eyes open or closed, intentionally swaying in the sagittal plane utilizing hip and knee strategies, or intentionally swaying in the sagittal plane utilizing the ankle strategy to the best of their ability.
There were very low correlations between marker and model for hip and knee strategies. For quiet stance with eyes open and eyes closed, marker and model were better correlated. Finally, intentional sway utilizing the ankle strategy led to very high correlations between marker and model. The fact that the correlations were not equal to one indicated the presence of error. The error was suggested to be the result of the body deviating from a rigid state above the ankle, and the error was shown to decrease as the ankle strategy was used to a greater extent. It was concluded that, while motion capture technology was an advantage, motion capture was not necessary for force plate analysis postural sway during static balance tasks. That is to say that in quiet stance the body does not behave as a perfect inverted pendulum, but it satisfies the model to the extent needed to characterize postural sway (not the location of the COM).

Perhaps more important than validating force plate measures based on their ability to properly represent movement of the COM is their correlation with the clinical status of a patient. Karlsson and Frykberg [119] studied the association between a commonly used clinical test, the Berg Balance Scale (BBS), and force plate measures during comfortable quiet stance. The BBS was chosen because of its previously validated reflection of postural stability. The subjects in this study were post-cerebral vascular accident patients. Only the standard deviation of the vertical ground reaction force was significantly correlated with the total score on the BBS. Only the sagittal mean velocity of the COP was moderately correlated with the static portion of the BBS. It was suggested that the BBS and various force plate measures quantify different aspects of postural control.
In addition to demonstrating moderate to high correlations with clinical tests of balance, force plate measures have also demonstrated the ability to discriminate between distinct populations. For example, Reid et al. [120] compared large fiber peripheral neuropathy patients to healthy controls utilizing force plates and a standard clinical neurological evaluation. For the posturographic examination, subjects stood with their feet together or apart with their eyes open or closed. Motor and sensory nerve conduction parameters were also assessed. Patients exhibiting large fiber peripheral neuropathy, as indicated by clinical examination, swayed more than healthy controls for all balance conditions tested. Patients also took longer to react. The total path length of the COP was suggested to be a useful initial screening tool for the detection of peripheral neuropathy. Force plate measures, specifically horizontal ground reaction forces, have also demonstrated the ability to discriminate between subjects with functional ankle instability and healthy controls [121].

Validation of Accelerometers for Analysis of Postural Control

In recent years, accelerometers have shown promise in their ability to properly assess postural sway. Most researchers have placed a single accelerometer at the small of the back near L3, the approximate height of the COM. Accelerometers, depending on the model, have the ability to measure COM accelerations along the anterior-posterior (AP), mediolateral (ML), and/or vertical axes. A potential problem to the use of accelerometers is that they only measure linear accelerations and the true movement of the COM is an arc. However, if the amplitude of horizontal movement is small relative to the height of the COM from the ground, as is assumed to be the case during quiet stance, this effect of
the arc may be negligible. However, any measurement or analysis errors should be avoided if possible.

Moe-Nilssen and Helbostad [24, 26] described methodology to remove or diminish a number of sources of variability in ACC data. One source of error in standard placement of the ACC is pelvic tilt. Pelvic tilt causes a deviation of the AP and vertical axes. Moe-Nilssen and Helbostad [26] was able to take advantage of a relatively constant vertical acceleration due to gravity. Knowing the true vertical acceleration due to gravity and what was being experimentally measured, a trigonometric correction was able to be applied thereby negating the effect of pelvic tilt. It was shown that simply correcting for pelvic tilt improved the sensitivity of the measurement dramatically. While other sources of variability were identified, correcting for them did not further improve the sensitivity of the measurement.

Like force plate measures, accelerometer measures have demonstrated the ability to discriminate between distinct subject groups. Kamen et al. [23] utilized a single-axis accelerometer, positioned to measure AP pelvic accelerations, to compare young and older adults. Subjects completed multiple trials of three balance tasks: floor with their eyes open, on foam with their eyes open inside a visual dome, and on foam with their eyes closed. Compared with young adults, older adults produced greater acceleration amplitudes and frequencies. The accelerometer demonstrated good reliability for all three conditions as well. They were unable to correct for pelvic tilt because of the lack of a vertical acceleration measure.
Moe-Nilssen and Helbostad [26] also compared young and older adult groups. Again, older adults exhibited greater acceleration amplitudes, AP and ML, than young adults. In addition, the tri-axial accelerometer proved capable of discriminating between balance conditions; pelvic accelerations increased as the task difficulty increased. In contrast to the reliability values reported by Kamen et al. [23], Moe-Nilssen [24] found the reliability to be low for less challenging balance tasks (i.e. standing on the floor with eyes open or wearing an opaque mask). Reliability for a more challenging condition, single leg stance with eyes open, was good. It was noted, however, that the absolute reliability was high for all three conditions. It was suggested that the range of values for the easier tasks was too small to produce good relative reliability.

Accelerometers have also been tested for concurrent validity with previously validated clinical tests. Cho and Kamen [25] compared the abilities of a single-axis ACC, ten meter straight line walk, functional reach, rapid stepping test, and standard Romberg to discriminate between fallers and non-fallers. For the accelerometer measures, subjects stood on a firm surface or foam surface with their eyes open or closed for a total of four different conditions. The accelerometer was able to discriminate between groups and balance conditions. There was no effect of group on the functional reach outcome. For the Romberg test, only standing on the right leg with eyes open condition showed a group difference. Both the rapid stepping test and the ten meter straight line walk were significantly different between fallers and non-fallers. No correlations between accelerometer and any of the clinical tests were presented or
discussed. Like Kamen et al. [23], there was no pelvic tilt correction because of the lack of a vertical acceleration measure.

O’Sullivan et al. [27] used a similar protocol to that used by Cho and Kamen to compare a tri-axial accelerometer to the BBS and three meter Timed Up-and-Go (TUG) in an older adult population (mean age = 78 years). The accelerometer was able to discriminate between balance tasks. Only for one condition, standing on foam with eyes open, was accelerometry able to discriminate fallers from non-fallers. However, that condition alone offered good sensitivity (58.3) and specificity (80.0) and was significantly correlated with the BBS (-0.829) and the TUG (0.621). The BBS and the TUG were also correlated with each other (-0.77).

**Force Plate and Accelerometer Comparison**

A few investigators have assessed balance with accelerometers and force plates simultaneously. Mayagiotia et al. [28] had subjects perform four different balance tasks on a force plate: comfortable feet placement or feet together with eyes open or closed. Both accelerometers and force plates exhibited the same patterns. Only in one case was the force plate more sensitive to condition than the accelerometer. Conversely, there were four cases where the accelerometer was more sensitive than the force plate.

In a very different study, Alderton et al. [29] employed the simultaneous use of an accelerometer and force plate to identify a change in postural strategy resulting from plantar flexor fatigue. Subjects were assessed under a single leg stance condition before and following a fatiguing set of calf raises. Both acceleration and COP amplitudes and amplitude variability increased following fatigue. The COP velocity decreased following
fatigue. Increases in amplitude accompanied by decreases in COP velocity indicated a change in postural strategy from baseline to fatigue. Importantly, the acceleration and COP data exhibited similar characteristics and were significantly correlated.

Finally, Whitney et al. [30] compared the ability of an accelerometer and force plate to assess balance in adults (age = 18-85) performing the SOT. The acceleration and force plate data were significantly correlated for all six conditions. However, the relationship changed with age indicating a change in postural strategy with age. Specifically, it was observed that older adults tend to use the ankle strategy less evidenced by increasing accelerations at the hip without a change in COP. Since older adults may often utilize a strategy that violates the assumption of an ankle strategy, it was argued that pelvic accelerations may be more valid than COP.

Overall, force plates and accelerometer measures have demonstrated reliability, concurrent validity, and discriminative validity. Furthermore, accelerometers and force plates contribute highly correlated information. Disadvantages of force plate measures for the assessment of postural control are that 1) the assumption that the body behaves as a rigid inverted pendulum may not be accurate, especially during more difficult balance challenges and in older adults, 2) the cost and lack of portability and accessibility may be prohibitive, and 3) different force plate measures appear to represent different aspects of postural control. The major disadvantage of accelerometers is that a correction for pelvic tilt must be made to achieve the greatest sensitivity. In addition to the lower cost and greater portability, the major advantage of an accelerometer over a force plate is that it more directly measures COM movement.
Fall Prevention

*The role of physical activity in fall prevention programs*

Falling in older adults has received much well deserved attention. One area of interest has been the identification of individuals at risk of falling. Physical inactivity; normal age-related declines in strength, power, and reaction times; and balance and gait abnormalities have all been independently associated with the risk of future falls \[17, 22, 57, 61, 122-125\]. It stands to reason, then, that interventions designed to address these deficiencies would reduce the risk of falls.

Research does generally support the use of multifaceted exercise programs for the purpose of fall prevention \[126\]. However, the effect is rather marginal with a Number Needed to Treat (NNT) of 16. While exercise interventions, on average, were shown to reduce fall risk, they did not significantly influence fall rates. Furthermore, a component analysis revealed that there was no difference between balance, endurance, flexibility, or strength interventions on fall risk.

One reason for the statistically significant but marginal effectiveness of exercise interventions might be study design. For example, Buchner et al. compared the effect of strength training, endurance training, or combined training on fall risk \[127\]. The endurance training was performed on a cycle ergometer while the strength training was performed in non-weight bearing positions using machines. While gains were seen in strength and cardiorespiratory fitness following training, no improvements in gait or balance were observed. This is not all that surprising considering neither mode of exercise challenged postural stability as would have been the case with treadmill walking.
and body weight bearing resistance exercises (i.e. chair stands, calf raises, standing leg curls …).

Others reporting marginal effects of exercise have concluded that the exercise intensity and study duration were insufficient to elicit the desired physical adaptations [128]. In contrast, even a high intensity lower extremity strength training program has proven ineffective at reducing fall risk [129].

The most likely reason for small effect size of exercise-based interventions is that fall risk is multifactorial and includes many components completely unrelated to physical fitness. These independent risk factors are things like environmental hazards (i.e. stairs, obstacles, slippery walking surface, loose rugs …), medication use (i.e. sedatives, anticonvulsants, antidepressants …), neurological disorders (i.e. vestibular dysfunction, diabetic neuropathy, Parkinson’s, Huntington’s …), visual impairment, cognitive impairment, and, as will be discussed later, confidence.

Chang et al. [126] demonstrated that multifactorial risk assessment and management programs were more effective than exercise interventions alone at reducing fall risk and rate of falls (NNT = 11). These programs first performed comprehensive assessments including orthostatic blood pressure, vision, balance and gait, drug review, activities of daily living (ADL), cognitive evaluation, and environmental hazards. Then, interventions were aimed at modifying each of these factors as needed for each individual.

From a research perspective, a multifactorial design is not ideal. First, each subject does not receive the same treatment, which makes the results less generalizable.
Second, the results of multifactorial interventions are more difficult to analyze (difficult to determine effect sizes of individual components). Finally, a multifactorial design is much more labor intensive and time consuming. From a clinical perspective, however, it appears that a multifactorial approach to fall prevention is the most effective and most appropriate strategy.

Exercise interventions are extremely difficult to administer, especially within a research construct, and investigators contributing to the literature on exercise interventions to reduce falls in older adults should be commended. These studies had several strengths. Intervention durations were generally three months to one year with an extensive prospective follow-up period. Sample sizes were often quite large which greatly improves the statistical power and potentially the generalizability of the findings. The exercise programs were mostly monitored rather than self-report. Finally, compliance and adherence were typically recorded and factored into the analyses.

There were also several limitations. The same exercise program was administered for all subjects, which does not take into account individual need. While intervention durations were significant in length, few studies monitored the level of ongoing physical activity during the follow-up period. Baseline measures of performance and functional abilities were not reassessed at the end of the follow-up period in most cases. Finally, in some cases the populations were very specific subsets of older adults (i.e. frail adults in nursing homes), which reduces the generalizability of the results.
The role of balance confidence in fall prevention

Low balance confidence (LBC), or fear of falling (FOF), has been indirectly and independently associated with falls [130]. Wijhuizen et al. [131] suggested a theoretical model characterizing the relationship between FOF and falls. In this model, FOF is associated to the occurrence of falls through a decreased level of physical activity. While decreased physical activity may decrease the incidence of risky behavior [59, 63], it also accelerates the aging related declines in muscle strength, power, and reaction times which are independent risk factors for falls [16, 66]. FOF is also suggested in this model to lead to hesitancy, a change from a fast mode of postural control to a slow mode of postural control. The fast mode is reactive and automatic, while the slow mode is cognitive and largely dependent on visual cues, which may increase the risk of falls.

Therefore, interventions aimed specifically at improving balance confidence may reduce the risk of falls. Rand et al. [132] reviewed the efficacy of such interventions. Interventions in this review were categorized into the following three groups: general exercise, Tai Chi, and multifactorial. Exercise interventions, including strength and functional balance exercises, significantly improved balance confidence, though the effect size was low. Multifactorial treatments, which included an exercise component in some studies, had similar effect sizes to exercise alone. In contrast, Tai Chi significantly improved balance confidence with a moderate effect size. Rand et al. suggested that the small effect sizes of exercise and multifactorial treatment were possibly due to balance confidence being a secondary aim to fall prevention. It was concluded that Tai Chi,
which challenges postural control through a reduced base of support, was the most beneficial intervention of those studied for improving balance confidence.

Integral components of a falls prevention intervention

When designing a fall prevention intervention a tradeoff between specificity and feasibility exists. It is clear that fall risk is multifactorial and varies greatly from one individual to the next. Therefore, ideally an intervention would begin with a comprehensive risk assessment of each individual followed by a very specific treatment plan. The treatment plan should include, as needed, general physical activity recommendations, specific balance and gait exercises, resistance training, environmental hazard modifications, minimal medication use, use of an assistive device, and a particular emphasis on improving balance confidence through activities like Tai Chi. As the treatment effect will vary by individual, it is important to periodically perform follow-up assessments and modify the treatment plan as needed.

In reality, though, interventions are more likely to be group-based and more generalized. Not only are group-based activities more feasible and cost effective, older adults are more likely to participate in group activities. Therefore, it is important to design comprehensive interventions that address the most common risks of falling in older adults.
Chapter 3: Reliability and Validity of a Wireless Accelerometer for the Assessment of Postural Sway

Introduction

Postural sway occurs spontaneously during static stance as a result of the unstable properties of the body [1]. In balance dysfunction, often a consequence of the normal aging process, the amplitude and frequency of this sway may exceed the limits of stability and contribute to a fall. Therefore, it is important to develop reliable and valid tools to assess and quantify postural sway. Postural sway assessments aim to measure deviations in the body center of mass (COM). Because the COM is difficult to measure directly, many investigators estimate its change in position over time by measuring changes in the center of applied pressure (COP) on a force plate. Importantly, the key assumption underlying the relationship between the COP and COM is that the body behaves as a rigid structure rotating only about the ankle in the sagittal and frontal planes like an inverted pendulum [118]. When this assumption is not met, as Whitney et al. [30] found to be the case in their older adult population, the path of the COP may no longer accurately reflect the path of the COM.

One way to avoid the assumption of a rigid body above the ankle is to place a sensor closer to the approximate COM. In recent years, accelerometers have shown promise in their ability to properly assess postural sway. Most researchers have placed a
single accelerometer at the small of the back near lumbar vertebrae 3, the approximate height of the COM. Accelerometers, depending on the model, have the ability to approximate linear COM accelerations along all three axes. Since the vertical position of the COM is not expected to change much during static balance tasks, the primary interest in postural sway studies is to quantitate the magnitude of the two horizontal accelerations, mediolateral (M-L) sway in the frontal plane and anterior-posterior (A-P) sway in the sagittal plane. Because accelerations are both positive and negative it is common practice to take the root mean square (RMS) of the acceleration signal to give the mean acceleration amplitude over time.

In addition to good reliability [23, 24], accelerometers have consistently demonstrated the ability to discriminate between distinct populations [23, 25-27], and balance tasks [23, 25-30]. In fact, Mayagoitia et al. [28] found that, for four out of five balance conditions, an accelerometer was better than a force plate at discriminating between balance conditions.

A wireless tri-axial accelerometer has recently come on the market that is smaller, less expensive, and more sensitive than those previously utilized in postural control studies. However, no reliability or validity studies on this device exist. Therefore, the purpose of the present study was to assess the new accelerometer’s test-retest reliability and ability to discriminate between balance tasks with alterations of sensory inputs. We first hypothesized that the within-subject trial-to-trial and block-to-block variability in the RMS of the horizontal pelvic accelerations would small indicated by highly correlated values. Postural sway amplitude and pelvic acceleration during static stance generally
increases with balance task difficulty [23, 25-30], such as altered vision or surface. Therefore, we also hypothesized that the RMS of the horizontal pelvic accelerations would be sensitive to changes in vision, supporting surface, and plane of sway. Knowledge gained from this study may support the continued use of this accelerometer in the assessment of postural sway in both a research and clinical setting.

Methods

Twenty community-dwelling healthy older adults (mean age 81±4 years: 8 male, 12 female) were recruited from a single independent living facility. Participants were included if they were ≥65 years of age and able to walk unassisted. Potential participants were excluded if they had experienced a fall within the previous 12 months, had ever been diagnosed with cardiopulmonary or neurological disease, had any documented orthopedic injury, had a diagnosed balance disorder, or if they used a walking aid or needed physical assistance to walk.

Measurement / Instrumentation

We utilized a wireless tri-axial accelerometer (YEI 3-Space Sensor™ Wireless, Yost Engineering Incorporation, Portsmouth, OH) complemented by a wireless dongle (Wireless 2.4GHz DSSS Dongle, Portsmouth, OH) to assess postural sway. The accelerometer had a range of ±2g, 14 bit resolution, and 0.00024g/digit sensitivity. It had the dimensions 35mm X 60mm X 15mm, and a weight of 28 grams, thereby making it completely portable and unobtrusive. The accelerometer was fixed to the lower back of
the participants (lumbar vertebrae 3) with a single 15 cm strip of surgical tape. Raw data was collected at approximately 250 Hz and low-pass filtered at 55Hz.

All data processing and analyses were performed with custom made Matlab programs (Matlab R2009a, Natick, MA, USA). The raw acceleration signal from each trial was transformed to correct for pelvic tilt and adjust for low frequency utilizing the protocol developed by Moe-Nilssen et al. [24, 26]. Following processing of raw data, we calculated the RMS for the A-P and M-L acceleration data.

*Procedures*

Following the informed consent process, the accelerometer was taped to the lower back of the participant and a gait belt for safety was positioned just below it. Participants completed three 30-second trials of balance conditions 1-4 from Table 1 in random order. The firm surface was a concrete floor with a thin carpet layer. The compliant surface was a gel mat (Wheelchair Skin Protection Cushion Gel E 3 Deluxe) with the dimensions 18 X 18 X 3 inches. For all conditions, participants stood barefoot with their feet together 1 m from a wall and facing that wall. For each test and trial, participants were asked to cross their arms over their chest and remain as motionless as possible for the duration of the test. A brief rest was provided between each trial. Participants completed two blocks of testing, with a 20 minute rest between testing blocks (the accelerometer was removed from the participant during the break). In all, a total of twenty-four 30-second trials were performed by each participant. Participants wore the gait belt and were guarded by a researcher throughout all testing to prevent falls.
Statistical Analyses

All statistical analyses were carried out using the Statistical Package for Social Sciences (SPSS) version 20.0. Intraclass Correlation Coefficients (ICC) were used to assess the trial-to-trial and block-to-block reliability. Two-way repeated measures ANOVA models (four conditions and six trials per condition as factors) were used to test for a main effect of balance condition on A-P and M-L acceleration RMS. Mauchley’s test revealed a violation of the sphericity assumption, so the Greenhouse-Geisser adjusted values from the ANOVA output were used to assess statistical significance. Pairwise comparisons of the four conditions were made post hoc with a Bonferroni correction for multiple comparisons (alpha adjusted from 0.05 to 0.008). Repeated measures ANOVA models (two axes and six trials per axis as factors) were also used to test for a main effect of acceleration axis on acceleration RMS for each balance condition. Significance was set a priori at alpha = 0.05.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Eyes Open</th>
<th>Eyes Closed</th>
<th>Firm</th>
<th>Compliant</th>
</tr>
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<tbody>
<tr>
<td>1</td>
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<tr>
<td>4</td>
<td></td>
<td>X</td>
<td></td>
<td>X</td>
</tr>
</tbody>
</table>

Table 1. Test conditions
Results

Table 2 shows the ICCs representing the trial-to-trial and block-to-block reliability for A-P and M-L accelerations for all four balance conditions. Coefficients were all good to excellent with values ranging from 0.736 to 0.972 for trial-to-trial and from 0.760 to 0.954 for block-to-block. Furthermore,

<table>
<thead>
<tr>
<th></th>
<th>Condition 1</th>
<th>Condition 2</th>
<th>Condition 3</th>
<th>Condition 4</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>A-P</td>
<td>M-L</td>
<td>A-P</td>
<td>M-L</td>
</tr>
<tr>
<td>Trial</td>
<td>0.841</td>
<td>0.874</td>
<td>0.847</td>
<td>0.972</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.832</td>
<td>0.870</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.736</td>
</tr>
<tr>
<td></td>
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<td></td>
<td></td>
<td>0.901</td>
</tr>
<tr>
<td>Block</td>
<td>0.911</td>
<td>0.929</td>
<td>0.772</td>
<td>0.930</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.846</td>
<td>0.760</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.830</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.954</td>
</tr>
</tbody>
</table>
Figure 1. Effect of balance condition on pelvic acceleration RMS. g = unit of gravity (9.8 m/s^2). Error bars represent one standard deviation from the mean. * indicates a significant condition effect at a level of 0.008. † indicates a significant acceleration axis effect within a given condition at a level of 0.05.

Figure 1 shows the relationship between balance condition and acceleration RMS. There was a significant main effect of balance condition on A-P acceleration RMS ($F(1.50, 28.47) = 22.92, p < 0.001$) and M-L acceleration RMS ($F(1.55, 29.36) = 59.60, p < 0.001$). There was also significant stepwise increase in A-P and M-L acceleration RMS from conditions 1 to 4. With respect to A-P sway, the mean acceleration RMS increases from conditions 1 to 2, 2 to 3, and 3 to 4 were $0.0031\pm0.0008g$, $0.0035\pm0.0011g$, and $0.0075\pm0.0022g$, respectively. Similarly, the M-L mean
acceleration RMS increased significantly from conditions 1 to 2, 2 to 3, and 3 to 4 (0.0025±0.0007g, 0.0068±0.0010g, and 0.0084±0.0018g, respectively). A-P and M-L acceleration RMS were statistically similar when standing on a firm surface (conditions 1 and 2). However, M-L acceleration RMS significantly exceeded A-P for condition 3 ($F(1, 19) = 4.61, p = 0.045$) and condition 4 ($F(1, 19) = 6.68, p < 0.018$).

**Discussion**

**Reliability**

The aim of the present study was to assess the test-retest reliability and ability of a new accelerometer to discriminate between balance tasks with different sensory conditions. In support of our first hypothesis, both the trial-to-trial and block-to-block reliability were good to excellent. This indicates that both the participant performance and acceleration measurement varied within an acceptable range.

The ICCs from the present study were substantially higher than those reported previously. ICCs for trial-to-trial reliability during two-foot static balance on a firm surface were reported by MoeNilssen (1998) to be in the range of 0.42 to 0.56. Importantly, their participants stood with their feet 10 cm apart creating a wider base of support (BOS) and better M-L stability than was afforded our participants standing with their feet together. While the ICCs were lower for two-foot wide stance, the rage of ICCs for a single leg stance (0.69 to 0.84) were much more in line with the present findings. Reliability has been shown to improve with increasing task difficulty [24, 27, 30] and that may have contributed the higher correlations presented here. An argument
against this explanation is that there was no apparent association between ICC magnitude and balance condition.

Ultimately, researchers and clinicians simply require a measurement that has an error associated with it that is significantly smaller than the expected change in a person following an intervention or spontaneous change in functional status. To our knowledge, there have been no longitudinal or randomized controlled trials utilizing accelerometry as a measure of postural sway. Therefore, we are currently unable to discuss the magnitude of the trial-to-trial and block-to-block variability in this practical context.

**Validity**

Though the most relevant form of validity is the ability to discriminate between clinically distinct populations, for the purpose of convenience, researchers will often investigate whether or not an assessment technique is able to discriminate between test conditions. In agreement with others [23, 25-30], the accelerometer investigated here was able to discriminate between visual and surface conditions and sway axis. Specifically, pelvic accelerations increased with the removal of vision and/or standing on a compliant mat.

In the present study, when standing on a firm surface, A-P and M-L accelerations were statistically similar; when standing on a mat, M-L accelerations were significantly greater than A-P accelerations. This contradicts the findings of others who consistently observed greater A-P than M-L pelvic accelerations [24, 28]. However, whereas participants in the present study stood with their feet together on a mat, participants in those studies stood with their feet 10 cm apart on a firm surface giving them a wider
BOS, thereby stabilizing M-L sway. In one study where the BOS was narrowed by balancing on one foot, there was little difference between A-P and M-L pelvic accelerations [24]. Horak and Nashner [90] demonstrated that a smaller BOS led to a change in balance strategy from an ankle strategy to a hip strategy and this likely accounts for the larger M-L accelerations observed in the present study.

**Study Limitations**

There were a number of limitations in the present study. First, our study population was fairly homogenous and healthy, so these results may not extend well to other more diverse groups (e.g., individuals with a lower functional status, neurologic disease, dementia, etc.). Second, the reliability and validity of this device are specific to the four balance conditions evaluated here and may not translate to the many variations that exist in common assessments of postural sway (e.g., standing on one foot, tandem stance, semi-tandem, etc.). Third, for feasibility, both blocks of tests were completed on the same day with only a 20 minute rest. Fatigue may have contributed to some of the variability between blocks for some participants. Finally, while wireless technology improved the versatility of the accelerometer, the lack of onboard memory necessitated the presence of a laptop for live retrieval of the data.

**Conclusion**

Both of our hypotheses were supported. The accelerometer exhibited good to excellent trial-to-trial and block-to-block reliability. Additionally, the accelerometer was sensitive to differences in visual and surface conditions and acceleration axis. Balance
task discrimination, while important, is only one part of the validation process.

Evaluating the ability of this accelerometer to discriminate between fallers and non-fallers is an area of continued research by these authors.
Chapter 4: Validation of an Accelerometer-based Balance Assessment as a Fall Risk Screening Instrument

Introduction

One out of every three older adults (those ≥ 65 years of age) fall each year [2]. Twenty to 30% of falls result in moderate to severe injuries, and older adults are hospitalized for these fall-related injuries five times more often than any other cause [2]. In addition to increased dependency and decreased quality of life, fall-related injuries are expected to result in annual medical costs of $55 billion by the year 2020 [2].

While falls may result from extrinsic and intrinsic factors, the relative contribution of each changes with age [3]; with aging there is a decrease in extrinsic causes due to less participation in risky behaviors, but there is also a concomitant increase in intrinsic causes due to normal age-related functional decline. Part of the normal aging process involves decrements in peripheral receptors and nerve function [4-11], cognition and attention [12-16], and muscle strength and power [17-21], all of which are highly involved in the regulation of upright posture. Not surprisingly, balance abnormality/impairment has been identified as a major risk factor for future falls [66].

Researchers and clinicians have become increasingly interested in the assessment of posture and balance, with the goal being to characterize movement of the center of mass (COM) in response to internal and external perturbations. Recently, several
investigators have suggested that in a clinical setting the use of accelerometry as a tool for approximating horizontal COM accelerations would facilitate fall risk stratification and enhanced care [133].

A wealth of research has shown that a single accelerometer placed on the lower back (at the estimated COM) is capable of discriminating between distinct populations and clinical conditions [23, 26-29, 134]. When compared with force plate center of pressure (COP) measurements (one of the most commonly used postural sway assessment tool), accelerometers have been as good as, or better at discriminating populations and tasks, such as a variety of balance conditions [28].

A wireless tri-axial accelerometer has recently come on the market that is smaller, less expensive, and more sensitive than those previously utilized in postural control studies. We suggest that an accelerometer-based balance assessment (ABA) using this device may serve as a convenient alternative or supplement in the assessment of postural control and fall risk stratification, both in a clinical and research setting. The ABA employs four different sensory conditions (standing on a firm floor or compliant mat with eyes open or closed) with and without a cognitive task (counting backward by 3’s).

Because no validity studies on this assessment protocol exist, the purpose of this study was to assess the discriminative validity and concurrent validity of the ABA with respect to its ability to identify individuals with a self-reported history of falls.

We tested the hypotheses that 1) compared with the Berg Balance Scale (BBS), TUG, and the Activities-specific Balance Confidence (ABC) scale, the ABA would be as good or better at discriminating fallers from non-fallers, and 2) the root mean squared
(RMS) of the horizontal pelvic accelerations achieved during the ABA would be significantly and inversely correlated with the average total scores on the BBS and on the ABC scale, and significantly and positively correlated with TUG times.

Methods

Study Design

Ninety-five independent living community-dwelling healthy older adults with ages ranging from 68 to 99 years (mean age 86±6 years; 29 male and 66 female) were recruited from multiple facilities. Only those ≥ 65 years of age and able to walk unassisted for at least 40 feet were included in this study. Residents were excluded if they had a diagnosed balance disorder or if they had a neurological condition that might negatively impact balance, such as Parkinson’s disease, multiple sclerosis, neuropathy, or vestibular or cerebellar disorders.

After obtaining informed consent, but prior to testing, participants filled out a medical history questionnaire where they reported their age, height, weight, gender, number of medications, activity level, arthritis, joint replacement, use of an assistive device, and the number of falls in the past six months. A fall was defined as a sudden unintentional change in position causing one to land on a lower level [135]. Participants wore a gait belt and were guarded by a researcher for the duration of the test session.

Participants then completed, in random order, the BBS, TUG, ABC scale, and ABA. The BBS, TUG, and ABC scale were administered by trained physical therapy students who had achieved an inter-rater reliability greater than 0.9. A different investigator administered the ABA. The average duration for a participant to complete all tests was 30-40 minutes.
Measurement / Instrumentation

We utilized a wireless tri-axial accelerometer (YEI 3-Space Sensor™ Wireless, Yost Engineering Incorporation, Portsmouth, OH) complemented by a wireless dongle (Wireless 2.4GHz DSSS Dongle, Portsmouth, OH) to assess postural sway. The accelerometer had a range of ±2g, 14 bit resolution, and 0.00024g/digit sensitivity. It had the dimensions 35mm X 60mm X 15mm, and a weight of 28 grams, thereby making it completely portable and unobtrusive. The accelerometer was fixed to the lower back of the participants (lumbar vertebrae 3) with a single 15 cm strip of surgical tape. Raw data was collected at approximately 250 Hz and low-pass filtered at 55Hz.

All data processing and analyses were performed with custom made Matlab programs (Matlab R2009a, Natick, MA, USA) and carried out by an unbiased 3rd party who had no access to the medical history or clinical test performance of the participants. The raw acceleration signal from each trial was transformed to correct for pelvic tilt and adjust for low frequency utilizing the protocol developed by Moe-Nilssen et al. [24, 26]. Following processing of raw data, we calculated the RMS for the A-P and M-L acceleration data.

Clinical Outcome Measures

The BBS [33] is a 14 item test battery that measures static and dynamic balance, sit-to-stand and stand-to-sit transitions, and functional ability to perform simple mobility tasks. Each item is scored 0-4 with four representing the highest level of function. The highest overall possible score is 56. The BBS is one of the most frequently used instruments to assess fall risk and has been considered the gold standard [34]. The TUG [136] measures the time taken for a person to stand from a chair, walk around a marker three meters away, and
return to a seated position in the same chair. A shorter time represents better function. The ABC scale [54] is a self-evaluation of balance confidence under a variety of normally encountered conditions. Each of these three instruments has demonstrated reliability and validity, and is commonly used in clinical and research settings.

*Accelerometer-based Balance Assessment Protocol*

Table 3 outlines the eight possible combinations of vision, surface, and attention utilized in this protocol. Participants completed no fewer than two and up to six 30-second trials of each condition in random order. For all conditions, participants stood barefoot with their feet together 1 m from a wall and facing that wall. Participants were asked to cross their arms over their chest and remain as motionless as possible for the duration of the test.

Conditions 5-8 required the participants to count backward by 3’s (serial subtraction) from a randomly generated starting number in the range of 100-500. A brief rest was provided between each trial. The goal was for each participant to achieve two successful trials for each condition taking ≤ 6 attempts. Taking a step, arms coming uncrossed, eyes opening, or need of assistance resulted in a failed attempt. The best attempt of two successful trials (the one with the lowest RMS values) for each condition was included in the data analysis [137].
Statistical Analyses

All statistical analyses were carried out using the Statistical Package for Social Sciences (SPSS) version 20.0. For the ABA, conditions 1-8 were progressively challenging. As such, there was an inverse relationship between task difficulty and the number of study participants able to complete one successful trial out of six possible attempts (Table 4). Missing data were addressed using the Impute function in SPSS.

Table 3. Accelerometer-based balance assessment test conditions

<table>
<thead>
<tr>
<th>Condition</th>
<th>Vision</th>
<th>Surface</th>
<th>Attention</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>2</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>3</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>4</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>5</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>6</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>7</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>8</td>
<td>X</td>
<td>X</td>
<td>X</td>
</tr>
</tbody>
</table>

SS = Serial Subtraction

Table 4. Percentage of participants with missing data by condition

<table>
<thead>
<tr>
<th>Condition</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
</tr>
</thead>
<tbody>
<tr>
<td>% Missing Data</td>
<td>0.0</td>
<td>5.4</td>
<td>7.5</td>
<td>28.0</td>
<td>1.1</td>
<td>6.5</td>
<td>5.4</td>
<td>23.7</td>
</tr>
</tbody>
</table>

The fall risk discriminative power of various combinations of the TUG, BBS, ABC, and/or the ABA was evaluated using logistic regression and chi-square analyses.
Logistic regression models had the binary outcome of a history of ≥ 1 fall vs. no falls, ≥ 2 falls vs. ≤ 1 fall, or ≥ 2 falls vs. no falls. Receiver operating characteristic (ROC) curves were created for each model and the corresponding area under the curve (AUC) subsequently calculated. Each model was also used to estimate the probability of one or multiple falls for each participant and, by setting different threshold probabilities, we utilized the estimated probability to predict whether each participant was a faller or not by observing whether the estimated probability exceeded the threshold probability. Contingency tables were then used to calculate the sensitivity, specificity, positive predictive value (PPV), negative predictive value (NPV), positive likelihood ratio (PLR), and negative likelihood ratio (NLR).

Finally, concurrent validity was evaluated by looking at the relationship between the acceleration RMS and the TUG, BBS, and ABC scale. For each participant 16 separate RMS values were recorded (one for each acceleration axis and for all eight balance conditions). The mean of those 16 RMS values for each participant was included in the correlation analysis. Spearman’s rank correlation coefficient was used to investigate the relationship between RMS, BBS, and the ABC scale measures. Pearson’s product-moment correlation was used to determine the relationship between RMS and TUG. For all analyses significance was set a priori at alpha = 0.05.
Results

**Discriminative Validity**

Perhaps the most important step in the validation of a fall risk screening tool is testing its ability to accurately discriminate fallers from non-fallers. Table 5 presents the AUC for the six logistic regression models evaluated here. Only the ABA alone or in combination with the clinical tests was able to discriminate participants with one or more falls from those with no history of falls. Inclusion of the clinical tests only improved the predictive ability of the model slightly (AUC = 0.805 for TUG, BBS, ABC, and ABA vs. AUC = 0.795 for ABA alone). No clinical test alone or in combination was able to discriminate between these groups.

For the comparison of multiple fallers vs. those with a history of one or fewer falls, no clinical test by itself was able to significantly predict fall status. While the combination of the TUG, BBS, and ABC scale was able to discriminate between groups, the addition of the ABA to the regression model improved the AUC by nearly 0.2 (AUC = 0.675 for TUG, BBS, and ABC vs. AUC = 0.871 for TUG, BBS, ABC, and ABA).

When comparing participants with multiple falls to those with no history of falls, all models were able to significantly predict fall status (AUC range = 0.716 to 0.897). However, as with the aforementioned group comparisons, the two models including the ABA exhibited the greatest AUC (≥ 0.841).
Table 6 shows several different quality measures commonly used to evaluate screening tools. While each is distinctly different, they are all calculated from the same 2 X 2 contingency table and therefore are all related. Briefly, sensitivity is the proportion of fallers who tested positive; specificity is the proportion of non-fallers who tested negative; PPV is the proportion of correctly identified fallers to all positive tests; NPV is the proportion of correctly identified non-fallers to all negative tests, PLR is the ratio of

<table>
<thead>
<tr>
<th>≥ One Fall vs. No Falls</th>
<th>Logistic Regression Model</th>
<th>AUC</th>
<th>Std. Error</th>
<th>P-value</th>
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</thead>
<tbody>
<tr>
<td>TUG, BBS, ABC, and ABA</td>
<td>0.805</td>
<td>0.050</td>
<td>0.000</td>
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</tr>
<tr>
<td>ABA</td>
<td>0.795</td>
<td>0.051</td>
<td>0.000</td>
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<tr>
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<td>0.109</td>
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<tr>
<td>BBS</td>
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<td>ABC</td>
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<table>
<thead>
<tr>
<th>≥ Two Falls vs. ≤ One Fall</th>
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<th>AUC</th>
<th>Std. Error</th>
<th>P-value</th>
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<tr>
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<tr>
<td>ABC</td>
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<tr>
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<td>0.069</td>
<td>0.408</td>
<td></td>
</tr>
<tr>
<td>TUG</td>
<td>0.449</td>
<td>0.072</td>
<td>0.489</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>≥ Two Falls vs. No Falls</th>
<th>Logistic Regression Model</th>
<th>AUC</th>
<th>Std. Error</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>TUG, BBS, ABC, and ABA</td>
<td>0.897</td>
<td>0.051</td>
<td>0.000</td>
<td></td>
</tr>
<tr>
<td>ABA</td>
<td>0.841</td>
<td>0.058</td>
<td>0.000</td>
<td></td>
</tr>
<tr>
<td>TUG, BBS, and ABC</td>
<td>0.750</td>
<td>0.067</td>
<td>0.001</td>
<td></td>
</tr>
<tr>
<td>TUG</td>
<td>0.740</td>
<td>0.068</td>
<td>0.001</td>
<td></td>
</tr>
<tr>
<td>BBS</td>
<td>0.731</td>
<td>0.069</td>
<td>0.002</td>
<td></td>
</tr>
<tr>
<td>ABC</td>
<td>0.716</td>
<td>0.065</td>
<td>0.004</td>
<td></td>
</tr>
</tbody>
</table>

ABA = regression model includes the RMS for each of the eight balance conditions.

Bold p-values are significant at a level of 0.05.
the proportion of fallers who tested positive to the proportion non-fallers who tested positive; and NLR is the ratio of the proportion of fallers who tested negative to the proportion of non-fallers who tested negative. For every measure but NLR, larger values represent a better screening tool. For NLR, the closer the value is to zero the better the screening tool.

The clinical value of each of these measures depends on the risk benefit ratio pertaining to the treatment. In the extreme case of treating cancer with chemotherapy, a therapy with many dangerous side effects, it is important to limit the number of false positives (e.g. higher PLR). In the case of fall risk, however, it may be more important to capture more fallers at the expense of treating individuals who may not need it (e.g. higher sensitivity).
In general, most of the models were better at correctly identifying non-fallers than fallers, indicated by substantially larger specificity than sensitivity values. Only for the comparison of multiple fallers to non-fallers did a model, the combination of the TUG, BBS, and ABC scale with the ABA, exhibit both good sensitivity (85.0) and specificity (82.5). No clinical test on its own or in combination exhibited good sensitivity (all ≤ 61.1). The combination of the TUG, BBS, and ABC scale resulted in excellent
specificity (91.2 to 93.0), but at the sacrifice of very poor sensitivity (5.0 to 22.2). For all three group comparisons, the ABA alone exhibited the best PPV (71.4 to 72.7), PLR (4.16 to 9.73), and NLR (0.63 to 0.18). For the comparisons of one or more falls to no falls and multiple falls to one or fewer falls, compared with the ABA alone, the addition of the clinical tests to the model did not improve the discriminative ability based on any measure evaluated here. For the comparison of multiple falls to no falls, the addition of the clinical tests to the ABA improved the sensitivity (58.0 with vs. 50.0 without), NPV (94.0 with vs. 84.1 without), and NLR (0.18 with vs. 0.54 without), but reduced the specificity, PPV, and PLR.

**Concurrent Validity**

We investigated the association between the ABA, TUG, BBS, and ABC scale to establish concurrent validity. Specifically, we were interested to know if the average RMS of the horizontal accelerations for the ABA 8 balance conditions were significantly correlated with scores for the clinical tests. Surprisingly, the associations were not only insignificant but extremely weak with correlation coefficients of 0.11, -0.04, and -0.02 for the TUG, BBS, and ABC scale, respectively (Table 7). As expected, the TUG, BBS, and ABC scale were all significantly correlated with each other and in the proper direction.
Discussion

The purpose of the present study was to assess the discriminative validity and concurrent validity of the ABA as a fall risk screening tool. Based on a variety of analyses, the ABA alone was better than the TUG, BBS, and/or ABC at discriminating fallers from non-fallers, and this was true for all three group comparisons. This result is somewhat in agreement with O’Sullivan et al. [27], who found that pelvic accelerations resulting from the task of standing on a mat with eyes open were significantly greater for fallers than non-fallers, while the BBS and TUG were not significantly different between groups. Cho and Kamen [25], on the other hand, found that head but not pelvic accelerations were greater in fallers. One potential reason for the discrepancy is that Cho and Kamen (1998) used a single axis accelerometer rather than the tri-axial accelerometers used in the present study and by O’Sullivan et al. (2009). Because of that,

![](image.png)

**Table 7. Correlation matrix for the association between clinical tests and the accelerometer-based balance assessment**

<table>
<thead>
<tr>
<th></th>
<th>Average RMS</th>
<th>TUG</th>
<th>BBS</th>
<th>ABC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average RMS</td>
<td>1.00</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TUG</td>
<td>0.11</td>
<td>1.00</td>
<td></td>
<td></td>
</tr>
<tr>
<td>BBS</td>
<td>-0.04</td>
<td>-0.78</td>
<td>1.00</td>
<td></td>
</tr>
<tr>
<td>ABC</td>
<td>-0.02</td>
<td>-0.51</td>
<td>0.48</td>
<td>1.00</td>
</tr>
</tbody>
</table>

*Values represent Pearson's or Spearman's (where appropriate) correlation coefficients. Bold values represent significance at a level of 0.05.*
they were unable to correct for pelvic tilt (the linear acceleration may not have been horizontal), which might call into question the validity of their data.

The sensitivity and specificity of the BBS (38.9 to 40.0 and 74.0 to 77.2, respectively) and the ABC scale (60.0 to 61.1 and 60.3 to 66.7) were much lower than those reported by Lajoie and Gallagher [35] (sensitivity and specificity for both the BBS and the ABC scale were > 80.0). There were, however, important differences that should be noted. First, they defined a faller as someone with ≥1 fall in the past year, whereas we only inquired about the past six months. Second, and potentially most important, their study population had a much younger mean age (approximately 74±3 vs. 86±6 years).

Though some studies have reported that the BBS was capable of retrospectively discriminating fallers from non-fallers [35, 36], others have found the instrument insensitive to subjects with a history of falls [37, 38]. In general, the specificity of the BBS far outweighs the sensitivity, which may make it a useful tool for ruling out high risk but not necessarily at properly categorizing individuals as high risk [39].

Prospective studies, too, have delivered mixed results. Brauer et al. [40] reported that the BBS was unable to prospectively predict falls. Muir et al. [41], on the other hand, found that the predictive ability of the BBS was dependent on the definition of a fall and how the BBS scores were categorized. For example, the BBS only moderately predicted multiple falls and was unable to significantly predict single and/or injurious falls. In agreement with our results, Nuels et al. [38] found that the BBS on its own was
unable to predict falls in older adults, and concluded that it should only be used as a supplement to screen for fall risk.

Our finding that the TUG consistently delivered poor sensitivity and specificity was not unexpected. Several studies argue against the use of TUG for fall prediction. Herman et al. [37] and Weiss et al. [43] reported the inability of the TUG to retrospectively discriminate fallers from non-fallers. The prospective fall studies of Hausdorff et al. [32] and Nordin et al. [44] reported no significant predictive ability of the TUG. Similar to the BBS, due to consistently high specificity but variable sensitivity, the TUG may be best suited to rule out rather than rule in high risk [44].

Though the TUG, BBS, and ABC scale were all significantly correlated with each other, our second hypothesis of a significant correlation between the ABA and the TUG, BBS, and ABC scale was not supported (all correlations were $\leq 0.11$). In a similar comparison between a postural sway assessment via force plate and the BBS, only the standard deviation of the vertical ground reaction force was significantly correlated with the total score on the BBS, and only the sagittal mean velocity of the COP was moderately correlated with the static portion of the BBS [119]. It was suggested that the BBS and various force plate measures quantify different aspects of postural control.

The correlation between the TUG and BBS observed in this study ($r = -0.78$) was very similar in magnitude to that reported by O’Sullivan et al. (2009) ($r = -0.77$) and by Steffen et al. [39] ($r = -0.76$). The correlation between the TUG and the ABC scale ($r = -0.51$) was also similar to that reported by Cho et al. ($r = -0.606$) [53]. To our knowledge, we are the first to report a correlation between the BBS and the ABC scale.
Study Limitations

There were several limitations. First, the prediction models evaluated based on the self-reported fall history of the participants may not equally predict future falls. Second, the best predictive models included the ABA along with all clinical tests. The time required to complete the four tests may be impractical in a clinical setting. Finally, while the process of the acceleration data collection is well within the ability of an inexperienced and untrained individual, the level of filtering and data processing required to achieve a usable quantity limits its clinical usefulness.

Conclusion

The ABA alone was better than the TUG, BBS, and/or ABC at discriminating fallers from non-fallers. Furthermore, the addition of the clinical tests to the ABA provided little improvement in most cases. While it is tempting to conclude that the ABA should replace the TUG, BBS, and the ABC scale, the importance of these tests should not be ignored. When the only outcome of interest is the assessment of fall risk, the ABA alone may be sufficient. However, clinicians are not simply interested in the degree of risk, but also the identification of specific impairment as it will aid them in the development of a treatment strategy. We therefore, suggest that the ABA be used in combination with other tests. Importantly, this study was based on a self-reported fall history and whether or not these results translate to the prediction of future falls is an area of ongoing research in our lab. We are also working to make the data analysis process
more user friendly such that clinicians with no technical expertise may use this accelerometer and interpret the results of the ABA.
Chapter 5: Conclusion

The accelerometer exhibited good to excellent trial-to-trial and block-to-block reliability. Additionally, the accelerometer was able to discriminate between visual and somatosensory conditions and acceleration axes.

The ABA alone was better than the TUG, BBS, and/or ABC at discriminating fallers from non-fallers. Furthermore, the addition of the clinical tests to the ABA provided little improvement in most cases. While it is tempting to conclude that the ABA should replace the TUG, BBS, and the ABC scale, the importance of these tests should not be ignored. When the only outcome of interest is the assessment of fall risk, the ABA alone may be sufficient. However, clinicians are not simply interested in the degree of risk, but also the identification of specific impairment as it will aid them in the development of a treatment strategy. We therefore, suggest that the ABA be used in combination with other tests. Importantly, this study was based on a self-reported fall history and whether or not these results translate to the prediction of future falls is an area of ongoing research in our lab. We are also working to make the data analysis process more user friendly such that clinicians with no technical expertise may use this accelerometer and interpret the results of the ABA.
References


