EVALUATION OF AN INERTIAL SENSOR TO QUANTIFY POSTURAL STABILITY ASSESSMENTS IN YOUNG HEALTHY INDIVIDUALS AND INDIVIDUALS WITH CHRONIC ANKLE INSTABILITY

by

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Measures of postural stability are utilized in clinical and research settings and are important for prevention and rehabilitation of musculoskeletal injuries. Force-plates are often used to quantify postural stability in research settings, however due to cost and size are not readily available in clinical settings. Clinical tests of postural stability require minimal equipment and are easily implemented, but are restricted due to reliability and sensitivity. Low-cost inertial sensors may be an effective alternative to force-plates for objective postural stability assessment. However, there is limited research determining which measures and assessments are most reliable, valid and discriminatory in populations with postural stability deficits related to musculoskeletal injury. For sensor-based postural stability assessments to be implemented in clinical settings, they must be reliable, valid, and discriminatory in desired target populations. The purposes of this dissertation were to (i) establish the reliability of accelerometry measures of postural stability, (ii) establish the concurrent validity of accelerometry measures compared to force plate measures of postural stability and their ability to detect differences in task difficulty, and (iii) determine the ability of accelerometry measures to discriminate postural stability deficits in individuals with chronic ankle instability (CAI). A total of 50 young, active individuals (25 control, 25 CAI) were recruited to address the study aims. Ten accelerometry measures were extracted from a waist-worn sensor during each of ten postural stability tasks of varying difficulty (eight static, two dynamic). Force-plate data were collected concurrently.

Several accelerometry measures of static and dynamic postural stability were found to be reliable within session and across three sessions in control and CAI groups. Within subject variability improved when at least three static or six dynamic trials were averaged. Static postural stability accelerometry measures showed weak ($r<0.5$) to strong ($r\leq0.75$) associations with force-plate measures, while dynamic postural stability associations ranged from weak to moderate ($0.5\leq r<0.75$). Accelerometry measures were sensitive to task difficulty and postural stability deficits in individuals with CAI. Overall, a subset of the accelerometer instrumented assessments provided reliable and valid, objective measures of postural stability. Integration with a mobile device will provide clinicians a low-cost, objective solution for postural stability assessment.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

PREFACE

Many incredible individuals have contributed to this work and have supported me throughout my graduate school career. When I started the Bioengineering PhD program, I knew that it would be demanding, but it was challenging in ways that I never could have imagined. The obstacles and opportunities presented during my time at the University of Pittsburgh have shaped me into the person I am today. For that I am grateful.

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1.0 INTRODUCTION

Measures of postural stability are important for prevention and rehabilitation of musculoskeletal injuries and for athletic performance optimization. Clinicians need objective measures of postural stability that are reliable, valid, and easy to implement. Force plate measures are the gold-standard for postural stability assessments, however this technology is expensive and not readily available in clinical settings. Therefore, clinicians rely on indirect, subjective assessments that are quick and easy to implement. More direct, objective measures may better identify patients with postural stability deficits and enhance clinical decision-making, but need to remain quick and easy to implement. Deficits in postural stability have been shown to occur following concussion[1–3] and lower extremity injuries,[4–6] and are associated with low back pain.[7] Postural stability deficits have also been shown to be predictive of ankle injury.[8–11] More than 80% of physical therapists assess static and dynamic postural stability regularly.[12] However, current clinical tests for postural stability are limited due to operator-dependency, i.e., measurement variability between testers, and have been shown to have insufficient sensitivity to mild balance impairments.[13–15]

Wearable inertial sensors provide a low-cost alternative to the traditional force plate measures of postural sway.[16,17] Accelerometry assessment is effective in differentiating among postural stability task difficulty, and a variety of time and frequency domain analyses have been utilized to quantify postural sway.[18,19] However, there is limited research determining which

accelerometry measures are most reliable, valid and discriminatory in populations with postural stability deficits related to musculoskeletal injury. A reliable, valid measure for detecting and monitoring these deficits throughout rehabilitation will enable a more complete recovery and reduced risk of reinjury.

1.1 BACKGROUND

Ankle sprains are the most common lower extremity musculoskeletal injury in healthy, physically active individuals, of which 80-90% are classified as a lateral ankle sprain (LAS).[20–22] Acute LAS is defined as acute traumatic injury to the lateral ligaments of the ankle as a result of high velocity inversion and internal rotation of the ankle/foot complex.[23,24]

1.1.1 Lateral ankle sprain and chronic ankle instability

Ankle injuries are among the most common musculoskeletal injuries, impacting athletes, military personnel, and the general population. Data from the National Collegiate Athletic Association (NCAA) Injury Surveillance System (ISS) indicate ankle sprains account for 15% of all reported sport-related injuries, with an overall incidence rate of 0.83 sprains per 1000 athletic exposures (AE).[24,25] Soccer and basketball athletes have the greatest incidence rates ranging from 1.15 – 1.3 ankle sprains per 1000 AE.[25] Military personnel also have an increased risk of ankle sprain, with incidence rates ranging from 34.95 - 45.14 sprains per 1000 person-years.[26,27] Gribble et al. suggest the incidence rate in military personnel translates to an estimated 0.35 - 0.45 ankle sprains per 1000 exposures.[24] Though incidence rates are lower in the general population (2.15

sprains per 1000 person-years) compared to athletes and military personnel, ankle sprains do have a significant impact and financial burden in the general population.[20]

In the United States, mean societal costs related to a joint sprain and ankle injury were reported to be $9,196 and $11,925, respectively.[28] These monetary values represent both short term costs directly related to healthcare consumed, and indirect or long term costs related to the impact of the sprain on productivity and quality of life. It is important to note, financial burden of LASs is often underestimated as less than 50% of individuals with LAS seek formal care.[29] As such, most of the burden associate with LAS is related to lost productivity associated with lost work days, lost playing time, as well as lost unpaid leisure time. An estimated 30-75% of individuals suffering from lateral ankle sprain report long term chronic impairment.[30,31] Individuals that suffer from chronic impairment comprise 70-85% of individuals that develop post-traumatic osteoarthritis (PTOA) and are much more likely to seek surgical intervention.[32,33]

The initial inflammatory phase of an acute LAS resolves in a relatively short period of time (about ten days after trauma). However, the remodeling or maturation phase can last up to one year after trauma.[34] After inflammatory symptoms subside and individuals return to activity, many report lingering symptoms of pain and decreased function.[35] Evidence suggests that during the one year period following acute LAS, athletes are twice as likely to experience a recurrent sprain.[36] This increased risk of injury may be due to associated sensorimotor,[24,37] postural stability,[6,38–40] and functional movement deficits.[41,42] It is thought that further damage of the already impaired ankle is a significant contributor to chronic ankle instability (CAI).[24]

Patients with CAI often experience ongoing pain, ankle instability, and feeling of the ankle joint giving way.[43] Hiller at al. proposed a model of CAI that identifies three contributing factors: mechanical instability, perceived instability, and recurrent sprain (Figure 1).[44]

Mechanical instability is a result of joint laxity experienced after ankle-ligament injury,[9] whereas perceived or functional instability is related to sensation of joint instability likely due to proprioceptive and neuromuscular deficits.[37] Individuals can have functional instability without showing signs of mechanical laxity or mechanical instability.[44] Delahunt et al. suggests both mechanical and perceived instability must persist for a minimum of one year post initial sprain for an individual to be classified as having CAI.[23] Development of CAI has severe consequences on an individual's quality of life and contributes to a faster progression towards PTOA. Residual pain and instability may lead to prolonged decrease in physical activity which has long term health implications.[31]

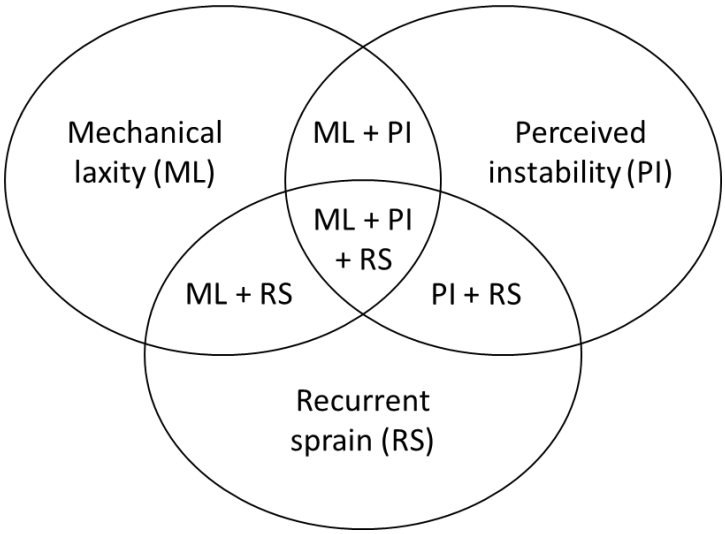

Figure 1. Chronic ankle instability subgroups. Figure adapted from Hiller et al.[44]

1.1.2 Postural stability

Postural stability is defined as the ability to maintain the body's center of mass (COM), or maintain equilibrium within the limits of stability, over the base of support.[45] Postural stability is a dynamic process that requires coordination of three sensory systems (visual, vestibular, and somatosensory) to detect motion, integrate sensorimotor information, and react to maintain the body's equilibrium over the base of support.[46] These multifactorial and complex interactions make assessment of postural stability difficult. There are both static and dynamic measures of postural stability utilized both in research and clinically. Static postural stability is defined as maintaining steadiness, or keeping the body as motionless as possible, on a fixed or unmoving base of support.[47] Dynamic postural stability has been defined as the ability to transfer the projection of the center of gravity around the supporting base during a change in position or location (single-leg jump or landing).[47,48] Postural stability impairments are prevalent following concussion[1] and various lower extremity injures,[4–6,49] and have been identified as a predictor of injury.[8–11]

Several clinical assessments have been developed to monitor postural stability in clinical settings. An estimated 80% of physical therapists assess postural stability regularly in the clinic.[12] These assessments often require minimal equipment and are evaluated by observation or measurement of an indirect parameter. The most commonly used assessment in orthopedic settings is the single-leg stance test, followed by the Berg Balance Scale and the Timed Up and Go test.[12] For athletic populations, the most common clinical assessments of postural stability include a timed single-leg stance,[50] the Balance Error Scoring System (BESS),[13,51] and the Star Excursion Balance Test (SEBT).[52,53] The timed single-leg stance is scored based on the length of time an individual can stand on one leg before the non-weight-bearing limb touches either the

weight-bearing limb or the floor.[50] The BESS test consists of three stance positions (feet together, single-leg, and tandem) on two surfaces (firm and foam) with eyes closed for 20 seconds each. The test is scored by an evaluator based on errors. The SEBT requires the individual to maintain a single-leg stance position while reaching with the contralateral leg to touch as far as possible in eight directions spaced in 45˚ increments. The SEBT has been proposed as a dynamic postural stability assessment, however the stable base of support does not simulate athletic tasks and may not be challenging enough to discriminate between healthy and injured populations or be predictive of future injury. Simpler tasks, like the single-leg stance, have sufficient interrater reliability, but have a ceiling effect.[54,55] Clinical tasks that are more challenging better differentiate postural stability deficits, but lack reliability.[13] Many of these assessments have been shown to differentiate injured populations when large deficits are present, but lack the fidelity to identify minor balance deficits that may lead to musculoskeletal injury.[56,57] Although clinicians will often assess postural stability, the results are underutilized in decision-making as clinicians often doubt the information gained.[58,59] Objective, standardized assessments that are reliable, valid, and easy to implement will give clinicians the information they need to differentiate balance deficits, ultimately optimizing injury prevention, rehabilitation, and performance training strategies.

Individuals with CAI have been shown to have postural stability deficits.[5,60–63] While these deficits can be difficult to detect with indirect and subjective clinical measures,[64] they are important in identifying risk of reinjury.[35] Postural stability deficits can occur in both the involved and uninvolved limbs following ankle sprain, which may be due to central changes that occur following injruy.[6,65] Thus, comparison to a matched control group rather than the uninvolved limb may provide a better basis for evaluation. Individuals with CAI also display

varying strategies of maintaining stability compared to healthy controls during static and dynamic assessment.[66–69] Individuals that have mechanical laxity and perceived instability exhibit worse postural control compared to those that report perceived instability but are mechanically stable.[61,70] Balance training has shown to be effective in improving proprioception and postural stability in individuals with CAI,[71] but it is important to identify neuromuscular deficits so that the proper rehabilitation can be administered.

Wearable inertial sensors provide a low-cost alternative to the traditional force plate measures of postural sway.[16,17] Researchers have started to explore ways to objectively assess postural stability using inertial sensors.[18,19] Some have shown that inertial sensor methods are sensitive to detecting neurological impairments,[72,73] vestibular disorders,[74] and concussion.[2] However, minimal research has been done to show if the inertial sensor based methods are able to detect postural stability deficits in populations that have suffered a musculoskeletal injury.

1.2 PROBLEM STATEMENT

Postural stability is an important factor in the rehabilitation of musculoskeletal injury, particularly in individuals that have developed CAI. Although advances in wearable sensors will be advantageous in objective assessment of postural stability in clinical practice, several limitations still remain. There is no consensus on which accelerometry-based postural stability measures should be utilized in clinical practice, particularly for individuals with a previous musculoskeletal injury, and it is often unknown how to translate these measures into clinically relevant and actionable data. For the data to be useful in a clinical setting, it must be reliable, valid, and able to identify balance deficits in the populations that will be tested.

1.3 STATEMENT OF PURPOSE

The purposes of this dissertation were to (i) establish the reliability of accelerometry measures of postural stability, (ii) establish the concurrent validity of accelerometry measures compared to force plate measures of postural stability and their ability to detect differences in task difficulty, and (iii) to determine the ability of accelerometry measures to discriminate postural stability deficits in individuals with CAI.

1.4 SPECIFIC AIMS AND HYPOTHESE

1.4.1 Specific aim 1

To establish the systematic bias, within subject variability, and test-retest reliability of static and dynamic postural stability assessed by accelerometry-based measures in healthy individuals and individuals with chronic ankle instability.

1.4.2 Specific aim 2

To establish the concurrent validity of accelerometry-based measures of static and dynamic postural stability postural stability compared to force plate derived measures across ten postural tasks of varying difficulty.

Hypothesis 2.1: The accelerometry-based measures of postural stability will be significantly correlated with the force plate measures with r-coefficients ranging from 0.7-0.9.

Hypothesis 2.2: The accelerometry-based measures will be able to differentiate among task difficulty.

1.4.3 Specific aim 3

To determine the discriminative validity of accelerometry-based measures of postural stability to differentiate healthy individuals from individuals with CAI during single-leg postural stability tasks of varying difficulty.

Hypothesis 3.1: Individuals with CAI will demonstrate diminished postural stability compared to healthy controls characterized by accelerometry-based measures of postural stability.

1.5 SIGNIFICANCE

The proposed study provides a foundation for implementing a low-cost inertial sensor in clinical practice for assessment of postural stability, particularly for identifying postural stability deficits in individuals with CAI. The results of this study will demonstrate the feasibility of using the sensor to track or assess progress in postural stability throughout rehabilitation. This is one of the first studies to look at the reliability and validity of an inertial sensor for assessment of postural stability in a group with a previous musculoskeletal injury.

1.6 ORGANIZATION OF DISSERTATION

The following chapters are arranged by Specific Aims: Chapter 2 explains the innovative aspects of this work and methodological considerations; Chapter 3 addresses Specific Aim 1: measures of reliability; Chapter 4 addresses Specific Aim 2: concurrent and discriminative validity; Chapter 5 addresses Specific Aim 3: discriminative validity in individuals with chronic ankle instability. Conclusions and future work are expressed in Chapter 6.

2.0 INNOVATION AND DEVELOPMENT

The development of portable, easy to administer, low-cost and objective postural stability assessments will improve detection and monitoring of postural stability deficits and enable clinicians to collect and analyze big data sets relative to postural stability in clinical practice. Ultimately, these tools will enable clinicians to improve their clinical decision-making and become more effective treating patients. Some researchers have begun to explore the use of accelerometry measures to detect postural stability deficits associated with neurological disorders, such as Parkinson's disease,[72,73,75] vestibular disorders,[74] and most recently concussion.[2] However, postural stability deficits associated with musculoskeletal injury likely impact sensory organization differently than these neurological disorders and injuries.[76–78] Thus, it is imperative to validate the use of inertial sensors in populations that may have postural stability deficits associated with musculoskeletal injury. With the high incidence of recurrent lateral ankle sprain (LAS) and neuromuscular impairments associated with chronic ankle instability (CAI), the purpose of this dissertation is to determine accelerometry-based measures of postural stability that are most reliable, valid compared to the gold-standard measures, and can differentiate between individuals with CAI and those without.

Dynamic assessments of postural stability may be more appropriate for active populations compared to static assessments,[79] and may be particularly useful in the clinic for active individuals that are nearing the end of their rehabilitation following musculoskeletal

injury. Dynamic postural stability tasks, quantified with force plate measures, are often used in sports medicine research and have been shown to effectively identify postural stability deficits in individuals with CAI. However, there is not a low-cost, reliable and sensitive method for assessing dynamic postural stability during jump-landing tasks in the clinic. Few studies have quantified dynamic postural stability during a jump-landing task using a wearable inertial sensor.[19] This dissertation will provide the basis of evidence needed to bring objective measures of static and dynamic postural stability from the laboratory to a clinical setting. Integration of the sensor with smart phone technology will provide clinicians a low-cost, objective solution for postural stability assessment.

2.1 SENSOR SELECTION

Inertial measurement units (IMUs) are low-powered microelectromechanical systems that use 3-dimensional accelerometers and 3-dimensional gyroscopes to measure linear acceleration and angular velocity. Often, a 3-dimensional magnetometer is included in the IMU to reduce sensor drift by continuously correcting the orientation of the sensor.[80] Many commercially available IMUs come with onboard processing utilizing the magnetometer to correct for errors such as sensor drift and also may employ onboard filtering and such as a Kalman filter. Several variations of commercially available accelerometers and IMUs exist. The overall goal of this dissertation was to identify reliable, valid, and discriminatory objective measures of static and dynamic postural stability that can be easily implemented in a clinical setting. This was taken into consideration when identifying the following criteria for sensor selection: accelerometer sensitivity and range, sampling frequency, cost, and device communication.

The accelerometer had to be triaxial and have the ability to capture stability characteristics during both static and dynamic tasks. It was important that the accelerometer be sensitive enough to pick up subtle differences in postural sway during the static postural stability tasks and have a broad enough range to capture peak accelerations during the jump-landing tasks. Sensitivity on the order of 0.001 *g* is required to differentiate between eyes open and eyes closed static stance conditions.[81] On the other extreme, tibia accelerations during jump-landing tasks have been shown to range from 3.5 – 6.5 *g* in the during a double-leg landing when jumping from 40% of one's height, which is similar to the dynamic postural stability task utilized in this dissertation.[82] Therefore, an accelerometer range of at least ±12 *g* with a sensitivity on the order of 0.001 *g*/digit was desired.

Peak frequency during gait in healthy, older adults has been reported to range from 1.56 – 1.81 Hz with a bandwidth of 6.26 – 7.89 Hz.[83] A sampling frequency of 100 - 500 Hz has been determined valid and reliable in assessing jumping performance using an accelerometer compared to force plate measures.[84,85] A sampling frequency of at least 250 Hz was desired to ensure capture of high frequency components of the signal. It was also important to consider cost and device communication for future conversion to a mobile application implementation in clinical settings. For easy integration into a mobile application, Bluetooth communication was preferred.

For the work presented in this dissertation, accelerations were collected at L5, near the center of mass (COM), using a YEI 3-Space Sensor Bluetooth (35 mm x 60 mm x 15 mm, 28 g; YOST Labs, Portsmouth, OH). Though this dissertation refers to data collected with the sensor placed over L5 as COM accelerations, it should be noted that the data collected is a surrogate measure of true COM acceleration. The sensor is comprised of triaxial gyroscope, accelerometer

and magnetometer. This specific sensor was selected because it is low cost ($320) and has a selectable accelerometer range (±6 *g*, ±12 *g*, ±24 *g*). The accelerometer has a 12-bit resolution and 0.003, 0.006, and 0.012 *g*/digit resolution, respective to the selected accelerometer range. The onboard processing scales, normalizes and compensates for drift error of the raw accelerometer, gyroscope, and magnetometer data and subsequently applies a Kalman filter. Caution should be taken when utilizing onboard processing methods. As such, the on board processing methods can be by-passed, recording only the raw sensor data. The Bluetooth option was favorable for future integration of the assessments outlined in the dissertation into a mobile application, and the sampling frequency with Bluetooth connection reaches up to 500 Hz. The sensor was factory calibrated for sensitivity and Zero-*g* level and are reset to these values when device is turned on.

2.2 METHODOLOGICAL CONSIDERATIONS

A pilot study was performed to identify optimal data processing techniques for the center of mass acceleration data collected during ten postural stability tasks of varying difficulty.

2.2.1 Postural stability assessments

As a pilot study, one healthy control participant completed ten postural stability tasks of varying difficulty. Center of mass accelerations were collected during each task with an IMU (YEI 3 – Space Sensor Bluetooth) positioned over L5 and secured with a belt. An additional neoprene belt was secured around the waist over the sensor to minimize artifact due to sensor motion during

the tasks (Figure 2). Due to low and inconsistent sampling frequencies observed with Bluetooth connection, the sensor was connected via 7.6 m cable to a computer for data logging, and only raw accelerations were logged. By-passing the onboard Kalman filter allowed for the data to be sampled at a higher frequency. Prior to the postural stability assessments, accelerometer data were collected during a five-second static capture where the participant was asked to stand upright with their back against a wall to minimize body sway.

Following the five-second static capture, the participant completed ten postural stability tasks of varying difficulty, eight static and two dynamic tasks (Figure 3). The static postural stability tasks included: double-leg static stance (DL) on a firm surface, double-leg stance on a

Figure 2. Sensor positioning and local coordinate system for the IMU. x-axis is red, y-axis is green, and z-axis is blue.

foam surface (Airex Pad, Airex Corp., Somersworth, NH) (DL-F), tandem stance (TAN), and a single-leg stance (SL) (Table 1). The participant was asked to hold the double-leg and tandem stance positions for 20 seconds based on the Balance Error Scoring System (BESS), a clinical test of postural stability often utilized in sports medicine research.[13] The participant was asked to hold the single-leg stance positions for ten seconds, which is common in force plate analyses of a single-leg stance task.[79] Each static task was performed with eyes open (EO) and eyes closed (EC) while barefoot and the participant was asked to complete five successful trials of each task. For the DL, DL-F, and TAN tasks, trials were marked unsuccessful if the participant removed their hands from their waist for greater than three seconds, if they stepped out of the stance position, or if they opened their eyes during the eyes closed trials. During the SL tasks, the participant was permitted to touch down on the ground with the non-test limb to maintain balance, but was instructed to promptly go back to the single-leg position. For the SL tasks, trials were marked unsuccessful if the participant's non-test leg touched the test leg or the ground outside of a 60 cm x 40 cm area, if the participant removed their hands from their waist for greater than three seconds, or if they opened their eyes during the eyes closed trials. These methods have been shown to be reliable in force plate and accelerometry analysis.[48,79,86] The reliability is described in detail in Section 3.1.

Two jump-landing tasks were performed: (i) forward jump where the participant initiated a jump from two feet at a distance equal to 40% of their height, cleared a 30.5 cm hurdle and landed on a single leg and (ii) lateral jump where the participant initiated a jump from two feet at a distance equal to 33% of their height, cleared a 15.2 cm and landed on a single leg (Table 1). The hurdle for each task was placed half way between the participant's takeoff and target landing positions. During both jump-landing tasks, the participant was asked to recover their balance on

the single leg and hold the single-leg position for five seconds. The participant was asked to complete twelve successful trials of the dynamic tasks in their own athletic shoes. Trials were marked unsuccessful if the participant took a hop or shifted their foot position after landing during the five-second stabilization period and/or if the participant touched down with the non-test limb. If the sensor shifted during the jumping tasks, the sensor was repositioned and another five-second static capture was taken. The methods utilized for the dynamic postural stability tasks have shown good reliability during force plate (ICC = 0.86 – 0.92) and accelerometry assessment (ICC = 0.84 – 0.92).[19,79] The reliability is described in greater detail in Section 3.1.

Table 1. Static and dynamic postural stability task descriptions

Task Position/Maneuver	Surface	Eyes Open	Eyes Closed	Task Description	Duration (s)
Double leg	Firm	DLEO	DLEC	Feet placed hips width apart, hands on waist, eyes focused eye-level straight ahead	20
Double leg	Foam	DLEO-F	DLEC-F	Feet placed hips width apart, hands on waist, eyes focused eye-level straight ahead	20
Tandem	Firm	TANEO	TANEC	Dominant (control) or involved (CAI) limb in front with the heel of front foot touching the toes of the rear foot, hands on waist, eyes focused eye-level straight ahead	20
Single leg	Firm	SLEO	SLEC	Stance limb was the dominant (control) or involved (CAI) limb, non-stance limb positioned beside but not touching stance limb or ground, hands on waist, eyes focused eye-level straight ahead	10
Forward jump-landing	Firm	DPS-AP	--	Initiated a jump from two feet at a distance equal to 40% of participant's height, cleared a 30.5 cm hurdle, landed on a single leg, regained balance and held single leg position	5
Lateral jump-landing	Firm	DPS-ML	--	Initiated a jump from two feet at a distance equal to 33% of participant's height, cleared a 15.2 cm hurdle, landed on a single leg, regained balance and held single leg position	5

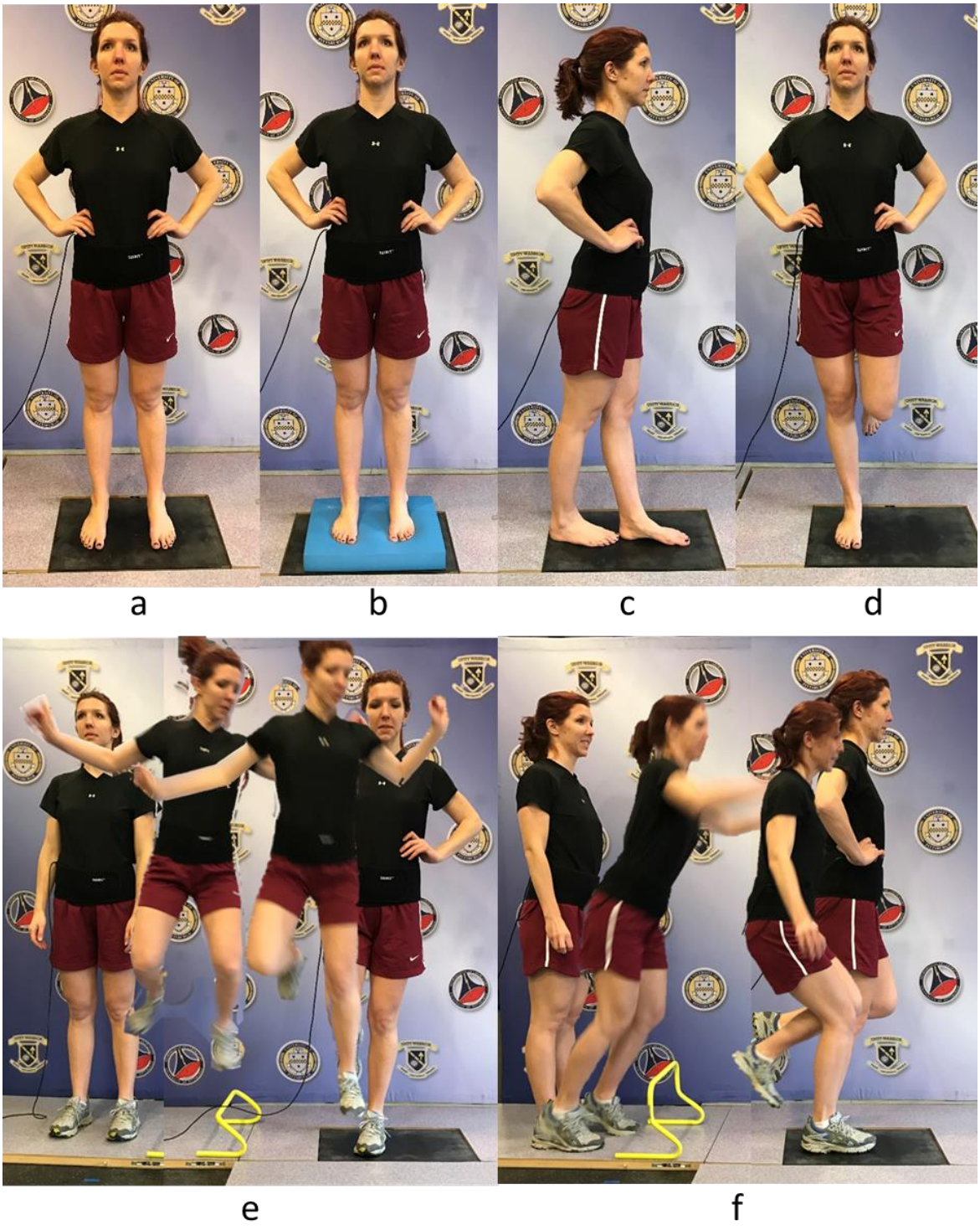

Figure 3. Static and dynamic postural stability tasks. (a) Double-leg stance completed with eyes open and eyes closed, (b) double-leg stance on foam completed with eyes open and eyes closed, (c) tandem stance completed with eyes open and eyes closed, (d) single-leg stance completed with eyes open and eyes closed, (e) lateral jump-landing and (f) forward jump-landing.

2.2.2 Filtering

Data from the pilot study were sampled at approximately 1200 Hz and down sampled to 1000 Hz using the *resample* function in Matlab (The MathWorks Inc., Natick, MA). Representative unfiltered data are shown in Figure 4. A power spectral density (PSD) analysis was performed on the anterior-posterior (AP) and medial-lateral (ML) acceleration time series across all ten postural stability tasks using the *pwelch* function in Matlab. The PSD plots were utilized to determine the optimal cutoff frequency for a low-pass Butterworth filter (Figure 5). A 20 Hz cutoff frequency was selected for the static tasks and a 50 Hz cutoff frequency was selected for the dynamic tasks. The transfer function coefficients for a second order low-pass digital Butterworth filters with cutoff frequencies normalized to 500 Hz were calculated in Matlab using the *butter* function. The coefficients were subsequently utilized in Matlab's *filtfilt* function to apply the low-pass Butterworth filters to the raw triaxial acceleration data collected during the static and dynamic tasks (Figure 6). The cutoff frequencies selected for this dissertation are similar to reported cutoff frequencies utilized for COM acceleration measures of static stance postural assessments in healthy populations (1.25 - 55 Hz).[16,18,87]

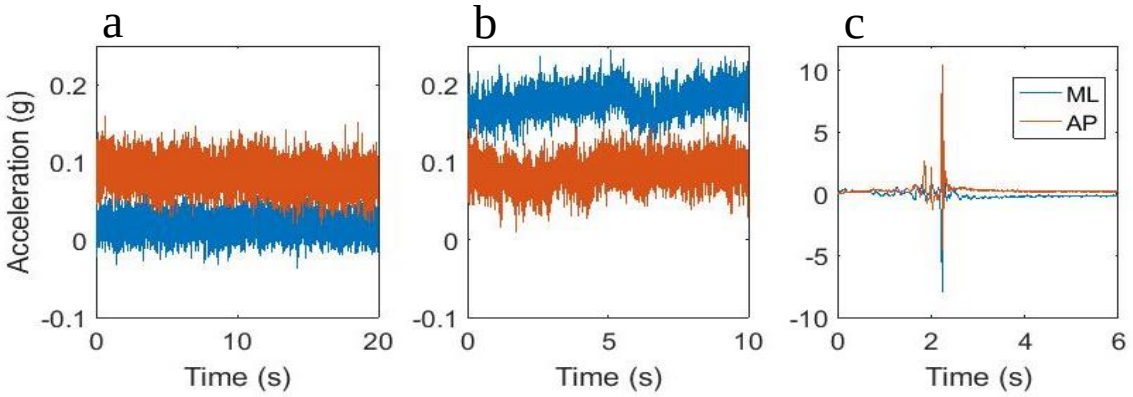

Figure 4. Unfiltered acceleration time series. Representative medial-lateral (blue) and anterior-posterior (red) acceleration time series data is shown for the (a) double-leg stance with eyes open, (b) single leg stance with eyes open and (c) forward jump landing.

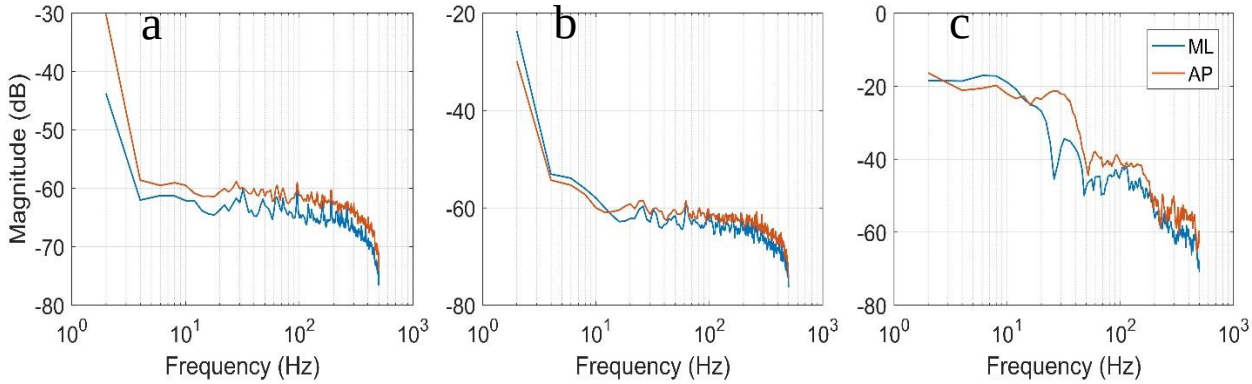

Figure 5. Power spectral density analysis of the unfiltered acceleration time series. Representative medial-lateral (blue) and anterior-posterior (red) power spectral density is shown for the (a) double-leg stance with eyes open, (b) single-leg stance with eyes open and (c) forward jump-landing.

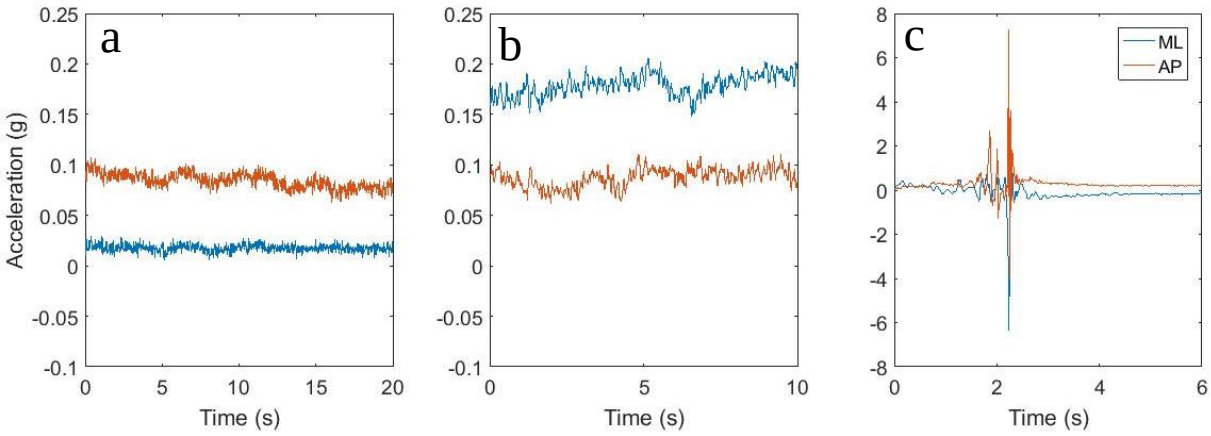

Figure 6. Filtered acceleration time series. Representative medial-lateral (blue) and anterior-posterior (red) acceleration time series data is shown for the (a) double-leg stance with eyes open low-pass filtered with a 20 Hz cutoff frequency, (b) single-leg stance with eyes open low-pass filtered with a 20 Hz cutoff frequency and (c) forward jump-landing low pass filtered with a 50 Hz cutoff frequency.

2.2.3 Quaternion rotation transformation

Correcting for accelerometer tilt may help discriminate between patient populations.[88] A quaternion rotation transformation described by Tundo et al. was used to adjust for arbitrary tilt of the sensor along the x, y, or z axes.[89] An initial gravity vector $\vec{V_i}$ was calculated by averaging the COM acceleration data over the five-second static capture to yield $\vec{V_i} = X\hat{\imath} + Y\hat{\jmath} + Z\hat{k}$ where X is linear acceleration in the medial-lateral direction, Y is linear acceleration in the anterior-

posterior direction, and Z is linear acceleration in the vertical direction. The desired gravity vector was $\overrightarrow{V_g} = X'\hat{\imath} + Y'\hat{\jmath} + Z'\hat{k}$ where $\overrightarrow{V_g} = (0,0,1)\ g$.

An axis vector $\vec{A}$ was calculated from the cross-product between the initial and desired gravity vectors:

$$\vec{A} = \overrightarrow{V_i} \times \overrightarrow{V_g} \tag{1}$$

Vector $\vec{A}$ described in equation (1) was then normalized by dividing by the magnitude of $\vec{A}$ to yield $\vec{A}_{norm}$.

$$\vec{A}_{norm} = \frac{\vec{A}}{\|\vec{A}\|} \tag{2}$$

The angle α between vectors was expressed as the cosine angle form the dot product of the initial and desired gravity vectors. Given X' and Y' = 0 and the magnitude of $\overrightarrow{V_g} = Z'$, the angle α is expressed as:

$$\alpha = \cos^{-1}\left(\frac{z}{\|\vec{V}_i\|}\right) \tag{3}$$

The axis-angle pair described in equations (2) and (3) were then used in the following quaternion rotation equations:

$$q_0 = \cos\left(\frac{\alpha}{2}\right),$$

$$q_1 = \sin\left(\frac{\alpha}{2}\right)\vec{A}_{norm,x},$$

$$q_2 = \sin\left(\frac{\alpha}{2}\right)\vec{A}_{norm,y},$$

$$q_3 = \sin\left(\frac{\alpha}{2}\right)\vec{A}_{norm,z},$$

$$R(q_0,q_1,q_2,q_3) = R = \begin{matrix} 1-2(q_2^2+q_3^2) & 2(q_1q_2+q_0q_3) & 2(q_0q_2+q_1q_3) \\ 2(q_1q_2+q_0q_3) & 1-2(q_1^2+q_3^2) & 2(q_2q_3+q_0q_1) \\ 2(q_1q_3+q_0q_2) & 2(q_0q_1+q_2q_3) & 1-2(q_1^2+q_2^2) \end{matrix} \tag{4}$$

The rotation matrix R, equation (4), was then applied to the filtered COM acceleration data $\vec{V}_i$ collected during each postural stability task to determine the final acceleration vector $\vec{V}_f$.

$$\vec{V}_f = R\vec{V}_i \quad (5)$$

Representative final acceleration vectors $\vec{V}_f$ for the AP and ML time series during the DLEO, SLEO, and DPS-AP tasks are shown in Figure 7. Comparing Figure 6 to Figure 7, the tilt due to accelerometer placement has been removed. Separation in the AP and ML time series in Figure 7b is due to tilt of the pelvis in the frontal plane during the SLEO task.

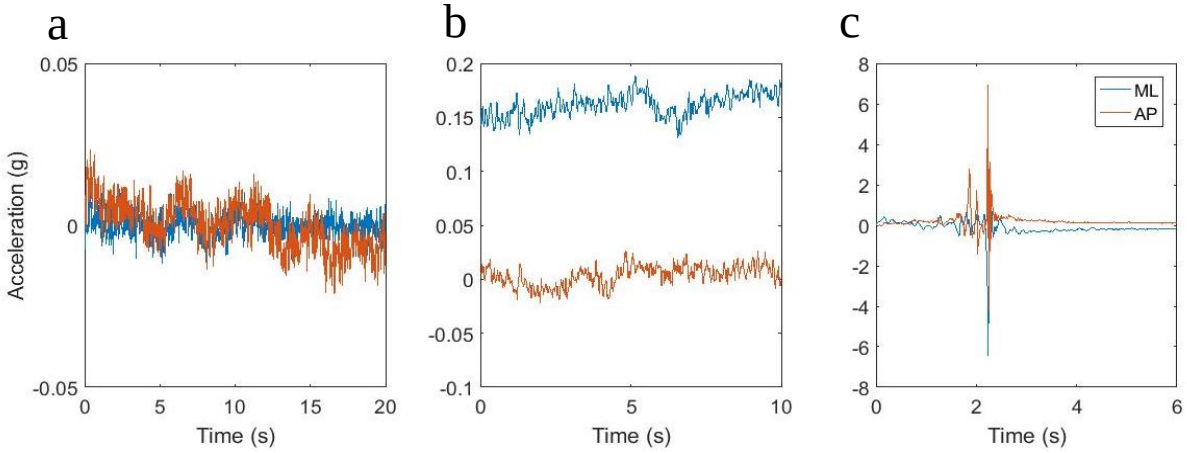

Figure 7. Representative quaternion rotation transformation. Representative medial-lateral (blue) and anterior-posterior (red) acceleration time series data after the quaternion rotation transformation is shown for the (a) double-leg stance with eyes open, (b) single-leg stance with eyes open and (c) forward jump-landing.

3.0 SPECIFIC AIM 1: SYSTEMATIC BIAS, WITHIN SUBJECT VARIABILITY AND INTERSESSION RELIABILITY OF ACCELEROMETRY MEASURES OF POSUTRAL STABILITY IN HEALTHY INDIVIDUALS AND INDIVIDUALS WITH CHRONIC ANKLE INSTABILITY

Postural stability is defined as an individual's ability to maintain their center of mass (COM) over a base of support. Postural stability can be assessed in static or dynamic states. Static postural stability assessments require a stationary base of support, and the demand placed on the postural control system varies based on visual input, base of support area, and support surface. Dynamic postural stability assessments require individuals to maintain their COM with the limits of their base of support while the base of support is perturbed. Static and dynamic postural stability are important for the prevention and rehabilitation of musculoskeletal injuries, and are often assessed with force plate technology in research settings. While force plate measures of static and dynamic postural stability provide greater fidelity compared to clinical assessments, they are not easily implemented clinical settings. Low-cost sensors may provide cost-effective, objective measures of postural stability for the prevention and rehabilitation of musculoskeletal injuries. To be adopted in a clinical setting, the reliability, concurrent validity and discriminative validity must be established in populations that have suffered from a musculoskeletal injury or may be at an increased risk of suffering a future injury. Specific Aim 1, presented in this chapter, establishes measures of reliability in healthy individuals and individuals with chronic ankle

instability (CAI). Concurrent and discriminative validity are later addressed as part of Specific Aims 2 and 3 (Chapters 4 and 5). The measures of reliability presented below establish important criteria for clinical implementation of sensor-based postural stability assessments and for subsequently examining measures of validity.

3.1 INTRODUCTION

Reliable, objective tools are needed in clinical and research settings to efficiently assess postural stability characteristics that may be related to musculoskeletal injury. Hopkins suggests systematic bias, within subject variation, and intersession reliability are the three most important measures in quantifying reliability particularly for measures of human performance.[90] The purpose of Specific Aim 1 is to establish the systematic bias, within subject variability, and intersession reliability of accelerometry-based measures of postural stability in healthy controls and individuals with CAI.

3.1.1 Systematic bias

Systematic bias is defined as a non-random change in a measure between trials that applies to all participants.[90] Examples of systematic bias include, but are not limited to, learning effects, training effects, and fatigue effects. Once identified, these biases can be mitigated by (i) providing familiarization trials or sessions to reduce learning effects and (ii) limiting session duration or providing appropriate rest periods to reduce fatigue effects. Postural sway parameters have been demonstrated to have learning effects when repeated on same or consecutive days

during double-leg stance tasks.[91] However, when repeated in one to two week intervals, learning effects were not detected.[92,93] Similarly, single-leg stance or more challenging static tasks such as a tandem stance do no exemplify learning effects between days.[94]

3.1.2 Within subject variability

Within subject variability is the random variation in a measure when one individual is tested repeatedly.[90] Sources of variability are largely driven by biological factors. Within subject standard deviation (S_w) and coefficient of variation (CV) can help to explain the within subject variability. The S_w of accelerometry-based measures of postural stability assessed during double-leg stance tasks has been shown to range from 0.0005 – 0.0134 *g* with CVs ranging from 12.9 – 54.4%.[86,87] Pagnacco et al. demonstrated within subject variations to be highly variable between subjects during double-leg stance tasks, which violates the assumption behind the intraclass correlation coefficient (ICC) that the variance of an individual subject is similar among subjects.[95] While measures of postural stability during static stance have also been shown to have a large amount of within subject variability in a single session, averaging multiple trials should decrease variability.[93]

Trial averaging is often used in human movement research due to the high variability of human performance. Measurement stability is thought to increase as the number of averaged trials increases.[96] To mitigate the high level of variability in human movement, researchers should consider the number of trials necessary to achieve performance stability. For static postural stability tasks, it has been found that participants' performance on the first trial is similar to performance on an average of three trials, suggesting that repeated trials are unnecessary.[16] An average of 3-5 trials is commonly accepted for jump-landing tasks used for dynamic postural

stability assessment.[79,97] However, there is limited data supporting the number of trials necessary to achieve measurement stability, particularly for novel dynamic postural stability tasks. Using a sequential averaging technique, researchers have shown an average of twelve trials to reach an acceptable predetermined level of performance stability during drop landing and vertical jumping tasks.[98,99]

3.1.3 Intersession reliability

Intersession, or test-retest, reliability represents how closely the measures of one trial or session track the measures of a repeated trial or session on an individual basis. ICC is defined in Specific Aim 1 as the proportion of true variance to total variance, where true variance is the difference between total variance and the variance due to error of measurement.[86] While few studies have considered systematic bias and within subject variability in accelerometry-based measures of postural stability, considerable amount of work has been done to assess intersession reliability in static postural stability assessments. Only one study has sought to determine the intersession reliability of accelerometry measures of dynamic postural stability.[19] A gap remains in assessing the intersession reliability in a population with CAI.

Accelerometry measures of postural stability have been shown to have poor ($ICC < 0.05$) to moderate ($0.5 \leq ICC < 0.75$) to good ($ICC \geq 0.75$) reliability during double-leg, single-leg, and jump-landing tasks (Table 2). The root mean square (RMS) derived from COM accelerations has been shown to have poor to moderate reliability during double-leg stance with and without visual input. ICC values during double-leg stance tasks range from 0.22-0.71, suggesting RMS during a double-leg stance may not be a reliable assessment of postural stability.[72,86] Other accelerometry-based time domain measures have been shown to have poor to good reliability

during double-leg stance including path length, normalized path length and peak-to-peak acceleration excursions with ICC values ranging from 0.47 – 0.88.[16,74,94] Frequency domain measures, such as mean frequency, also show moderate test-retest reliability.[72] Some researchers have considered intersession reliability in populations with postural stability impairments, for example in individuals with Parkinson's disease or those with vestibular impairments.[72,74] It is valuable to consider the reliability of an assessment in the desired test population. However, limited research has established the reliability of accelerometry measures of postural stability during double-leg stance tasks in young, healthy individuals and individuals with CAI.

Root mean square and path length values calculated from COM accelerations have also been shown to be reliable between days during static single-leg tasks, with ICC values ranging from 0.69-0.85.[86,94] The other measures described for the double-leg tasks (i.e., normalized path length, peak-to-peak acceleration excursions, and mean frequency) have not been evaluated for reliability during single-leg tasks. However, these measures are often derived from force plate data and have been utilized in assessing individuals with CAI. These measures may be useful in developing clinic-friendly assessments, but have not been assessed for reliability using COM accelerations.

Root mean square derived from COM accelerations has been shown to be reliable between days for the dynamic postural stability jump-landing task.[19] Other time and frequency domain measures have not been examined during this task, but may be equally or more effective in differentiating postural stability deficits. The intersession reliability of accelerometry measures of postural stability is unknown in a population with CAI. Pathologic populations may display altered movement patterns and may have greater variability in task performance. Understanding systematic bias, within subject variability, and intersession reliability of postural

stability tasks in specific populations is a critical step in determining the usefulness of an assessment particularly if repeated measures are collected. Measures of reliability are also useful in determining the magnitude of change required to be meaningful or clinically relevant. The purpose of this study was (i) to determine systematic bias among sessions and trials to establish guidelines for familiarization trials; (ii) to determine within subject variability of accelerometry-based measures of postural stability; and (iii) to establish the intersession reliability.

Table 2. Review of accelerometry-based measures of postural stability

Study	Participants	Measure	ICC Values by Task								
			DLEO	DLEC	DLEOF	DLECF	TANEO	TANEC	SLEO	SLEC	DPS-AP
Heebner et al., 2015	Recreationally active Age = 24.3 ± 4.2 years n = 10	RMS									0.835 - 0.924
Meo-Nilssen, 1998	Healthy Age = 22.9 ± 1.9 years n = 19	RMS	0.20 - 0.58	0.42 - 0.52					0.69 - 0.84		
Marchetti et al., 2013	Healthy Age = 47.4 ± 30 years n = 48 - 84 -- Vestibular disorder Age = 60.4 ± 8.5 years n = 4 - 17	NPL	0.86 -- 0.87	0.85 -- 0.67	0.74 -- 0.74	0.82 -- 0.46	0.83 -- 0.74	0.28 -- 0.80			
Mancini et al., 2012	Healthy Age = 60.2 ± 8.2 years n = 12 -- Parkinson's Disease Age = 60.4 ± 8.5 years n = 13	RMS	0.71 -- 0.83								
Saunders et al., 2015	Healthy Age = 81 ± 4.3 years	RMS	0.84 - 0.87	0.85 - 0.97	0.83 - 0.87	0.74 - 0.90					
Williams et al., 2016	Healthy Age = 28.8 ± 8.7 years n = 30	Path length, RMS	0.27 - 0.44	0.15 - 0.57			0.07 - 0.57	0.02 - 0.43	0.15 - 0.80	0.71 - 0.95	
Whitney et al., 2011	Healthy Age = 47.8 ± 21.2 years n = 81	NPL, RMS, P2P	0.16 - 0.72	0.46 - 0.72							

DLEO = double leg stance, eyes open; DLEC = double leg stance, eyes closed; DLEOF = double leg stance on foam, eyes open; DLECF = double leg stance of foam, eyes closed; TANEO = tandem stance, eyes open; TANEC = tandem stance, eyes closed; SLEO = single leg stance, eyes open; SLEC = single leg stance, eyes closed; DPS-AP = forward jump-landing maneuver; RMS = root mean square; NPL = normalized path length; P2P = peak-to-peak

3.2 MATERIAS AND METHODS

3.2.1 Participants

A total of 20 participants, ten healthy controls and ten individuals with CAI, were recruited and enrolled to assess measures of reliability for Specific Aim 1 of this dissertation. An equal proportion of men and women were recruited for each group. The 20 participants recruited to address Specific Aim 1 are a subset of a group of 50 participants recruited to address Specific Aims 2 and 3 of this dissertation. Participant recruitment and screening procedures are detailed in Appendix A and described briefly below. The demographic information for participants recruited to address Specific Aim 1 is described in Table 3. All participants engaged in physical activity for at least 30 minutes at a given time, three days per week. Participants were excluded if they self-reported history of fracture or surgery to the lower extremity, head injury within three months prior to test date, low back pain, or other known disorder (vestibular, neurological, or orthopedic) that could affect postural stability. Women were excluded if they were knowingly pregnant.

Table 3. Participant demographics for Specific Aim 1

Group	Gender	Height (cm)	Weight (kg)	BMI	Age (years)	CAIT	Ankle Sprains	Time since last sprain (months)
Control	5 M; 5 F	169.6 ± 10.5	66.1 ± 10.5	22.9 ± 2.3	22.8 ± 3.4	29.5 ± 0.8	0.0 ± 0.0	-
CAI	5 M; 5 F	176.0 ± 8.9	10.2 ± 7.5	22.7 ± 2.0	22.8 ± 3.4	19.1 ± 5.3	5.2 ± 3.5	21.6 ± 28.4
p value	-	0.52	0.19	0.077	0.82	0.00*	-	-

*Significant p value

Participants meeting inclusion and exclusion criteria completed three additional screening procedures for group assignment: self-reported ankle sprain history, Cumberland Ankle Instability Tool (CAIT) questionnaire, and talar tilt test (Appendix A, Figure 20). A certified Athletic Trainer (ATC) completed the talar tilt test. For this study, lateral ankle sprain (LAS) was defined as injury to the lateral ligaments of the ankle caused by rolling over on or "twisting" the ankle that resulted in disruption of normal physical activity for at least three days.[43] Participants were included in the control group if they had no prior history of self-reported LAS, scored ≥ 28 on the CAIT questionnaire, and showed no lateral mechanical laxity as measured by the talar tilt test. Participants were included in the CAI group if they self-reported a first incident LAS greater than one year prior to test date, had no subsequent LAS within three months prior to test date, scored ≤ 24 on the CAIT questionnaire, and had a positive sign of mechanical laxity as measured by the talar tilt test.[100]

3.2.2 Study design

A repeated-measures study design was used to determine any systematic bias, to establish within subject variability, and to assess the intersession reliability during ten postural stability tasks of varying difficulty. Each participant completed three test sessions on three separate days. The control group had an average of 7.5 ± 1.0 days between sessions 1 and 2 and 7.4 ± 1.3 days between sessions 2 and 3. The CAI group had similar average days between each session with an average of 8.7 ± 1.4 days between sessions 1 and 2 and 7.8 ± 1.1 days between sessions 2 and 3. Participants were asked to refrain from drinking caffeine and alcohol 24 hours prior to each testing session.

3.2.3 Experimental protocol

During each testing session, participants were asked to complete ten postural stability tasks of varying difficulty, eight static tasks and two dynamic tasks. The methods utilized for the postural stability testing are detailed in Section 2.2 and are described briefly below. A inertial measurement unit (IMU) equipped with a triaxial accelerometer (YEI 3-Space Sensor, YOST Labs, Portsmouth, OH) was secured with a belt so that the center of the sensor was positioned over L5, approximately at the COM. A neoprene belt was positioned over the sensor and secured around the participant's waist to limit vibration of the sensor during motion (Section 2.2.1, Figure 2). All tasks were performed on a constrained area (60 cm x 40 cm) located 2.5 m from a wall. COM accelerations were sampled at approximately 1200 Hz.

Prior to the postural stability assessments, accelerometer data was collected during a five-second static capture where the participants were asked to stand with their back against a wall to minimize body sway. Static tasks were performed barefoot and participants did not perform any familiarization trials. Participants were asked to complete five successful trials of each of the static postural stability tasks. Dynamic tasks were performed in the participants' own athletic shoes, and participants did not perform any familiarization trials. Participants were given unlimited attempts to successfully complete twelve trials of each dynamic task. Participants took a minimum of 30 seconds rest between trials of a task and two minutes rest between each task to minimize fatigue. The order of tasks was randomized for each participant using Latin square design. For consistency, tasks during sessions 2 and 3 were completed in the same order as session 1. All trials of a given task were completed before moving onto the next task.

Static double-leg tasks included double-leg stance (DL), double-leg stance on an Airex Pad (Airex Corp., Somersworth, NH) (DL-F), and tandem stance (TAN). Each of the three

double-leg static task positions were executed with eyes open (EO) and eyes closed (EC) (Section 2.2.1, Figure 3). Each of these tasks lasted for a duration of 20 seconds. A static single-leg stance (SL) task was also completed with EO and EC. The SL task lasted for a duration of ten seconds. Participants completed this task on their dominant limb (Control) or involved limb (CAI). During the SL tasks, the participant was permitted to touch down on the force plate with the non-test leg to maintain stability through the duration of the test.

The forward (DPS-AP) and lateral (DPS-ML) jump-landing dynamic postural stability tasks were initiated from a distance equal to 40% and 33% of the participant's height, respectively.[79] Participants were asked to initiate each jump from two limbs, clear a 30.5 cm (DPS-AP) or 15.2 cm (DPS-ML) hurdle, and land on their dominant limb (Control) or involved limb (CAI) on the constrained area. Upon landing, participants were asked to recover their balance and hold the single-leg position for five seconds.

3.2.4 Data reduction

The acceleration time series was resampled from 1200 Hz to 1000 Hz and filtered using a low-pass Butterworth filter with a cutoff frequency of 20 Hz for static tasks and 50 Hz for dynamic tasks. The low pass filter was selected based on a power spectral density analysis described in Section 2.2.2.

An alignment procedure was performed to correct for misplacement of the sensor along the vertical and transverse axes.[89] The mean accelerations in the x, y, and z directions were calculated during the five-second static capture. The static capture position was utilized to determine the sensors orientation relative to gravity. Then, a quaternion rotation transformation was applied to the filtered data.[89] Twenty seconds of data were analyzed for the double-leg tasks

and ten seconds of data were analyzed for the single-leg tasks. For the jump-landing tasks, a three second window during landing was used for analysis. The analysis window was set to begin where peak vertical acceleration during landing was identified.

Root mean square (RMS), normalized path length (NPL), peak to peak (P2P), stability indices (SI), and mean power frequency (MPF) were extracted from the transformed COM acceleration data.[16,97] Each variable was calculated along the anterior-posterior (AP) and medial-lateral (ML) axes as follows:

$$RMS = \sqrt{\frac{1}{N}\sum_{j=1}^{N-1}\left(a_j - a_{avg}\right)^2} \tag{6}$$

where *N* is the number of samples, a_j is acceleration data at time sample *j* in either the AP or ML direction, and a_{avg} is the average across the acceleration time series in either the AP or ML direction.

$$NPL = \frac{1}{t}\sum_{j=1}^{N-1}\left|a_{j+1} - a_j\right| \tag{7}$$

where *N* is the number of samples, t is the time duration, and a_j is acceleration data at time sample *j* in either the AP or ML direction.

$$SI = \frac{1}{m}\sqrt{\frac{1}{N}\sum_{j=1}^{N-1}\left(a_j\right)^2} \tag{8}$$

where *N* is the number of samples, a_j is acceleration data at time sample *j* in either the AP or ML direction, and *m* is the participant's body mass.

$$MPF = \sum_{j=1}^{M} f_j P_j / \sum_{j=1}^{M} P_j \quad (9)$$

where f_j is the frequency value of the acceleration data power spectrum at the frequency bin *j*, P_j is the acceleration data power spectrum at the frequency bin *j*, and M is the length of the frequency bin.

P2P was calculated as the difference between the maximum and minimum acceleration across the acceleration time series in either the AP or ML direction.

3.2.5 Statistical analysis

Statistical analyses were performed using SPSS software (v23; SPSS; Chicago, IL). Data were tested for normality and sphericity using the Shapiro-Wilk and Mauchly's tests, respectively. Data that were not normally distributed were transformed using 100x× natural logarithm of the observed value. If data were not normally distributed following the transformation, data were evaluated using a non-parametric test. An alpha level of 0.05, two sided, was set a priori.

3.2.5.1 Systematic bias

First, data were evaluated for systematic bias within session 1 by comparing means or medians between each trial for static (n = 5) and dynamic (n = 12) tasks. Trials that exhibited learning effects were excluded from further analysis in Sections 3.2.5.2 and 3.2.5.3. Second, data were evaluated for systematic bias among sessions by comparing means or medians from the average of remaining trials for each session (n = 3). Normally distributed data were evaluated using a repeated-measures analysis of variance (RM ANOVA). One 1-way RM ANOVA was completed

for each independent variable to determine any significant differences among the mean values for each trial. When the sphericity assumption was violated, a Greenhouse-Geisser correction was used. Any significant main effects were assessed further using pairwise comparisons with Bonferroni correction. P-values were adjusted to a 0.05 α-level within the SPSS software based on the number of comparisons. Data that were not normally distributed after transformation were evaluated using a Friedman test. Significant main effects were assessed further using Wilcoxon Signed Ranks Test with Bonferroni correction.

3.2.5.2 Within subject variability

The systematic bias analyses indicated learning effects may be present in repeated trials in a single session during the DLEC-F task measured using RMSap. However, many of the measures across tasks did not exemplify within session learning effects. Due to participants not receiving any familiarization prior to collection of the first trial, participants often had to be coached or reminded of the proper positioning during their first attempt. Therefore, some individuals had actually received some familiarization prior to the first successful trial. To control for some individuals receiving familiarization and others not, the first successful trial of each task was marked as a familiarization trial and excluded from further analysis.

Within subject variability was calculated using a sequential averaging technique for trials from static (n = 4) and dynamic (n = 11) tasks using similar methods to Connaboy et al. and Hopkins.[90,101] Within subject variability was reported as the typical error (TE_n) and coefficient of variation (CV) calculated as

$$\mathrm{TE}_n = \frac{s_{diff}}{\sqrt{2}} \tag{10}$$

$$\mathrm{CV} = 100\left(\frac{\mathrm{TE}_n}{M_n}\right) \tag{11}$$

where s_{diff} is the standard deviation of the difference in means of *n* and *n*-1 repeated cycles, TE_n is the TE from *n* repeated cycles, and M_n is the mean of the same *n* repeated cycles. 95% confidence interval were calculated for TE_n and CV.[102]

3.2.5.3 Intersession reliability

Intersession reliability was assessed using ICC(2,1). This model was chosen as each participant was assessed by the same rater (sensor) and the sensor utilized was the only sensor of interest in this study.[103] As suggested by Portney and Watkins, ICC values above 0.75 indicate good reliability, between 0.5 and 0.75 indicate moderate reliability, and below 0.5 indicate poor reliability.[104] The standard error of measurement (SEM) was calculated as

$$SEM = SD\sqrt{1 - ICC} \tag{12}$$

where SD is the standard deviation and ICC is the ICC(2,1).

3.3 RESULTS

3.3.1 Systematic Bias

Results from the RM ANOVAs and Friedman's tests indicate some accelerometry measures exhibited systematic bias across trials and/or across sessions while other accelerometry measures did not (Appendix B.1, Table 10). The results for each task are described in detail below.

Double-leg stance, eyes open: There were no significant main effects found across trials in the control and CAI groups. RMSap, P2Pap and APSI each had a significant main effect among sessions in the control group (RMSap: $p = 0.01$, P2Pap: $p = 0.05$, APSI: $p = 0.05$). Pairwise comparison indicated P2Pap in session 1 was significantly different from session 2 ($p = 0.04$), however no differences were found between sessions 1 and 3 or sessions 2 and 3 (Figure 8). The CAI group showed no significant main effects among sessions.

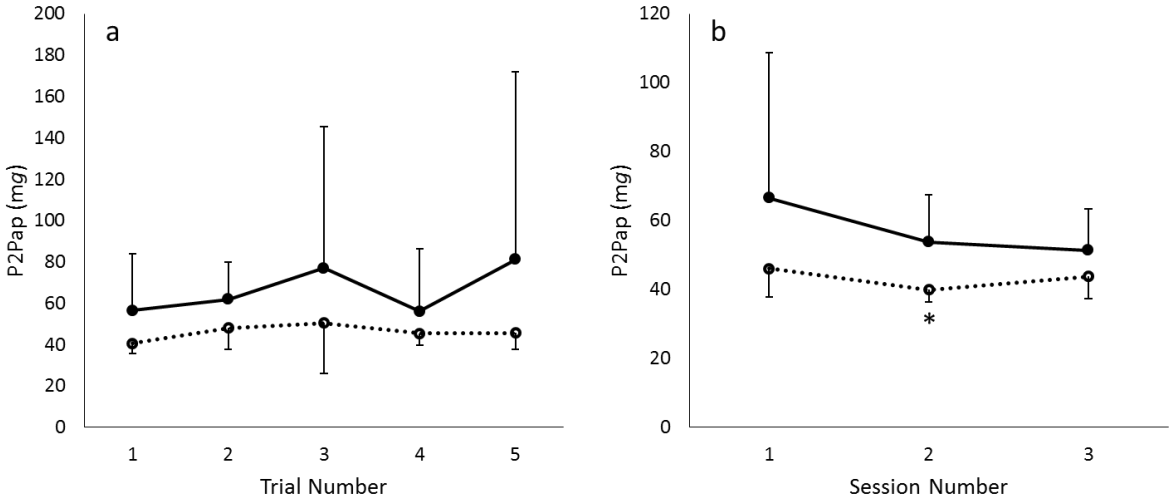

Figure 8. Systematic bias across trials and sessions for the double-leg stance with eyes open task. Anterior-posterior peak-to-peak measures (P2Pap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from session 1 ($p < 0.05$).

Double-leg stance, eyes closed: There were significant main effects across trials in the CAI group for RMSap ($p = 0.05$), NPLap ($p = 0.04$), P2Pap ($p = 0.04$), and APSI ($p = 0.02$). Pairwise comparisons for RMSap only showed a significant difference between trials 1 and 2 ($p = 0.01$). No other pairwise comparisons were found for RMSap (Figure 9). Due to this observed

learning effect, RMSap of trial 1, session 1 was excluded from further analyses. Pairwise comparisons for NPLap showed a significant difference only between trials 4 and 5 within the CAI group ($p < 0.01$). All trials for NPLap were included in further analyses as there was no clear learning or fatigue effect (Figure 10). There were no significant pairwise comparisons for P2Pap or APSI, therefore all trials were included in further analyses. There were no significant main effects found across trials in the control group. No main effects were found across sessions for the control and CAI groups.

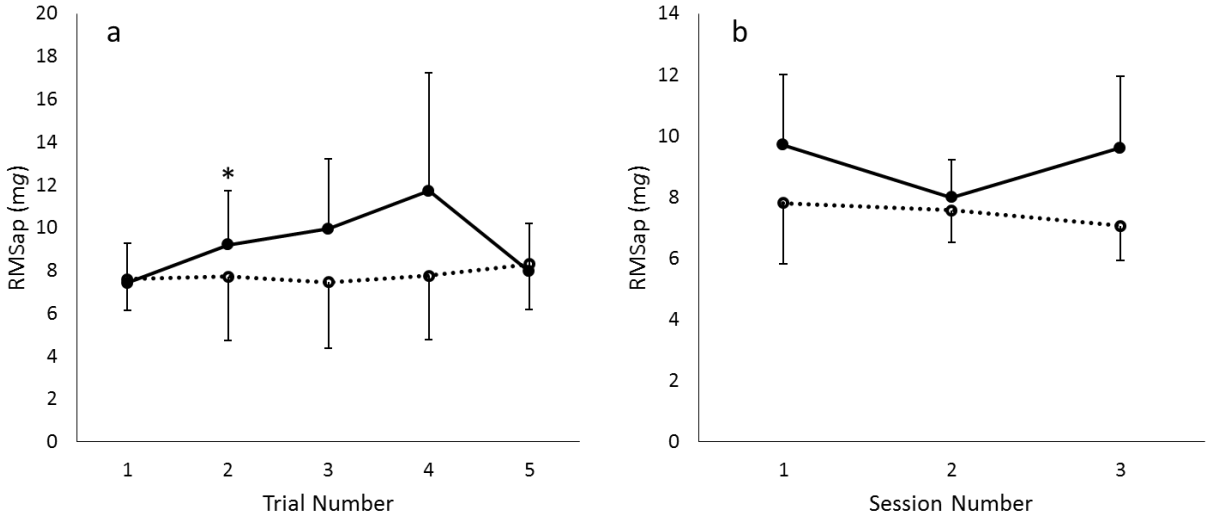

Figure 9. Systematic bias in root mean square across trials and sessions for the double-leg stance with eyes closed task. Anterior-posterior root mean square measures (RMSap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from trial 1 ($p < 0.05$).

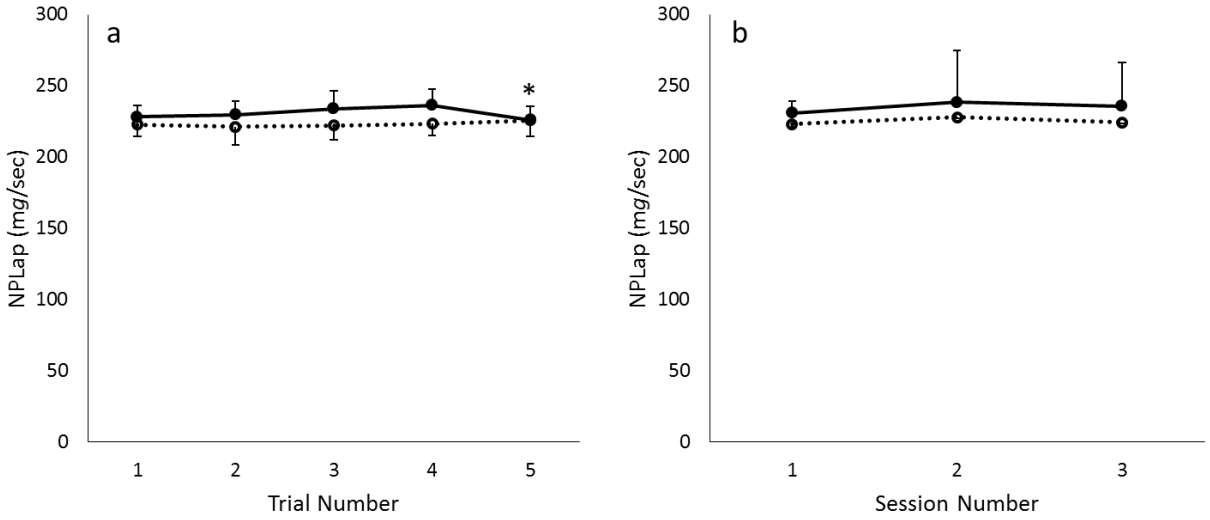

Figure 10. Systematic bias in normalized path length across trials and sessions for the double-leg stance with eyes closed task. Anterior-posterior normalized path length measures (NPLap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from trial 4 ($p < 0.05$).

Double-leg stance on foam, eyes open: There was a significant main effect across trials in the CAI group for RMSml ($p = 0.04$). Pairwise comparison revealed trial 3 to be significantly different from trial 4 ($p = 0.02$) and trial 4 to be significantly different from trial 5 ($p < 0.01$) (Figure 12). With no clear systematic bias, significant differences may be attributed to performance variability. To test this hypothesis, all trials were included in further analyses of within subject variability. There were no other significant main effects across trials for the control and CAI groups. Analysis across sessions revealed a significant main effect for MPFml in the CAI group ($p = 0.02$). Pairwise comparison showed session 1 was significantly different from sessions 2 ($p = 0.04$) and 3 ($p = 0.02$) (Figure 11). There was no difference between sessions 2 and 3. Due to this observed learning effect, MPFml from session 1 trials was not included for further analyses in the CAI group.

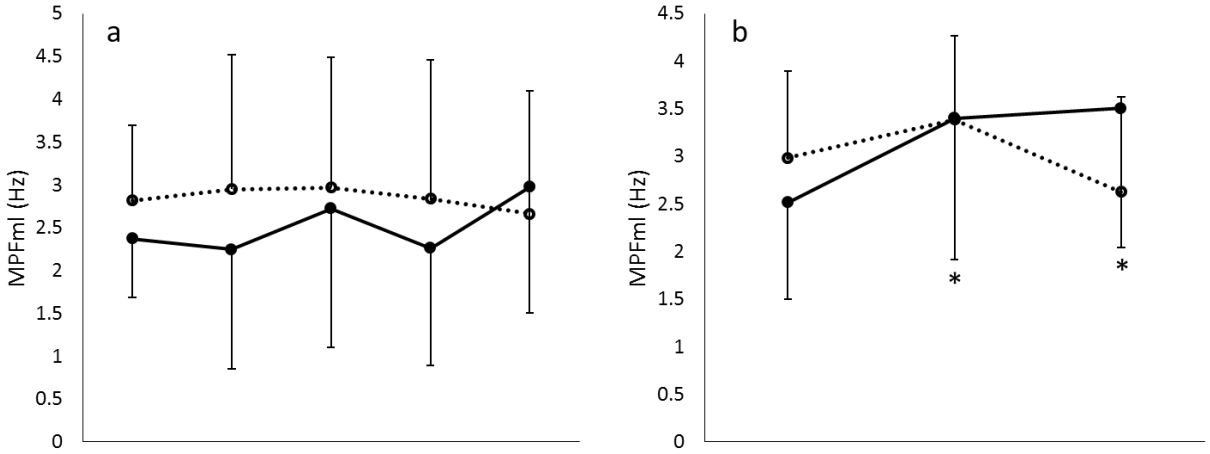

Figure 11. Systematic bias in mean frequency across trials and sessions for the double-leg stance on foam with eyes open task. Medial-lateral mean frequency (MPFml) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from session 1 ($p < 0.05$).

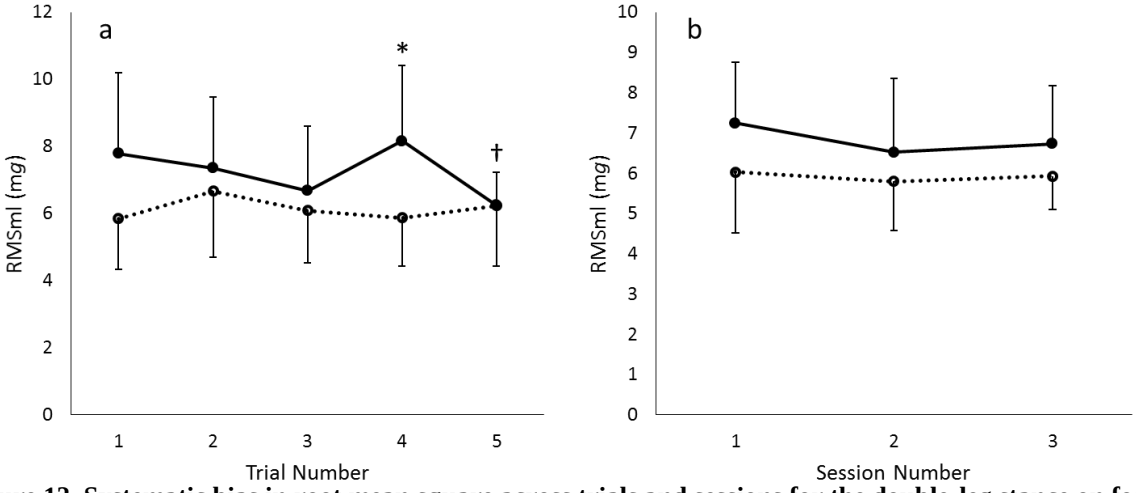

Figure 12. Systematic bias in root mean square across trials and sessions for the double-leg stance on foam with eyes open task. Medial-lateral root mean square measures (RMSml) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from trial 3 ($p < 0.05$). †Significantly different from trial 4.

Double-leg stance on foam, eyes closed: There was a significant main effect across trials for NPLap in the CAI group ($p = 0.03$), however there were no significant pairwise comparisons (Figure 13). No other significant main effects were found across trials for control and CAI groups. Analysis among sessions showed no significant differences for both control and CAI groups.

Tandem stance, eyes open: No significant main effects were found across trials for the control and CAI groups. In the control group, significant main effects were found across sessions for MLSI ($p = 0.04$) and MPFml ($p = 0.01$). In the CAI group, significant main effects were found across sessions for MPFap ($p = 0.04$). Pairwise comparisons showed no significant differences between sessions for either group.

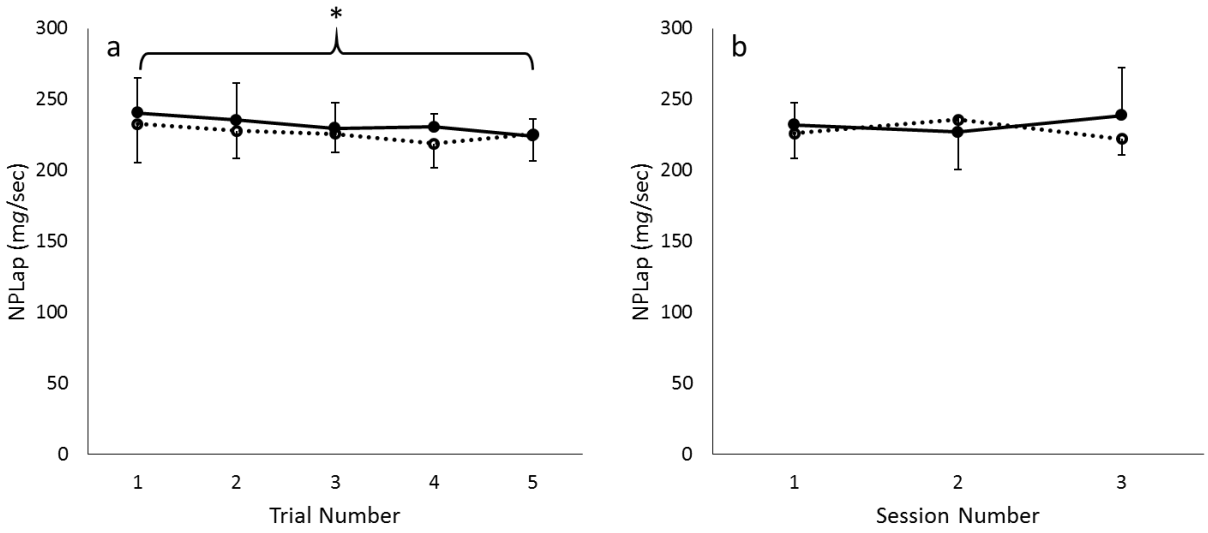

Figure 13. Systematic bias in normalized path length across trials and sessions for the double-leg stance on foam with eyes closed task. Anterior-posterior normalized path length (NPLap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significant main effect ($p < 0.05$).

Tandem stance, eyes closed: No significant main effects were found across trials or sessions for the control and CAI groups.

Single-leg stance, eyes open: No significant main effects were found across trials or sessions for the control and CAI groups.

Single-leg stance, eyes closed: No significant main effects were found across trials or sessions for the control and CAI groups.

Forward jump-landing: No significant main effects were found across trials for both control and CAI groups. A significant main effect was found for RMSap in the CAI group across sessions. Pairwise comparison showed session 3 was significantly different from sessions 1 ($p < 0.01$) and 2 ($p < 0.01$) (Figure 14). MPFml had a significant main effect across sessions in the control group ($p = 0.05$), however no significant pairwise comparisons were found between sessions.

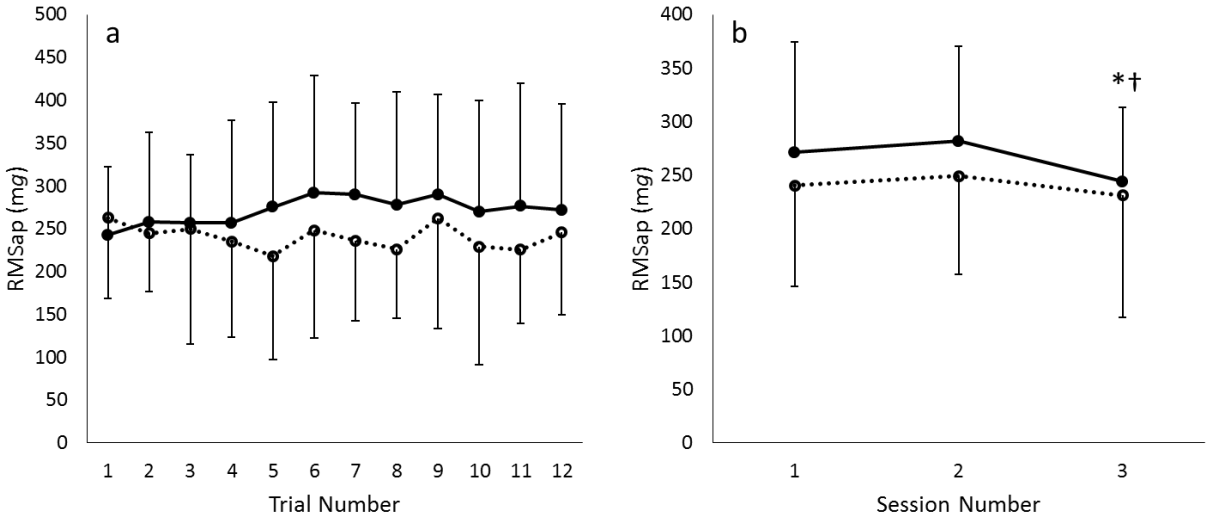

Figure 14. Systematic bias in root mean square across trials and sessions for the forward jump-landing task. Anterior-posterior root mean square (RMSap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from session 1 ($p < 0.05$). †Significantly different from session 2 ($p < 0.05$).

Lateral jump-landing: No significant main effects were found across trials for both control and CAI groups. Similar to the forward jump-landing, a significant main effect was found for RMSap in the CAI group across sessions. Pairwise comparison showed session 3 was significantly different from sessions 1 ($p < 0.01$) and 2 ($p = 0.02$) (Figure 16). MPFml had a significant main effect across sessions in the control group ($p = 0.01$). Pairwise comparison showed trials 1 and 2 were significantly different ($p = 0.01$), however no other pairwise differences were significant (Figure 15).

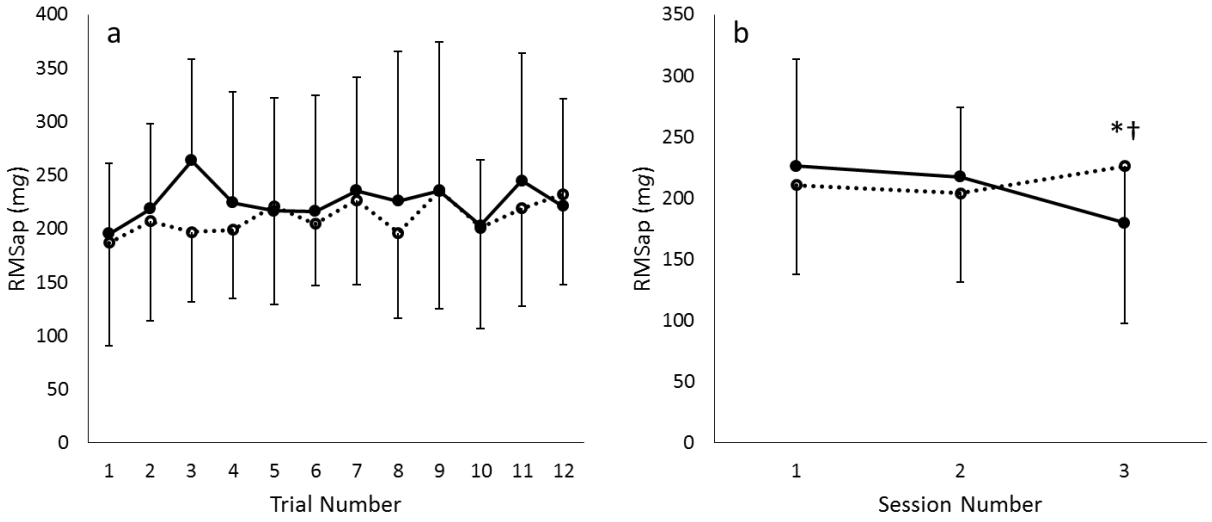

Figure 16. Systematic bias in root mean square across trials and sessions for the lateral jump-landing task. Anterior-posterior root mean square measures (RMSap) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from session 1 ($p < 0.05$). †Significantly different from session 2 ($p < 0.05$).

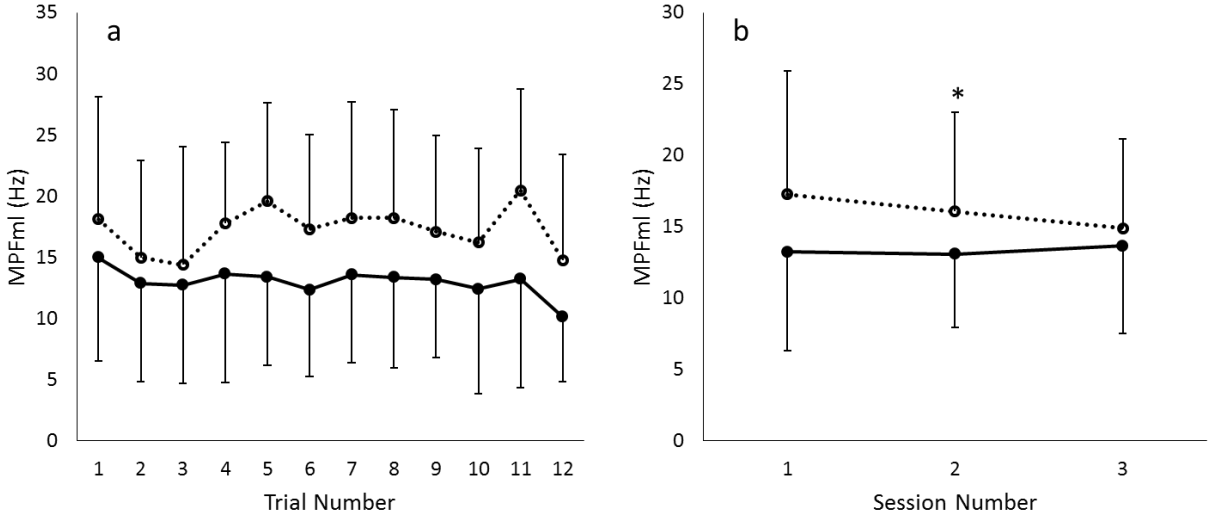

Figure 15. Systematic bias in mean frequency across trials and sessions for the lateral jump-landing task. Medial-lateral mean frequency measures (MPFml) are shown as means and standard deviations for the control (unfilled markers) and chronic ankle instability (filled markers) groups. Means are shown for (a) trials within session 1 and (b) sessions. *Significantly different from session 1 ($p < 0.05$).

3.3.2 Within subject variability

The results from the sequential averaging of trials for each of the postural stability tasks and accelerometry measures are presented in Table 11 through Table 28 in Appendix B.2. The results for each task are described in detail below.

For most variables derived from COM accelerations during the static tasks, the reliability was found to improve as the number of trials averaged increased. There were some exceptions to this pattern (i.e., DLEO P2Pml) where %CV increased as the number of trials averaged increased. In most instances, the largest return in decreasing $\%CE_{TE}$ was observed when comparing the average of three trials to the average of two trials. For the dynamic tasks, most variables reached a threshold point around n = 5 or n = 6 trials and any additional trials averaged yielded minimal return in reduction of the %CV.

For a given variable, TE_n increased with increasing task demand. This trend was observed in both control and CAI groups. The coefficient of variation, %CV, for a given variable and number of trials averaged varied across tasks however, there was no apparent trend observed with increasing task difficulty. Dynamic tasks demonstrated similar random error (%CV) compared to static tasks for a given value of n.

Across all static tasks NPLap and NPLml demonstrated the smallest %CV suggesting these variables were subject to the least amount of random error. MPFap and MPFml demonstrated the greatest %CV across all static tasks. Less variability in %CV was observed during the dynamic tasks. At n = 6, DPS-AP APSI had the smallest %CV for both control (2.82) and CAI (0.45) groups. For the control and CAI groups, the greatest %CV was observed in P2Pml (8.85) and MPFml (4.57), respectively. A similar range of %CV was observed for the DPS-ML task across variables at n = 6, however, RMSap and NPLml exhibited the smallest

%CV in control and CAI groups, (2.93 and 1.87, respectively), and MPFap exhibited the greatest %CV in both groups (6.50 and 7.15, respectively).

Overall, the control and CAI groups demonstrated similar trends described above, however, there were instances where the CAI group demonstrated a greater extent of random error (i.e., DLEO P2Pml). There were also instances where the control group demonstrated a greater extent of random error (i.e., TANEC P2Pml).

3.3.3 Intersession reliability

Intraclass correlation coefficients and SEM values for all measures extracted from the ten postural stability tasks in both the control and CAI groups are presented in Table 29 through Table 33 in Appendix B.3. The results are described in detail below.

Root mean square values derived from COM accelerations of both groups during the static tasks showed moderate to good ICC values ranging from 0.50 – 0.88 in the AP direction and 0.57 – 0.91 in the ML direction. DPS-AP and –ML RMS values showed moderate to good ICC values within the control group with values ranging from 0.61 – 0.94. The CAI group demonstrated moderate to good reliability with ICC values ranging from 0.74 – 0.94. Standard error in measurement for RMS values during static tasks ranged from 0.05 – 1.58 m*g* in the control group and 0.13 - 0.66 m*g* in the CAI group. DPS-AP and –ML had RMS SEM values ranging from 1.39 – 21.69 m*g* in the control group and 1.15 – 12.50 in the CAI group.

Normalized path length ICC values from the static tasks were poor to good ranging from -0.16 – 0.95 for the control group and -0.60 – 0.96 for the CAI group. ICC values were poor to moderate for the DL and DL-F tasks and moderate to good for the TAN and SL tasks. Static tasks had SEM values ranging from 0.67 – 15.41 m*g*/s for the control group and 0.90 – 11.75

m*g*/s for the CAI group. NPL measures from the dynamic tasks showed moderate to good ICC values ranging from 0.53 – 0.90 (SEM: 61.6 – 198.6 m*g*/s) within the control group and 0.81 – 0.96 (SEM: 11.6 – 279.4 m*g*/s) within the CAI group.

Peak to peak values during all postural stability tasks except the DLEC, TANEO, and SLEO tasks showed moderate to good reliability with ICC values ranging from 0.58 – 0.94 for the control group and 0.52 – 0.95 for the CAI group. DELC, TANEO, SLEO P2P values demonstrated poor to good intersession reliability with ICC values ranging from -0.30 – 0.85 and SEM values ranging from 0.59 – 9.31 m*g*. SEM values ranged from 0.18 – 12.58 m*g* for the static tasks and 9.34 – 347.75 m*g* for the dynamic tasks.

Stability indices in the AP and ML directions showed poor to good reliability across tasks with ICCs ranging from 0.30 – 0.93 in the control group and 0.25 – 0.90 in the CAI group. APSI and MLSI derived from the DPS-ML task showed large discrepancies between control and CAI ICC values with the control group demonstrating good reliability (ICC: 0.81 – 0.88) and the CAI group demonstrating poor reliability (ICC: 0.25 – 0.37). SEM values ranged from 0.01 – 0.20 m*g*/kg.

Mean power frequency demonstrated poor to good reliability for the static tasks with ICCs ranging from -0.19 – 0.85 (SEM: 0.06 – 0.53 Hz) for the control group and 0.14 – 0.86 (0.02 – 0.31 Hz) for the CAI group. There was a trend toward increased reliability as task difficulty increased. However, MPF also demonstrated poor to good reliability for the DPS-AP and –ML tasks with ICC values ranging from 0.55 – 0.95 (SEM: 0.25 – 0.78 Hz) for the control group and 0.46 – 0.89 (SEM: 0.12 – 1.70 Hz) for the CAI group.

3.4 DISCUSSION

The purpose of Specific Aim 1 was to examine the systematic bias, within subject variability, and intersession reliability of accelerometry-based measures of various postural stability assessments in young, healthy individuals and in individuals with CAI. Systematic bias assessments were utilized to test for any significant trends in the data that may suggest effects of learning or fatigue across trials within a single session and/or across sessions. Within subject variability assessments were performed to describe the variability associated with each postural stability task and measure and to identify an optimal number of trials to average for an assessment. Finally, intersession reliability was established to better understand which measures and postural stability tasks provided similar results across sessions, which is critical when performing repeated measures of an assessment in both research and clinical settings. Measures of reliability may vary in individuals with CAI as they may have different postural control strategies compared to healthy individuals. Establishing these measures of reliability is a critical step in determining the usefulness of an assessment particularly if repeated measures are collected. This work demonstrates that several COM acceleration measures of static and dynamic postural stability tasks are reliable between days, have no effect of learning or fatigue, and are similar between control and CAI populations.

Systematic Bias. Most accelerometry measures extracted from the various postural stability tasks indicated little to no learning effects or fatigue effects within a single session and across three sessions with the exception of the DLEO, DLEC, and DLEOF assessments (Table 4). The control group showed differences across sessions within the DLEO task, however not in a distinguishable, systematic manner. The average of trials within session 2 was significantly different from session 1, but session 3 was similar to session 1. Other researchers have

Table 4. Systematic bias in sensor-based measures of postural stability

DLEO	DLEC	DLEO-F	DLEC-F	TANEO	TANEC	SLEO	SLEC	DPS-AP	DPS-ML
RMSap	RMSap**	RMSap	RMSap	RMSap	RMSap	RMSap	RMSap	RMSap*	RMSap*
RMSml	RMSml	RMSml**	RMSml	RMSml	RMSml	RMSml	RMSml	RMSml	RMSml
NPLap	NPLap**	NPLap	NPLap	NPLap	NPLap	NPLap	NPLap	NPLap	NPLap
NPLml	NPLml	NPLml	NPLml	NPLml	NPLml	NPLml	NPLml	NPLml	NPLml
P2Pap*	P2Pap	P2Pap	P2Pap	P2Pap	P2Pap	P2Pap	P2Pap	P2Pap	P2Pap
P2Pml	P2Pml	P2Pml	P2Pml	P2Pml	P2Pml	P2Pml	P2Pml	P2Pml	P2Pml
APSI	APSI	APSI	APSI	APSI	APSI	APSI	APSI	APSI	APSI
MLSI	MLSI	MLSI	MLSI	MLSI	MLSI	MLSI	MLSI	MLSI	MLSI
MPFap	MPFap	MPFap	MPFap	MPFap	MPFap	MPFap	MPFap	MPFap	MPFap
MPFml	MPFml	MPFml*	MPFml	MPFml	MPFml	MPFml	MPFml	MPFml	MPFml*

*Significant main effect across sessions ($p < 0.05$). **Significant main effect across trials within session 1 ($p < 0.05$).

demonstrated that there are no learning effects in a similar double leg stance assessment of postural stability during repeated sessions.[105] It is possible this finding is due to variability of task performance rather than a systematic bias as less challenging postural stability tasks have been reported to have worse reliability between sessions.[16,86] This finding is also supported by the poor to moderate ICC values found for the measures extracted from the DLEO task in this study. Variability from session to session may be attributed to lack of focus during easier tasks.[16,86] Variability may also be attributed to young, active individuals having less regularity in sway during double-leg stance tasks which may allow them to be more adaptable to perturbations.[106] It is also possible the values may have stabilized if more trials and sessions were included in the analysis.

During the DLEC task, the CAI group had a significant difference between trials 1 and 2 in the RMSap measure. There was also a main effect across sessions in the CAI group performance on the DLEOF tasks where pairwise comparison revealed MPFml during session 1 was significantly different from sessions 2 and 3 with no difference between sessions 2 and 3.

These are possible learning effects given the difference observed was in the 1st trial or session and all remaining trials or sessions were consistent. Postural sway parameters have been demonstrated to have learning effects when repeated on same or consecutive days during double leg stance tasks, which may explain the observed learning effect within session.[91] However, when repeated in one to two week intervals, learning effects were not detected.[92,93] It is also possible this difference was due to chance or variability task performance, particularly since the difference was only observed in one of ten measures.

No systematic biases were found for the single leg stance tasks, the tandem stance tasks, or the DLECF task. In a study performed by Diamantopoulos et al., participants were instructed to practice the tandem stance task with eyes open and eyes closed over a consecutive ten day period.[107] Mean path length of center of pressure (COP) was measured for each participant on 5 occasions during the ten day period and showed no significant improvement suggesting lack of short-term learning effects.[107] Other studies have found learning effects to be greatest in tasks that remove visual input, but find the effects of learning decrease as days between sessions increase.[108]

Root mean square in the AP direction extracted from the DPS-AP task was significantly different during session 3 compared to sessions 1 and 2 in the CAI group. The average RMSap during session 3 was greater than in sessions 1 and 2. A similar pattern was observed in the RMSap variable during the DPS-ML task in the CAI group. This result is surprising as RMS is thought to decrease with improved stability.[86] It is therefore unclear if the increased RMSap values in session 3 are an effect of learning or a factor that was not tested. Nibali et al. studied systematic bias of kinetic and kinematic variables during a vertical jump in an athletic population and found familiarization trials were not necessary.[109] Participants in the Nibali et al. study

completed 2 – 6 testing sessions. More than three repeated testing sessions may be needed to identify or rule out learning effects in these more challenging tasks.

Within subject variability. For a given number of averaged trials, CVs and typical errors increased with increasing task difficulty indicating greater within subject variability with the more challenging tasks. Results of within subject variability showed similar trends between the control and CAI groups. Within subject variability has been shown to be greater in more challenging stance conditions with CVs ranging from 18.0 – 23.0% for accelerometry measures of double-leg stance task with eyes open on a firm surface and CVs ranging from 23.6 - 54.4% for a double leg stance with eyes closed on a foam surface.[86,87] The results of Specific Aim 1 were similar for the double-leg static tasks with CVs ranging from 0.94 – 61.51% when two trials were averaged. CVs reported from accelerometry derived RMS measures of postural stability during a single leg stance with eyes open range from 15.1 – 15.5%.[86] Similarly, CVs reported in the current study of the RMS measure during the single leg stance task with eyes open ranged from 7.04 – 12.75% when averaging two trials.

Within subject variability was reduced when the number of trials averaged was increased. Averaging four trials of the double-leg static tasks yielded much lower CVs (0.34 - 18.9%). Similarly, CVs from the single-leg static tasks dropped to 0.96 – 4.01% when averaging four trials. However, diminishing returns were found when averaging three trials of the static tasks compared to an average of four trials. For the dynamic tasks, diminishing returns in the CVs were found when averaging greater than six trials. Thus, an average of three static trials and six dynamic trials is suggested to minimize within subject variability in measures of postural stability. Performing six jump-landing trials in a clinical setting may be unrealistic due to time constraints. However, clinicians report perceived value of information gathered from a balance

assessment to be more important than testing time.[58] It is possible that the atypical performances are the ones of greatest interest, but without several repeated trials, it is difficult to distinguish typical from atypical for a given individual.

Intersession reliability. Interestingly, NPL measures demonstrated the least amount of within subject variability among trials within a single session, however, demonstrated poor to good intersession reliability during the less challenging tasks with ICCs ranging from -0.60 – 0.80. Intersession reliability improved with increasing task difficulty. Researchers have demonstrated similar findings of improved reliability with increasing task difficulty and suggest this may be a result of the participants being less focused on the easier tasks.[16,86] However, other researchers have found lower ICC values with increasing task difficulty, though the participants in the study fell within a much larger age range.[74] NPL and P2P measures have been shown previously to have better reliability compared to RMS measures.[16] ICCs of RMS measures from a DPS-AP task have been reported to range from 0.835 – 0.841 in the AP and ML directions.[19] ICCs of the RMS measures during the DPS-AP task ranged from 0.61- 0.91 and were slightly better in the DPS-ML task 0.74 – 0.94. ICCs during a DPS-ML task have not been previously reported.

The work presented addressing Specific Aim 1 has several limitations. The systematic bias was determined by first evaluating systematic bias of trials within session 1 and then between sessions. Therefore, it is unknown if learning effects, if any were the same during sessions 2 and 3 as observed during session 1. Future studies may consider an iterative approach to consider the origin of systematic bias when biases are present across sessions. Also, time between sessions was restricted to a 7 – 10 day window, however, time of day was not controlled for. Postural stability may not be affected by time of day,[110] but it is possible other intrinsic

factors such as tiredness could affect performance. This study was conducted with young, active individuals and the results should not be extrapolated to older adult populations. The results from the CAI group are specific to CAI and may not be similar in individuals that have suffered other musculoskeletal injuries. Only three testing sessions were conducted. For the more challenging jump-landing tasks, it may take more than three testing sessions to observe any effects due to learning.

3.5 CONCLUSION

Low-cost sensors may be an effective alternative to force plates for objective assessment of postural stability for the prevention and rehabilitation of musculoskeletal injuries. To be readily adopted in an orthopedic clinical setting, the accelerometry measures must be reliable, valid compared to the gold-standard and discriminatory in populations that have suffered from musculoskeletal injury. The purpose of Specific Aim 1 was to establish measures of reliability in control and CAI groups. Several accelerometry measures of postural stability were found to be reliable across sessions and did not show learning or fatigue effects within or across sessions. More challenging tasks such as the single-leg stance or jump-landing tasks showed good reliability and small CVs suggesting they may be more appropriate to use for a repeated measures study design and may be best when comparing young, active populations. The results presented in Specific Aim 1 suggest averaging at least three static postural stability trials and at least six dynamic postural stability trials to minimize within subject variability. The criteria established are important to consider when implementing these postural stability assessments in a

clinical setting, and have been taken into consideration for the measures of validity examined in Specific Aims 2 and 3.

4.0 SPECIFIC AIM 2: CONCURRENT AND DISCRIMINATIVE VALIDITY OF ACCELEROMETRY MEASURES OF POSTURAL STABILITY IN HEATHY INDIVIDUALS

Great advancements have been made in accelerometry-based measures of postural stability in the past decade. However, the majority of the work to date has been in quantifying postural stability during static stance, either in a double- or single-leg stance position and on various surfaces.[16,18,74,86,87] In Chapter 3, Specific Aim 1 was addressed, establishing the reliability of accelerometry measures of static and dynamic postural stability during ten tasks of varying difficulty in healthy individuals and individuals with chronic ankle instability (CAI). To be useful in a clinical setting, new measures should be compared against the gold-standard, criterion measures and should have discriminative ability. This chapter addresses Specific Aim 2 and two types of validity are established: concurrent validity and discriminative validity. Concurrent validity is determined by comparing accelerometry measures of postural stability to the gold-standard force plate measures of postural stability that are measured concurrently. Discriminative validity can be assessed in several ways. For Specific Aim 2, discriminative validity is assessed by determining the ability of accelerometry measures of postural stability to differentiate between tasks of varying difficulty. Specific Aim 3, presented in Chapter 5, establishes another type of discriminative validity by comparing accelerometry measures of postural stability in healthy individuals compared to individuals with CAI. The results from Specific Aim 2 will be

useful in determining how well the accelerometry measures correspond to the well-established force plate measures and will determine the sensitivity of the measures to changes in task demand. This will allow for development of a continuum of postural stability assessments that can be used in rehabilitation or performance training settings.

4.1 INTRODUCTION

Postural stability deficits have been associated with history of musculoskeletal injury and are able to predict lower extremity injury.[8,38,60,61] Postural stability is commonly assessed in clinical and research settings and numerous protocols and assessments exist. Clinical assessments typically require few resources, are inexpensive, and are easy to implement. However, clinical assessments are often subjective and have limited resolution.[13,54] Laboratory assessments often involve force plate technology which provides an objective measure, but is expensive and not easily implemented in a clinical setting. Low-cost accelerometers are an excellent alternative to force plates, providing the opportunity to achieve objective assessments of postural stability at a low cost. However, the clinical usefulness of an assessment increases when the concurrent and discriminative validity have been established. Therefore, the purpose of Specific Aim 2 was to establish the concurrent validity of accelerometry measures of postural stability compared to force plate measures during concurrent analysis, and to determine the discriminative validity of accelerometry measures by assessing their ability to differentiate between tasks of varying difficulty.

To assess concurrent validity, two different measurement procedures are carried out simultaneously and the new measures are compared to the criterion measures. For this study, the

new measures are the accelerometry measures of postural stability and the criterion measures are force plate measures of postural stability. Discriminative validity is defined as the ability of a measure to differentiate between two groups. For Specific Aim 2, discriminative validity is assessed by determining the ability of accelerometry measures to differentiate among tasks of varying difficulty.

Force plates are often used to assess postural stability, particular in research settings and can be considered the gold-standard. For athletic populations, center of pressure (COP) or ground reaction forces (GRFs) collected during a single-leg stance are most commonly used to assess static postural stability and ground reaction forces following a single leg jump-landing are often utilized to assess dynamic postural stability.[79] Measures calculated from force plate analysis during these tasks often have greater discriminatory ability compared to clinical tests. Mean velocity and distance from the mean COP differentiate between individuals with and without CAI in a single-leg stance task.[60,67,70] Dynamic postural stability index measures have also been shown to differentiate injured and healthy populations.[61,111,112] In a prospective study, force plate measures of postural stability during a single leg stance have been shown to be predictive of ankle sprain.[8] While objective measures of postural stability have high value in injury risk assessment, the cost and size of force plates limit their portability and utility in a clinical setting.

Researchers have investigated the validity of objective, accelerometry-based measures of postural stability compared to clinical tests. An instrumented version of the Balance Error Scoring System (BESS) test shows that during stance on foam, objective accelerometry-based measures accurately predicted BESS scores assigned by a rater.[113] However, the algorithm did not accurately predict BESS scores during stance on a firm surface. This is consistent with force

plate measures collected during the BESS test that show higher association between force plate measures and BESS scores during the more challenging conditions.[56]

Researchers have also investigated the relationship between center of mass (COM) accelerations and COP measures of postural stability. Though these instruments measure different aspects of postural sway, a link between measures has been established using the inverted pendulum model.[114] Whitney et al. observed significant associations between COM acceleration and force plate derived center of pressure measures during computerized dynamic posturography.[16] Several time and frequency domain accelerometry measures of postural stability have been shown to be significantly correlated with force plate measures of postural stability during a double-leg stance as well as a tandem stance ($r = -0.54 - 0.89$).[18,19,72] Correlation coefficients ranged from -0.16 - 0.75 when comparing a measure of amplitude between accelerometry and force plate measures during a single leg stance.[19,115]

Jump-landing dynamic postural stability assessments quantified using the RMS of COM accelerations showed low to moderate correlations with the stability indices calculated from concurrently measured ground reaction forces. Correlation coefficients ranged from -0.291 – 0.703.[19] Though there is limited research on accelerometry measures of this jump-landing assessment of dynamic postural stability, researchers have investigated various performance parameters during jump landing tasks including jump height, landing impact, velocity, and power.[84,116] Elvin et al. found strong associations between accelerometer and force plate assessment of jump height and landing impact, however Choukou et al. found accelerometer and force plate assessments of jump performance to be different.[84,116]

Measures of postural control should be sensitive enough to detect differences in postural control as task difficulty increases. Goldie et al. demonstrated this phenomenon using force plate

measures of postural stability during double leg, tandem, and single leg stance positions.[48] Accelerometry measures of postural stability have also been shown to have the required sensitivity to detect these differences.[18,19,87] Though during simple double-leg stance tasks with eyes open and eyes closed, differences are on the order of 0.001 *g*.[81,86]

Several studies have considered the validity of various static assessments of postural stability, but few have developed and validated methods for assessing dynamic postural stability during a jump-landing task. The purpose of Specific Aim 2 is (i) to determine the ability of accelerometry-based measures to differentiate between tasks of various difficulty levels and (ii) to establish the relationship between accelerometry and force plate measures of postural stability. RMS of COM accelerations have been shown to increase with increasing task difficulty.[19,87] Therefore, it was hypothesized other COM acceleration derived measures would also be able to differentiate among task difficulty. It was also hypothesized the measures would show significant correlations with force plate measures of postural stability.

4.2 MATERIALS AND METHODS

4.2.1 Participants

A total of 25 young, healthy participants (13 men and 12 women) were recruited and enrolled in the study to assess concurrent and discriminative validity of accelerometry-based measures of postural control (age: 22.6 ± 3.0 years; height: 173.1 ± 9.9 cm; weight: 67.8 ± 10.5 kg; BMI: 22.6 ± 2.4). Participant recruitment and screening procedures are detailed in Appendix A. All participants engaged in physical activity for at least 30 minutes at a given time, three days per

week. Participants were excluded if they self-reported history of fracture or surgery to the lower extremity, head injury within three months prior to test date, low back pain, or other known disorder (vestibular, neurological, or orthopedic) that could affect postural stability. Women were excluded if they were knowingly pregnant.

4.2.2 Study design

A cross-sectional cohort study design was utilized (i) to assess the ability of accelerometry-based measures of postural stability to distinguish between tasks of various difficulties and (ii) to establish the relationship between accelerometry and force plate measures of postural stability. All testing for this specific aim was completed during one testing session. Participants were asked to refrain from drinking caffeine and alcohol 24 hours prior to each testing session.

4.2.3 Experimental protocol

Participants were asked to complete ten postural stability tasks of varying difficulty, eight static tasks and two dynamic tasks. The postural stability tasks are explained in detail in Section 2.2.1 and briefly below. A triaxial accelerometer (YEI 3-Space Sensor, YOST Labs, Portsmouth, OH) was secured with a belt so that the center of the sensor was positioned over L5, approximately at the COM. A neoprene belt was positioned over the sensor and secured around the participant's waist to limit vibration of the sensor during motion. All tasks were performed on a force plate (Type 9286BA, 60 cm x 40 cm platform; Kistler Instrument Corp. Amherst, NY) located 2.5 m from a wall. Center of mass accelerations and GRFs were collected concurrently and were sampled at approximately 1200 and 1000 Hz, respectively.

Prior to the postural stability assessments, accelerometer data was collected during a five-second static capture where the participants were asked to stand with their back against a wall to minimize body sway. Prior to each trial, the participants performed three heel taps to allow the COM acceleration and GRF data to be later synchronized in Matlab. Static tasks were performed barefoot and the participants performed one familiarization trial. Participants were asked to complete four successful trials of each of the static postural stability tasks following the familiarization trial. Dynamic tasks were performed in the participant's own athletic shoes and participants performed familiarization trials until the participant had successfully completed one trial. Participants were then given unlimited attempts to successfully complete eleven additional trials of each given dynamic task. Participants took a minimum of 30 seconds rest between trials of a task and two minutes rest between each task to minimize fatigue. The order of tasks was randomized for each participant using Latin square design. All trials of a given task were completed before moving onto the next task.

Static double-leg tasks included double-leg stance (DL), double-leg stance on an Airex Pad (Airex Corp., Somersworth, NH) (DL-F), and tandem stance (TAN). Each of the three double-leg static task positions will be executed with eyes open (EO) and eyes closed (EC). Each of these tasks lasted for a duration of 30 seconds, ten seconds for the heel taps and 20 seconds static stance. A static single-leg stance (SL) task was also completed with EO and EC. The SL task lasted for a duration of 20 seconds, ten seconds for the heel taps and ten seconds static stance. Participants completed this task on their dominant limb (Control) or involved limb (CAI). During the SL tasks, the participants were permitted to touch down on the force plate with the non-test limb to maintain stability through the duration of the test.

The anterior-posterior (DPS-AP) and medial-lateral (DPS-ML) dynamic postural stability tasks were initiated from a distance equal to 40% and 33% of the participant's height, respectively.[79] Participants were asked to initiate each jump from two limbs, clear a 30.5 cm (DPS-AP) or 15.2 cm (DPS-ML) hurdle, and land on their dominant limb (Control) or involved limb (CAI) on the force plate.

4.2.4 Data reduction

The acceleration time series was resampled from approximately 1200 Hz to 1000 Hz and filtered using a low-pass Butterworth filter with a cutoff frequency of 20 Hz for static tasks and 50 Hz for dynamic tasks. The low pass filter was selected based on a power spectral density analysis described in Section 2.2.2. Ground reaction forces were also filtered using a low-pass Butterworth filter with cutoff frequencies of 20 Hz and 50 Hz for static and dynamic tasks, respectively.

An alignment procedure was performed to correct for misplacement of the sensor along the vertical and transverse axes.[89] The mean accelerations in the x, y, and z directions were calculated during the five-second static capture. The static capture position was utilized to determine the sensors orientation relative to gravity. Then, a quaternion rotation transformation was applied to the filtered data.[89]

For the static tasks, data were synchronized in Matlab using the cross-correlation function, *xcorr*. Five-second regions of the original signals where the heel taps occurred were selected and compared using cross-correlation (Figure 17). The signals were aligned using the estimated lag difference, which was the delay between signals when the cross-correlation was at a maximum (Figure 18). For the double-leg static tasks, the first ten seconds of the trial were

removed and the remaining 20 seconds were used for further analysis. For the single-leg static tasks, the first ten seconds of the trial were removed and the remaining ten seconds were used for analysis. The dynamic tasks were not synchronized using the cross-correlation method. Rather, the start of the three second window was identified at the sample number where peak vertical acceleration or peak vertical GRF occurred.

Root mean square (RMS), normalized path length (NPL), peak to peak (P2P), stability indices (SI), and mean power frequency (MPF) were extracted from the transformed COM acceleration data and the ground reaction forces.[16,97] Each variable was calculated along the anterior-posterior (AP) and medial-lateral (ML) axes using equations (6) – (9) and methods described in Section 3.2.4.

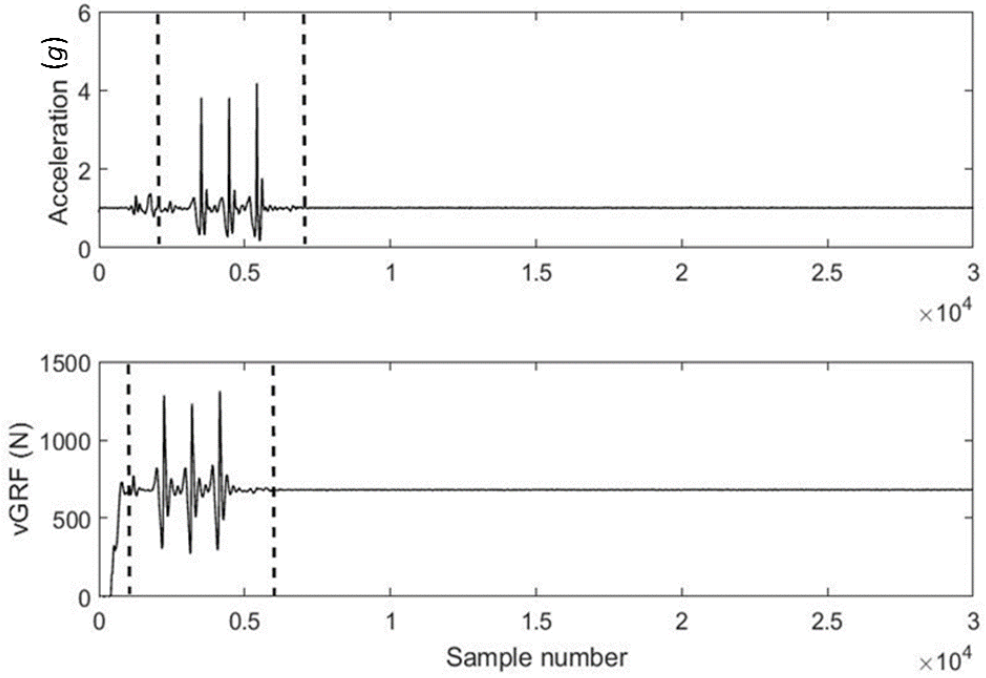

Figure 17. Vertical acceleration and ground reaction force time series prior to signal synchronization.

Representative vertical acceleration and ground reaction force (vGRF) time series from the double leg stance with eyes open task. Vertical bars represent the 5 second time window utilized for cross-correlation.

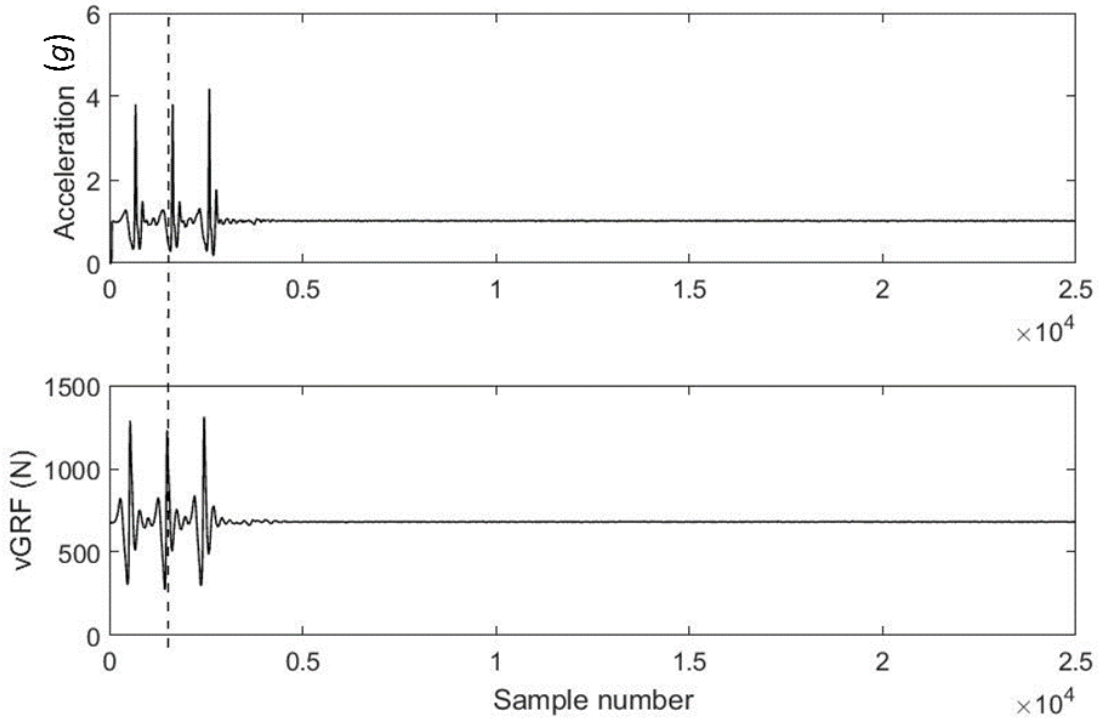

Figure 18. Vertical acceleration and ground reaction force time series following signal synchrozation.

Representative vertical acceleration and ground reaction force (vGRF) time series from the double leg stance with eyes open task. Vertical bar is placed for reference at sample number n = 1500.

4.2.5 Statistical analysis

Statistical analyses were be performed using SPSS software. Descriptive statistics including means and standard deviations as well as medians and interquartile ranges were calculated where appropriate. Data were tested for normality using a Shapiro-Wilk test. Data were not normally distributed, so nonparametric statistical tests were used to test the hypotheses. A Friedman's test with Bonferroni correction was used to determine if the COM acceleration measures could

distinguish between tasks of various difficulties and Spearman's ranked correlation (ρ) to determine the relationship between the COM accelerations and force plate measures. ρ values greater than 0.75 indicated strong correlation, those between 0.5 and 0.75 indicate moderate correlation, and those less than 0.5 indicate weak correlation. An alpha level of 0.05, two sided, was set a priori.

4.3 RESULTS

Each variable calculated from COM accelerations showed significant main effects across all static tasks and across all tasks in both the AP and ML directions (Figure 19). The results for the post hoc pairwise comparisons are presented in Appendix C (Table 34). No differences were observed between the DLEO and DLEC tasks for any of the COM acceleration variables. Few differences between tasks were observed in the MPFap and MPFml variable analysis. Only NPLml showed a significant difference between DLECF and TANEO tasks. Significant differences were found between most tasks in the RMS, NPL, and P2P measures in both the AP and ML directions. MLSI also showed significant differences between most tasks.

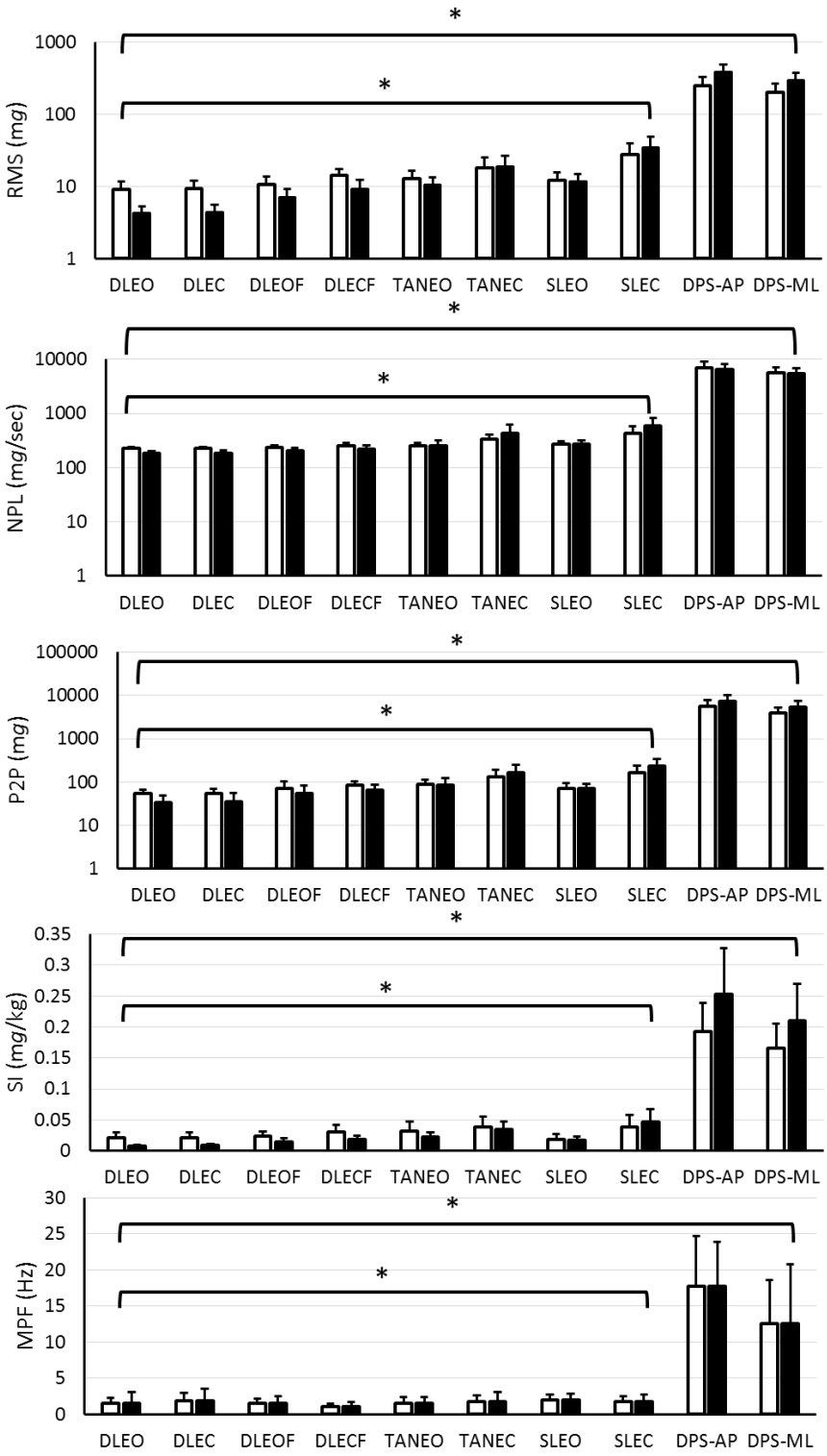

Figure 19. Between task comparisons for accelerometry measures of postural stability. Means and standard deviations for anterior-posterior (unfilled bars) and medial-lateral (filled bars) are presented for each task and measure. *Significant main effect ($p < 0.05$).

Associations between the measures derived from COM accelerations and ground reaction forces assessed using Spearman's ranked correlations ranged from weak to strong (Table 5). Across all static tasks, NPL and P2P showed the highest correlations in both the AP and ML with Spearman's rho correlation coefficients ranging from 0.467 – 0.925. MPF (AP and ML) showed the weakest correlations with Spearman's rho correlation coefficients ranging from -0.542 – 0.673. The DPS-AP task showed no significant correlations between COM acceleration and ground reaction force measures. The DPS-ML task showed weak to moderate correlations with the strongest correlations in the RMSap, NPLap, and P2Pap measures.

Table 5. Correlation between COM acceleration and force plate measures of postural stability

	Task									
	DLEO	DLEC	DLEOF	DLECF	TANEO	TANEC	SLEO	SLEC	DPS-AP	DPS-ML
RMSap	0.385	0.368	0.381	0.620*	0.698*	0.701*	0.502*	0.689*	0.218	0.590*
RMSml	0.809*	0.751*	0.688*	0.834*	0.569*	0.795*	0.448*	0.652*	0.237	0.153
NPLap	0.467*	0.462*	0.655*	0.749*	0.874*	0.887*	0.888*	0.840*	0.244	0.525*
NPLml	0.533*	0.665*	0.818*	0.925*	0.890*	0.887*	0.783*	0.648*	0.177	0.346
P2Pap	0.450*	0.511*	0.665*	0.685*	0.825*	0.885*	0.705*	0.815*	0.061	0.517*
P2Pml	0.662*	0.693*	0.748*	0.860*	0.797*	0.871*	0.644*	0.777*	0.296	0.161
APSI	0.239	-0.034	0.063	0.374	0.456*	0.490*	0.256	0.633*	0.007	0.128
MLSI	-0.075	0.116	0.179	0.512*	0.398*	0.581*	0.342	0.568*	0.108	0.208
MPFap	-0.098	-0.244	0.030	0.151	0.040	-0.031	-0.005	0.538*	-0.198	0.155
MPFml	-0.518	-0.542	-0.150	-0.302	0.485*	0.673*	0.385	0.340	0.302	0.092

*Significant correlation ($p < 0.05$)

4.4 DISCUSSION

For accelerometry measures of postural stability to be useful in a clinical setting, it is important to assess their concurrent validity against the gold-standard and their discriminative validity. Thus, the purpose of Specific Aim 2 was to determine the relationship between COM acceleration measures and GRF measures of postural stability when measured concurrently and to assess the ability of accelerometry measures of postural stability to differentiate task difficulty. All COM acceleration measures showed significant main effects across the ten postural stability tasks. Pairwise comparisons revealed that some measures were better at distinguishing task difficulty than others and none of the measures were able to distinguish between the two simplest tasks: double-leg stance with eyes open and double-leg stance with eyes closed. Many of the COM acceleration measures showed strong correlations with the force plate measures during the static tasks, however, associations were weak to moderate during the dynamic tasks. The results of Specific Aim 2 suggest a waist worn accelerometer provides valid measures of a continuum of static postural stability assessments. While dynamic measures of postural stability were found to be more challenging, COM accelerations during this task may be measuring a different mechanism of postural stability than the ground reaction forces.

Normalized path length, P2P, and RMS in the AP and ML directions as well as MLSI are most sensitive to differences in task difficulty. Though many AP measures showed differences between tasks of varying difficulty, ML measures of sway demonstrated greater sensitivity across tasks. Heebner et al. found RMS in the ML direction to better differentiate task difficulty compared to RMSap among a similar continuum of static tasks.[19] Greater sensitivity in the ML direction has also been demonstrated during force plate analysis of postural stability tasks.[48,79] Our study showed greater sensitivity to changes in difficulty compared to the study by Heebner

et al, but similar to a study by Neville et al.[19] Given that each of these studies used similar populations, the difference in results may be due to accelerometer sensitivity and range. Neville et al. utilized an accelerometer with ± 1.7 *g* range, while Heebner et al. utilized an accelerometer with a ± 16 *g* range to capture accelerations during the dynamic postural stability tasks.[18,19] The current study used an accelerometer with a selectable range which may have provided greater sensitivity during the static tasks while also providing the range to capture high accelerations during the dynamic tasks. However, the measures did not detect differences between the DLEO and DLEC tasks. This result may be attributed to the sensitivity of the accelerometer utilized and/or may be a result of high performance of the young, healthy population.

The strongest associations between accelerometry and force plate assessment of postural stability were found in the NPL and P2P measures. RMS measures also showed moderate to strong associations. These strong associations suggest a link exists between the measures.[114] This link is important as many force plate measures of static postural stability have been shown to be effective in differentiating between individuals with minor balance deficits due to musculoskeletal injuries such as ankle injuries,[5,60–63] and measures derived from COM accelerations that show strong associations with force plate measures may have similar capabilities. RMS measures from COM accelerations and ground reaction forces have been shown previously to have moderate associations ranging from 0.63 - 0.74.[18,72] Similar to the results presented in the current study, Whitney et al. found NPL to have the greatest association between accelerometry and force plate based measures compared to RMS and P2P during computerized dynamic posturography.[16]

Among static tasks, greater associations between COM acceleration and GRF measures were found as task difficulty increased, and associations were also found to be greater in the ML

direction than in the AP direction. The increased association observed during the more challenging static conditions corresponds to what has been reported in the literature.[16] One possible explanation for this trend is a change in strategy of balance in the more challenging conditions. Hip strategy can lead to greater accelerations at the COM that may not be reflected in GRF measures.[16,115] Similar to the current study, Adlerton et al. also found greater associations for a measure of amplitude between measures in the ML times series compared to the AP time series.[115] This could be attributed to differences in hip versus ankle strategy. Hip strategy would lead to increased COM accelerations in primarily the AP direction whereas the ML accelerations would be less affected by this strategy.

Associations between accelerometry and force plate measures were weak to moderate in the dynamic postural stability tasks. This result is not surprising as strength of the relationship between acceleration and criterion measure decreases with increasing movement intesity.[117] The acceleration data from these tasks were low pass filtered with a cutoff frequency of 50 Hz in the current study. Low pass filtering accelerometer data with lower cutoff frequencies (8-10 Hz) may aid in mitigating these differences.[117] The weak association between measures may also be a result of differences in the identified peak vertical acceleration or vertical GRF during landing leading to a slightly different window of data being analyzed.[84] Sell et al. showed that static balance tasks are not significantly correlated with dynamic tasks suggesting that the tasks require different control mechanisms.[79] This may explain why the measures were well correlated with the static but not dynamic tasks. With a lack of strong association between the measures, it is unknown if the accelerometry measures of dynamic postural stability will be as effective as force plate measures in discriminating minor balance deficits.

Mean power frequency was the least effective measure in differentiating task difficulty and showed weak associations between accelerometry and force plate measures of postural stability. Though there were significant main effects across tasks, there were few significant pairwise comparisons. This is not surprising due to the large amount of within subject variability identified in Specific Aim 1 (Chapter 3). Other researchers have found that mean frequency derived from COM accelerations has weak associations with force plate derived measures.[72] Other frequency domain measures such as centroid frequency have been shown to differentiate among task difficulty and show moderate associations between accelerometry and force plate measures of postural stability.[18]

The methodology utilized to address Specific Aim 2 has several limitations that must be acknowledged. During simple double-leg stance tasks with eyes open and eyes closed, differences in RMS have been reported to be on the order of 0.001 *g*.[81,86] The accelerometer utilized had a sensitivity of 0.003 *g* which may have affected the ability of the measures to differentiate between the DLEO and DLEC tasks. Although an accelerometer with a selectable range was utilized to account for the sensitivity and range requirements for both static and dynamic tasks, it may be more effective to employ two different accelerometers to achieve improved sensitivity in the static tasks. Additionally, other factors may influence correlation of the accelerometry measures including sensor placement and movement artifact. To limit error due to these factor, the same tester placed the sensor over L5, and accelerometer tilt was corrected for through a rotation transformation. A neoprene belt was positioned over the sensor to limit movement artifacts. Though the sensor was positioned over L5, true COM varies among individuals.[118] Future research should consider alternative algorithms for analyzing COM

accelerations during dynamic postural stability tasks that will be more closely associated with the gold-standard force plate measures.

4.5 CONCLUSION

Static and dynamic postural stability is important for the prevention and rehabilitation of musculoskeletal injuries and is important for sport performance optimization. In research settings, force plates are the gold-standard for objective postural stability assessment, however force plate assessments are not easily implemented in clinical settings. Thus, objective measures of postural stability are underutilized in clinical settings. Low-cost sensors offer an alternative objective assessment of postural stability, but must be reliable, valid compared to gold-standard measures, and discriminatory to be implemented in clinical settings. Specific Aim 1 established accelerometry measures of postural stability are reliable across tasks of varying difficulty. Specific Aim 2, addressed in this chapter, showed that accelerometry measures of static postural stability were correlated with force plate measures under certain conditions. The accelerometry measures were also able to discriminate differences in task difficulty. Dynamic postural stability assessments, such as jump-landing tasks, provide a greater challenge to the postural control system, but do not show a strong association to the force plate measures. This result does not necessarily indicate the measures will not be effective in differentiating individuals with balance deficits. Specific Aim 3 (Chapter 5) determines the ability of the accelerometry measures to discriminate between healthy controls and individuals that have suffered from previous musculoskeletal injury.

5.0 SPECIFIC AIM 3: DISCRIMINATIVE VALIDITY OF ACCELEROMETRY MEASURES OF POSTURAL STABILITY IN HEALTHY INDIVIDUALS AND INDIVIDUALS WITH CHRONIC ANKLE INSTABILITY

Waist worn, low-cost sensors offer objective assessments of postural stability that may be a cost-effective alternative to the gold-standard force plate measures. To be adopted in clinical settings, the sensor measures of postural stability must be reliable, valid compared to the gold standard and discriminative in the desired target population. Researchers have begun to explore the use of accelerometry-based measures to detect postural stability deficits associated with neurological disorders, such as Parkinson's disease,[72,73,75] vestibular disorders,[74] and most recently concussion.[2] However, postural stability deficits associated with musculoskeletal injury likely impact sensory organization differently than these disorders and injuries.[76–78] Thus, it is imperative that accelerometry measures of postural stability be validated in populations that have postural stability deficits associated with musculoskeletal injury. Specific Aim 1 established accelerometry measures are reliable in young, healthy individuals and individuals with chronic ankle instability (CAI). Specific Aim 2 demonstrated accelerometry measures of static postural stability are correlated with gold-standard force plate measures, while dynamic tasks showed weak to moderate relationships. Specific Aim 2 also demonstrated discriminative validity of the accelerometry measures in their ability to detect differences in task difficulty. Specific Aim 3 addresses a different kind of discriminative validity where accelerometry measures of static and

dynamic postural stability are examined to determine their ability to differentiate between healthy controls and individuals with CAI. Measures and assessments that are able to differentiate individuals with minor balance deficits associated with previous musculoskeletal injury will create an effective tool for monitoring postural stability for injury prevention and rehabilitation purposes.

5.1 INTRODUCTION

Lateral ankle sprains are the most common musculoskeletal injury among active populations and have a high prevalence in the general population.[20,24] In the United States, mean societal costs related to a single LAS incident are approximately $11,925.[28] Injury often results in lost work days, lost playing time, and lost leisure time. An estimated 30-75% of individuals suffering from lateral ankle sprain report long term chronic impairment, which greatly impacts an individual's quality of life.[30,31] Individuals that suffer from chronic impairment comprise 70-85% of individuals that develop post-traumatic osteoarthritis (PTOA) and are much more likely to seek surgical intervention.[32,33] Also, individuals with CAI often have deficits in strength,[119] proprioception,[120] dorsiflexion range of motion,[121,122] and postural stability[123] which may increase their risk for reinjury.

Several clinical postural stability assessments have been shown to discriminate between individuals with CAI and healthy individuals. Static clinical tests that have shown to identify individuals with CAI include the foot-lift test, the single-leg stance scored as part of the balance error scoring system (BESS), and timed single-leg stance with eyes closed.[124] Individuals with CAI have been shown to lift the foot more times during the foot-lift test, make more errors

during the BESS, and demonstrate shorter times on the single-leg stance with eyes closed compared to healthy individuals.[124,125] Functional clinical tests that have been shown to identify individuals with CAI include the star excursion balance test (SEBT) and side-hop test.[124] Individuals with CAI have shorter reach distances during the SEBT,[124] which may be due to limited dorsiflexion range of motion rather than a postural stability deficit.[126,127] During the side hop test, CAI individuals take longer to complete ten repetitions compared to healthy individuals. Though these clinical tasks have been shown to be effective in identifying individuals with CAI, they have limitations in interrater reliability and have a ceiling effect.[13,54] Due to these limitations, researchers have developed methods of assessing static and dynamic postural stability using a force plate.

Several force plate measures of postural stability have been examined during a static single-leg stance in individuals with CAI. In a meta-analysis center of pressure (COP) velocity was shown to be most sensitive to postural stability deficits in individuals with CAI.[128] COP velocity, time-to-boundary, and distance from the mean COP have also been shown to differentiate between healthy individuals and those with CAI during a single-leg stance.[60,67,70,124] These instrumented static assessments have greater reliability compared to clinical tests, but still may not have the sensitivity to detect sensorimotor deficits associated with balance.[67,123] An instrumented functional test may be more sensitive and specific to identifying individuals with CAI.

It is important to not only consider tasks that require individuals to maintain balance in static tasks, but also to consider performance of voluntary movement.[123] The challenge in developing these tasks arise in finding balance between attaining reliable measures and maintaining “life-like” situations. Depending on the type of activity, non-contact mechanisms of

injury have been reported to be more common than contact mechanisms of injury.[21] For example, landing has been identified as the most common mechanism of LASs in basketball players.[29] Researchers have developed a single leg jump-landing task that requires individuals to jump a set distance, land on a single leg, and stabilize their center of mass over their base of support. Both forward and lateral jumping protocols have been utilized. The task provides a greater challenge compared to static tasks, and the inclusion of a jump-landing may be more ecologically valid for active populations.

A few different force plate measures have been proposed to quantify dynamic postural stability during the jump-landing maneuvers including time to stabilization (TTS) and the dynamic postural stability index (DPSI).[97,129] While TTS has been shown to differentiate individuals with ankle instability,[66,69] it is highly influenced by skill level.[130] The DPSI during a forward jump-landing task has been shown to identify individuals with both perceived instability and mechanical laxity.[61] Brown et al., showed that both the forward and lateral jump-landing tasks identified differences in individuals with CAI and those without.[111] Other studies have also shown the DPSI during a forward jump-landing task to be different between individuals with and without CAI.[112,131] While this dynamic assessment of postural stability appears to be effective in differentiating individuals with CAI, the need for force plate technology limits adoption of the assessment into clinical settings. Limited work has been done to develop new methods of quantifying dynamic postural stability during a jump-landing task using a waist-worn accelerometer.

Accelerometry measures of postural stability have been validated many populations showing they are sensitive to balance deficits in individuals with Parkinson's disease,[72,73,75] vestibular disorders,[74] older adults,[132] concussion,[2] and are sensitive to muscular fatigue.[115] In a

recent study, Chiu et al. established smart phone based accelerometry measures of postural stability that differentiate between individuals with CAI and healthy controls.[133] In the study by Chiu et al., the smart phone was fixed to the middle of the shin and participants completed a single leg stance with eyes open and eyes closed, and the extracted measure was an average of the acceleration time series.[133] This previously published work demonstrates the feasibility in utilizing accelerometry measures of postural stability to identify individuals with CAI. However, no studies have examined which accelerometry measures are most sensitive to balance deficits in a population with CAI and the ability of accelerometry measures to differentiate between individuals with and without CAI during dynamic postural stability tasks is unknown. Therefore, the purpose of Specific Aim 3 was to identify static and dynamic postural stability measures derived from COM accelerations that discriminate between healthy controls and individuals with CAI.

5.2 MATERIALS AND METHODS

5.2.1 Participants

A total of 50 participants, 25 healthy controls and 25 individuals with CAI, were recruited and enrolled to assess the ability of accelerometry-based measures to identify individuals with CAI. An equal proportion of men and women were recruited for each group. Participant recruitment and screening procedures are detailed in Appendix A and described briefly below. Participant demographics for Specific Aim 3 are described in Table 6. All participants engaged in physical activity for at least 30 minutes at a given time, three days per week. Participants were excluded if

they self-reported history of fracture or surgery to the lower extremity, head injury within three months prior to test date, low back pain, or other known disorder (vestibular, neurological, or orthopedic) that could affect postural stability. Women were excluded if they were knowingly pregnant.

Table 6. Participant demographics for Specific Aim 3

Group	Gender	Height (cm)	Weight (kg)	BMI	Age (years)	CAIT	Ankle Sprains	Time since last sprain (months)
Control	13 M; 12 F	173.1 ± 9.9	67.8 ± 10.5	22.6 ± 2.4	22.6 ± 3.0	29.4 ± 0.7	0.0 ± 0.0	-
CAI	13 M; 12 F	175.5 ± 8.9	73.5 ± 9.6	23.8 ± 2.3	22.2 ± 4.7	19.6 ± 3.8	4.8 ± 3.5	20.0 ± 20.6
p value	-	0.61	0.36	0.70	0.15	0.00*	-	-

*Significant p value

Participants meeting inclusion and exclusion criteria completed three additional screening procedures for group assignment: self-reported ankle sprain history, Cumberland Ankle Instability Tool (CAIT) questionnaire, and talar tilt test (Appendix A, Figure 20). A certified Athletic Trainer (ATC) completed the talar tilt test. For this study, lateral ankle sprain (LAS) was defined as injury to the lateral ligaments of the ankle caused by rolling over on or "twisting" the ankle that resulted in disruption of normal physical activity for at least three days.[43] Participants were included in the control group if they had no prior history of self-reported LAS, scored ≥ 28 on the CAIT questionnaire, and showed no mechanical lateral laxity as measured by the talar tilt test. Participants were included in the CAI group if they self-reported a first incident LAS greater than one year prior to test date, had no subsequent LAS within three months prior to test date, scored ≤ 24 on the CAIT questionnaire, and had a positive sign of mechanical laxity as measured by the talar tilt test.[100]

5.2.2 Study design

A cross-sectional cohort study design was utilized to establish the ability of accelerometry-based measures of postural stability to discriminate between individuals with CAI and healthy controls. All testing for this specific aim was completed during one testing session. Participants were asked to refrain from drinking caffeine and alcohol 24 hours prior to each testing session.

5.2.3 Experimental protocol

All participants completed four tasks of postural stability including a single leg stance with eyes open (SLEO) and eyes closed (SLEC) and forward and lateral jump-landing tasks (DPS-AP and DPS-ML, respectively). Methods were similar to those described in Section 3.2.3. During the testing session, participants were asked to complete the static and dynamic postural stability tasks with a triaxial accelerometer (YEI 3-Space Sensor, YOST Labs, Portsmouth, OH) positioned over L5. A neoprene belt was positioned over the sensor and secured around the participant's waist to limit vibration of the sensor during motion (Section 2.2.1, Figure 2). All tasks were performed on a constrained area (60 cm x 40 cm) located 2.5 m from a wall. COM accelerations were sampled at approximately 1200 Hz.

A five-second static capture was performed where accelerations were recorded as the participants stood upright with their back against a wall. Static tasks were performed barefoot and participants performed one familiarization trial of each task. Participants were asked to complete four additional successful trials of each of the static postural stability tasks. Dynamic tasks were performed in the participant's own athletic shoes and participants performed one successful familiarization trial of each dynamic task. Participants were given unlimited attempts

to successfully complete eleven additional trials of each dynamic task. Participants took a minimum of 30 seconds rest between trials of a task and two minutes rest between each task to minimize fatigue. The order of tasks was randomized for each participant using Latin square design. All trials of a given task were completed before moving onto the next task.

Static single-leg stance tasks lasted for a duration of ten seconds. Participants completed this task on their dominant limb (Control) or involved limb (CAI). During the SL tasks, the participant was permitted to touch down on the floor with the non-test leg to maintain stability through the duration of the test.

The anterior-posterior (DPS-AP) and medial-lateral (DPS-ML) dynamic postural stability tasks were initiated from a distance equal to 40% and 33% of the participant's height, respectively.[79] Participants were asked to initiate each jump from two limbs, clear a 30.5 cm (DPS-AP) or 15.2 cm (DPS-ML) hurdle, and land on their dominant limb (Control) or involved limb (CAI) on the force plate.

5.2.4 Data reduction

The acceleration time series was resampled from approximately 1200 Hz to 1000 Hz and filtered using a low-pass Butterworth filter with a cutoff frequency of 20 Hz for static tasks and 50 Hz for dynamic tasks. The low pass filter was selected based on a power spectral density analysis described in Section 2.2.2. Ground reaction forces were also filtered using a low-pass Butterworth filter with cutoff frequencies of 20 Hz and 50 Hz for static and dynamic tasks, respectively.

An alignment procedure was performed to correct for misplacement of the sensor along the vertical and transverse axes.[89] The mean accelerations in the x, y, and z directions were

calculated during the five-second static capture. The static capture position was utilized to determine the sensors orientation relative to gravity. Then, a quaternion rotation transformation was applied to the filtered data.[89] Twenty seconds of data were analyzed for the double-leg tasks and ten seconds of data were analyzed for the single-leg tasks. For the jump-landing tasks, a three second window during landing was used for analysis. The analysis window was set to begin where peak vertical acceleration was identified.

Root mean square (RMS), normalized path length (NPL), peak to peak (P2P), stability indices (SI), and mean power frequency (MPF) were extracted from the transformed COM acceleration data and the ground reaction forces.[16,97] Each variable was calculated along the anterior-posterior (AP) and medial-lateral (ML) axes using equations 6 – 9 and methods described in Section 3.2.4.

5.2.5 Statistical analysis

Statistical analyses were performed using SPSS software. Descriptive statistics including means and standard deviations as well as medians and interquartile ranges were calculated where appropriate. The ability of each measure of postural stability across the 4 tasks to accurately identify individuals with CAI was estimated using a Receiver Operating Characteristic (ROC) curve. The area under the ROC curve (AUC) was tested against 0.50. An AUC of 1 indicates perfect discrimination and accuracy of the test and an AUC of 0.5 indicates random guessing. An alpha level of 0.05 was set a priori.

5.3 RESULTS

Means and standard deviations (SD) as well as medians and interquartile ranges (IQR) of each COM acceleration measure were calculated for the single-leg stance tasks (Table 7) and jump-landing tasks (Table 8) within the control and CAI groups. ROC curve analyses were performed to identify measures of postural stability that accurately classify individuals with CAI (Appendix D, Figure 25 and Figure 26). RMS, NPL, and P2P in both the AP and ML directions had significant AUC values for the single-leg stance task with eyes open ($p = 0.002 – 0.047$). Only NPLap ($p = 0.026$) and P2Pap ($p = 0.039$) had significant AUC values for the single-leg stance task with eyes closed. No significant AUC values were found for COM acceleration measures from the jump-landing tasks. AUC values closer to 1 indicate greater ability of the assessment to correctly classify individuals with and without CAI.

Table 7. ROC curve analysis for single-leg stance postural stability tasks

	Control				CAI				ROC Curve	
	Mean	SD	Median	IQR	Mean	SD	Median	IQR	AUC	p value
SLEO										
RMSap (mg)	12.14	3.55	11.34	5.36	17.49	8.33	15.49	7.52	0.74	0.003*
RMSml (mg)	11.79	2.98	11.64	5.82	17.76	9.70	14.33	11.25	0.68	0.026*
NPLap (mg/s)	275.7	36.3	271.2	55.6	296.3	37.7	293.2	53.4	0.66	0.047*
NPLml (mg/s)	269.0	47.8	259.6	74.5	329.4	91.4	297.3	132.2	0.71	0.011*
P2Pap (mg)	70.6	22.3	63.3	29.2	96.8	41.3	94.6	33.1	0.74	0.003*
P2Pml (mg)	70.1	18.9	68.1	26.3	110.1	54.6	96.2	62.5	0.76	0.002*
APSI (mg/kg)	0.02	0.008	0.015	0.011	0.022	0.010	0.019	0.148	0.66	0.056
MLSI (mg/kg)	0.02	0.006	0.016	0.009	0.021	0.011	0.017	0.014	0.59	0.273
MPFap (Hz)	1.94	0.82	1.97	1.12	1.60	0.71	1.48	0.91	0.37	0.123
MPFml (Hz)	2.16	0.89	2.36	1.24	2.26	1.09	2.30	1.57	0.51	0.915
SLEC										
RMSap (mg)	27.92	12.02	25.3	14.54	31.29	12.4	28.91	16.79	0.60	0.211
RMSml (mg)	35.12	14.31	39.98	20.06	46.69	25.83	41.85	25.69	0.65	0.076
NPLap (mg/s)	437.1	138.8	406.4	160.9	518.6	162.5	493.7	224.1	0.68	0.026*
NPLml (mg/s)	599.0	237.9	524.0	241.0	705.5	236.8	674.5	373.8	0.63	0.105
P2Pap (mg)	165.6	70.1	157.7	99.8	203.9	66.4	202.3	109.5	0.67	0.039*
P2Pml (mg)	237.9	101.8	232.9	151.6	314.2	143.4	256.1	212.9	0.65	0.073
APSI (mg/kg)	0.039	0.019	0.036	0.018	0.039	0.014	0.035	0.021	0.54	0.594
MLSI (mg/kg)	0.047	0.021	0.043	0.037	0.055	0.026	0.045	0.034	0.58	0.357
MPFap (Hz)	1.71	0.75	1.62	1.22	1.82	0.81	1.74	1.09	0.53	0.677
MPFml (Hz)	2.31	1.04	2.3	1.57	2.04	1.01	1.98	1.86	0.43	0.377

*Significant AUC ($p < 0.05$)

Table 8. ROC curve analysis for dynamic postural stability tasks

	Control				CAI				ROC Curve	
	Mean	SD	Median	IQR	Mean	SD	Median	IQR	AUC	p value
DPS-AP										
RMSap (mg)	252.2	80.4	235.4	142.2	279.8	110.8	256.7	148.0	0.56	0.467
RMSml (mg)	387.5	110.3	385.8	159.5	361.1	118.9	347.7	182.8	0.43	0.399
NPLap (mg/s)	7045.4	2167.8	6623.6	3335.0	7746.8	3150.1	7580.1	4220.7	0.54	0.594
NPLml (mg/s)	6561.0	1556.1	6530.3	2159.3	6596.4	1781.3	6557.8	2255.8	0.51	0.946
P2Pap (mg)	5667.3	2159.5	5356.5	4129.0	6337.5	3080.6	5870.8	4291.6	0.55	0.528
P2Pml (mg)	7446.4	2584.9	7174.2	4148.2	6996.9	2609.7	6593.1	3195.4	0.44	0.491
APSI (mg/kg)	0.192	0.046	0.193	0.061	0.181	0.059	0.177	0.080	0.43	0.388
MLSI (mg/kg)	0.253	0.075	0.236	0.108	0.224	0.072	0.204	0.124	0.40	0.204
MPFap (Hz)	17.72	7.04	17.98	9.73	21.73	11.58	23.41	15.65	0.64	0.101
MPFml (Hz)	17.40	6.22	16.46	7.67	16.22	4.70	17.30	6.45	0.48	0.839
DPS-ML										
RMSap (mg)	203.8	63.4	206.1	70.3	203.8	63.4	206.1	63.6	0.52	0.839
RMSml (mg)	293.5	78.7	276.1	102.3	293.5	78.7	276.1	95.5	0.60	0.240
NPLap (mg/s)	5653.4	1417.7	5599.4	2415.3	5653.4	1417.7	5599.4	2690.3	0.53	0.705
NPLml (mg/s)	5520.2	1247.6	5386.5	1874.5	5520.2	1247.6	5386.5	1568.5	0.59	0.273
P2Pap (mg)	3935.7	1386.1	4049.5	1884.1	3935.7	1386.1	4049.5	2490.9	0.59	0.290
P2Pml (mg)	5257.5	2093.1	5063.4	2554.0	5257.5	2093.1	5063.4	2491.6	0.58	0.318
APSI (mg/kg)	0.166	0.040	0.171	0.063	0.166	0.040	0.171	0.058	0.66	0.059
MLSI (mg/kg)	0.210	0.059	0.204	0.073	0.210	0.059	0.204	0.095	0.60	0.211
MPFap (Hz)	12.57	6.07	11.75	11.90	12.57	6.07	11.75	12.40	0.50	0.977
MPFml (Hz)	15.85	8.26	14.88	14.29	15.85	8.26	14.88	8.38	0.61	0.190

5.4 DISCUSSION

Accelerometry measures of postural stability were determined reliable in Specific Aim 1 and discriminative among task difficulty in Specific Aim 2. Accelerometry measures of static postural stability were shown to correlate with force plate measures, however this relationship was not observed in measures of dynamic postural stability. The purpose of Specific Aim 3 was to establish another type of discriminative validity by determining the ability of accelerometry measures to correctly identify individuals that have suffered a previous musculoskeletal injury. Assessment methods that are reliable and sensitive enough to detect balance deficits in this population are needed to accurately identify individuals that are at greater risk for future injury and will benefit from balance training interventions. The results suggest several measures of static postural stability during a single leg stance are sensitive to postural stability deficits. However, the methods utilized in this study to quantify dynamic postural stability during a single-leg jump landing task were not effective in differentiating between control and CAI groups.

Several COM acceleration measures from the SLEO task accurately identified individuals with CAI including RMS, NPL, and P2P in the AP and ML directions. Chiu et al. also found differences in accelerometry measures of static postural control in individuals with CAI during single leg stance tasks in both eyes open and eyes closed conditions.[133] The measure utilized by Chiu et al. was an average of the recorded accelerations of the shin and did not differentiate AP

and ML directions. The current study demonstrates COM acceleration measures are also sensitive to postural stability deficits in CAI populations.

Surprisingly, there were more measures that accurately identified individuals with CAI during the SLEO task than the SLEC task. During the SLEC task only NPLap and P2Pap accurately identified individuals with CAI. Individuals with CAI often perform worse in eyes closed conditions as they rely more heavily on visual cues for balance due to diminished somatosensory feedback.[134] This theory is supported by findings from a study by Ross et al. that demonstrated ML force plate measures during static stance to have the greatest discriminatory ability.[135] In the current study, greater sensitivity in the AP accelerometry measures may be explained by the CAI group adopting more of a hip dominant strategy during the task which tends to increase sway in the AP rather than ML direction.

The dynamic postural stability tasks were included in the study as previous research suggests that static test may not be sensitive enough to detect differences between healthy controls and individuals with CAI and suggests that functional tests have better discriminatory capability.[67,123] Contrarily, COM acceleration measures during the single-leg jump landing task were not able to identify individuals with CAI. It was unexpected to find the static postural stability measures to be more effective at identifying individuals with CAI than the dynamic postural stability measures, particularly given the prior research that has established force plate measures of a similar task to be effective in differentiating between CAI and healthy or coper populations.[61,111,112,131] However, similar to the results presented in the current study, other researchers have also found static tests to be just as effective or more effective at identifying individuals with CAI.[124,128,135,136] Although the dynamic postural stability measures utilized in this study were not effective in differentiating between groups, there may be other data

signatures or data processing methods that would increase the discriminatory ability of the assessment.

There are several possible explanations for the dynamic postural stability accelerometry measures not being able to identify individuals with CAI. The COM accelerations may be capturing a different mechanism of postural stability that is not effective in differentiating between groups. Specific Aims 1 and 2 (Chapters 3 and 4, respectively), identified reliable accelerometry measures of dynamic postural stability, however the accelerometry measures had weak to moderate associations to the force plate derived measures. Another possible explanation for the lack of findings is that only the AP and ML directions were considered during analysis. The vertical component of the signal may have important data signatures that better differentiate between groups as CAI populations have been shown to have higher impact peak forces and increased loading rates during running compared to healthy controls.[137] Finally, accelerations recorded at the COM may wash out any differences in altered movement strategies that may contribute to postural stability and risk of future injury as instability at the ankle could be compensated for at the knee or hip. It is known that individuals with CAI display altered movement strategies compared to copers and healthy controls.[42,138–140] Individuals tend to land with less plantarflexion and inversion, more knee and hip flexion, and less hip abduction.[42,138] It is possible that these differences are not reflected in the individual's control at their COM and that other lower extremity compensations aid in stabilizing the COM. An accelerometer placed on tibia may eliminate some variability and better isolate any differences seen at the ankle.

The lack in significant findings related to dynamic postural stability, though unlikely, may also be related to the jump-landing task or the CAI population. Though the jump-landing task utilized has limitations, performance on similar dynamic tasks has been suggested to be

related to ankle instability as the ankle stabilizers are required to restrain excessive joint motion during the landing period.[124,141] The task also requires plantarflexion which forces the foot and ankle into an unstable position and requires dorsiflexion flexion to absorb landing forces which is restricted in individuals with CAI.[127] The DPS-ML task in particular was thought to provide a greater stress to the lateral ankle stabilizers mimicking the injury mechanism for LAS.[141] Additionally, it is possible that the CAI sample population did not have the same level of impairment as other CAI populations that have shown dynamic postural stability deficits. However, in a study by Brown et al., individuals that had both perceived instability and mechanical laxity had a CAIT score of 19.4 ± 5.1, self-reported history of 3.4 ± 2.2 ankle sprains, and reported on average 24.4 ± 22.6 months since last ankle sprain incident.[61] The population recruited for this study was very similar with an average CAIT score of 19.6 ± 3.8, self-reported history of 4.8 ± 3.5 ankle sprains, and reported on average 20.0 ± 20.6 months since last ankle sprain incident. Therefore, we think it is unlikely that task or population restricted our findings.

The analyses addressing Specific Aim 3 have several limitations to acknowledge. Postural stability decreases with increasing BMI.[142] Though BMI was not statistically different between groups, the BMI in the CAI group was 5% greater than the control group. It is possible that this contributed to the observed static balance deficits, however this is representative of the population. Individuals with CAI tend to have a higher BMI and report lower levels of physical activity compared to age-matched controls with no history of injury.[143,144] Also, postural stability is associated with athletic ability.[106] Individuals included in the study were recreationally active for at least 30 minutes three days per week, but there may have been differences in athletic

ability between the groups. Future studies should include stricter requirements for matching athletic ability between groups.

5.5 CONCLUSION

Postural stability is important for the prevention and rehabilitation of musculoskeletal injuries. Specific Aims 1 and 2 established the reliability, concurrent validity and discriminative validity of several accelerometry measures of static and dynamic postural stability. Specific Aim 3 established another type of discriminative validity indicating measures during a static single-leg stance are sensitive to balance deficits in individuals with CAI, with the single-leg stance with eyes open showing the greatest discriminatory ability. Dynamic measures of postural stability were unable to differentiate between control and CAI groups. The portable, objective measures of static postural stability provide a low-cost method for assessing postural stability in clinical settings for prevention and rehabilitation of musculoskeletal injuries.

6.0 CONCLUSIONS AND FUTURE WORK

Postural stability is important for injury prevention, rehabilitation, and performance optimization. Several laboratory and clinical assessments of postural stability exist. However, clinical tests lack interrater reliability and have a ceiling effect. Laboratory tests have greater reliability and sensitivity, but utilize force plate technology which is expensive, cumbersome and not easily implemented in clinical settings. Low-cost accelerometers have been shown to be an effective tool for quantifying postural stability during static tasks. However, little work has been done to establish accelerometry measures that are reliable, valid compared to the gold-standard measure, and are able to detect differences in postural stability in a population with CAI. Also, little work has been done to establish methods for assessing dynamic postural stability using a waist worn accelerometer. The purpose of this dissertation was to establish the reliability, gold-standard criterion validity, and discriminatory ability of accelerometry-based measures of postural stability across ten tasks of varying difficulty.

Some accelerometry measures and tasks were found to be have greater intersession reliability, gold-standard criterion validity, and discriminatory ability than others. Normalized path length (NPL) and peak-to-peak (P2P) values demonstrated the greatest reliability, gold-standard criterion validity, and discriminatory ability across tasks compared to the root mean square (RMS), stability index (SI) and mean frequency (MPF). The simpler tasks, such as a double leg stance with eyes open or eyes closed, have high variability between sessions.

Therefore, when assessing young, active individuals, it is suggested to implement more challenging tasks. The results from this work also suggest averaging at least three static postural stability trials and at least six dynamic postural stability trials to minimize within subject variability. Accelerometry measures, particularly NPL and P2P, provide a valid assessment of static postural stability tasks. The work presented establishes a continuum of postural stability tasks of increasing difficulty. Dynamic postural stability assessments, such as jump-landing tasks, provide a greater challenge to the postural control system, but do not show a strong association to the force plate measures. NPL and P2P in the AP direction are sensitive to balance deficits in individuals with CAI during static single-leg stance tasks with eyes open and eyes closed. Based on these findings, NPL and P2P are suitable measures of static postural stability and can be implemented in a clinical setting when assessing postural stability in individual with CAI. Sensor specifications must be taken into consideration if implementing any of these assessments in a clinical setting with a generic inertial sensor and any onboard processing methods should be carefully evaluated. Integration of the inertial sensor with smart phone technology will provide clinicians a low-cost, objective solution for postural stability assessment.

The accelerometry measures of dynamic postural stability utilized in this study, while reliable, were not valid or sensitive to balance deficits. Future work should consider other data processing methods that may better differentiate between healthy and CAI populations. It may also be of value to collect accelerations of the lower limb during the dynamic postural stability jump-landing tasks. Future research should establish reliability and validity of similar COM acceleration measures of postural stability in populations that have suffered from other lower extremity musculoskeletal injuries, such as an anterior-cruciate ligament (ACL) tear. Other metrics that can be quantified using the inertial sensor such as flexibility, range of motion,

biomechanics and agility should also be considered as musculoskeletal injuries are often multifactorial in nature.

APPENDIX A

STUDY PARTICIPANTS

Power analyses were performed using G*Power (v.3.1) and Pass 14 (v.14.0.6; NCSS Statistical Software; Kaysville, UT) to determine the sample sizes necessary to adequately power each Specific Aim. The power analysis indicated a total of 24 participants would be required for Specific Aim 2 to achieve 80% power and to detect an interclass correlation of 0.9 under the alternative hypothesis when the null hypothesis is set to 0.7. A total of 42 individuals, 21 healthy controls and 21 individuals with chronic ankle instability (CAI), would be required to achieve 81% power to detect a difference of 0.2 between the area under the ROC curve (AUC) with a null hypothesis of 0.7 and an alternative hypothesis of 0.9.

A total of 50 participants, 25 healthy controls and 25 individuals with CAI, were recruited and enrolled in this dissertation study. An equal proportion of men and women (13 men: 12 women) were recruited for each group. All participants engaged in physical activity for at least 30 minutes at a given time, three days per week. Participants were excluded if they self-reported history of fracture or surgery to the lower extremity, head injury within three months prior to test date, low back pain, or other known disorder (vestibular, neurological, or orthopedic) that could affect postural stability. Women were excluded if they were knowingly pregnant.

Participants meeting these criteria completed three additional screening procedures for group assignment: self-reported ankle sprain history, Cumberland Ankle Instability Tool (CAIT) questionnaire, and talar tilt test (Figure 20). Participants were included in the control group if they had no prior history of self-reported LAS, scored ≥ 28 on the CAIT questionnaire, and showed no mechanical lateral laxity as measured by the talar tilt test. Participants were included in the CAI group if they self-reported a first incident LAS greater than one year prior to test date, had no subsequent LAS within three months prior to test date, scored ≤ 24 on the CAIT questionnaire, and had a positive sign of mechanical laxity as measured by the talar tilt test. Screening criteria are detailed below.

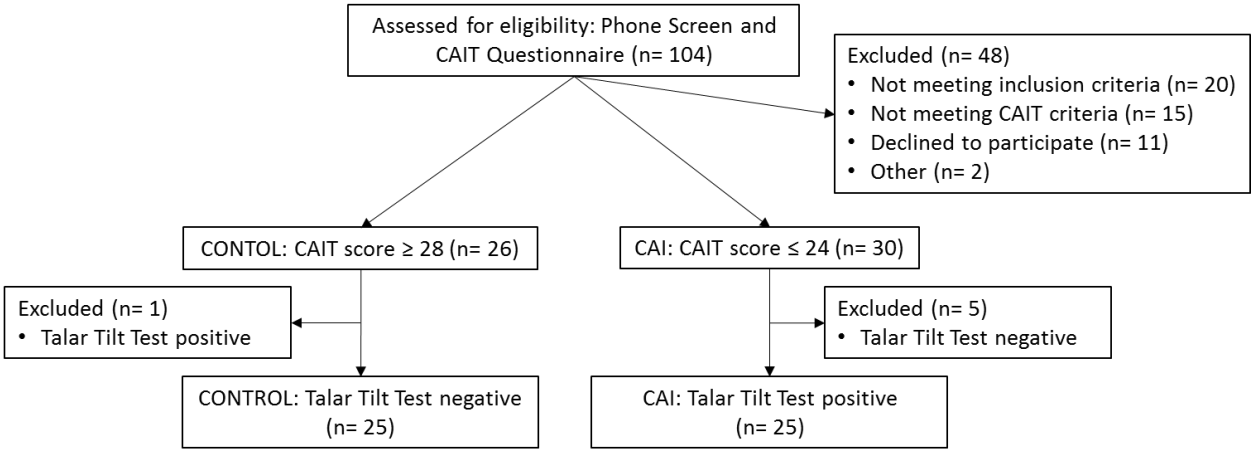

Figure 20. Study enrollment consort diagram.

A.1 SELF REPORTED ANKLE SPRAIN HISTORY

Lateral ankle sprain (LAS) was defined as injury to the lateral ligaments of the ankle caused by rolling over on or "twisting" the ankle that resulted in disruption of normal physical activity for at least three days. Participants must have reported a first incident LAS greater than one year prior to test date. Participants were excluded if they had a subsequent sprain within three months prior to test date.[43]

A.2 CUMBERLAND ANKLE INSTABILITY TOOL

Participants completed the CAIT questionnaire as part of a phone screening procedure (Table 9). To be included in the CAI group, participants must have scored ≤ 24 on the CAIT on their previously injured limb, and control participants must have scored ≥ 28 on their dominant limb. The CAIT has been shown to have a sensitivity of 0.83 and specificity of 0.74 with excellent test-retest reliability (ICC = 0.96).[100]

Table 9. The Cumberland Ankle Instability Tool

Please tick the ONE statement in EACH question that BEST describes your ankles.

		LEFT	RIGHT	Score
1.	I have pain in my ankle			
	Never	☐	☐	5
	During sport	☐	☐	4
	Running on uneven surfaces	☐	☐	3
	Running on level surface	☐	☐	2
	Walking on uneven surfaces	☐	☐	1
	Walking on level surfaces	☐	☐	0
2.	My ankle feels UNSTABLE			
	Never	☐	☐	4
	Sometimes during sport (not every time)	☐	☐	3
	Frequently during sport (every time)	☐	☐	2
	Sometimes during daily activity	☐	☐	1
	Frequently during daily activity	☐	☐	0
3.	When I make SHARP turns, my ankle feels UNSTABLE			
	Never	☐	☐	3
	Sometimes when running	☐	☐	2
	Often when running	☐	☐	1
	When walking	☐	☐	0
4.	When going down the stairs, my ankle feels UNSTABLE			
	Never	☐	☐	3
	If I go fast	☐	☐	2
	Occasionally	☐	☐	1
	Always	☐	☐	0
5.	My ankle feels UNSTABLE when standing on ONE leg			
	Never	☐	☐	2
	On the ball of my foot	☐	☐	1
	With my foot flat	☐	☐	0
6.	My ankle feels UNSTABLE when			
	Never	☐	☐	3
	I hop from side to side	☐	☐	2
	I hop on the spot	☐	☐	1
	When I jump	☐	☐	0
7.	My ankle feels UNSTABLE when			
	Never	☐	☐	4
	I run on uneven surfaces	☐	☐	3
	I jog on uneven surfaces	☐	☐	2
	I walk on uneven surfaces	☐	☐	1
	I walk on a flat surface	☐	☐	0
8.	TYPICALLY, when I start to roll over (or "twist") on my ankle, I can stop it			
	Immediately	☐	☐	3
	Often	☐	☐	2
	Sometimes	☐	☐	1
	Never	☐	☐	0
	I have never rolled over on my ankle	☐	☐	3
9.	After a TYPICAL incident of my ankle rolling over, my ankle returns to "normal"			
	Almost immediately	☐	☐	3
	Less than one day	☐	☐	2
	1-2 days	☐	☐	1
	More than 2 days	☐	☐	0
	I have never rolled over on my ankle	☐	☐	3

*Note that the score column will not be present on the subject's copy

A.3 TALAR TILT TEST

The talar tilt test was completed by a certified Athletic Trainer (ATC). Both instrumented and manual talar tilt tests have been shown to have low sensitivity (0.36 and 0.49, respectively), but good to excellent specificity (0.72-0.94 and 0.78-0.88, respectively).[145] While not sufficient as a stand-alone screening for individuals with CAI, the talar tilt test is useful in ruling in the condition. Based on a study by Rosen and Brown, there is little benefit to using the instrumented talar tilt test over the manual talar tilt test.[145]

APPENDIX B

RELIABILITY TABLES

The results from the systematic bias, within subject variability, and intersession reliability presented here are explained in detail in Chapter 3. Reliability measures were assessed across ten postural stability tasks including eight static tasks: double leg stance on a firm surface (DL), double leg stance on a foam surface (DL-F), tandem stance (TAN), single leg stance (SL) performed with eyes open (EO) and eyes closed (EC), and two dynamic tasks: forward jump-landing (DPS-AP) and lateral jump-landing (DPS-ML). Five measures were calculated from the center of mass (COM) accelerations in the anterior-posterior (AP) and medial-lateral (ML) directions including root mean square (RMS), normalized path length (NPL), peak to peak (P2P), stability index (SI), and mean power frequency (MPF).

B.1 SYSTEMATIC BIAS

Systematic bias was assessed across trials within session 1 and across three repeated sessions using a repeated measures ANOVA when data were normally distributed and Friedman's test

when data were not normally distributed. The p values from these assessments are presented in Table 10. The results are described in detail in Section 3.3.1.

Table 10. Systematic bias across postural stability tasks for control and CAI groups

		RMSap	RMSml	NPLap	NPLml	P2Pap	P2Pml	APSI	MLSI	MPFap	MPFml
		p values									
DLEO											
Control	Trial (n = 5)	0.26	0.25	0.16	0.22	0.44	0.68	0.37	0.59	0.33	0.74
	Session (n = 3)	0.01*	0.13	0.88	0.40	0.05*	0.30	0.05*	0.27	0.06	0.24
CAI	Trial (n = 5)	0.21	0.75	0.68	0.78	0.41	0.81	0.30	0.87	0.39	0.66
	Session (n = 3)	0.74	0.50	1.00	0.67	0.74	0.41	0.72	0.46	0.19	0.76
DLEC											
Control	Trial (n = 5)	0.58	0.10	0.85	0.72	0.77	0.05	0.55	0.87	0.11	0.83
	Session (n = 3)	0.12	0.91	0.43	0.08	0.34	0.31	0.50	0.31	0.44	0.50
CAI	Trial (n = 5)	0.05*	0.71	0.04*	0.17	0.04*	0.34	0.02*	0.32	0.09	0.17
	Session (n = 3)	0.33	0.41	0.67	0.50	0.42	0.50	0.32	0.12	0.50	0.11
DLEOF											
Control	Trial (n = 5)	0.69	0.34	0.20	0.92	0.89	0.67	0.33	0.88	0.46	0.98
	Session (n = 3)	0.99	0.72	0.91	0.06	0.89	0.74	0.27	0.20	0.49	0.13
CAI	Trial (n = 5)	0.49	0.04*	0.24	0.31	0.37	0.53	0.71	0.26	0.68	0.39
	Session (n = 3)	0.76	0.31	0.41	0.12	0.67	0.74	0.95	0.45	0.50	0.02*
DLECF											
Control	Trial (n = 5)	0.64	0.62	0.19	0.10	0.72	0.33	0.53	0.26	0.72	0.52
	Session (n = 3)	0.10	0.41	0.91	0.50	0.08	0.61	0.38	0.91	0.19	0.50
CAI	Trial (n = 5)	0.41	0.26	0.03*	0.16	0.27	0.62	0.14	0.91	0.57	0.93
	Session (n = 3)	0.17	0.50	0.74	0.34	0.15	0.91	0.06	0.50	0.14	0.35
TANEO											
Control	Trial (n = 5)	0.21	0.11	0.23	0.58	0.22	0.06	0.73	0.17	0.90	0.11
	Session (n = 3)	0.32	0.74	0.27	0.91	0.30	1.00	0.15	0.04*	0.20	0.01*
CAI	Trial (n = 5)	0.98	0.92	0.65	0.46	0.85	0.39	0.81	0.98	0.53	0.69
	Session (n = 3)	0.08	0.32	0.69	0.15	0.27	0.50	0.27	0.41	0.04*	0.46

*Significant main effect ($p < 0.05$)

Table 10. Continued

		RMSap	RMSml	NPLap	NPLml	P2Pap	P2Pml	APSI	MLSI	MPFap	MPFml
		p values									
TANEC											
Control	Trial (n = 5)	0.17	0.14	0.17	0.28	0.06	0.08	0.45	0.11	0.73	0.92
	Session (n = 3)	0.44	0.48	0.50	0.50	0.67	0.57	0.67	0.79	0.53	0.18
CAI	Trial (n = 5)	0.94	0.26	0.99	0.88	0.90	0.98	0.90	0.50	0.92	0.20
	Session (n = 3)	0.66	0.25	0.50	0.50	0.99	0.47	0.46	0.50	0.91	0.43
SLEO											
Control	Trial (n = 5)	0.42	0.13	0.67	0.34	0.43	0.28	0.65	0.33	0.34	0.28
	Session (n = 3)	0.29	0.68	0.27	0.15	0.74	0.58	0.50	0.85	0.16	0.99
CAI	Trial (n = 5)	0.23	0.99	0.95	0.79	0.25	0.87	0.28	0.26	0.38	0.10
	Session (n = 3)	0.62	0.97	0.90	0.74	0.50	0.64	0.43	0.07	0.85	0.86
SLEC											
Control	Trial (n = 5)	0.07	0.40	0.10	0.34	0.21	0.26	0.65	0.40	0.71	0.23
	Session (n = 3)	0.12	0.30	0.12	0.30	0.12	0.15	0.27	0.06	0.27	0.29
CAI	Trial (n = 5)	0.52	0.64	0.28	0.27	0.29	0.50	0.99	0.66	0.56	0.19
	Session (n = 3)	0.65	0.90	0.31	0.77	0.63	0.76	0.56	0.77	0.58	0.76
DPS-AP											
Control	Trial (n = 12)	0.80	0.48	0.86	0.59	0.77	0.64	0.51	0.61	0.81	0.45
	Session (n = 3)	0.30	0.24	0.74	0.40	0.67	0.49	0.35	0.18	0.84	0.05*
CAI	Trial (n = 12)	0.68	0.47	0.94	0.63	0.88	0.46	0.78	0.19	0.33	0.35
	Session (n = 3)	0.00*	0.20	0.37	0.90	0.34	0.07	0.47	0.77	0.80	0.19
DPS-ML											
Control	Trial (n = 12)	0.34	0.54	0.66	0.68	0.49	0.44	0.30	0.77	0.64	0.39
	Session (n = 3)	0.67	0.33	0.50	0.59	1.00	0.78	0.89	0.50	0.91	0.01*
CAI	Trial (n = 12)	0.58	0.74	0.70	0.67	0.75	0.94	0.39	0.85	0.48	0.65
	Session (n = 3)	0.01*	0.93	0.09	0.45	0.35	0.74	0.12	0.43	0.92	0.92

*Significant main effect ($p < 0.05$)

B.2 WITHIN SUBJECT VARIABILITY

Typical errors (TE_n) for *n* averaged trials and coefficients of variation (%CV) for each postural stability task and measure are presented in Table 11 through Table 28. 95% confidence intervals were calculated for TE_n and CV and are reported as lower (LCL) and upper (UCL) confidence limits. The results are described in detail in Section 3.1.2.

Table 11. Within subject variability of accelerometry measures of postural stability during double leg stance with eyes open

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	1.01	0.68	1.93	±	13.21	8.75	26.84	0.89	0.61	1.63	±	9.72	6.46	19.44
	3	±	0.25	0.17	0.47	±	3.10	2.08	6.02	0.86	0.59	1.57	±	8.50	5.66	16.91
	4	±	0.24	0.16	0.46	±	3.67	2.51	6.80	0.50	0.35	0.92	±	4.71	3.21	8.76
RMSml (m*g*)	2	±	0.22	0.15	0.41	±	4.92	3.36	9.17	1.23	0.85	2.25	±	20.33	13.57	40.19
	3	±	0.14	0.10	0.26	±	3.49	2.39	6.46	0.38	0.26	0.69	±	6.06	4.13	11.35
	4	±	0.09	0.06	0.16	±	2.09	1.43	3.85	0.39	0.27	0.72	±	6.35	4.32	11.89
NPLap (m*g* /s)	2	±	2.07	1.42	3.78	±	0.94	0.64	1.72	6.20	4.27	11.32	±	2.65	1.81	4.88
	3	±	1.55	1.06	2.82	±	0.70	0.48	1.27	2.60	1.79	4.75	±	1.17	0.80	2.14
	4	±	0.88	0.60	1.60	±	0.39	0.27	0.72	3.80	2.61	6.93	±	1.49	1.02	2.73
NPLml (m*g* /s)	2	±	3.22	2.22	5.88	±	1.82	1.25	3.36	10.01	6.88	18.27	±	5.15	3.51	9.60
	3	±	1.72	1.18	3.14	±	0.94	0.64	1.72	2.62	1.80	4.78	±	1.40	0.96	2.58
	4	±	1.43	0.98	2.60	±	0.73	0.50	1.33	5.88	4.04	10.73	±	2.65	1.82	4.90
P2Pap (m*g*)	2	±	7.99	5.50	14.59	±	13.29	8.96	25.59	24.15	16.61	44.10	±	26.45	17.52	53.48
	3	±	2.47	1.70	4.50	±	3.53	2.42	6.54	5.35	3.68	9.76	±	8.41	5.71	15.89
	4	±	1.67	1.15	3.05	±	3.00	2.05	5.54	10.77	7.41	19.66	±	7.66	5.21	14.42
P2Pml (m*g*)	2	±	1.03	0.71	1.89	±	3.54	2.42	6.55	22.66	15.59	41.37	±	36.52	23.88	76.52
	3	±	1.11	0.76	2.02	±	3.83	2.62	7.09	5.70	3.92	10.41	±	10.39	7.03	19.77
	4	±	2.53	1.74	4.63	±	5.23	3.57	9.75	8.77	6.03	16.01	±	13.01	8.78	25.02
APSI (m*g* /kg)	2	±	0.02	0.01	0.03	±	3.54	2.42	6.55	0.03	0.02	0.05	±	15.56	10.46	30.21
	3	±	0.02	0.01	0.03	±	3.83	2.62	7.09	0.02	0.01	0.03	±	9.56	6.48	18.14
	4	±	0.04	0.03	0.07	±	5.23	3.57	9.75	0.02	0.01	0.03	±	6.80	4.63	12.75
MLSI (m*g* /kg)	2	±	0.01	0.01	0.02	±	11.78	7.96	22.55	0.02	0.01	0.04	±	19.87	13.28	39.21
	3	±	0.01	0.01	0.01	±	8.64	5.86	16.33	0.01	0.00	0.01	±	5.08	3.47	9.46
	4	±	0.00	0.00	0.01	±	3.76	2.57	6.98	0.01	0.01	0.02	±	10.65	7.21	20.29
MPFap (Hz)	2	±	0.82	0.56	1.50	±	61.51	39.06	139.93	0.50	0.34	0.90	±	35.19	23.05	73.40
	3	±	0.23	0.16	0.42	±	11.70	7.91	22.39	0.25	0.17	0.46	±	18.76	12.55	36.87
	4	±	0.32	0.22	0.58	±	18.90	12.64	37.17	0.17	0.11	0.30	±	13.21	8.91	25.42
MPFml (Hz)	2	±	0.78	0.54	1.42	±	22.13	14.74	44.04	0.67	0.46	1.23	±	16.91	11.34	33.00
	3	±	0.56	0.38	1.02	±	16.53	11.09	32.21	0.36	0.25	0.65	±	8.95	6.07	16.93
	4	±	0.32	0.22	0.59	±	7.44	5.06	14.00	0.36	0.25	0.65	±	13.56	9.14	26.14

Table 12. Within subject variability of accelerometry measures of postural stability during double leg stance with eyes closed

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	0.58	0.40	1.06	±	8.64	5.76	17.22	1.60	1.10	2.92	±	16.64	10.96	34.31
	3	±	0.95	0.65	1.74	±	10.75	7.14	21.61	1.12	0.77	2.04	±	9.55	6.36	19.10
	4	±	0.41	0.28	0.74	±	4.53	3.09	8.41	0.49	0.33	0.89	±	5.96	4.06	11.14
RMSml (m*g*)	2	±	0.25	0.17	0.45	±	5.43	3.70	10.13	0.58	0.40	1.06	±	10.69	7.23	20.37
	3	±	0.14	0.09	0.25	±	3.24	2.21	5.99	0.41	0.28	0.75	±	5.69	3.88	10.64
	4	±	0.08	0.05	0.14	±	1.75	1.20	3.21	0.34	0.23	0.61	±	6.00	4.09	11.23
NPLap (m*g* /s)	2	±	4.60	3.16	8.40	±	2.04	1.40	3.76	2.63	1.81	4.80	±	1.13	0.78	2.07
	3	±	1.47	1.01	2.68	±	0.65	0.44	1.18	1.38	0.95	2.51	±	0.57	0.39	1.05
	4	±	1.68	1.15	3.06	±	0.75	0.51	1.37	1.62	1.11	2.95	±	0.69	0.47	1.26
NPLml (m*g* /s)	2	±	5.16	3.55	9.41	±	2.68	1.84	4.95	5.54	3.81	10.11	±	2.89	1.98	5.33
	3	±	2.05	1.41	3.73	±	1.13	0.77	2.07	1.39	0.96	2.54	±	0.75	0.51	1.37
	4	±	0.89	0.61	1.63	±	0.47	0.32	0.86	2.94	2.02	5.36	±	1.52	1.05	2.80
P2Pap (m*g*)	2	±	3.00	2.06	5.48	±	6.67	4.54	12.51	28.28	19.45	51.63	±	19.38	12.95	38.17
	3	±	3.34	2.29	6.09	±	7.11	4.84	13.36	4.32	2.97	7.88	±	6.11	4.16	11.44
	4	±	1.24	0.86	2.27	±	2.65	1.81	4.89	9.43	6.49	17.22	±	8.81	5.98	16.67
P2Pml (m*g*)	2	±	4.10	2.82	7.49	±	10.69	7.23	20.37	12.95	8.91	23.64	±	16.92	11.35	33.03
	3	±	7.90	5.44	14.43	±	14.55	9.79	28.14	7.38	5.08	13.48	±	8.93	6.06	16.90
	4	±	1.68	1.15	3.06	±	3.31	2.26	6.12	8.01	5.51	14.63	±	11.13	7.53	21.24
APSI (m*g* /kg)	2	±	0.03	0.02	0.05	±	19.89	13.29	39.26	0.05	0.04	0.10	±	27.00	17.87	54.71
	3	±	0.03	0.02	0.05	±	12.29	8.30	23.56	0.05	0.03	0.08	±	13.30	8.97	25.60
	4	±	0.02	0.01	0.03	±	8.52	5.78	16.09	0.01	0.01	0.03	±	6.03	4.11	11.28
MLSI (m*g* /kg)	2	±	0.01	0.00	0.01	±	5.79	3.94	10.81	0.02	0.02	0.04	±	18.33	12.28	35.98
	3	±	0.01	0.00	0.01	±	7.54	5.13	14.20	0.02	0.02	0.04	±	14.21	9.57	27.45
	4	±	0.01	0.00	0.01	±	5.40	3.68	10.07	0.01	0.01	0.02	±	8.31	5.64	15.68
MPFap (Hz)	2	±	0.80	0.55	1.45	±	42.17	27.38	90.10	0.71	0.49	1.29	±	57.40	36.62	128.90
	3	±	0.26	0.18	0.48	±	23.95	15.91	47.99	0.20	0.14	0.36	±	23.85	15.85	47.77
	4	±	0.22	0.15	0.39	±	16.10	10.81	31.33	0.27	0.19	0.50	±	14.19	9.56	27.41
MPFml (Hz)	2	±	0.34	0.23	0.61	±	9.29	6.30	17.61	0.66	0.45	1.20	±	33.31	21.86	69.01
	3	±	0.49	0.34	0.90	±	14.18	9.55	27.40	0.63	0.43	1.14	±	25.95	17.20	52.37
	4	±	0.34	0.23	0.61	±	11.08	7.49	21.14	0.36	0.24	0.65	±	10.50	7.11	19.99

Table 13. Within subject variability of accelerometry measures of postural stability during double leg stance on foam with eyes open

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	1.40	0.96	2.56	±	15.68	10.34	32.18	0.89	0.61	1.62	±	9.70	6.46	19.42
	3	±	0.58	0.40	1.07	±	5.72	3.83	11.24	0.59	0.41	1.08	±	7.26	4.85	14.37
	4	±	0.63	0.43	1.14	±	5.79	3.95	10.82	0.46	0.32	0.85	±	5.19	3.54	9.67
RMSml (m*g*)	2	±	0.50	0.34	0.91	±	6.66	4.54	12.49	0.74	0.51	1.35	±	10.20	6.91	19.41
	3	±	0.32	0.22	0.58	±	4.72	3.22	8.78	0.37	0.26	0.68	±	5.45	3.72	10.18
	4	±	0.17	0.12	0.31	±	2.36	1.62	4.35	0.19	0.13	0.35	±	2.67	1.83	4.93
NPLap (m*g* /s)	2	±	2.74	1.89	5.00	±	1.17	0.80	2.15	3.40	2.34	6.20	±	1.41	0.97	2.59
	3	±	2.34	1.61	4.28	±	1.02	0.70	1.88	5.80	3.99	10.58	±	2.21	1.52	4.08
	4	±	1.56	1.07	2.84	±	0.65	0.45	1.19	1.61	1.11	2.95	±	0.64	0.44	1.18
NPLml (m*g* /s)	2	±	2.38	1.64	4.34	±	1.16	0.80	2.13	5.14	3.54	9.39	±	2.44	1.67	4.50
	3	±	1.58	1.08	2.88	±	0.81	0.56	1.48	5.67	3.90	10.34	±	2.60	1.78	4.80
	4	±	0.65	0.45	1.19	±	0.34	0.23	0.62	2.17	1.49	3.97	±	1.04	0.71	1.90
P2Pap (m*g*)	2	±	6.85	4.71	12.51	±	11.94	8.06	22.86	4.77	3.28	8.70	±	8.56	5.81	16.17
	3	±	3.48	2.39	6.35	±	6.03	4.11	11.29	7.02	4.83	12.82	±	9.99	6.77	18.99
	4	±	2.26	1.56	4.13	±	3.55	2.43	6.58	2.16	1.49	3.95	±	3.50	2.39	6.48
P2Pml (m*g*)	2	±	4.87	3.35	8.89	±	8.31	5.64	15.69	4.33	2.98	7.91	±	9.34	6.33	17.70
	3	±	3.02	2.07	5.50	±	5.72	3.90	10.68	11.63	8.00	21.23	±	14.78	9.95	28.62
	4	±	1.60	1.10	2.92	±	3.06	2.09	5.65	3.40	2.34	6.20	±	5.56	3.79	10.39
APSI (m*g* /kg)	2	±	0.02	0.02	0.04	±	15.93	10.70	30.98	0.03	0.02	0.06	±	14.93	10.04	28.92
	3	±	0.02	0.01	0.03	±	7.78	5.29	14.67	0.03	0.02	0.05	±	14.95	10.06	28.97
	4	±	0.02	0.02	0.04	±	8.71	5.91	16.47	0.01	0.01	0.02	±	8.07	5.48	15.21
MLSI (m*g* /kg)	2	±	0.02	0.02	0.05	±	13.62	9.18	26.25	0.02	0.01	0.03	±	15.87	10.67	30.86
	3	±	0.01	0.01	0.03	±	10.07	6.82	19.15	0.01	0.00	0.01	±	4.54	3.10	8.44
	4	±	0.01	0.00	0.01	±	3.88	2.65	7.20	0.01	0.00	0.01	±	4.86	3.32	9.05
MPFap (Hz)	2	±	0.52	0.36	0.96	±	30.90	20.34	63.48	0.46	0.32	0.84	±	33.73	22.13	70.00
	3	±	0.17	0.12	0.31	±	15.65	10.52	30.40	0.42	0.29	0.77	±	27.39	18.12	55.57
	4	±	0.21	0.14	0.38	±	17.45	11.70	34.13	0.25	0.17	0.45	±	16.78	11.26	32.74
MPFml (Hz)	2	±	0.49	0.34	0.90	±	26.38	17.47	53.32	0.71	0.49	1.29	±	29.82	19.66	61.03
	3	±	0.47	0.32	0.86	±	19.15	12.81	37.70	0.17	0.12	0.31	±	7.01	4.77	13.17
	4	±	0.18	0.12	0.33	±	8.10	5.50	15.28	0.18	0.12	0.32	±	10.89	7.37	20.78

Table 14. Within subject variability of accelerometry measures of postural stability during double leg stance on foam with eyes closed

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	1.64	1.13	2.99	±	12.92	8.55	26.20	1.42	0.97	2.58	±	9.45	6.29	18.88
	3	±	0.55	0.38	1.01	±	4.54	3.05	8.88	1.47	1.01	2.68	±	9.59	6.38	19.17
	4	±	0.43	0.30	0.78	±	3.43	2.35	6.36	0.52	0.36	0.95	±	3.66	2.50	6.78
RMSml (m*g*)	2	±	0.78	0.54	1.43	±	10.24	6.93	19.47	0.77	0.53	1.40	±	10.95	7.41	20.89
	3	±	0.36	0.25	0.66	±	4.26	2.91	7.92	0.35	0.24	0.65	±	4.81	3.29	8.96
	4	±	0.26	0.18	0.47	±	3.16	2.16	5.85	0.46	0.32	0.84	±	5.37	3.66	10.02
NPLap (m*g*/s)	2	±	2.94	2.02	5.37	±	1.22	0.84	2.24	6.62	4.55	12.08	±	2.43	1.67	4.48
	3	±	2.07	1.42	3.78	±	0.88	0.60	1.60	3.64	2.50	6.64	±	1.35	0.93	2.48
	4	±	1.56	1.07	2.84	±	0.62	0.43	1.13	1.66	1.14	3.04	±	0.67	0.46	1.23
NPLml (m*g*/s)	2	±	6.27	4.31	11.45	±	2.89	1.98	5.35	6.81	4.68	12.42	±	2.95	2.02	5.46
	3	±	2.18	1.50	3.99	±	1.01	0.69	1.84	3.83	2.63	6.99	±	1.76	1.21	3.24
	4	±	1.69	1.16	3.09	±	0.80	0.55	1.46	2.65	1.83	4.84	±	1.26	0.87	2.32
P2Pap (m*g*)	2	±	7.61	5.23	13.89	±	9.63	6.53	18.28	7.33	5.04	13.38	±	9.24	6.27	17.52
	3	±	3.66	2.52	6.68	±	4.83	3.30	8.99	8.76	6.03	15.99	±	10.14	6.87	19.28
	4	±	2.13	1.46	3.89	±	2.79	1.91	5.16	1.85	1.27	3.38	±	2.49	1.71	4.59
P2Pml (m*g*)	2	±	6.31	4.34	11.51	±	10.32	6.99	19.64	10.33	7.11	18.87	±	17.36	11.64	33.93
	3	±	2.25	1.55	4.11	±	3.76	2.57	6.96	4.74	3.26	8.66	±	7.76	5.28	14.62
	4	±	2.31	1.59	4.22	±	4.58	3.13	8.51	2.81	1.93	5.12	±	4.88	3.33	9.08
APSI (m*g*/kg)	2	±	0.06	0.04	0.11	±	22.90	15.24	45.70	0.03	0.02	0.05	±	8.84	6.00	16.72
	3	±	0.02	0.01	0.04	±	9.44	6.40	17.90	0.07	0.05	0.12	±	14.53	9.78	28.09
	4	±	0.01	0.01	0.02	±	4.71	3.22	8.77	0.02	0.01	0.04	±	6.02	4.10	11.26
MLSI (m*g*/kg)	2	±	0.03	0.02	0.05	±	18.37	12.30	36.05	0.02	0.02	0.04	±	19.29	12.90	37.98
	3	±	0.01	0.01	0.03	±	7.36	5.01	13.85	0.01	0.01	0.02	±	6.34	4.32	11.88
	4	±	0.01	0.01	0.01	±	4.73	3.23	8.81	0.01	0.01	0.02	±	6.22	4.24	11.65
MPFap (Hz)	2	±	0.35	0.24	0.63	±	49.82	32.06	109.18	0.18	0.12	0.33	±	17.79	11.92	34.85
	3	±	0.26	0.18	0.48	±	21.42	14.28	42.52	0.16	0.11	0.29	±	19.60	13.10	38.66
	4	±	0.11	0.08	0.20	±	9.34	6.33	17.70	0.10	0.07	0.19	±	12.28	8.29	23.54
MPFml (Hz)	2	±	0.58	0.40	1.06	±	35.22	23.06	73.47	0.74	0.51	1.35	±	39.98	26.03	84.77
	3	±	0.17	0.12	0.31	±	14.64	9.85	28.33	0.26	0.18	0.48	±	13.11	8.85	25.23
	4	±	0.11	0.08	0.20	±	8.08	5.49	15.24	0.17	0.12	0.32	±	10.34	7.00	19.68

Table 15. Within subject variability of accelerometry measures of postural stability during tandem stance with eyes open

			Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	1.88	1.27	3.60	±	12.60	8.35	25.53	2.72	1.87	4.96	±	14.73	9.91	28.51
	3	±	0.58	0.39	1.12	±	4.96	3.32	9.71	1.75	1.21	3.20	±	10.59	7.17	20.17
	4	±	0.45	0.30	0.86	±	3.79	2.54	7.39	0.98	0.68	1.80	±	6.42	4.37	12.03
RMSml (m*g*)	2	±	3.78	2.55	7.24	±	20.11	13.18	42.06	1.28	0.88	2.34	±	11.62	7.85	22.22
	3	±	1.13	0.76	2.16	±	5.28	3.54	10.36	0.70	0.48	1.28	±	6.23	4.24	11.66
	4	±	0.65	0.44	1.24	±	4.29	2.93	7.98	1.02	0.70	1.86	±	6.32	4.31	11.84
NPLap (m*g* /s)	2	±	17.49	11.82	33.51	±	4.89	3.28	9.58	12.01	8.26	21.93	±	3.96	2.71	7.34
	3	±	6.24	4.21	11.95	±	1.73	1.16	3.33	5.74	3.95	10.47	±	1.98	1.36	3.64
	4	±	4.95	3.35	9.49	±	1.39	0.95	2.54	3.02	2.08	5.51	±	1.12	0.77	2.06
NPLml (m*g* /s)	2	±	51.27	34.63	98.22	±	10.37	6.89	20.81	19.02	13.08	34.72	±	5.76	3.93	10.77
	3	±	16.89	11.41	32.35	±	3.21	2.16	6.24	12.40	8.53	22.64	±	3.50	2.40	6.49
	4	±	9.06	6.12	17.36	±	1.73	1.18	3.17	6.96	4.79	12.71	±	1.98	1.36	3.64
P2Pap (m*g*)	2	±	26.79	18.10	51.33	±	12.54	8.31	25.40	13.71	9.43	25.03	±	12.62	8.52	24.24
	3	±	12.23	8.26	23.43	±	8.44	5.62	16.79	7.38	5.07	13.47	±	7.20	4.90	13.53
	4	±	7.56	5.10	14.47	±	5.76	3.93	10.77	5.94	4.08	10.84	±	6.05	4.12	11.31
P2Pml (m*g*)	2	±	51.09	34.51	97.88	±	23.85	15.55	50.66	12.26	8.43	22.38	±	14.53	9.78	28.11
	3	±	17.73	11.97	33.96	±	7.43	4.96	14.71	4.46	3.07	8.14	±	6.12	4.17	11.45
	4	±	10.26	6.93	19.66	±	6.66	4.54	12.49	13.32	9.16	24.31	±	8.93	6.06	16.91
APSI (m*g* /kg)	2	±	0.06	0.04	0.12	±	22.46	14.67	47.43	0.04	0.03	0.08	±	15.45	10.39	30.00
	3	±	0.02	0.01	0.04	±	8.41	5.61	16.73	0.05	0.03	0.09	±	14.53	9.78	28.11
	4	±	0.02	0.01	0.03	±	8.11	5.51	15.30	0.03	0.02	0.05	±	8.86	6.01	16.76
MLSI (m*g* /kg)	2	±	0.08	0.05	0.15	±	24.89	16.20	53.09	0.04	0.03	0.07	±	20.22	13.51	39.97
	3	±	0.02	0.01	0.04	±	5.38	3.61	10.57	0.01	0.01	0.02	±	6.24	4.25	11.69
	4	±	0.02	0.01	0.04	±	8.36	5.68	15.79	0.02	0.01	0.03	±	7.23	4.92	13.60
MPFap (Hz)	2	±	0.44	0.30	0.85	±	44.22	28.06	101.68	0.37	0.26	0.68	±	29.41	19.40	60.11
	3	±	0.34	0.23	0.65	±	19.66	12.89	41.04	0.38	0.26	0.70	±	24.87	16.51	50.01
	4	±	0.22	0.15	0.41	±	16.81	11.28	32.79	0.20	0.13	0.36	±	16.29	10.94	31.73
MPFml (Hz)	2	±	0.39	0.26	0.74	±	28.34	18.35	61.27	0.73	0.50	1.33	±	34.93	22.88	72.79
	3	±	0.24	0.16	0.46	±	11.33	7.52	22.82	0.27	0.19	0.49	±	9.90	6.71	18.81
	4	±	0.26	0.18	0.50	±	14.76	9.93	28.58	0.29	0.20	0.54	±	10.78	7.30	20.55

Table 16. Within subject variability of accelerometry measures of postural stability during tandem stance with eyes closed

			Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	6.25	4.30	11.41	±	26.15	17.32	52.81	2.63	1.81	4.79	±	13.80	9.30	26.63
	3	±	3.61	2.48	6.59	±	11.57	7.82	22.13	1.10	0.76	2.00	±	6.88	4.69	12.92
	4	±	1.96	1.35	3.57	±	7.51	5.11	14.14	0.73	0.50	1.33	±	3.67	2.51	6.80
RMSml (m*g*)	2	±	4.55	3.13	8.31	±	22.51	14.99	44.87	1.91	1.32	3.49	±	8.26	5.61	15.59
	3	±	3.03	2.09	5.54	±	11.15	7.54	21.29	1.90	1.31	3.47	±	9.00	6.11	17.04
	4	±	1.30	0.90	2.38	±	5.58	3.80	10.41	0.99	0.68	1.81	±	4.45	3.04	8.27
NPLap (m*g*/s)	2	±	33.53	23.06	61.22	±	6.69	4.56	12.55	21.48	14.77	39.21	±	5.18	3.54	9.66
	3	±	22.78	15.67	41.59	±	5.81	3.96	10.86	15.66	10.77	28.58	±	4.11	2.81	7.63
	4	±	9.44	6.49	17.23	±	2.46	1.68	4.53	9.14	6.29	16.68	±	2.62	1.79	4.83
NPLml (m*g*/s)	2	±	62.31	42.86	113.75	±	9.75	6.61	18.51	45.52	31.31	83.11	±	6.53	4.45	12.24
	3	±	42.66	29.34	77.87	±	7.61	5.18	14.33	30.90	21.26	56.41	±	6.08	4.15	11.38
	4	±	23.07	15.87	42.12	±	4.08	2.79	7.58	17.87	12.29	32.63	±	3.78	2.59	7.01
P2Pap (m*g*)	2	±	31.21	21.47	56.98	±	22.50	14.98	44.84	29.47	20.27	53.80	±	19.69	13.16	38.84
	3	±	37.16	25.56	67.85	±	11.20	7.57	21.39	10.69	7.35	19.52	±	8.52	5.79	16.11
	4	±	19.84	13.64	36.21	±	8.85	6.01	16.74	7.31	5.03	13.34	±	4.77	3.26	8.89
P2Pml (m*g*)	2	±	28.38	19.52	51.81	±	18.97	12.69	37.32	36.25	24.93	66.18	±	19.44	13.00	38.31
	3	±	61.48	42.29	112.24	±	14.26	9.61	27.56	18.50	12.72	33.77	±	10.82	7.32	20.63
	4	±	21.45	14.76	39.16	±	8.40	5.71	15.87	13.94	9.59	25.45	±	5.96	4.06	11.14
APSI (m*g*/kg)	2	±	0.12	0.08	0.22	±	34.57	22.66	71.95	0.10	0.07	0.19	±	22.11	14.73	43.99
	3	±	0.07	0.05	0.13	±	17.28	11.59	33.78	0.04	0.03	0.08	±	11.23	7.59	21.44
	4	±	0.04	0.03	0.07	±	10.95	7.41	20.89	0.03	0.02	0.05	±	7.85	5.33	14.78
MLSI (m*g*/kg)	2	±	0.09	0.07	0.17	±	23.82	15.83	47.71	0.06	0.04	0.11	±	14.46	9.73	27.95
	3	±	0.06	0.04	0.11	±	16.73	11.22	32.63	0.03	0.02	0.05	±	7.44	5.06	13.99
	4	±	0.02	0.02	0.04	±	5.59	3.81	10.43	0.02	0.02	0.04	±	6.51	4.43	12.21
MPFap (Hz)	2	±	0.72	0.50	1.32	±	63.79	40.41	146.16	0.43	0.30	0.79	±	38.41	25.05	81.02
	3	±	0.32	0.22	0.58	±	27.80	18.38	56.48	0.28	0.19	0.52	±	21.02	14.02	41.66
	4	±	0.26	0.18	0.48	±	17.28	11.58	33.77	0.15	0.10	0.27	±	11.59	7.83	22.16
MPFml (Hz)	2	±	0.51	0.35	0.94	±	31.54	20.75	64.94	0.49	0.34	0.90	±	31.33	20.62	64.47
	3	±	0.34	0.23	0.62	±	26.43	17.50	53.44	0.17	0.12	0.31	±	11.47	7.76	21.93
	4	±	0.12	0.08	0.23	±	6.05	4.12	11.32	0.19	0.13	0.35	±	8.12	5.52	15.31

Table 17. Within subject variability of accelerometry measures of postural stability during single leg stance with eyes open

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	1.59	1.09	2.89	±	12.75	8.61	24.50	1.66	1.14	3.02	±	11.74	7.93	22.46
	3	±	0.70	0.48	1.28	±	6.38	4.35	11.96	1.66	1.14	3.02	±	7.85	5.33	14.79
	4	±	0.41	0.28	0.78	±	4.01	2.69	7.82	1.39	0.96	2.54	±	6.42	4.37	12.03
RMSml (m*g*)	2	±	0.96	0.66	1.76	±	9.71	6.58	18.44	0.97	0.67	1.77	±	7.04	4.79	13.23
	3	±	1.09	0.75	1.98	±	10.50	7.11	20.00	1.24	0.85	2.27	±	6.97	4.74	13.09
	4	±	0.52	0.35	0.99	±	5.11	3.42	10.01	1.52	1.04	2.77	±	6.95	4.73	13.06
NPLap (m*g* /s)	2	±	4.03	2.77	7.36	±	1.64	1.13	3.02	13.03	8.96	23.78	±	4.11	2.81	7.63
	3	±	1.94	1.33	3.54	±	0.76	0.52	1.39	9.97	6.85	18.19	±	3.06	2.09	5.65
	4	±	3.63	2.45	6.95	±	1.44	0.97	2.77	5.61	3.86	10.23	±	1.87	1.29	3.45
NPLml (m*g* /s)	2	±	8.64	5.94	15.77	±	2.92	2.00	5.40	19.36	13.32	35.35	±	5.79	3.95	10.82
	3	±	6.72	4.62	12.26	±	2.19	1.50	4.04	16.12	11.09	29.44	±	4.24	2.90	7.88
	4	±	13.10	9.01	23.92	±	3.43	2.35	6.35	11.06	7.61	20.20	±	3.45	2.36	6.38
P2Pap (m*g*)	2	±	4.96	3.41	9.05	±	8.10	5.50	15.28	9.73	6.69	17.76	±	10.53	7.13	20.06
	3	±	2.71	1.87	4.95	±	4.67	3.19	8.69	3.48	2.40	6.36	±	3.56	2.43	6.59
	4	±	1.88	1.27	3.60	±	3.70	2.48	7.20	5.73	3.94	10.45	±	4.97	3.39	9.26
P2Pml (m*g*)	2	±	5.24	3.60	9.56	±	9.54	6.47	18.10	7.06	4.85	12.88	±	9.02	6.12	17.08
	3	±	4.62	3.18	8.43	±	7.75	5.27	14.59	11.47	7.89	20.94	±	8.09	5.50	15.26
	4	±	2.57	1.74	4.93	±	4.15	2.78	8.10	8.90	6.12	16.25	±	7.86	5.34	14.80
APSI (m*g* /kg)	2	±	0.02	0.02	0.04	±	12.50	8.44	24.00	0.06	0.04	0.11	±	19.30	12.91	38.01
	3	±	0.01	0.01	0.03	±	8.70	5.91	16.46	0.04	0.03	0.07	±	9.74	6.60	18.49
	4	±	0.01	0.01	0.02	±	8.20	5.47	16.30	0.03	0.02	0.06	±	10.25	6.94	19.51
MLSI (m*g* /kg)	2	±	0.01	0.01	0.03	±	11.88	8.03	22.75	0.03	0.02	0.06	±	12.25	8.28	23.49
	3	±	0.02	0.01	0.04	±	15.39	10.35	29.86	0.01	0.01	0.02	±	5.88	4.01	10.99
	4	±	0.01	0.01	0.02	±	7.14	4.77	14.13	0.02	0.01	0.04	±	7.98	5.42	15.05
MPFap (Hz)	2	±	0.37	0.25	0.68	±	23.60	15.69	47.23	0.50	0.34	0.92	±	40.54	26.38	86.14
	3	±	0.24	0.16	0.43	±	15.72	10.56	30.54	0.21	0.15	0.39	±	21.10	14.07	41.83
	4	±	0.23	0.16	0.45	±	14.39	9.51	29.38	0.22	0.15	0.40	±	17.14	11.49	33.48
MPFml (Hz)	2	±	0.56	0.38	1.02	±	23.58	15.68	47.19	0.46	0.32	0.85	±	28.66	18.93	58.43
	3	±	0.42	0.29	0.76	±	29.31	19.34	59.89	0.11	0.07	0.19	±	7.50	5.10	14.12
	4	±	0.26	0.17	0.49	±	12.34	8.18	24.97	0.23	0.16	0.42	±	14.24	9.59	27.51

Table 18. Within subject variability of accelerometry measures of postural stability during single leg stance with eyes closed

			Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	5.03	3.46	9.19	±	19.80	13.23	39.07	2.09	1.44	3.81	±	8.49	5.76	16.03
	3	±	5.81	4.00	10.61	±	17.08	11.46	33.37	2.31	1.59	4.22	±	7.87	5.35	14.83
	4	±	1.79	1.23	3.27	±	5.73	3.91	10.71	1.04	0.72	1.91	±	4.29	2.93	7.97
RMSml (m*g*)	2	±	6.97	4.80	12.73	±	20.19	13.48	39.90	7.95	5.47	14.51	±	17.45	11.70	34.13
	3	±	4.56	3.14	8.32	±	14.90	10.03	28.86	3.91	2.69	7.15	±	11.35	7.67	21.68
	4	±	4.11	2.83	7.50	±	11.36	7.68	21.70	5.69	3.91	10.38	±	12.52	8.45	24.02
NPLap (m*g* /s)	2	±	18.45	12.69	33.67	±	4.92	3.36	9.16	22.36	15.38	40.82	±	4.91	3.35	9.15
	3	±	21.87	15.05	39.93	±	4.71	3.22	8.77	32.22	22.16	58.82	±	5.30	3.62	9.90
	4	±	8.57	5.89	15.64	±	2.21	1.52	4.08	18.39	12.65	33.56	±	4.14	2.83	7.69
NPLml (m*g* /s)	2	±	54.77	37.67	99.99	±	9.08	6.16	17.19	55.11	37.90	100.60	±	8.87	6.02	16.79
	3	±	35.90	24.69	65.54	±	7.65	5.20	14.41	48.00	33.01	87.62	±	6.95	4.73	13.05
	4	±	18.69	12.86	34.12	±	3.39	2.32	6.27	34.01	23.39	62.08	±	5.49	3.75	10.26
P2Pap (m*g*)	2	±	21.29	14.64	38.87	±	15.43	10.38	29.96	14.84	10.21	27.09	±	10.84	7.34	20.68
	3	±	32.73	22.51	59.75	±	16.55	11.11	32.25	14.06	9.67	25.66	±	6.71	4.57	12.60
	4	±	9.88	6.79	18.03	±	5.58	3.81	10.42	12.40	8.53	22.63	±	8.02	5.45	15.13
P2Pml (m*g*)	2	±	32.45	22.32	59.24	±	17.22	11.55	33.66	47.97	32.99	87.57	±	16.92	11.35	33.03
	3	±	54.27	37.33	99.08	±	19.06	12.75	37.50	28.90	19.88	52.76	±	12.07	8.16	23.14
	4	±	23.74	16.33	43.35	±	10.46	7.08	19.91	27.02	18.58	49.32	±	11.83	8.00	22.65
APSI (m*g* /kg)	2	±	0.06	0.04	0.12	±	20.21	13.49	39.93	0.04	0.03	0.08	±	11.49	7.77	21.96
	3	±	0.07	0.05	0.13	±	16.11	10.82	31.35	0.04	0.02	0.06	±	10.27	6.96	19.54
	4	±	0.03	0.02	0.06	±	9.09	6.17	17.21	0.02	0.01	0.03	±	4.72	3.22	8.79
MLSI (m*g* /kg)	2	±	0.11	0.08	0.20	±	21.98	14.65	43.73	0.08	0.05	0.15	±	17.25	11.57	33.71
	3	±	0.06	0.04	0.11	±	16.72	11.22	32.62	0.04	0.03	0.08	±	9.51	6.45	18.04
	4	±	0.09	0.06	0.16	±	15.62	10.50	30.34	0.06	0.04	0.11	±	11.89	8.04	22.77
MPFap (Hz)	2	±	0.42	0.29	0.76	±	27.19	17.99	55.12	0.45	0.31	0.83	±	23.13	15.39	46.21
	3	±	0.24	0.16	0.43	±	20.45	13.65	40.45	0.22	0.15	0.40	±	13.52	9.12	26.06
	4	±	0.27	0.19	0.50	±	14.89	10.02	28.85	0.30	0.20	0.54	±	15.73	10.57	30.57
MPFml (Hz)	2	±	0.60	0.42	1.10	±	30.90	20.35	63.49	0.41	0.28	0.75	±	26.52	17.56	53.64
	3	±	0.33	0.23	0.61	±	15.90	10.68	30.92	0.16	0.11	0.30	±	9.84	6.67	18.69
	4	±	0.23	0.16	0.41	±	21.43	14.29	42.54	0.20	0.14	0.37	±	13.86	9.34	26.74

Table 19. Within subject variability of root mean square measures of postural stability during a forward jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	34.40	23.66	62.80	±	14.51	9.77	28.06	25.92	17.83	47.32	±	10.07	6.83	19.15
	3	±	11.65	8.01	21.26	±	6.47	4.41	12.12	13.38	9.20	24.42	±	5.33	3.64	9.95
	4	±	7.86	5.41	14.35	±	3.81	2.61	7.07	10.07	6.93	18.38	±	4.20	2.87	7.80
	5	±	4.79	3.30	8.75	±	2.19	1.50	4.03	8.54	5.88	15.60	±	3.42	2.34	6.33
	6	±	10.27	7.06	18.75	±	4.59	3.14	8.54	2.36	1.62	4.30	±	1.11	0.76	2.03
	7	±	5.69	3.91	10.39	±	2.43	1.67	4.49	4.57	3.14	8.34	±	1.77	1.21	3.25
	8	±	5.80	3.99	10.58	±	2.68	1.83	4.94	3.42	2.35	6.24	±	1.42	0.98	2.61
	9	±	6.23	4.28	11.36	±	3.32	2.27	6.14	3.63	2.50	6.63	±	2.31	1.58	4.25
	10	±	2.70	1.86	4.93	±	1.39	0.96	2.56	7.59	5.22	13.85	±	3.15	2.16	5.83
	11	±	2.13	1.47	3.89	±	0.97	0.67	1.78	6.14	4.06	12.49	±	2.57	1.69	5.30
RMSml (m*g*)	2	±	47.03	32.35	85.85	±	10.72	7.25	20.42	25.74	17.70	46.99	±	11.06	7.48	21.11
	3	±	17.90	12.31	32.68	±	5.11	3.48	9.52	15.42	10.60	28.14	±	6.10	4.16	11.42
	4	±	11.93	8.21	21.79	±	4.08	2.79	7.57	13.46	9.26	24.58	±	4.25	2.91	7.90
	5	±	8.53	5.87	15.57	±	2.55	1.74	4.70	12.10	8.33	22.10	±	4.74	3.24	8.82
	6	±	15.04	10.35	27.47	±	4.57	3.12	8.50	2.55	1.75	4.66	±	0.92	0.63	1.68
	7	±	10.95	7.53	19.98	±	2.98	2.04	5.51	4.95	3.41	9.04	±	1.73	1.19	3.19
	8	±	8.17	5.62	14.91	±	2.42	1.66	4.45	7.25	4.99	13.23	±	2.33	1.60	4.30
	9	±	7.51	5.17	13.72	±	3.23	2.21	5.98	4.41	3.03	8.05	±	1.53	1.05	2.82
	10	±	4.81	3.31	8.77	±	1.06	0.73	1.95	8.68	5.97	15.84	±	2.74	1.88	5.05
	11	±	3.71	2.55	6.78	±	0.98	0.68	1.80	5.95	3.93	12.11	±	2.49	1.64	5.14

Table 20. Within subject variability of normalized path length measures of postural stability during a forward jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
NPLap (mg/s)	2	±	923.87	635.47	1686.63	±	14.30	9.63	27.63	804.93	553.66	1469.49	±	12.26	8.28	23.52
	3	±	364.01	250.38	664.53	±	7.62	5.18	14.35	365.09	251.12	666.51	±	3.81	2.60	7.06
	4	±	240.66	165.54	439.36	±	3.75	2.56	6.95	287.57	197.80	524.99	±	3.75	2.57	6.96
	5	±	154.76	106.45	282.52	±	2.44	1.67	4.49	282.28	194.16	515.34	±	3.85	2.63	7.14
	6	±	344.69	237.09	629.27	±	5.31	3.62	9.91	119.20	81.99	217.60	±	1.34	0.92	2.47
	7	±	185.90	127.87	339.38	±	2.84	1.94	5.24	113.06	77.77	206.41	±	1.66	1.14	3.05
	8	±	178.61	122.86	326.08	±	2.72	1.86	5.02	99.13	68.18	180.97	±	1.55	1.06	2.84
	9	±	169.59	116.65	309.61	±	3.10	2.12	5.73	78.16	53.76	142.68	±	1.40	0.96	2.57
	10	±	87.83	60.41	160.33	±	1.36	0.93	2.50	191.27	131.56	349.18	±	2.64	1.81	4.87
	11	±	65.77	45.24	120.07	±	0.99	0.68	1.82	133.00	87.94	270.69	±	1.57	1.04	3.22
NPLml (mg/s)	2	±	895.10	615.68	1634.10	±	11.92	8.05	22.82	417.51	287.18	762.21	±	8.79	5.97	16.62
	3	±	299.47	205.99	546.72	±	4.71	3.22	8.77	260.11	178.91	474.86	±	3.13	2.14	5.79
	4	±	93.28	64.16	170.30	±	1.37	0.94	2.51	295.33	203.14	539.15	±	4.61	3.15	8.57
	5	±	207.46	142.70	378.75	±	2.76	1.89	5.10	186.31	128.15	340.13	±	3.03	2.07	5.60
	6	±	229.19	157.65	418.42	±	3.78	2.59	7.02	79.24	54.50	144.65	±	1.42	0.97	2.60
	7	±	193.44	133.05	353.14	±	2.91	2.00	5.38	90.90	62.53	165.95	±	1.57	1.08	2.89
	8	±	105.95	72.87	193.42	±	1.87	1.28	3.44	73.98	50.89	135.06	±	1.35	0.93	2.48
	9	±	117.25	80.65	214.06	±	2.59	1.77	4.77	66.77	45.92	121.89	±	1.25	0.86	2.29
	10	±	70.54	48.52	128.77	±	1.00	0.69	1.84	166.82	114.74	304.55	±	2.51	1.72	4.63
	11	±	56.35	38.76	102.88	±	0.81	0.56	1.49	79.47	52.55	161.75	±	1.39	0.92	2.85

Table 21. Within subject variability of peak to peak measures of postural stability during a forward jump-landing task

			Control						CAI					
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
P2Pap (m*g*)	2	±	786.64	541.08	1436.10	±	19.71	13.17	38.89	534.43	367.60	975.66	±	12.85	8.67	24.69
	3	±	392.63	270.07	716.79	±	10.22	6.92	19.44	470.33	323.51	858.64	±	8.31	5.64	15.68
	4	±	211.24	145.30	385.65	±	8.95	6.07	16.94	303.49	208.75	554.05	±	5.76	3.93	10.76
	5	±	136.04	93.57	248.35	±	3.51	2.40	6.50	263.95	181.55	481.87	±	7.07	4.81	13.28
	6	±	346.43	238.29	632.45	±	7.93	5.39	14.95	89.68	61.68	163.71	±	2.15	1.47	3.96
	7	±	244.46	168.15	446.28	±	5.20	3.55	9.70	155.83	107.18	284.48	±	2.50	1.72	4.62
	8	±	172.61	118.73	315.11	±	4.64	3.17	8.63	97.01	66.73	177.11	±	1.99	1.36	3.66
	9	±	210.67	144.91	384.60	±	5.76	3.92	10.76	113.64	78.17	207.47	±	3.17	2.17	5.86
	10	±	86.03	59.18	157.06	±	1.84	1.26	3.38	205.68	141.47	375.49	±	3.65	2.49	6.76
	11	±	91.48	62.92	167.01	±	1.97	1.35	3.62	189.10	125.03	384.87	±	3.69	2.43	7.66
P2Pml (m*g*)	2	±	1150.48	791.34	2100.32	±	14.06	9.47	27.15	563.03	387.27	1027.88	±	18.19	12.18	35.67
	3	±	467.56	321.61	853.59	±	8.74	5.93	16.53	313.15	215.39	571.69	±	4.98	3.40	9.29
	4	±	291.88	200.77	532.87	±	10.26	6.95	19.51	309.57	212.93	565.15	±	5.50	3.75	10.26
	5	±	256.79	176.63	468.81	±	4.63	3.16	8.62	324.96	223.52	593.25	±	9.91	6.72	18.83
	6	±	372.15	255.98	679.41	±	8.85	6.01	16.74	75.88	52.19	138.52	±	1.19	0.82	2.19
	7	±	342.79	235.78	625.79	±	6.65	4.53	12.47	130.27	89.60	237.82	±	2.90	1.98	5.35
	8	±	268.68	184.81	490.50	±	5.47	3.73	10.22	128.76	88.56	235.06	±	2.36	1.62	4.35
	9	±	199.08	136.93	363.43	±	5.32	3.63	9.92	124.51	85.64	227.30	±	2.49	1.71	4.59
	10	±	89.05	61.25	162.57	±	1.32	0.91	2.43	224.24	154.24	409.37	±	3.86	2.64	7.15
	11	±	90.34	62.14	164.92	±	2.06	1.42	3.80	135.46	89.57	275.71	±	4.25	2.79	8.85

Table 22. Within subject variability of stability index measures of postural stability during a forward jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
APSI (m*g* /kg)	2	±	0.19	0.13	0.35	±	13.08	8.82	25.16	0.14	0.10	0.26	±	8.05	5.47	15.18
	3	±	0.10	0.07	0.19	±	7.84	5.33	14.77	0.06	0.04	0.11	±	3.25	2.22	6.01
	4	±	0.07	0.05	0.12	±	4.74	3.23	8.81	0.05	0.04	0.10	±	2.85	1.95	5.26
	5	±	0.04	0.03	0.07	±	2.92	2.00	5.39	0.09	0.06	0.17	±	5.45	3.71	10.17
	6	±	0.06	0.04	0.11	±	4.13	2.82	7.67	0.01	0.01	0.02	±	0.65	0.45	1.19
	7	±	0.05	0.03	0.08	±	3.12	2.14	5.78	0.02	0.02	0.04	±	1.26	0.86	2.31
	8	±	0.05	0.03	0.09	±	3.04	2.08	5.61	0.02	0.01	0.04	±	1.25	0.86	2.30
	9	±	0.04	0.03	0.07	±	3.15	2.16	5.83	0.02	0.01	0.04	±	2.04	1.40	3.76
	10	±	0.02	0.01	0.03	±	1.08	0.74	1.98	0.04	0.03	0.08	±	2.71	1.86	5.00
	11	±	0.01	0.01	0.02	±	0.95	0.65	1.73	0.05	0.03	0.10	±	2.79	1.83	5.76
MLSI (m*g* /kg)	2	±	0.27	0.18	0.49	±	10.61	7.18	20.20	0.17	0.11	0.30	±	11.22	7.59	21.43
	3	±	0.12	0.08	0.21	±	5.58	3.80	10.41	0.15	0.10	0.28	±	9.48	6.43	17.99
	4	±	0.09	0.06	0.17	±	5.34	3.64	9.97	0.07	0.05	0.13	±	4.24	2.90	7.87
	5	±	0.05	0.03	0.09	±	2.50	1.71	4.61	0.08	0.05	0.14	±	5.00	3.41	9.32
	6	±	0.10	0.07	0.18	±	4.97	3.39	9.26	0.06	0.04	0.10	±	2.08	1.43	3.83
	7	±	0.06	0.04	0.11	±	2.69	1.84	4.97	0.07	0.05	0.12	±	2.43	1.66	4.47
	8	±	0.05	0.03	0.08	±	2.12	1.46	3.91	0.05	0.03	0.09	±	1.88	1.29	3.46
	9	±	0.05	0.03	0.09	±	2.97	2.04	5.50	0.04	0.03	0.08	±	1.39	0.96	2.56
	10	±	0.03	0.02	0.06	±	1.40	0.96	2.57	0.05	0.04	0.10	±	1.86	1.28	3.42
	11	±	0.02	0.02	0.04	±	1.04	0.71	1.91	0.04	0.03	0.08	±	2.47	1.63	5.09

Table 23. Within subject variability of mean power frequency measures of postural stability during a forward jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
MPFap (Hz)	2	±	2.99	2.05	5.45	±	21.00	14.01	41.62	2.44	1.68	4.45	±	14.91	10.03	28.88
	3	±	2.37	1.63	4.33	±	14.78	9.95	28.62	2.19	1.50	3.99	±	10.18	6.90	19.37
	4	±	1.81	1.25	3.31	±	12.72	8.59	24.43	0.94	0.64	1.71	±	6.27	4.27	11.74
	5	±	1.06	0.73	1.93	±	12.73	8.59	24.45	0.72	0.49	1.31	±	4.15	2.83	7.70
	6	±	1.17	0.80	2.14	±	6.94	4.73	13.04	0.60	0.41	1.09	±	3.63	2.48	6.72
	7	±	1.22	0.84	2.22	±	7.61	5.17	14.32	0.79	0.54	1.45	±	3.75	2.57	6.96
	8	±	0.78	0.54	1.43	±	10.43	7.06	19.85	0.42	0.29	0.77	±	2.55	1.75	4.71
	9	±	0.73	0.50	1.32	±	5.60	3.82	10.46	0.41	0.29	0.76	±	2.38	1.63	4.39
	10	±	0.50	0.34	0.91	±	4.07	2.78	7.55	0.63	0.43	1.14	±	2.89	1.98	5.35
	11	±	0.49	0.33	0.89	±	3.73	2.55	6.92	0.50	0.33	1.01	±	2.39	1.57	4.92
MPFml (Hz)	2	±	1.69	1.16	3.08	±	11.80	7.97	22.59	1.31	0.90	2.40	±	22.38	14.90	44.59
	3	±	1.07	0.74	1.96	±	6.60	4.50	12.39	1.22	0.84	2.23	±	13.95	9.40	26.92
	4	±	0.99	0.68	1.81	±	8.44	5.73	15.94	0.74	0.51	1.35	±	7.20	4.90	13.52
	5	±	0.85	0.59	1.56	±	7.41	5.04	13.94	0.92	0.63	1.68	±	8.92	6.05	16.87
	6	±	0.64	0.44	1.17	±	5.77	3.93	10.78	0.52	0.35	0.94	±	4.57	3.12	8.51
	7	±	0.77	0.53	1.40	±	5.87	4.00	10.97	0.17	0.12	0.32	±	1.16	0.80	2.13
	8	±	0.59	0.41	1.08	±	5.20	3.55	9.69	0.39	0.27	0.71	±	3.52	2.41	6.52
	9	±	0.48	0.33	0.88	±	4.13	2.83	7.68	0.29	0.20	0.53	±	3.45	2.36	6.39
	10	±	0.28	0.19	0.52	±	2.32	1.59	4.28	0.43	0.30	0.79	±	5.25	3.58	9.79
	11	±	0.26	0.18	0.48	±	2.32	1.59	4.27	0.34	0.23	0.70	±	4.72	3.10	9.84

Table 24. Within subject variability of root mean square measures of postural stability during a lateral jump-landing task

			Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
RMSap (m*g*)	2	±	19.70	13.55	35.97	±	10.26	6.95	19.51	16.40	11.28	29.94	±	6.44	4.39	12.07
	3	±	10.39	7.14	18.96	±	5.79	3.95	10.82	13.64	9.38	24.90	±	6.94	4.73	13.04
	4	±	6.07	4.17	11.07	±	2.64	1.81	4.87	16.41	11.29	29.97	±	7.54	5.13	14.20
	5	±	5.81	3.99	10.60	±	2.93	2.00	5.41	5.91	4.06	10.78	±	3.96	2.70	7.34
	6	±	6.65	4.57	12.13	±	2.93	2.00	5.41	8.36	5.75	15.27	±	3.47	2.37	6.42
	7	±	3.06	2.10	5.58	±	1.80	1.24	3.31	8.41	5.79	15.36	±	3.52	2.41	6.52
	8	±	7.24	4.98	13.22	±	2.42	1.66	4.46	5.47	3.76	9.98	±	2.08	1.43	3.84
	9	±	3.89	2.68	7.11	±	1.95	1.34	3.59	4.24	2.92	7.75	±	1.97	1.35	3.62
	10	±	2.37	1.63	4.34	±	1.06	0.73	1.94	4.53	3.06	8.68	±	2.32	1.56	4.50
	11	±	4.29	2.95	7.82	±	2.04	1.40	3.76	2.20	1.38	5.41	±	1.27	0.79	3.15
RMSml (m*g*)	2	±	48.07	33.06	87.75	±	21.07	14.05	41.77	39.76	27.35	72.58	±	15.99	10.74	31.10
	3	±	20.63	14.19	37.66	±	7.24	4.92	13.61	19.75	13.59	36.06	±	7.74	5.26	14.59
	4	±	18.34	12.61	33.48	±	6.06	4.13	11.35	16.56	11.39	30.24	±	5.73	3.91	10.70
	5	±	12.27	8.44	22.39	±	4.45	3.04	8.28	9.46	6.51	17.28	±	6.04	4.11	11.29
	6	±	14.10	9.70	25.74	±	4.49	3.07	8.35	7.77	5.35	14.19	±	2.66	1.82	4.92
	7	±	8.48	5.83	15.48	±	3.13	2.14	5.79	9.64	6.63	17.61	±	2.89	1.98	5.33
	8	±	6.42	4.42	11.73	±	2.33	1.60	4.30	7.42	5.10	13.55	±	2.03	1.39	3.74
	9	±	5.83	4.01	10.65	±	1.98	1.36	3.64	4.52	3.11	8.26	±	1.59	1.09	2.93
	10	±	6.70	4.61	12.24	±	2.45	1.68	4.53	3.07	2.08	5.89	±	1.03	0.70	1.99
	11	±	5.31	3.65	9.69	±	1.91	1.31	3.52	3.30	2.06	8.09	±	1.16	0.72	2.88

Table 25. Within subject variability of normalized path length measures of postural stability during a lateral jump-landing task

			Control							CAI					
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
NPLap (mg/s)	2	±	572.60	393.86	1045.35	±	12.95	8.74	24.90	700.01	481.49	1277.95	±	10.43	7.06	19.86
	3	±	234.08	161.01	427.34	±	5.21	3.56	9.72	273.16	187.89	498.68	±	4.62	3.15	8.59
	4	±	212.77	146.35	388.43	±	3.31	2.27	6.13	448.93	308.79	819.57	±	7.40	5.03	13.91
	5	±	125.54	86.35	229.18	±	2.17	1.49	4.00	145.94	100.38	266.43	±	2.86	1.96	5.28
	6	±	188.77	129.85	344.63	±	3.49	2.39	6.46	229.97	158.18	419.84	±	2.89	1.98	5.35
	7	±	77.99	53.64	142.38	±	1.28	0.88	2.34	222.30	152.91	405.84	±	2.87	1.97	5.31
	8	±	242.31	166.67	442.36	±	3.24	2.22	6.00	165.03	113.51	301.28	±	1.66	1.14	3.05
	9	±	72.03	49.54	131.50	±	1.54	1.06	2.83	65.27	44.90	119.16	±	1.02	0.70	1.87
	10	±	71.58	49.23	130.67	±	1.58	1.08	2.90	98.25	66.36	188.23	±	1.65	1.11	3.18
	11	±	99.03	68.12	180.79	±	1.96	1.35	3.62	44.03	27.48	107.98	±	0.99	0.62	2.44
NPLml (mg/s)	2	±	786.76	541.16	1436.32	±	15.56	10.46	30.22	740.15	509.10	1351.22	±	14.53	9.78	28.10
	3	±	267.79	184.19	488.87	±	4.91	3.35	9.14	298.04	205.01	544.11	±	5.56	3.79	10.37
	4	±	325.12	223.63	593.54	±	5.24	3.58	9.78	249.54	171.64	455.55	±	4.44	3.03	8.25
	5	±	110.50	76.00	201.73	±	2.18	1.49	4.01	139.87	96.21	255.34	±	3.66	2.50	6.78
	6	±	200.58	137.96	366.18	±	3.49	2.39	6.46	106.62	73.34	194.65	±	1.87	1.28	3.44
	7	±	96.45	66.34	176.09	±	1.67	1.14	3.06	166.37	114.44	303.73	±	2.68	1.84	4.95
	8	±	76.10	52.35	138.94	±	1.65	1.13	3.03	123.87	85.20	226.14	±	1.69	1.16	3.12
	9	±	97.99	67.40	178.90	±	1.78	1.22	3.26	56.65	38.97	103.42	±	1.10	0.75	2.01
	10	±	64.57	44.41	117.87	±	1.46	1.00	2.68	73.32	49.53	140.47	±	1.37	0.92	2.63
	11	±	93.81	64.53	171.27	±	1.67	1.15	3.07	52.91	33.03	129.76	±	0.94	0.59	2.32

Table 26. Within subject variability of peak to peak measures of postural stability during a lateral jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
P2Pap (m*g*)	2	±	783.39	538.84	1430.16	±	29.23	19.29	59.70	404.57	278.28	738.59	±	15.89	10.68	30.90
	3	±	241.50	166.11	440.89	±	9.37	6.35	17.76	318.11	218.81	580.75	±	7.51	5.11	14.13
	4	±	190.74	131.19	348.21	±	5.88	4.01	11.00	394.83	271.58	720.80	±	9.85	6.68	18.71
	5	±	154.85	106.51	282.69	±	5.00	3.42	9.33	135.35	93.10	247.09	±	5.51	3.76	10.28
	6	±	235.21	161.79	429.41	±	5.30	3.61	9.88	210.41	144.73	384.12	±	4.14	2.83	7.69
	7	±	89.96	61.88	164.23	±	3.00	2.05	5.55	194.99	134.12	355.97	±	5.31	3.62	9.90
	8	±	119.30	82.06	217.79	±	3.25	2.22	6.01	137.16	94.34	250.40	±	2.88	1.97	5.32
	9	±	109.28	75.17	199.50	±	2.94	2.01	5.44	73.25	50.38	133.72	±	1.43	0.98	2.63
	10	±	72.96	50.19	133.20	±	2.16	1.48	3.97	89.62	60.53	171.69	±	3.51	2.35	6.82
	11	±	98.80	67.96	180.36	±	3.29	2.25	6.09	42.94	26.80	105.32	±	1.86	1.15	4.61
P2Pml (m*g*)	2	±	1337.78	920.17	2442.26	±	50.29	32.34	110.38	613.79	422.19	1120.55	±	25.83	17.12	52.11
	3	±	461.33	317.32	842.21	±	10.97	7.42	20.93	328.86	226.20	600.37	±	9.63	6.53	18.27
	4	±	407.86	280.54	744.58	±	9.09	6.16	17.21	280.37	192.84	511.84	±	6.58	4.48	12.33
	5	±	237.68	163.49	433.92	±	6.44	4.39	12.07	190.36	130.93	347.52	±	7.62	5.18	14.35
	6	±	330.47	227.31	603.31	±	6.27	4.27	11.75	178.93	123.07	326.65	±	3.93	2.69	7.29
	7	±	180.94	124.46	330.32	±	5.01	3.42	9.34	217.95	149.91	397.89	±	3.50	2.39	6.47
	8	±	174.42	119.97	318.42	±	3.86	2.64	7.16	153.36	105.49	279.98	±	2.86	1.96	5.29
	9	±	165.31	113.70	301.79	±	3.41	2.34	6.32	90.78	62.44	165.72	±	1.73	1.19	3.19
	10	±	139.29	95.81	254.29	±	3.19	2.18	5.89	61.59	41.60	117.98	±	1.71	1.15	3.30
	11	±	100.74	69.29	183.91	±	2.63	1.80	4.85	63.44	39.60	155.58	±	1.20	0.75	2.97

Table 27. Within subject variability of stability index measures of postural stability during a lateral jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
APSI (mg /kg)	2	±	0.14	0.09	0.25	±	9.52	6.46	18.06	0.10	0.07	0.18	±	7.24	4.92	13.60
	3	±	0.10	0.07	0.18	±	5.89	4.01	11.01	0.09	0.06	0.16	±	5.50	3.75	10.27
	4	±	0.05	0.03	0.09	±	2.81	1.92	5.18	0.10	0.07	0.17	±	6.86	4.67	12.87
	5	±	0.07	0.05	0.12	±	3.56	2.44	6.60	0.06	0.04	0.11	±	5.84	3.98	10.91
	6	±	0.04	0.03	0.08	±	2.95	2.02	5.44	0.06	0.04	0.10	±	3.29	2.25	6.10
	7	±	0.02	0.02	0.04	±	1.80	1.23	3.31	0.05	0.03	0.09	±	2.90	1.98	5.35
	8	±	0.06	0.04	0.11	±	2.84	1.94	5.24	0.04	0.03	0.08	±	2.78	1.91	5.14
	9	±	0.03	0.02	0.06	±	2.11	1.45	3.89	0.03	0.02	0.06	±	2.10	1.44	3.87
	10	±	0.01	0.01	0.03	±	0.80	0.55	1.46	0.02	0.02	0.04	±	1.89	1.27	3.66
	11	±	0.05	0.03	0.09	±	2.49	1.71	4.59	0.01	0.01	0.02	±	0.89	0.55	2.19
MLSI (mg /kg)	2	±	0.33	0.23	0.61	±	21.00	14.01	41.63	0.24	0.16	0.44	±	16.32	10.96	31.78
	3	±	0.13	0.09	0.23	±	6.56	4.47	12.30	0.12	0.08	0.22	±	6.32	4.30	11.83
	4	±	0.14	0.09	0.25	±	6.41	4.37	12.02	0.10	0.07	0.18	±	5.98	4.08	11.19
	5	±	0.09	0.06	0.16	±	4.92	3.36	9.17	0.13	0.09	0.24	±	6.94	4.72	13.03
	6	±	0.09	0.06	0.16	±	4.35	2.97	8.08	0.05	0.04	0.10	±	2.92	2.00	5.40
	7	±	0.06	0.04	0.12	±	3.42	2.34	6.32	0.05	0.04	0.10	±	2.95	2.02	5.44
	8	±	0.04	0.03	0.08	±	2.57	1.76	4.74	0.04	0.03	0.08	±	2.31	1.58	4.26
	9	±	0.06	0.04	0.11	±	2.49	1.71	4.59	0.05	0.03	0.09	±	2.60	1.78	4.79
	10	±	0.04	0.03	0.07	±	2.19	1.50	4.04	0.03	0.02	0.05	±	1.35	0.91	2.61
	11	±	0.05	0.03	0.08	±	2.03	1.39	3.73	0.04	0.03	0.10	±	0.96	0.60	2.36

Table 28. Within subject variability of mean power frequency measures of postural stability during a lateral jump-landing task

		Control							CAI						
Variable	Trials (n)		TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}	TE_n	TE_{LCL}	TE_{UCL}		%CV	CV_{LCL}	CV_{UCL}
MPFap (Hz)	2	±	3.38	2.33	6.17	±	37.87	24.72	79.73	1.99	1.37	3.63	±	27.09	17.93	54.91
	3	±	1.40	0.97	2.56	±	11.57	7.82	22.12	0.97	0.67	1.77	±	10.42	7.06	19.84
	4	±	1.46	1.00	2.66	±	11.76	7.94	22.50	0.88	0.61	1.61	±	5.66	3.86	10.57
	5	±	0.65	0.45	1.19	±	10.98	7.43	20.95	1.02	0.70	1.86	±	7.85	5.33	14.79
	6	±	0.67	0.46	1.23	±	6.50	4.42	12.18	0.94	0.65	1.72	±	7.15	4.87	13.44
	7	±	0.39	0.27	0.70	±	4.14	2.83	7.69	0.38	0.26	0.69	±	4.73	3.23	8.79
	8	±	0.83	0.57	1.52	±	5.98	4.07	11.18	0.26	0.18	0.48	±	2.79	1.91	5.16
	9	±	0.34	0.23	0.62	±	2.61	1.79	4.82	0.35	0.24	0.63	±	4.32	2.95	8.02
	10	±	0.43	0.30	0.79	±	4.78	3.27	8.90	0.38	0.26	0.73	±	4.96	3.33	9.73
	11	±	0.40	0.28	0.73	±	5.41	3.69	10.10	0.33	0.20	0.80	±	3.47	2.15	8.73
MPFml (Hz)	2	±	4.01	2.76	7.32	±	45.48	29.41	98.24	1.42	0.98	2.60	±	20.70	13.81	40.97
	3	±	1.75	1.20	3.20	±	12.42	8.39	23.83	1.11	0.76	2.02	±	12.71	8.58	24.42
	4	±	0.79	0.54	1.43	±	6.15	4.19	11.51	0.86	0.59	1.57	±	8.29	5.63	15.65
	5	±	0.62	0.43	1.14	±	5.00	3.42	9.32	0.30	0.20	0.54	±	3.83	2.62	7.10
	6	±	0.83	0.57	1.51	±	6.22	4.24	11.64	0.42	0.29	0.76	±	4.09	2.80	7.60
	7	±	0.59	0.41	1.08	±	4.45	3.04	8.28	0.32	0.22	0.58	±	3.63	2.48	6.72
	8	±	0.61	0.42	1.11	±	4.23	2.89	7.86	0.40	0.27	0.73	±	3.47	2.38	6.43
	9	±	0.29	0.20	0.53	±	1.91	1.31	3.51	0.28	0.19	0.50	±	4.00	2.73	7.42
	10	±	0.41	0.28	0.75	±	2.67	1.83	4.92	0.22	0.15	0.42	±	4.65	3.12	9.10
	11	±	0.27	0.19	0.50	±	2.42	1.66	4.46	0.08	0.05	0.20	±	1.11	0.69	2.74

B.3 INTERSESSION RELIABILITY

Intersession reliability was assessed across sessions using intraclass correlation coefficients, ICC(2,1). ICCs, 95% confidence intervals (CI), and standard error in the measurement (SEM) values for all measures extracted from the ten postural stability tasks in both the control and CAI groups are presented in Table 29 through Table 33. The results are described in detail in section 3.3.3.

Table 29. Intersession reliability for accelerometry measures of double leg stance on firm surface tasks

	Control				CAI			
	ICC	95% CI		SEM	ICC	95% CI		SEM
DLEO								
RMSap	0.50	-0.05	0.85	0.54	0.77	0.35	0.94	0.25
RMSml	0.82	0.50	0.95	0.05	0.79	0.41	0.94	0.16
NPLap	0.29	-1.37	0.82	0.81	0.26	-1.22	0.80	5.10
NPLml	0.68	0.35	0.90	1.14	0.48	-0.41	0.86	6.57
P2Pap	0.58	-0.03	0.88	2.55	0.58	-0.14	0.88	6.15
P2Pml	0.65	0.08	0.90	0.96	0.80	0.42	0.95	1.49
APSI	0.71	0.22	0.92	0.01	0.67	0.01	0.91	0.01
MLSI	0.79	0.42	0.94	0.00	0.73	0.24	0.92	0.01
MPFap	0.29	-0.45	0.77	0.45	0.54	-0.37	0.88	0.13
MPFml	-0.19	-2.19	0.67	0.53	0.56	-0.35	0.88	0.16
DLEC								
RMSap	0.80	0.44	0.95	0.16	0.69	0.13	0.91	0.33
RMSml	0.57	-0.21	0.88	0.08	0.57	-0.16	0.88	0.27
NPLap	-0.16	-3.18	0.71	1.24	-0.24	-3.70	0.69	3.60
NPLml	0.80	0.46	0.95	1.32	-0.60	-4.75	0.60	9.97
P2Pap	0.75	0.28	0.93	0.59	-0.30	-2.81	0.65	9.31
P2Pml	0.09	-1.25	0.74	2.48	0.04	-1.89	0.74	6.21
APSI	0.84	0.53	0.96	0.00	0.60	-0.09	0.89	0.01
MLSI	0.80	0.45	0.95	0.00	0.87	0.63	0.97	0.00
MPFap	0.80	0.43	0.94	0.10	0.15	-1.67	0.78	0.16
MPFml	0.61	-0.17	0.90	0.16	0.78	0.39	0.94	0.31

Table 30. Intersession reliability for accelerometry measures of double leg stance on foam surface tasks

	Control				CAI			
	ICC	95% CI		SEM	ICC	95% CI		SEM
DLEOF								
RMSap	0.88	0.64	0.97	0.07	0.77	0.31	0.94	0.22
RMSml	0.91	0.76	0.98	0.07	0.83	0.51	0.95	0.10
NPLap	0.29	-1.08	0.81	5.11	0.07	-1.46	0.74	9.05
NPLml	0.34	-0.94	0.82	6.35	0.09	-1.39	0.74	11.75
P2Pap	0.90	0.69	0.97	0.18	0.79	0.36	0.94	0.59
P2Pml	0.84	0.56	0.96	0.89	0.73	0.22	0.93	1.67
APSI	0.68	0.04	0.92	0.01	0.43	-0.90	0.85	0.00
MLSI	0.93	0.80	0.98	0.00	0.87	0.63	0.97	0.00
MPFap	0.55	-0.29	0.88	0.12	0.66	0.06	0.91	0.17
MPFml	0.78	0.37	0.94	0.12	0.86	0.59	0.96	0.18
DLECF								
RMSap	0.76	0.36	0.93	0.38	0.77	0.37	0.94	0.46
RMSml	0.82	0.48	0.95	0.10	0.85	0.56	0.96	0.13
NPLap	0.33	-0.97	0.82	4.86	0.50	-0.40	0.86	4.49
NPLml	0.49	-0.43	0.86	5.99	0.52	-0.40	0.87	4.41
P2Pap	0.78	0.40	0.94	1.98	0.88	0.64	0.97	0.94
P2Pml	0.86	0.59	0.96	0.22	0.79	0.38	0.94	0.94
APSI	0.48	-0.46	0.86	0.01	0.52	-0.29	0.87	0.02
MLSI	0.74	0.23	0.93	0.01	0.76	0.32	0.93	0.01
MPFap	0.41	-0.60	0.83	0.18	0.28	-0.64	0.78	0.19
MPFml	0.74	0.29	0.93	0.14	0.74	0.25	0.93	0.08

Table 31. Intersession reliability for accelerometry measures of tandem stance tasks

	Control				CAI			
	ICC	95% CI		SEM	ICC	95% CI		SEM
TANEO								
RMSap	0.65	0.02	0.90	0.42	0.73	0.27	0.93	0.66
RMSml	0.66	0.02	0.91	0.44	0.79	0.40	0.94	0.54
NPLap	0.84	0.51	0.96	1.00	0.69	0.08	0.92	3.52
NPLml	0.74	0.22	0.93	4.51	0.82	0.50	0.95	8.68
P2Pap	0.34	-0.72	0.81	7.33	0.58	-0.20	0.89	4.69
P2Pml	0.39	-0.77	0.84	6.69	0.85	0.55	0.96	2.39
APSI	0.39	-0.53	0.83	0.03	0.71	0.23	0.92	0.02
MLSI	0.58	-0.05	0.88	0.02	0.84	0.54	0.96	0.01
MPFap	0.56	-0.31	0.88	0.12	0.62	0.05	0.89	0.26
MPFml	0.81	0.46	0.95	0.20	0.85	0.58	0.96	0.10
TANEC								
RMSap	0.88	0.67	0.97	0.55	0.81	0.44	0.95	0.20
RMSml	0.85	0.57	0.96	0.46	0.91	0.76	0.98	0.58
NPLap	0.95	0.87	0.99	1.34	0.88	0.66	0.97	4.53
NPLml	0.95	0.87	0.99	1.36	0.96	0.88	0.99	4.96
P2Pap	0.79	0.40	0.94	6.12	0.69	0.01	0.92	0.92
P2Pml	0.83	0.52	0.95	7.02	0.88	0.66	0.97	4.31
APSI	0.86	0.60	0.96	0.01	0.51	-0.60	0.87	0.00
MLSI	0.77	0.33	0.94	0.01	0.75	0.30	0.93	0.03
MPFap	0.79	0.41	0.94	0.08	0.51	-0.55	0.87	0.05
MPFml	0.78	0.36	0.94	0.09	0.86	0.59	0.96	0.06

Table 32. Intersession reliability for accelerometry measures of single leg stance tasks

	Control				CAI			
	ICC	95% CI		SEM	ICC	95% CI		SEM
SLEO								
RMSap	0.50	-0.45	0.86	0.40	0.65	0.00	0.91	0.62
RMSml	0.64	-0.03	0.90	0.36	0.84	0.52	0.96	0.11
NPLap	0.90	0.73	0.97	0.67	0.88	0.64	0.97	0.91
NPLml	0.89	0.68	0.97	3.16	0.88	0.65	0.97	3.21
P2Pap	0.30	-1.39	0.82	0.73	0.76	0.32	0.93	2.90
P2Pml	0.69	0.11	0.92	1.96	0.73	0.22	0.93	3.78
APSI	0.12	-1.60	0.76	0.01	0.55	-0.22	0.88	0.02
MLSI	0.49	-0.48	0.86	0.01	0.75	0.29	0.93	0.01
MPFap	0.62	-0.02	0.90	0.15	0.14	-1.86	0.78	0.12
MPFml	0.76	0.28	0.94	0.06	0.71	0.13	0.92	0.07
SLEC								
RMSap	0.83	0.52	0.95	1.26	0.77	0.30	0.94	0.43
RMSml	0.86	0.60	0.96	1.58	0.70	0.07	0.92	0.41
NPLap	0.89	0.67	0.97	7.10	0.84	0.62	0.95	6.96
NPLml	0.91	0.75	0.98	15.41	0.95	0.84	0.99	0.90
P2Pap	0.88	0.65	0.97	5.33	0.61	-0.22	0.90	4.06
P2Pml	0.85	0.57	0.96	12.58	0.52	-0.47	0.87	10.54
APSI	0.74	0.28	0.93	0.02	0.73	0.23	0.93	0.01
MLSI	0.81	0.48	0.95	0.03	0.78	0.34	0.94	0.01
MPFap	0.65	-0.02	0.91	0.09	0.87	0.64	0.97	0.08
MPFml	0.85	0.57	0.96	0.08	0.93	0.80	0.98	0.02

Table 33. Intersession reliability for accelerometry measures of jump-landing tasks

	Control				CAI			
	ICC	95% CI		SEM	ICC	95% CI		SEM
DPS-AP								
RMSap	0.91	0.73	0.98	1.40	0.77	0.33	0.94	9.31
RMSml	0.61	-0.05	0.89	21.69	0.90	0.69	0.97	1.15
NPLap	0.90	0.71	0.97	74.60	0.83	0.51	0.95	223.30
NPLml	0.53	-0.41	0.87	198.62	0.91	0.74	0.98	34.67
P2Pap	0.82	0.47	0.95	43.60	0.72	0.21	0.92	347.75
P2Pml	0.66	0.01	0.91	293.88	0.90	0.70	0.97	102.03
APSI	0.86	0.60	0.96	0.04	0.64	-0.05	0.90	0.05
MLSI	0.27	-0.88	0.79	0.20	0.48	-0.63	0.86	0.07
MPFap	0.55	-0.40	0.88	0.78	0.89	0.66	0.97	0.31
MPFml	0.92	0.76	0.98	0.42	0.46	-0.42	0.85	1.70
DPS-ML								
RMSap	0.94	0.82	0.98	3.03	0.74	0.29	0.93	12.50
RMSml	0.76	0.25	0.94	1.39	0.94	0.82	0.98	2.43
NPLap	0.87	0.62	0.96	154.43	0.81	0.48	0.95	279.37
NPLml	0.87	0.61	0.96	61.69	0.96	0.90	0.99	11.56
P2Pap	0.94	0.82	0.98	42.07	0.87	0.64	0.97	121.06
P2Pml	0.84	0.52	0.96	58.30	0.95	0.85	0.99	9.34
APSI	0.88	0.63	0.97	0.01	0.37	-0.67	0.82	0.09
MLSI	0.81	0.41	0.95	0.01	0.25	-1.32	0.80	0.11
MPFap	0.57	-0.37	0.89	0.25	0.87	0.60	0.96	0.15
MPFml	0.95	0.87	0.99	0.23	0.89	0.66	0.97	0.12

APPENDIX C

PAIRWISE COMPARISONS OF POSTURAL STABILITY TASKS

Significant main effects were observed in the Friedman's test across each of the accelerometry measures when analyzing all static tasks together and when analyzing all tasks (Section 4.3). Post hoc pairwise comparisons were performed between tasks for each of the COM acceleration measures (Table 34). Significance level was adjusted based on nine comparisons.

Table 34. Pairwise comparisons of accelerometry measures of postural stability

		RMSap	RMSml	NPLap	NPLml	P2Pap	P2Pml	APSI	MLSI	MPFap	MPFml
Tasks Compared		p values									
DLEO	DLEC	0.989	0.638	0.861	0.459	0.696	0.798	0.183	0.143	0.109	0.158
DLEC	DLEOF	0.006	0.000*	0.000*	0.000*	0.000*	0.000*	0.264	0.000*	0.104	0.000*
DLEOF	DLECF	0.000*	0.000*	0.000*	0.001*	0.000*	0.004*	0.003*	0.000*	0.003*	0.002*
DLECF	TANEO	0.020	0.221	0.122	0.001*	0.313	0.021	0.716	0.040	0.009	0.581
TANEO	TANEC	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.008	0.000*	0.288	0.026
TANEC	SLEO	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.397	0.042
SLEO	SLEC	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.339	0.657
SLEC	DPS-AP	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*	0.000*
DPS-AP	DPS-ML	0.001*	0.000*	0.001*	0.001*	0.000*	0.000*	0.011	0.003*	0.000*	0.150

*Significant difference bweteen tasks ($p < 0.006$)

APPENDIX D

DISCRIMINATORY ANALYSIS IN CONTROL AND CAI GROUPS

D.1 ACCELERATION TRACES IN CONTROL AND CAI GROUPS ACROSS STATIC POSTURAL STABILITY TASKS

Anterior-posterior (AP) and medial-lateral (ML) acceleration traces across eight static postural stability tasks are shown in Figure 21 through Figure 24. The traces are from one representative control and one representative CAI participant. Data shown have been filtered and transformed according to methods described in Section 2.2.2 and Section 2.2.3.

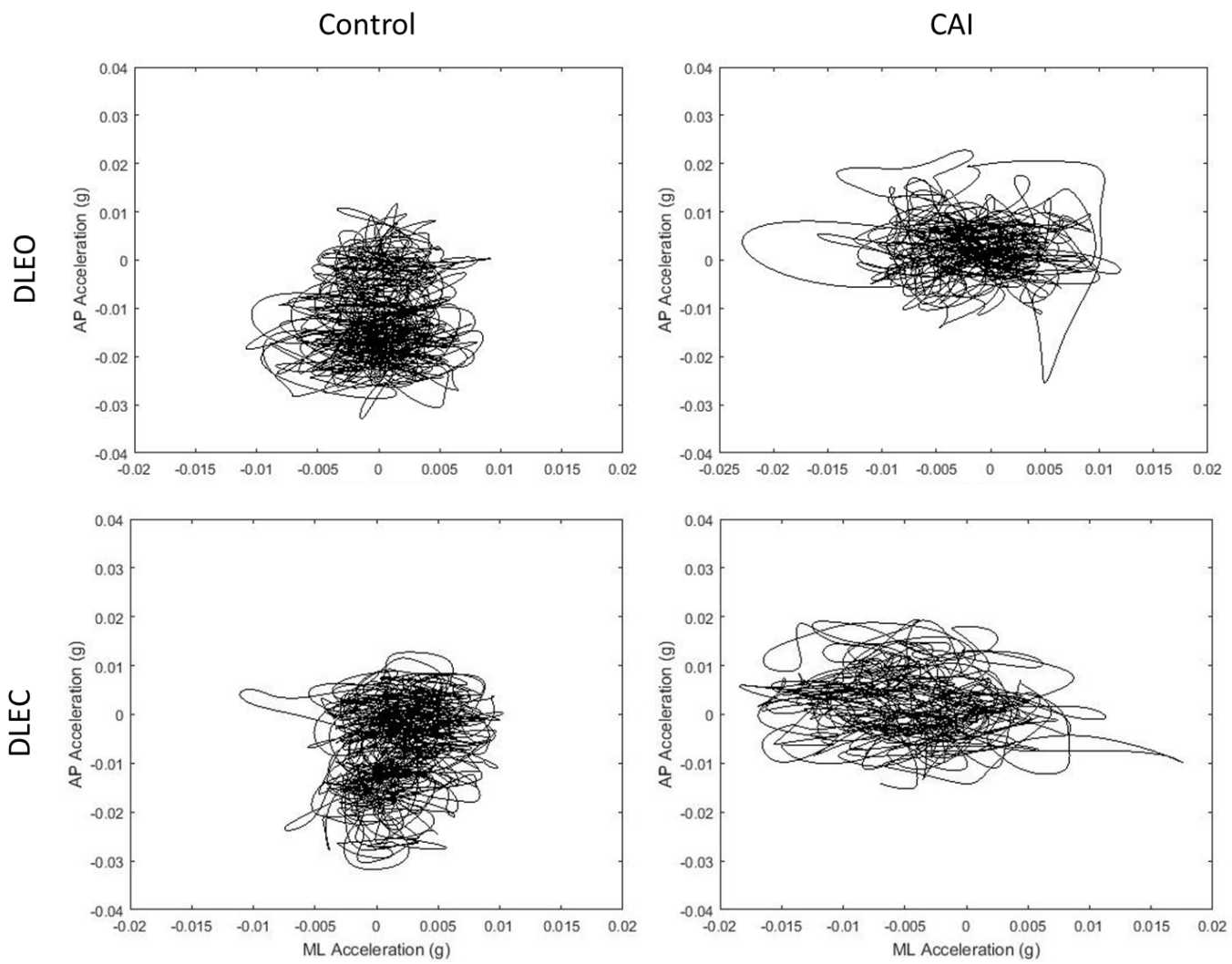

Figure 21. Acceleration trace in double-leg stance on firm surface tasks. Representative acceleration traces from one control participant and one CAI participant during the double-leg stance with eyes open (DLEO) and double-leg stance with eyes closed (DLEC) tasks.

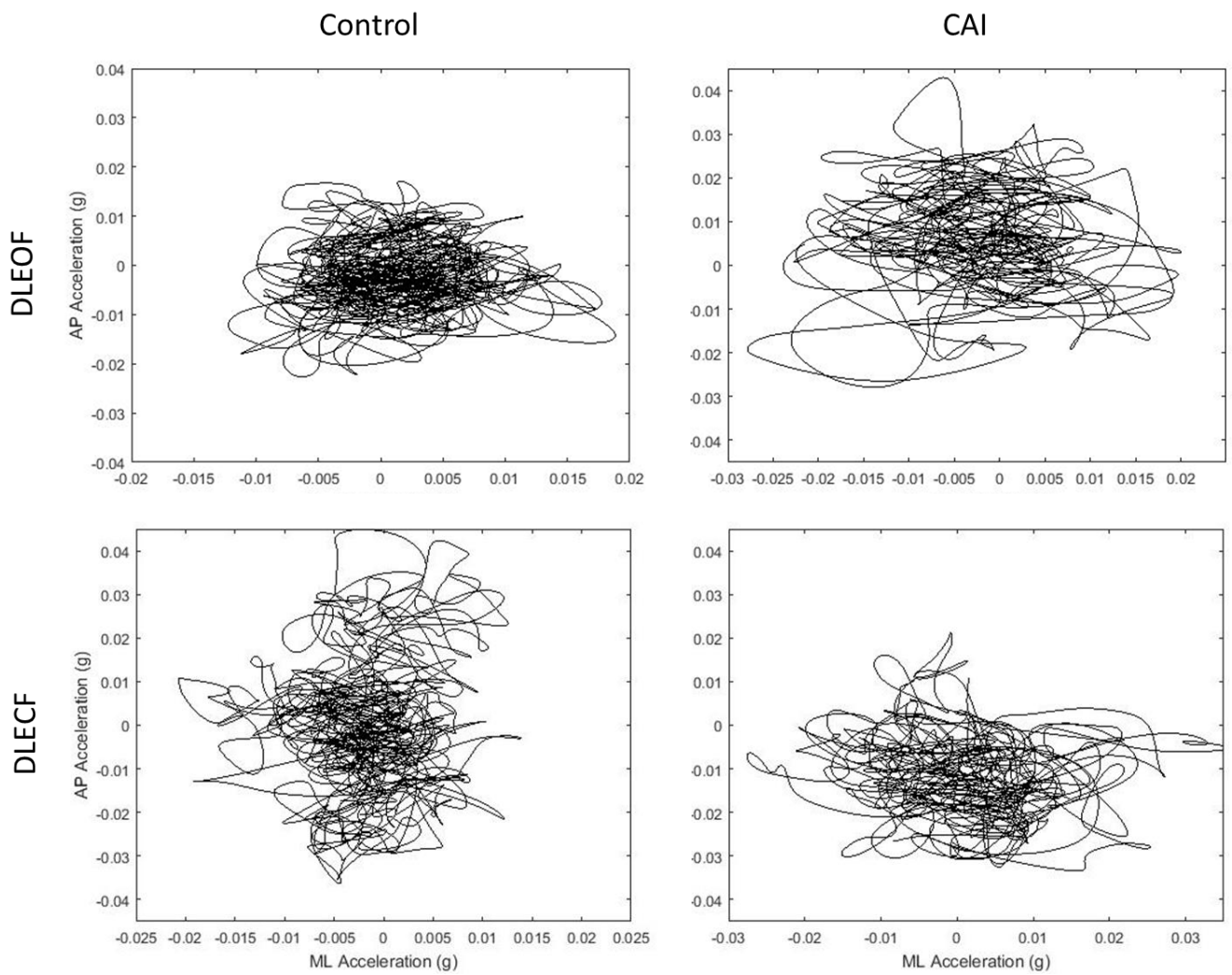

Figure 22. Acceleration trace in double-leg stance on foam surface tasks. Representative acceleration traces from one control participant and one CAI participant during the double-leg stance with eyes open on foam (DLEOF) and double-leg stance with eyes closed on foam (DLECF) tasks.

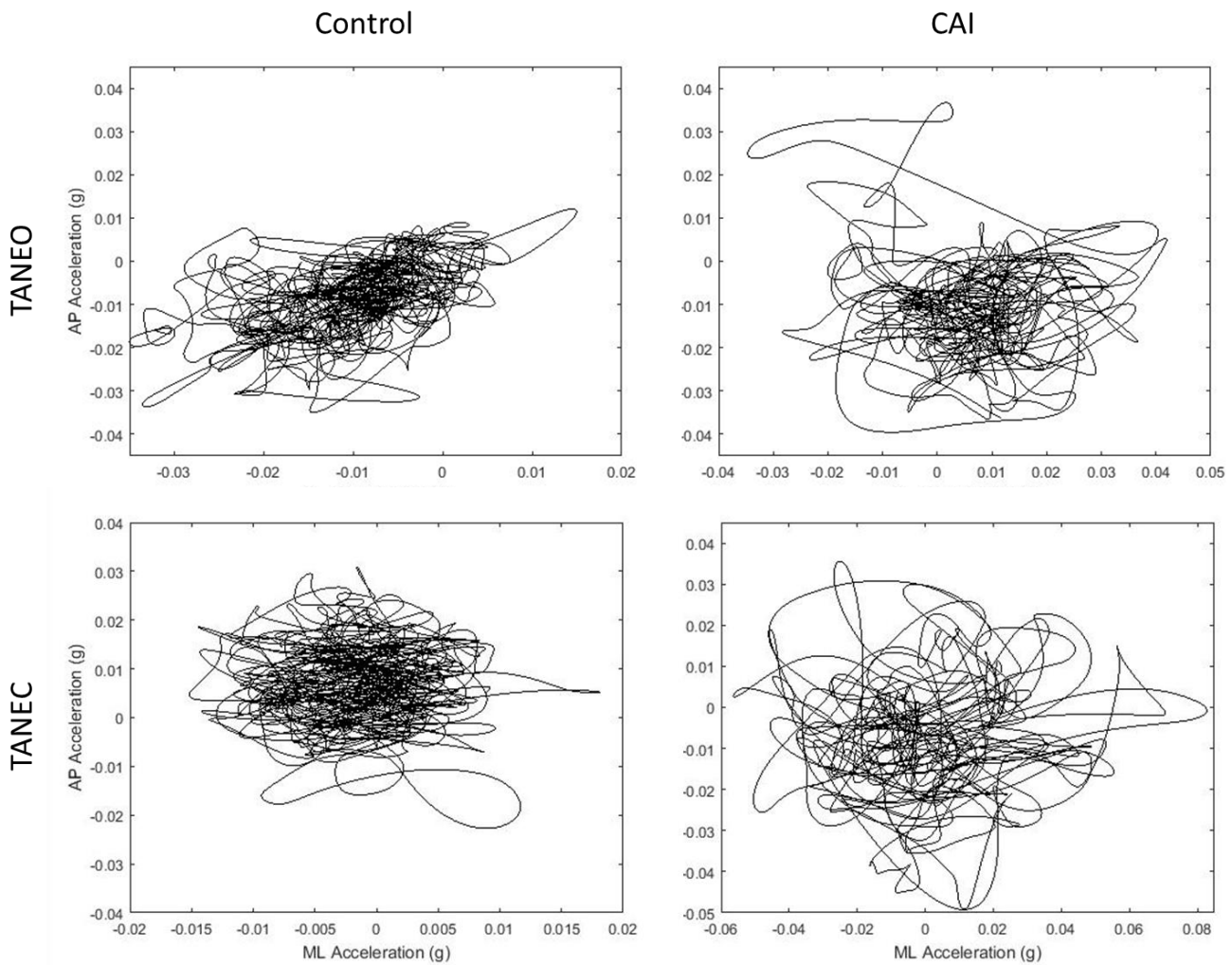

Figure 23. Acceleration trace in tandem stance tasks. Representative acceleration traces from one control participant and one CAI participant during the tandem stance with eyes open (TANEO) and tandem stance with eyes closed (TANEC) tasks.

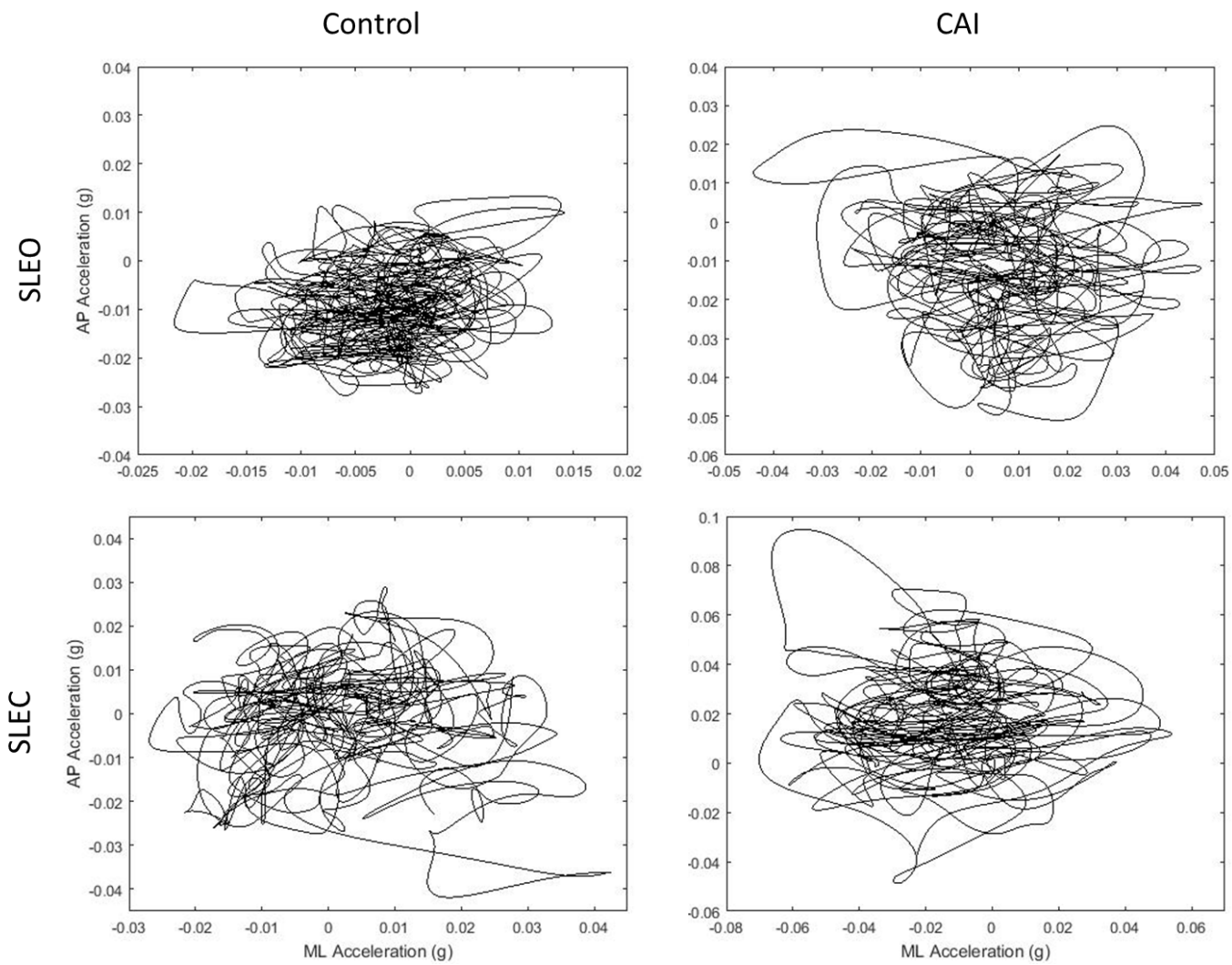

Figure 24. Acceleration trace in single-leg stance tasks. Representative acceleration traces from one control participant and one CAI participant during the single-leg stance with eyes open (SLEO) and single-leg stance with eyes closed (SLEC) tasks.

D.2 RECEIVER OPERATING CHARACTERISTIC CURVES

Receiver operating characteristic (ROC) curves for the single leg stance with eyes open (SLEO) and eyes closed (SLEC) and the single-leg jump-landing tasks in the anterior-posterior (DPS-AP) and medial-lateral (DPS-ML) directions were generated from the sensitivity and 1-specificity of the postural stability measures (Figure 25 and Figure 26). Area under the curve was found for each measure and task and is described in detail in Section 5.3.

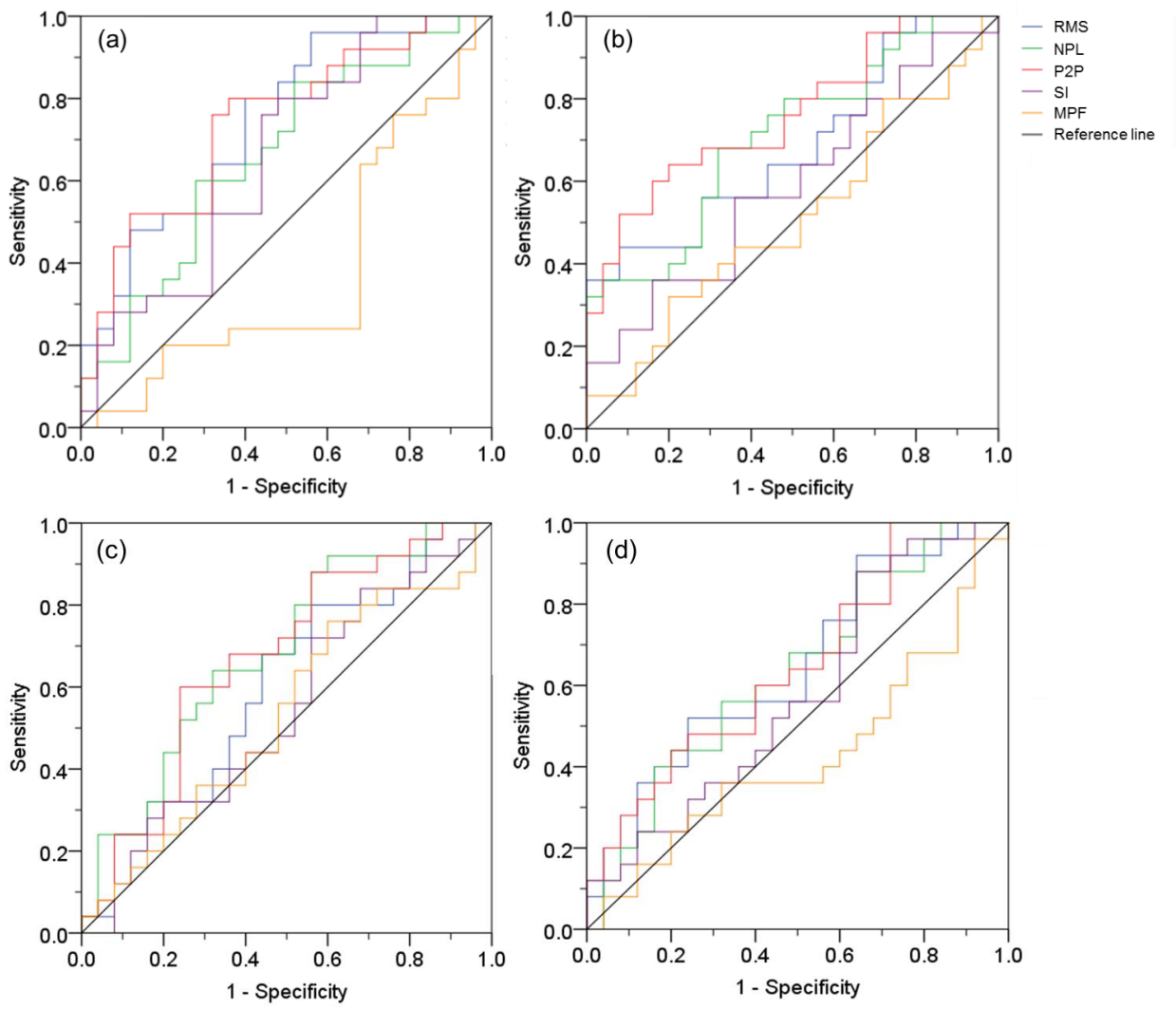

Figure 25. ROC curves for accelerometry measures during single-leg stance tasks. (a) Measures derived from the anterior-posterior acceleration signal during a single-leg stance with eyes open. (b) Measures derived from the medial-lateral acceleration signal during a single-leg stance with eyes open. (c) Measures derived from the anterior-posterior acceleration signal during a single-leg stance with eyes closed. (d) Measures derived from the medial-lateral acceleration signal during a single-leg stance with eyes closed.

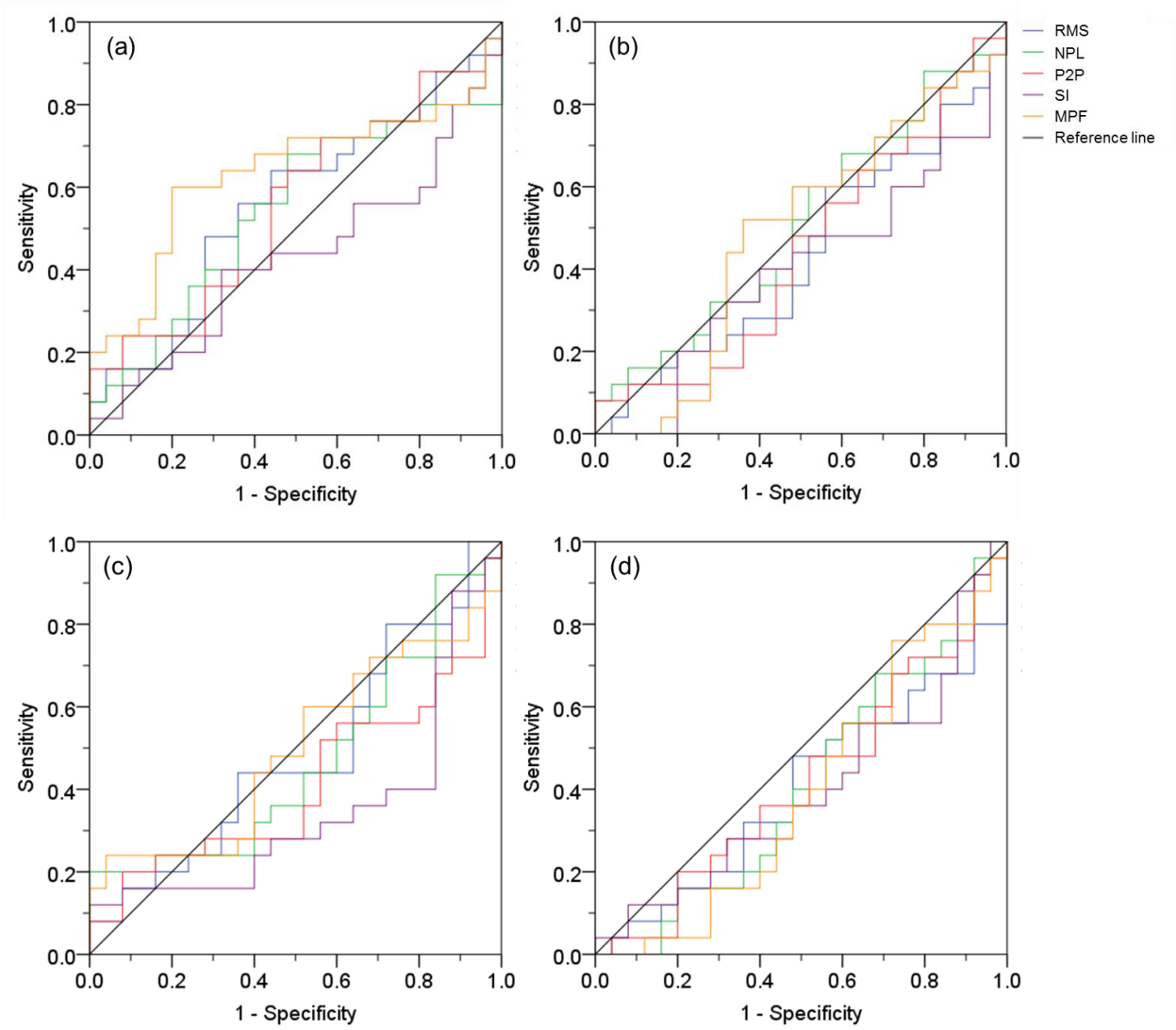

Figure 26. ROC curves for accelerometry measures during jump-landing tasks. (a) Measures derived from the anterior-posterior acceleration signal during a forward jump-landing dynamic postural stability task. (b) Measures derived from the medial-lateral acceleration signal during a forward jump-landing dynamic postural stability task. (c) Measures derived from the anterior-posterior acceleration signal during a lateral jump-landing dynamic postural stability task. (d) Measures derived from the medial-lateral acceleration signal during a lateral jump-landing dynamic postural stability task.

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