Full length Article

Reliability and validity of an accelerometry based measure of static and dynamic postural stability in healthy and active individuals

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ABSTRACT

Postural stability is an important measure in both research and clinical practice. A portable, easy to use device that can provide higher resolution than current clinical tests may allow for better identification of patients or athletes with postural stability deficits. The purpose of this study was to evaluate the ability of a tri-axial accelerometer to quantify postural stability in a healthy athletic population. Ten subjects were recruited to determine the reliability of the accelerometer to measure dynamic postural stability and thirteen were recruited to compare the accelerometer measures across tasks of varying difficulty. Subjects were asked to complete four static postural stability tasks with eyes open and eyes closed and two dynamic postural stability tasks for a total of ten tasks. During each task postural stability was measured using a tri-axial accelerometer and force platform. Differences between postural stability scores between tasks and the correlation between the two measures were assessed. The accelerometer demonstrated moderate to good test–retest reliability (ICC = 0.732 to 0.899). Only the medial–lateral axis of the accelerometer showed significant differences between static tasks but all directions were able to show significant differences between static and dynamic tasks. Additionally, Spearman's ranked correlations showed little to no correlation between the accelerometer and force platform scores. Accelerometers are a reliability tool for postural stability that measure low difficulty tasks best in the medial–lateral direction. Low correlation between the accelerometer and force platform suggest that these two methods are not measuring the same components of postural stability.

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1. Introduction

Postural stability has been defined as the ability to maintain and control one's center of gravity, or maintain equilibrium within the limits of stability, over a base of support [1]. The success of this process is the result of coordination and synergy between the vestibular, visual, and somatosensory systems [1–3]. Decrements in postural stability have been shown to have a relationship with a history of musculoskeletal and/or neurological injury [4–6] and have been able to predict lower extremity injury [2,7]. However, the complex nature of postural stability, specifically dynamic postural stability, has made it difficult to measure in a clinical setting. Current clinical methods and instrumentation for measuring dynamic postural stability do exist; however, they have limitations [8]. Instrumentation and methods that offer portability and ease of use but also provides the necessary resolution and discriminatory ability for quantifying dynamic postural stability would be beneficial to both clinicians and researchers.

Postural stability can be categorized into static and dynamic postural stability. Static postural stability has been referred to as a person's ability to maintain a steady standing posture over a static base of support [9]. Conversely, dynamic postural stability can be defined as the ability to transfer and control the projection of one's center of mass over a base of support while transitioning from a dynamic to static state [9,10]. Previous research has demonstrated no relationship between different static and dynamic measures of postural stability, suggesting that static and dynamic tasks may be targeting different afferent and efferent pathways [11,12]. For this reason it is important to use both static and dynamic assessments when evaluating possible deficits in postural stability.

Force platforms are frequently used to assess and quantify static postural stability using a broad range of algorithms [13]. In an athletic population, static postural stability is commonly evaluated by measuring the excursion of the center of pressure (COP) [9,14] or the standard deviations of ground reaction forces (GRFs)

[9,10,12,15]. Dynamic postural stability has been evaluated by measuring time to stabilization (TTS) [15–18] or calculating the dynamic postural stability index (DPSI) based on force platform data [4,10,19]. Although these measures are reliable and valid, force platform systems are expensive and reduce researchers' ability to test outside of laboratory settings [9,10].

Other methods of evaluating postural control that are more portable and simple to use have been validated and are currently used in clinical practice. Two commonly used postural stability tests that have been shown to be reliable and are used in sports medicine practice are the balance error scoring system (BESS) [20,21] and the star excursion balance test (SEBT) [8,22]. The BESS test uses a single-leg balance task and counts the number of corrective actions occurring which is scored as an ordinal rather than a continuous variable. The SEBT also uses a single-leg stance but measures the cumulative reach distance of the opposite foot in three directions [22]. This task does provide a continuous measure, offering increased measure resolution, and has been proposed to be a dynamic measure of postural stability. However, it does not necessarily simulate athletic tasks and can be limited by factors that may not be directly related to postural control systems such as available ankle dorsiflexion range of motion. Although some dynamic tests have been validated in a clinical population, it may not provide enough resolution or discriminatory capabilities to identify risk of injury in healthy or athletic populations. There appears to be a need for technology or assessment tools capable of quantifying dynamic postural stability that are portable, easy to use, and quick. Accelerometers may be an appropriate instrument to meet these demands.

The use of trunk accelerations to measure static postural stability has been shown to be reliable [23,24] and has been previously used to quantify postural stability [25–27]. Most recently, Dalton et al. [26] determined that an accelerometer-based sensor is capable of distinguishing differences in static postural stability between manifest and pre-manifest Huntington's disease groups and healthy control. In order to use this technology in a different population, such as athletes, similar studies using athletic individuals are necessary. Additionally, athletic individuals may require more challenging and sport-specific tasks, such as balancing upon landing, to better discern those with poor postural stability. The ability of an accelerometer-based sensor to measure such dynamic postural stability is still unknown.

A portable, easy to use device that can provide higher resolution than current clinical tests of postural stability may allow for better identification of patients or athletes with dynamic postural stability deficits. Therefore, the purpose of this study was to evaluate the ability of a tri-axial accelerometer to quantify postural stability in a healthy and athletic population. The first aim of this study was to determine if an accelerometry-based measure of dynamic postural stability is reliable. The second aim of this study is to assess the validity of data collected by an accelerometer placed at the approximate center of mass (COM) by assessing its ability to distinguish between balance tasks of varying difficulty. Validity was further assessed by measuring the relationship between postural stability concurrently collected with the tri-axial accelerometer and a force platform. It was hypothesized that the dynamic measure of postural stability with the accelerometer will demonstrate good to excellent test–retest reliability. It was also hypothesized that the accelerometer measures would be able detect differences in postural stability scores between tasks of varying difficulty and the accelerometer and force platform postural stability scores would display good correlation coefficients [28]. If the hypotheses of the current study are correct then researchers will have a tool capable of quantifying postural stability in an athletic population and increase the ability to conduct testing outside of the laboratory setting.

2. Methods

2.1. Subjects

Subjects for this study were recruited from a recreationally active population defined as participating in physical activity a minimum of 3 days per week for at least 30 min each session. Any subject who reported a history of lower extremity fracture or surgery to their dominant leg, lower extremity injury within the last six months, or sustained a concussion within the last three months was excluded. For the reliability aim of this study ten healthy male subjects were recruited. Previous research has demonstrated ICC values of 0.74 for RMS calculations at the center of mass using an accelerometer [24]. Based on sample size estimation tables from Walter et al. [29] at least 10 subjects will be needed to fins significant ICC values of 0.7 with two testing sessions [29]. For the validity portion of aim of this study thirteen healthy males were recruited. All subjects completed a written informed consent prior to participation in accordance with the Institutional Review Board. Demographics for each group are shown in Table 1.

2.2. Materials

Ground reaction force data were collected using a force platform (Type 9286BA, 60 cm × 40 cm platform; Kistler Instrument Corp., Amherst, NY) with an onboard amplifier. The analog signal from the force platform was converted to a digital signal and acquired using Nexus software (Vicon Motion Systems, Centennial, CO) at 1000 Hz. The force platform was mounted flush with the surrounding custom-built flooring from which the subjects were asked to jump from during the dynamic postural control tasks.

Center of mass accelerations were measured using a wireless, custom-built accelerometer (ZeroPoint Technologies, Johannesburg, South Africa). The device weighs approximately 31 g and is 4.2 × 3.9 × 1.2 cm in size. This tri-axial accelerometer sensor node consists of three orthogonal (±16 × g) uni-axial microelectromechanical system (MEMS) accelerometers (Model: ADXL78; Analog Devices Inc., Norwood, MA), a buffer, amplification unit, and microcontroller. An on-board SD-card gives the accelerometer the capability of storing up to 2 h of continuous data. Acceleration data was also collected at 1000 Hz. The accelerometer was attached to the subject using a Neoprene belt and positioned over L5 so the vertical axis was in-line with the spine and the horizontal axes (anterior–posterior and medial–lateral) aligned in the transverse plane of the pelvis.

2.3. Procedure

Previous research as already established the reliability of force platform measures of static and dynamic postural stability [9,12]. Goldie et al. [9] determined that the reliability of static postural stability tasks used in the current study have test–retest interclass correlation coefficients (ICC) of 0.31–0.85. Sell et al. [12] determined that the reliability of anterior–posterior (AP) and medial–lateral (ML) DPSI using force platforms have ICC values of 0.86 and 0.90, respectively, with a SEM of 0.01 [12]. Additionally,

Table 1
Subject demographics.

	Age (years)	Height (cm)	Weight (kg)
	Mean (SD)	Mean (SD)	Mean (SD)
Reliability group (n = 10)	24.3 (4.2)	176.7 (4.8)	76.2 (9.7)
Validity group (n = 13)	24.1 (3.1)	180.4 (3.8)	81.2 (8.8)

Mean and standard deviation (SD) values by group.

the reliability of static postural stability using COM accelerations has been previously established to have ICC values ranging from 0.20 to 0.84 [23].

The subjects enrolled in the reliability portion of this study completed three successful DPSI-AP tasks on 2 different days separated by a minimum of 48 h. The DPSI-AP task requires a forward broad jump of a distance equal to 40% of the subject's height over a 30.5 cm hurdle and a single leg landing [12]. Each subject was given at least three practice trials to familiarize him or herself with each task. Trials were discarded and repeated if the subject hopped, touched down with the opposite foot, or pivoted the landing foot during the landing. To control for a potential order affect the order of the tasks was randomized.

To examine validity each subject completed five successful trials of eight different static postural stability tasks and two dynamic postural stability tasks. Static postural stability tasks included double-leg stance (DL), double leg stance on an Airex® Pad (Airex Corp., Somersworth, NH) (DL-F), tandem stance (TAN), and single-leg (SL) stance and were completed with eyes open (EO) and with eyes closed (EC). Dynamic postural stability was performed as described in the reliability testing with the addition of a DPSI-ML task [12]. Static and dynamic postural stability tasks are shown in Fig. 1.

2.4. Data reduction

All data from the force platform and accelerometer were processed using a custom Matlab (MathWorks, Natick, MA) script and filtered using low-pass Butterworth filters with cut off frequencies of 20 Hz and 50 Hz, respectively. The standard deviation of normalized ground reaction forces (% body weight) were used to measure static postural stability with the force platform. The dynamic postural stability index (DPSI) was calculated for the filtered force platform data based on Wikstrom et al. [10] for the forward jump (DPSI-AP) and lateral jump (DPSI-ML). This calculation creates a stability index for each anatomical direction and the resultant vector of all three directions over a three second period beginning with initial contact with the force platform, defined by a threshold of greater than 5.0% of body weight.

$$\text{DPSI} = \left(\frac{\sqrt{\sum\left\{(0 - GRFx)^2 + \sum(0 - GRFy)^2 + \sum(body\ weight - GRFz)^2\right\}/number\ of\ samples}}{body\ weight} \right) \tag{1}$$
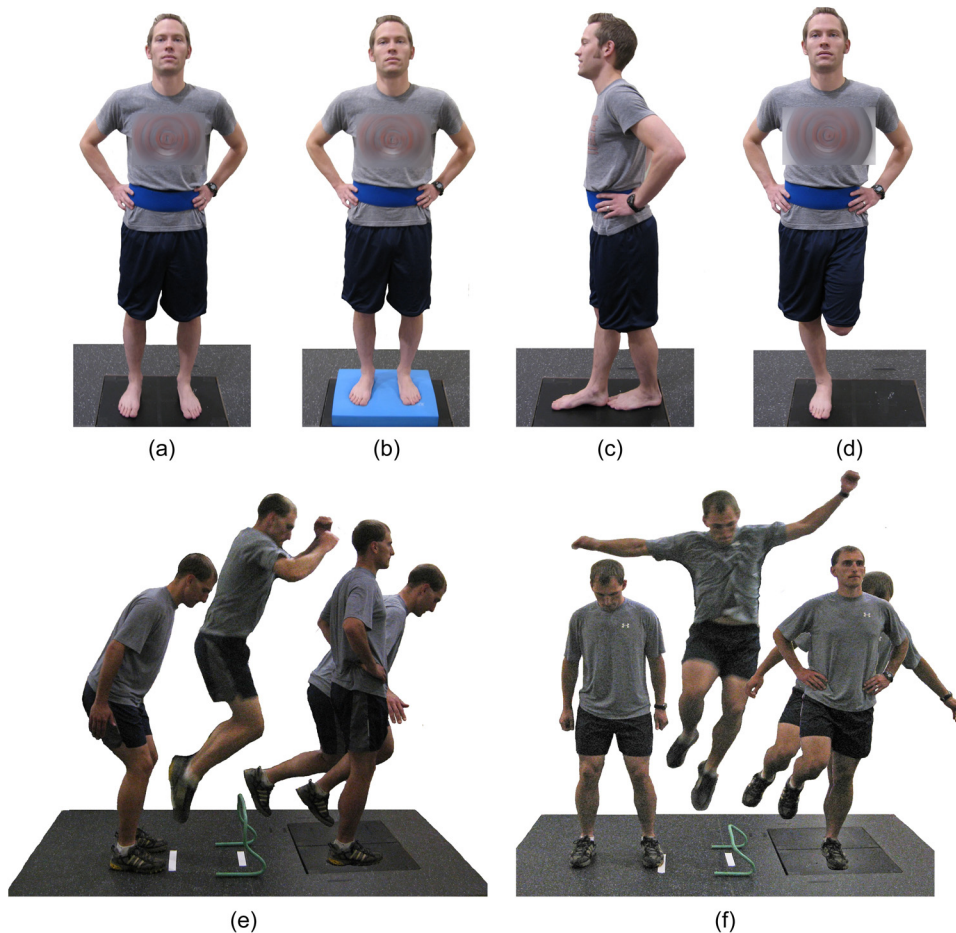

Fig. 1. Postural stability tasks. (a) Double-leg stance completed with eyes open and eyes closed, (b) double-leg stance on foam completed with eyes open and eyes closed, (c) tandem stance completed with eyes open and eyes closed, (d) single-leg stance completed with eyes open and eyes closed, (e) forward jump landing (DPSI-AP), and (f) lateral jump landing (DPSI-ML).

The root mean square (RMS) was used for all accelerometer data. This measure was calculated for each orthogonal direction and resultant vector for each task. The RMS of acceleration for the static tasks were calculated over the ten second trial where the dynamic tasks were calculated during a three second window beginning at the time of peak vertical acceleration.

2.5. Statistical analysis

Test–retest reliability was assessed using an interclass correlation coefficient calculation (2,1). Multivariate analysis and correlation tests were used to determine if the COM acceleration measure could distinguish between tasks of various difficulties and if a relationship exists between accelerometer and force platform measures. The Kruskal–Wallis test was used to determine the ability of the RMS of acceleration to distinguish between tasks because the data was not all normally distributed. If a significant difference was found subsequent pairwise analysis was used to determine specific differences between tasks. Spearman's ranked correlations were also used to determine the relationship between the accelerometer measures and force platform measures. An alpha level of 0.05 was set a priori.

3. Results

The first specific aim of this study was to determine the reliability of an accelerometer based measure of dynamic postural stability. The reliability for the RMS calculations during dynamic postural stability was shown to be good, demonstrating ICC values ranging from 0.835 to 0.924 and SEM values of 0.028–0.053 (Table 2).

The second specific aim of this study was to evaluate the validity of the accelerometer-based postural stability measure by comparing across tasks and with the force platform measures of postural stability. The means and standard deviations of the postural stability using an accelerometer and using a force platform can be found online (Supplementary Table 1) with the between task comparison results shown in Table 3. The Kruskal Wallis tests revealed significant differences between static and dynamic tasks when postural stability was measured using an accelerometer ($p < 0.001$) or a force platform ($p < 0.001$) in all directions. However, only the medial–lateral direction of the accelerometer was able to show significant differences between static tasks. The between task pairwise comparison can be found online (Supplementary Table 2). Spearman's ranked correlations between the accelerometer and GRF measures showed low to moderate relationships in the different axes. The ML axis of the accelerometer was only significantly correlated to the ML direction of the GRF data during the single-lag static balance tasks (Table 4). The vertical axis of the accelerometer was only significantly correlated to the vertical GRF data during the DPSI-AP and DPSI-ML tasks. Lastly, the resultant axis was only significantly correlated to the DPSI-ML task.

4. Discussion

The purpose of this study was to evaluate the reliability and validity of an accelerometer-based measure of static and dynamic postural stability. It was hypothesized that an accelerometer-based measure of dynamic postural stability would demonstrate good to excellent test–retest reliability. This hypothesis was supported as the ICC values for the test–retest reliability ranged from 0.732 to 0.899. We also hypothesized that the accelerometer measure of postural stability would detect differences in postural

Table 2
Reliability of RMS postural stability measures during DPSI-AP.

Task	ICC	95% CI			SEM
RMS–AP	0.835	0.334	–	0.959	0.031
RMS–ML	0.841	0.358	–	0.960	0.053
RMS–V	0.893	0.570	–	0.973	0.028
RMS–R	0.924	0.696	–	0.981	0.029

Interclass correlation coefficient (ICC), 95% confidence interval (CI) and standard error of measurement (SEM) results. RMS = root mean square, AP = anterior–posterior, ML = medial–lateral, V = vertical, R = resultant.

Table 3
Between task comparisons for accelerometer and ground reaction forces.

	Axis	Static comparison	All task comparison
		p-Value	*p*-Value
Accelerometer	AP	0.756	<0.001
	ML	<0.001	<0.001
	Vert	0.576	<0.001
	Resultant	1.000	<0.001
GRF	AP	<0.001	<0.001
	ML	<0.001	<0.001
	Vert	<0.001	<0.001
	Resultant	<0.001	<0.001

Kruskal–Wallis test results from the accelerometer and ground reaction forces (GRFs) in the anterior–posterior (AP), medial–lateral (ML), vertical (Vert), and resultant direction comparing only static tasks and all tasks.

stability between tasks of varying difficulty and display good correlations with the force platform measure of postural stability. This hypothesis was partially supported by the results of this study. The accelerometer measure of postural stability was able to detect differences between tasks of different difficulty; however, these measures did not correlate well with the force platform measure of postural stability.

Moe-Nilssen [23] investigated the reliability of static postural stability using accelerometry and found that the medial–lateral direction and resultant vector (medial–lateral and anterior–posterior) during single-leg stance were the only values that were reliable [23]. Very low ICC values in double-leg stance and single-leg stance in the anterior–posterior direction were attributed to the low range of values between subjects and the majority of postural sway occurred in the medial lateral direction during single-leg stance [23]. The current study was the first to examine postural stability using COM accelerations during a landing task and was able to demonstrate that all directions of the accelerometer-based measure of dynamic postural stability were reliable.

The current study also examined the ability of a COM accelerometer to measure postural stability and distinguish between tasks of varying difficulty. This study demonstrated that the accelerometer measure of postural stability was able to distinguish between tasks. Goldie et al. [9] conducted a similar study evaluating the validity of postural control measures using a force platform. This study utilized a similar design by evaluating tasks of varying difficulty. They found a similar trend in difficulty as the current study suggests with both the accelerometer and force platform. Goldie et al. [9] also determined that in double-leg stance conditions the best predictors of postural control were ML and AP directions, however, during single-leg stance ML was the best predictor of postural control. This may explain why the current study also found the ML scores were most correlated to force platform measures in single leg stance and static postural stability scores were only distinguishable in the ML direction, specifically between the double-leg and single-leg tasks.

The current study also found little to moderate correlations between accelerometer and GRF measures of postural stability. The overall lack of correlations is most likely because the devices are measuring different aspects of postural stability [25]. The COM mounted accelerometer is capturing accelerations at the person's approximate center of mass. The GRFs are measuring reactive forces at the ground-foot interface that may be more reflective of the corrective actions taken by the individual to maintain postural stability instead of body sway. Winter [3] proposed that acceleration measures at the COM may be better as it is an approximate measure of body sway about the COM [3].

We acknowledge that this study does have limitations that need to be considered. The sample size was small compared to other reliability studies,[9,24] however, we were able to establish a

Table 4
Correlations between accelerometer and GRF measures of postural stability.

	DLEO	DLEC	DLEO-F	DLEC-F	TEO	TEC	SLEO	SLEC	DPSI-AP	DPSI-ML
AP	−0.536	−0.201	−0.206	−0.171	0.006	−0.028	0.050	−0.022	−0.291	0.176
ML	0.033	0.243	0.261	0.143	0.489	−0.060	0.626*	0.731*	0.126	0.258
Vert	0.088	0.099	0.555	0.099	0.000	0.253	−0.017	0.039	0.681*	0.703*
Resultant	−0.165	−0.110	−0.209	−0.434	0.379	0.039	0.148	−0.159	0.269	0.593*

* p-Values were significant ($p < 0.05$).

reliable measure using the accelerometer. Additionally, there are other factors that can affect the reliability and correlation of the accelerometer measure that include accelerometer placement between subjects and movement artifact. Each subject was fitted by the same examiner and the jump landing task was for dynamic postural stability was always measured last to minimize any effect of the jump landings on the position of the device during the static postural stability task. Another limitation of the current study is that the range of the accelerometer was set to $\pm 18 \times g$. This range was selected because two of the tasks were jump landing tasks and would elicit higher accelerations. Previous studies that have investigated only static tasks have used lower range accelerometers such as $\pm 1.5 \times g$ that may provide better resolution during static tasks [30]. Lastly, the acceleration data was not transformed to horizontal–vertical axes as previous studies have and only used one stability metric [25]. Our method was chosen to retain an approximate anatomical coordinate system during the dynamic postural stability tasks. Future research should investigate the use of more than one metric and the utility of coordinate system transformation during dynamic postural stability tasks.

5. Conclusion

The accelerometer measure of postural stability was reliable and able to distinguish between static and dynamic tasks. However, differences between individual static tasks were only identified in the medial–lateral direction. Based on the current findings the medial–lateral accelerations seem to be most distinguishing compared to other directions. Although there was a lack of correlation between accelerometer and force platform measures of postural stability, except the ML direction in single-leg stance, the accelerometer still showed a similar pattern in postural stability scores by task as seen in the force platform in this study and previous studies. Future research should investigate the use of an accelerometry-based postural stability measure using a pathological population.

Conflict of interest statement

The authors of this manuscript do not have any conflicts of interest to disclose.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2014.12.009.

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