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Leveling the playing field: Evaluation of a portable instrument for quantifying balance performance

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ABSTRACT

Balance is a complex, sensorimotor task requiring an individual to maintain the center of gravity within the base of support. Quantifying balance in a reliable and valid manner is essential to evaluating disease progression, aging complications, and injuries in clinical and research settings. Typically, researchers use force plates to track motion of the center of gravity during a variety of tasks. However, limiting factors such as cost, portability, and availability have hindered postural stability evaluation in these settings. This study compared the "gold standard" for assessing postural stability (i.e., the laboratory-grade force plate) to a more affordable and portable assessment tool (i.e., BTrackS balance plate) in healthy young adults. Correlations and Bland-Altman plots between the center of pressure outcome measures derived from these two instruments were produced. Based on the results of this study, the measures attained from the portable balance plate objectively quantified postural stability with high validity on both rigid and compliant surfaces, demonstrated by thirty-five out of thirty-eight observed postural stability metrics in both surface conditions with a correlation of 0.98 or greater. The low cost, portable system performed similarly to the lab-grade force plate indicating the potential for practitioners and researchers to use the BTrackS balance plate as an alternative to the more expensive force plate option for assessing postural stability, whether in the lab setting or in the field.

1. Introduction

Postural steadiness, or postural stability, is defined as the ability of an individual to maintain their balance during quiet standing. More specifically, maintaining balance means keeping the center of gravity within the base of support (Bronstein and Pavlou, 2013; Horak, 2009; Marcolin et al., 2016; Prieto et al., 1996). Postural stability can be quantified with measures of vertical and horizontal reaction forces, center-of-pressure (CoP) displacement, or lumbar horizontal displacement (Prieto et al., 1996). Measures of force and displacement are typically calculated via instruments such as force plates, balance platforms, or accelerometers.

For decades, laboratory-grade force plates have set the standard by which measures of postural stability are quantified. However, the large expense of these devices, costing upwards of ~\$5000–\$75,000 or more, lack of portability, and requirement for external power sources, often preclude this option for individuals conducting assessments in the clinical or field settings (Goble et al., 2016; Whitney and Wrisley, 2004). In addition to an AC power requirement, the laboratory-grade force plate (FP) is also required to be fixed (bolted) to a surrounding structure. To address this problem, a more cost effective (\$~795, plus software) and lighter (<7 kg) option has been developed by Balance Tracking Systems Inc., identified here as the BTrackS Balance Plate (BBP). This device has shown to have high accuracy and precision, as well as near perfect inter-device reliability for both X and Y CoP directions when compare to a laboratory-grade force plate (Goble et al., 2018).

In their initial evaluation of the BBP, O'Connor and colleagues (O'Connor et al., 2016) utilized an inverted pendulum model to provide a proof of concept that the BBP is a valid device for measuring CoP (Goble et al., 2016; O'Connor et al., 2016). These researchers demonstrated this validity through the comparison of the signals from the BBP and a FP. However, prior validation studies did not break the CoP signal into separate dependent variables for discrete evaluation of medio-lateral and antero-posterior motion. Further, previous work has only reported linear regression associations between output variables, only reflecting a snapshot

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of the validity of the BBP, particularly because correlations only show an association between variables, and not that the values are equal (O'Connor et al., 2016). Further, output of the BBP has yet to be directly compared against a laboratory-grade force plate for multiple postural stability metrics during actual balance tasks in healthy adults, rather than using a more abstract inverted pendulum model.

The aim of this study was to compare the "gold standard" for assessing postural stability to a more affordable and portable assessment tool in healthy young adults. Using derived postural metrics from the CoP time series, calculations were performed according to the methods of Prieto and colleagues (Prieto et al., 1996). This included both time dependent variables and combined distance and time-dependent variables. We hypothesized that CoP-based metrics of postural stability as measured by the BBP and FP would be similar. Furthermore, we hypothesized that by utilizing Bland-Altman plots, an interpretation of the results could be made beyond simple correlations between the two devices and validation provided that there is true agreement in the output(s) signals.

2. Methods

2.1. Participants

Twenty healthy college adults (10 males and 10 females; 26 ± 4 years, 1.7 ± 0.1 m, 66.68 ± 9.35 kg, body mass index 22.86 ± 1.58 kg m^{-2}) were recruited to participate in this study. Eligible participants were young adults between 18 and 30 years who were (1) able to stand on two feet for at least an hour, (2) free from neurological disorders or recent musculoskeletal injuries that would impact balance, and (3) not taking medications known to impact balance. Approval for this study was given by the local Institutional Review Board at Colorado State University and all participants provided written informed consent before participation.

2.2. Experimental procedure

All assessments occurred within a single testing session. Each participant's postural stability was simultaneously measured using an embedded force plate (Bertec Corporation, Columbus, OH) with Vicon Nexus (VICON, Englewood, CO) and via the BBP (Balance Tracking Systems, Inc., San Diego, CA). The BBP was placed on top of the FP with the long axis of the BBP in line with the long axis of the FP (see Fig. 1). For reference the alignment of the device axes were aligned via the outer edge (posterior (heel) side) of the BBP parallel with the edge of the FP.

Each system was zeroed in this configuration prior to each trial. Participants performed two 30-s trials of standing quietly. Prior to any participation all participants were given verbal and visual instructions by the lead researcher. Once the participant acknowledged understanding of these instructions, the researcher began collecting FP and BBP data. First, participants stood on a rigid surface (i.e., directly on the BBP) with eyes open and looking at a fixed target 4.37 m away. Second, a compliant, 7 cm thick Elite Balance pad (Airex, Sins, Switzerland) was placed on top of the BBP and participants again attempted to stand quietly while focusing on the same fixed target. In each condition, participants were instructed to stand in the base position as quietly/still as possible with their hands on their hips and their feet together (see Fig. 2). Participants were given a "step up" cue and then stepped on to the BBP, assuming the base position. Participants remained as still as possible until the researcher indicated the trial was over (~35 s after the participant first stepped onto the BBP). The participant was then asked to step back off the plate. Then, this procedure was repeated for the second condition of the protocol. The FP and BBP instruments continuously collected data prior to and after each quiet standing trial. This was necessary for time synchronizing the two separate recordings during post-processing. Since the aim of this study was not to compare differences between the testing surfaces, the order of the testing was kept consistent throughout the entirety of the protocol.

2.3. Data analysis

During all trials, ground reaction forces were collected at 25 Hz by the BBP and 100 Hz by the force plate. Text data files from both systems were exported to MATLAB (MathWorks, Natick, MA, version R2017a) for processing. BTS, Inc.'s proprietary software filters the CoP data prior to export using a second order, low-pass Butterworth filter with a cutoff frequency of 4 Hz. All force plate data

Fig. 1. Orientation of the BTrackS Balance Plate (atop the force plate) to the Bertec force plate. The directional difference was accounted for in the devised MATLAB script when calculating postural stability metrics.

Fig. 2. Postural stability assessments occurred on both rigid (A) and compliant (B) surface conditions.

were filtered in a similar fashion (i.e., a second order, low-pass Butterworth with $F_c = 4$ Hz) in MATLAB after export from Vicon Nexus. It is also important to note that although the plates collected at different sampling rates, the force plate data was down-sampled to 25 Hz to match the BBP sampling rate prior to calculating outcome measures. A 25 Hz sampling rate satisfies the Nyquist theorem for the slow (<10 Hz) CoP changes that we measured for this study (Goble et al., 2017).

Time synchronization was achieved using a custom MATLAB script. First, the time index of initial foot contact on the FP within a trial was identified as the point at which the vertical ground reaction force (F_z) exceeded 5% body weight. A five-second delay from this time index was then applied to ensure participants had adequate time to adopt a stable position in the prescribed testing stance. Trial data analyzed were the subsequent 30 s of signal following this initial 5-s delay. This ensured any shifting and moving to assume the base position was not included in the trial data. Eqs. (1) and (2) below were used to derive the CoP coordinates for the anterior-posterior (AP) and medio-lateral (ML) directions from the FP signals for each trial.

$$CoP_{AP} = \frac{M_X}{F_Z} \quad (1)$$

$$CoP_{ML} = -1\left(\frac{-M_Y}{F_Z}\right) \quad (2)$$

where M_Y is the ML moment, M_X is the AP moment, and F_Z the vertical reaction force. Note that due to the orientation of the participant on the FP, the X coordinates of the CoP represent anterior-posterior (AP) motion and Y coordinates represent mediolateral (ML) motion.

To correct these CoP coordinates calculated from the FP signal for the height of the BBP's surface above the force plate surface, the following corrections were applied (Eqs. (3) and (4)).

$$CoP_{AP} = \frac{((COP_{AP} * F_z) + (h * COP_{AP}))}{F_Z * (-1)} \quad (3)$$

$$CoP_{ML} = \frac{((COP_{ML} * F_z) + (h * COP_{ML}))}{F_Z * (-1)} \quad (4)$$

where h = height of the BBP in mm. Finally, the CoP coordinates were converted from mm to cm to match the units for the CoP coordinates exported from the BBP.

A similar procedure was applied to the BBP data to determine when a trial began. However, since the BBP's output does not include ground reaction forces, the CoP coordinate data for the mediolateral direction were used. When unloaded, the cells of the filtered BBP CoP_{ML} data column are all zero. The MATLAB script sequentially searched until a nonzero CoP_{ML} coordinate was identified. This instance indicated the participant had stepped onto the plate. A five second delay followed by 30 s of trial data were then similarly identified from the BBP data. When overlaid, the signals output from the BBP and FP displayed matched CoP signal tracings (demonstrated in Fig. 3).

The equations developed by Prieto and colleagues (Prieto et al., 1996) for quantifying measures of postural stability were applied to both the BBP and FP data for each 30-s trial. This resulted in a set of outcome variables from both the BBP and FP in the rigid condition, and likewise one set of outcome variables per system in the compliant condition. Pearson product moment correlations and Bland-Altman plots were then calculated for all performance variables obtained from the BBP and FP in the rigid and compliant conditions, respectively. The product moment correlation coefficients (r) establishes the relationship between BBP outcome metrics and metrics from the FP. However, when assessing the comparability between two separate devices it is important to establish the differences, of which simple correlations are not able to discern. The implementation of Bland-Altman plots signifies quantification of the agreement between two devices and can be reproduced for each of the quantitative postural stability metrics. Utilizing this approach enables the analysis to go beyond simple relationships and further explains the mean differences between the devices in each derived metric, thereby constructing the limits of agreement (Giavarina, 2015). All statistical analyses were completed using R (R Foundation for Statistical Computing, Vienna, Austria) with risk of type I error set at $\alpha = .05$.

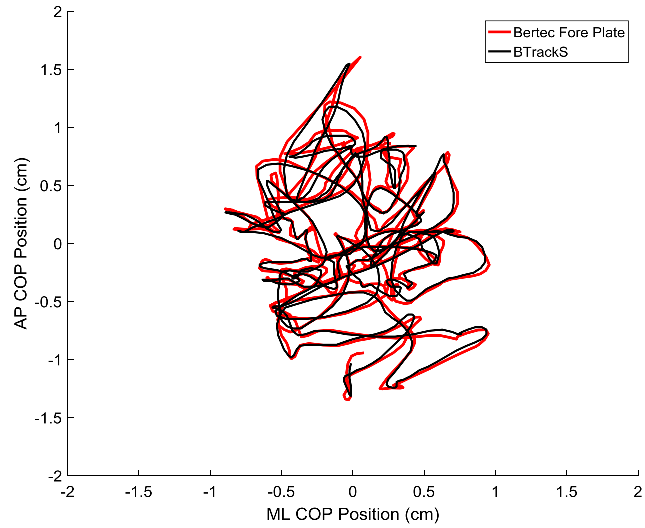

Fig. 3. CoP measures during the eyes open rigid surface testing condition, showing an overlay of the FP atop the BBP.

3. Results

Means and standard deviations for all the dependent variables for the rigid surface and compliant surface conditions are presented in Table 1.

3.1. Rigid surface

Positive correlation coefficients for all postural stability dependent variables between systems were significant at a p-value < 0.001 level in the rigid surface testing condition (Table 1).

3.2. Compliant surface

Positive correlation coefficients for the following postural stability dependent variables between systems were significant at a p-value < 0.001 level in the compliant surface testing condition (Table 1).

3.3. Bland-Altman analysis

The Bland-Altman plots indicated strong agreement between the devices in both surface conditions (i.e., rigid and compliant). The mean difference, between the two devices for all postural stability metrics ranged between $-6.439e^{-5}$ cm to 1.271 cm and $-9.32e^{-4}$ cm to -1.110 cm for the rigid and compliant surfaces, respectively (seen in Figs. 4 and 5). Taken together, regression and Bland-Altman analyses demonstrate strong agreement and minimal difference in the output of the two devices studied.

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Table 1
Summary of means (standard deviations) and correlations for postural stability metrics on both the rigid and compliant surface testing conditions.

Dependent variable	Rigid surface				Compliant surface			
	Bertec	BTrackS	Mean diff	r	Bertec	BTrackS	Mean diff	r
Mean path length (cm)	44.63 (14.39)	43.36 (15.51)	1.27	0.999*	78.91 (31.08)	80.02 (33.12)	−1.11	0.999*
Path length AP (cm)	24.00 (8.79)	23.46 (9.21)	0.55	1.000*	50.43 (25.04)	50.62 (26.21)	−0.19	0.999*
Path length ML (cm)	32.65 (10.40)	31.35 (11.52)	1.3	0.997*	49.71 (14.87)	50.78 (16.44)	−1.07	0.996*
Mean velocity (cm s^{-1})	1.49 (0.48)	1.45 (0.52)	0.04	0.999*	2.63 (1.04)	2.67 (1.1)	−0.04	0.999*
Velocity AP (cm s^{-1})	0.80 (0.29)	0.78 (0.31)	0.02	1.000*	1.68 (0.84)	1.69 (0.87)	−0.001	0.999*
Velocity ML (cm s^{-1})	1.09 (0.35)	1.05 (0.38)	0.04	0.998*	1.66 (0.50)	1.69 (0.55)	−0.04	0.996*
Mean distance (cm)	0.63 (0.20)	0.62 (0.20)	0.01	0.999*	0.91 (0.21)	0.90 (0.20)	0.01	0.998*
Mean frequency (Hz)	0.39 (0.10)	0.38 (0.11)	0.009	0.990*	0.46 (0.14)	0.47 (0.14)	−0.008	0.997*
Frequency AP (Hz)	0.36 (0.13)	0.36 (0.14)	0.002	0.998*	0.47 (0.20)	0.48 (0.20)	−0.006	0.999*
Frequency ML (Hz)	0.54 (0.14)	0.51 (0.14)	0.03	0.962*	0.58 (0.13)	0.59 (0.14)	−0.01	0.991*
RMS (cm)	0.71 (0.22)	0.70 (0.22)	0.009	0.999*	1.03 (0.23)	1.02 (0.23)	0.008	0.998*
RMS AP (cm)	0.53 (0.21)	0.52 (0.20)	0.01	1.000*	0.79 (0.21)	0.78 (0.21)	0.01	0.999*
RMS ML (cm)	0.46 (0.13)	0.46 (0.13)	0.002	0.996*	0.65 (0.15)	0.65 (0.15)	0.001	0.990*
Range AP (cm)	2.52 (0.77)	2.44 (0.76)	0.08	0.986*	3.96 (1.11)	3.94 (1.11)	0.02	0.998*
Range ML (cm)	2.57 (0.80)	2.50 (0.79)	0.07	0.934*	3.56 (0.90)	3.51 (0.80)	0.05	0.950*
AD AP (cm)	0.44 (0.18)	0.43 (0.18)	0.009	1.000*	0.64 (0.18)	0.64 (0.17)	0.009	0.999*
AD ML (cm)	0.37 (0.10)	0.37 (0.10)	−0.00006	0.998*	0.51 (0.12)	0.51 (0.12)	−0.009	0.996*
95% CC (cm^2)	4.68 (2.87)	4.56 (2.81)	0.12	0.999*	9.35 (4.12)	9.35 (4.12)	0.17	1.000*
Sway area (cm^2/s)	0.32 (0.18)	0.31 (0.18)	0.01	0.997*	0.80 (0.46)	0.81 (0.49)	−0.01	0.997*

Note: ML = mediolateral, AP = anterior-posterior, RMS = Root mean squared, AD = absolute distance.
* significant value at p-value < .001.

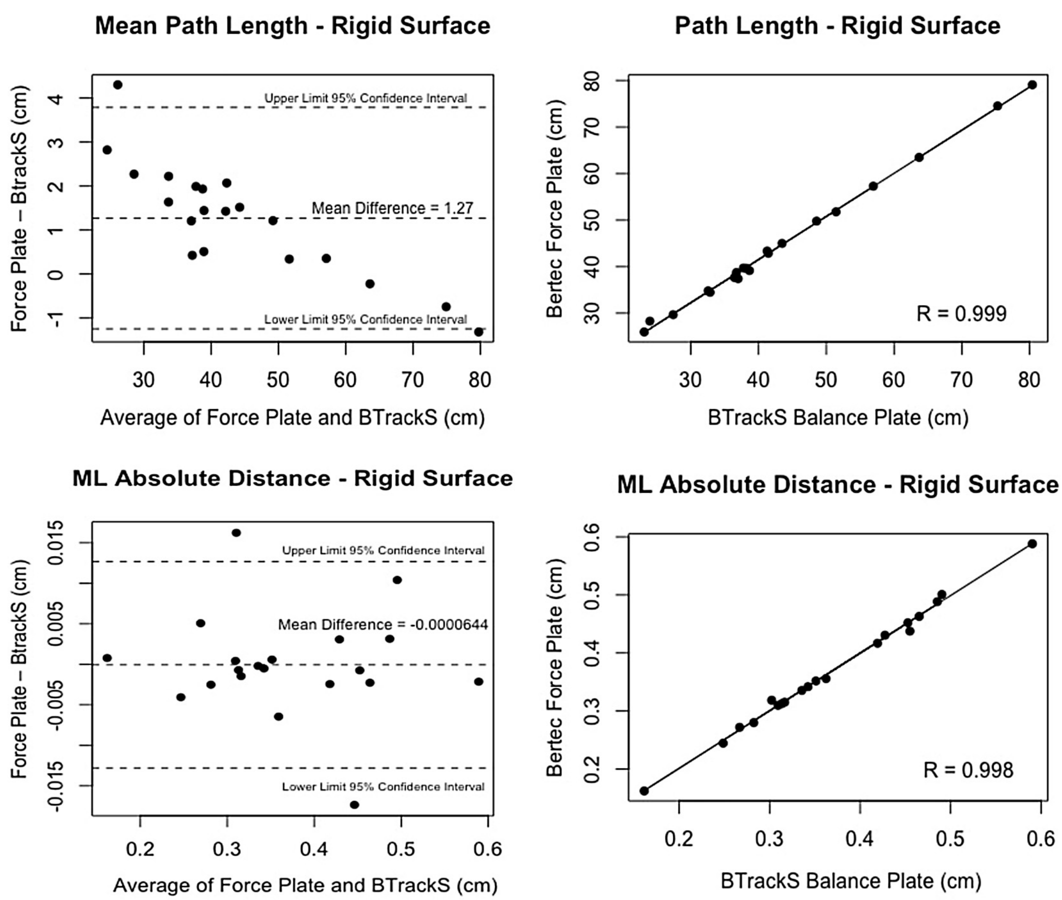

Fig. 4. Representative Bland-Altman plots identifying the agreement between the FP and BBP on the rigid surface. The top row identifies the maximum difference between devices and the bottom row identifies the minimum differences between devices with their associated correlation plot. All of the additional rigid surface postural metric differences fell between the maximum and minimum differences demonstrated in this figure.

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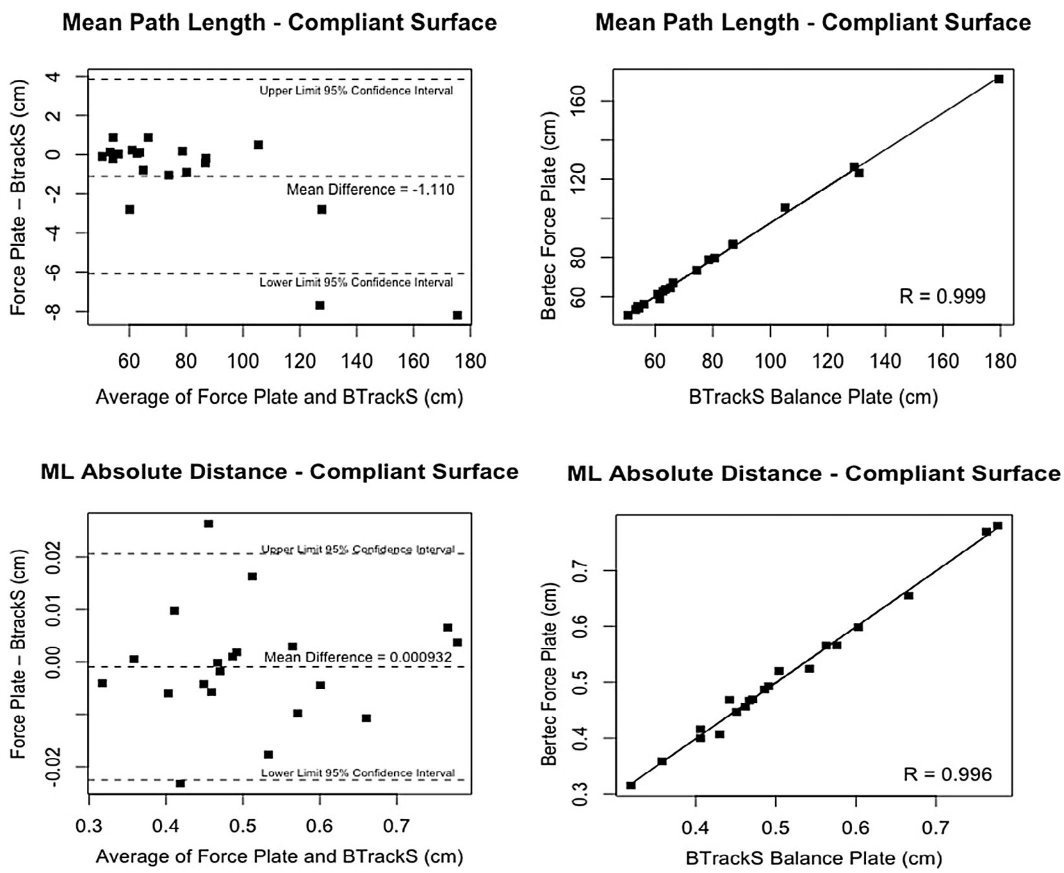

Fig. 5. Representative Bland-Altman plots identifying the agreement between the FP and BBP on the compliant surface. The top row identifies the maximum difference between devices and the bottom row identifies the minimum differences between devices with their associated correlation plot. All of the additional compliant surface postural metric differences fell between the maximum and minimum differences demonstrated in this figure.

4. Discussion

Our hypothesis that the CoP based metrics of postural stability as measured by the BBP and FP would be similar was supported. This suggests that the BBP is a valid alternative to a traditional force plate for quantifying balance. However, it should be noted that although there were strong relationships demonstrated between dependent variables, the magnitudes and variability of these measures were not precisely equivalent between systems. The BBP was consistent in underestimating each outcome variable in comparison to the FP in the rigid surface condition, with the exception of ML RMS, AP frequency, and the AP absolute distance. In these cases, mean values were equal between devices. Alternatively, in the compliant surface condition the BBP was greater in magnitude than the FP with all postural stability dependent variable measures excluding the mean RMS and Range in both AP and ML. These exclusions were opposed to the trend that the BBP was of larger magnitudes than the FP, to which the RMS ML, 95% confidence circle area, and absolute distance in both directions had the equivalent means for both devices.

The strong correlations between the BBP and FP devices is consistent with previous literature concerning the BBP (O'Connor et al., 2016), the correlations were as strong as r = 1.000 with a minimum correlation of r = 0.934 (see Table 1). Furthermore, the agreement between devices was strong in both the rigid and compliant surface conditions, the path length was consistent in both surface conditions for showing the poorest agreement with a mean difference of 1.271 cm and −1.110 cm, respectively (see Figs. 4 and 5). Despite this small mean difference in device output for path length, in both conditions this outcome measure still maintained very strong association between devices, with a correlation of r = 0.999. This factor could be due to a larger spread of the data based on the unit of measurement of the variable. It is possible that the magnitude of difference may increase with time, and should be taken into account in the event of cross-referencing data collections from a force plate to the BBP, however over the course of a 30 s trial, the current results demonstrate very limited practical differences between device output. This trend opposes what has been produced from Nintendo Wii Balance Board research, where they saw overestimates in both 'mean CoP sway' and 'CoP path velocity' in comparison to simultaneous collection by a force plate (Huurnink et al., 2013).

The ability of the BBP to consistently show near perfect agreement with the FP are in accordance with BBP conclusions drawn by O'Conner et al. (O'Connor et al., 2016). The utilization of the inverted pendulum identified in the O'Conner (O'Connor et al., 2016) study of the BBP has the capabilities to demonstrate near perfect signal agreement, accuracy, and precision with the signal output from a laboratory-grade force plate. The increased reliability of the portable force platform coincides with the inter-device reliability established in the BBP by Goble et al. (Goble et al., 2018). Our findings are in agreement with both of these authors' outcomes, allowing us to conclude that the BBP is a valid, cost effective, and portable alternative to the "gold standard" laboratory-grade force plate (Chang et al., 2014; O'Connor et al., 2016).

Any of the marginally lower correlations observed in the resulting postural stability variables of this study could be explained by two possible explanations, the first being issues with complex derivatives and/or unaccounted complications from the devices offset. As with any mathematical measure there is always the small chance of errors being introduced, and these errors can compound on each other as more complex derivatives progress. Further, when processing a signal, any error inherent within the signal will be magnified when squared. This is evident when two systems are not providing exactly the same data. It is important to understand that when doing more complex and derivative postural analysis, errors in the output will be multiplied. Although, these compound errors do need to be taken into consideration when identifying the primary outcomes of the study, the BBP capabilities could change research in the clinical settings as well as the availability within the laboratory setting. The portability of this instrumentation could allow for objective quantification of postural stability in settings where this would have been nearly impossible to quantify beforehand such as clinics, small academic institutions, athletic field settings, or even in-home healthcare.

Technical limitations and future directions to this study include time synchronization, noise of the BBP signal in postural stability, evaluation of sway entropy, task-based assessment and BBP long term reliability. Due to inabilities to introduce a time Transistor-Transistor Logic (TTL) pulse into each of the devices, we were unable to time synchronize the system, allotting for the chance of small error. Minimal error was introduced into each collection due to the deficiency of time synchronization. Portable force platforms are inherently susceptible to both mechanical noise and vibrations, affecting the overall collection frequency of the signal. Future examinations that systematically target these sources of data contamination are planned. Beyond signal assessment, a more extensive analysis of the CoP signal including non-linear measures such as entropy and random walk CoP (Collins and De Luca, 1993) variables are worth exploring in future validations of the BBP. The nature of the task being observed was also limiting in this study, as the validation of these instruments exclusively focused on eyes open conditions and did not include conditions where the eyes were closed. This was done in an attempt to limit any confounding variables outside of the validation of the instruments. Lastly, although the acute accuracy of the BBP was confirmed, we did not assess the long-term reliability of the BBP.

We conclude that the BBP is a valid alternative to the "gold standard" laboratory-grade force plate for quantifying postural stability. Researchers should be aware that small differences in CoP coordinates recorded by the BBP may occur and should consider whether these differences are meaningful for their particular purpose. The portability and cost-effective nature of the BBP are a benefit, regardless of its somewhat lower precision. Thus, researchers evaluating postural stability outside of the laboratory should consider the BBP a valid instrument for quantifying balance in those settings. This device will allow a wider range of populations to utilize it as an objective way to quantify postural stability where before subjective tests had been utilized. The BBP can provide lab-quality balance assessment to clinicians and researchers outside of the lab setting.

Conflict of interest statement

D Goble holds an equity stake (i.e. stock options) in the parent company for the BTrackS Balance Plate. None of the other authors have any conflicts of interest to declare.

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