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Short communication

Comparison of a laboratory grade force platform with a Nintendo Wii Balance Board on measurement of postural control in single-leg stance balance tasks

Arnold Huurnink*, Duncan P. Fransz, Idsart Kingma, Jaap H. van Dieën

Research Institute MOVE, Faculty of Human Movement Sciences, VU University Amsterdam, Amsterdam, The Netherlands

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ABSTRACT

Training and testing of balance have potential applications in sports and medicine. Laboratory grade force plates (FP) are considered the gold standard for the measurement of balance performance. Measurements in these systems are based on the parameterization of center of pressure (CoP) trajectories. Previous research validated the inexpensive, widely available and portable Nintendo Wii Balance Board (WBB). The novelty of the present study is that FP and WBB are compared on CoP data that was collected simultaneously, by placing the WBB on the FP. Fourteen healthy participants performed ten sequences of single-leg stance tasks with eyes open (EO), eyes closed (EC) and after a sideways hop (HOP). Within trial comparison of the two systems showed small root-mean-square differences for the CoP trajectories in the x and y direction during the three tasks (mean $\pm$ SD; EO: 0.33 ± 0.10 and 0.31 ± 0.16 mm; EC: 0.58 ± 0.17 and 0.63 ± 0.19 mm; HOP: 0.74 ± 0.34 and 0.74 ± 0.27 mm, respectively). Additionally, during all 420 trials, comparison of FP and WBB revealed very high Pearson's correlation coefficients (r) of the CoP trajectories (x: 0.999 ± 0.002; y: 0.998 ± 0.003). A general overestimation was found on the WBB compared to the FP for 'CoP path velocity' (EO: 5.3 ± 1.9%; EC: 4.0 ± 1.4%; HOP: 4.6 ± 1.6%) and 'mean absolute CoP sway' (EO: 3.5 ± 0.7%; EC: 3.7 ± 0.5%; HOP: 3.6 ± 1.0%). This overestimation was highly consistent over the 140 trials per task ($r > 0.996$). The present findings demonstrate that WBB is sufficiently accurate in quantifying CoP trajectory, and overall amplitude and velocity during single-leg stance balance tasks.

1. Introduction

Training and testing of balance have an important place in research on prevention of falls in the elderly, rehabilitation of neurological or orthopedic patients, improvement of sports performance, and reduction of injury risk (Gil-Gómez et al., 2011; Hrysomallis, 2007, 2011; Melzer et al., 2010). Parameters derived from center of pressure (CoP) trajectories measured by a laboratory grade force plate (FP) are the gold standard for balance performance. Given the large potential for application of balance assessments, an inexpensive, widely available, portable and accurate force plate would be a tremendous advancement. The Wii Balance Board (WBB) (Nintendo, Kyoto, Japan) is designed to serve as a video game controller, but might satisfy these criteria (Clark et al., 2010). Compared to a FP, limitations are a low sample rate, the unavailability of horizontal forces, a larger amount of noise, an inconsistent sampling interval, occasional glitches in the data, and a manufacturer advised maximum load of 1962 N (Pagnacco et al., 2011). Previous comparison between the assessment of balance with a FP and a WBB showed encouraging results, but was performed with separate trials for both instruments (Clark et al., 2010). In the present study, the WBB was positioned on top of a FP to enable simultaneous measurements of the CoP, which eliminated within subject variability.

2. Methods

2.1. Participants

Fourteen healthy volunteers were recruited from the members of staff at the Faculty of Human Movement Sciences (6 males, 8 females; mean (range); age 28.0 (24–34) years; height 175.3 (165–197) cm; body weight 67.2 (55–85) kg). The local Ethics Committee approved the study and all participants provided informed consent.

2.2. Measurement setup

A WBB (45×26.5 cm in the x and y direction, respectively) was placed upon a FP (Kistler model 9218B), which measured 60×40 cm and was mounted flush with the laboratory floor. The WBB has four strain gauge load sensors and is only

* Correspondence to: Research institute MOVE, Faculty of Human Movement Sciences, VU University, Van der Boechorststraat 9, 1081 BT Amsterdam, The Netherlands. Tel.: +31205988492; fax: +31205988529.
E-mail address: a.huurnink@gmail.com (A. Huurnink).

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able to measure vertical forces. The WBB was connected wirelessly to a laptop computer via Bluetooth. The WBB data were sampled at 35 sample · s^{-1} (Pagnacco et al., 2011) and retrieved with custom-written software (Labview 8.5 National Instruments, Austin, TX, U.S.A.). The FP data were sampled at 1000 sample · s^{-1}.

The validity of static CoP measurement was verified with a 10 kg load for twelve known positions (25 mm in between) and with a 60 kg load for nine positions (50 mm in between). This resulted in root-mean-square (RMS) differences between the two systems in *x* and *y* directions of 0.59 mm and 0.67 mm for the 10 kg load, and 0.73 mm and 0.58 mm for the 60 kg load.

2.3. Procedures

Prior to our measurements, dead weight noise tests of 55 to 80 kg were performed to estimate the amount of noise disturbance of the FP and WBB (Pagnacco et al., 2011). Participants performed ten series of three balance tasks of 10 s in a single session: single-leg stance with eyes open (EO) and closed (EC), and single-leg stance after a short sideways hop (HOP). The hop was performed in lateral direction and was commenced while standing with both legs next to the WBB. Participants were instructed to perform all tasks on their preferred leg and barefoot. Participants were asked to stand as still as possible, to keep their hands on their hips and to focus on a point 2 m ahead. A trial was considered invalid if participants displaced their standing leg or touched the floor with the contralateral leg.

2.4. Data processing

A custom MATLAB (The Mathworks, Natick, RI, USA) program was designed for data reduction. Our data showed that the time interval between samples of WBB data was inconsistent (SD: 0.42 ms), therefore linear interpolation of the raw signals for the four WBB load sensors was applied to obtain a regular sample rate of 1000 sample · s^{-1} (Pua et al., 2012). The data were filtered with a second order Butterworth low-pass filter. Close examination of the WBB CoP signal frequency distribution and the WBB noise effects have led to an estimated optimal cutoff frequency of 12 Hz, which was similar to Clark et al. (2010). FP signals processed with a low-pass filter up to a cutoff frequency of 12 Hz, faithfully represented single-leg stance balance assessments (Ross et al., 2009). This was corroborated in our data. For instance, the FP 'CoP path velocity' value derived from the signal below 12 Hz, was 98% of the 'CoP path velocity' value derived from the signal up to 20 Hz. Therefore, we decided to be consistent across both systems with regards to the low-pass filter cutoff frequency. To synchronize the onset of measurements, the time lags between FP and WBB were corrected with a time series covariance function of the vertical force data. CoP calculations for the WBB were performed with vertical forces of the four load sensors, and the distance between a load sensor and the middle of the WBB (21.6 × 11.8 cm, in the *x* (anteroposterior) and *y* (mediolateral) direction, respectively). CoP calculations for the FP were based on vertical and horizontal forces in accordance with the manufacturer's manual. However, the moment arm of the horizontal forces was adapted to take the height of the WBB into account. Within each trial, for both FP and WBB, the mean CoP was considered as the origin.

2.5. Data analysis

The total trial length was analyzed for the EO and EC tasks, while for the HOP task only the data from 0.5 to 3.5 s after initial contact (vertical force > 10 N) were taken into account. To quantify the horizontal ground reaction force (GRF), both mean and maximum value of the (absolute) total vector horizontal GRF were calculated per trial.

The RMS error and Pearson's correlation coefficient (*r*) between FP and WBB were calculated for CoP trajectories in *x* and *y* direction during each trial (Derrick et al., 1994). Additionally, two common CoP parameters (i.e., 'CoP path velocity' and 'mean CoP sway'), which are axis independent, were calculated for the FP and WBB (Clark et al., 2010; Melzer et al., 2010). The 'mean CoP sway' represents the mean absolute distance from the CoP trajectory to the origin. Furthermore, for each balance task, *r* was calculated over all trials between the FP and WBB CoP parameter values.

3. Results

Table 1 presents the results of the dead weight noise testing, which showed the effect of noise on the FP and WBB systems for 'mean CoP sway' (0.03–0.04 and 0.12–0.16 mm, respectively) and 'CoP path velocity' (0.5–1.0 and 4.1–5.7 mm s^{-1}, respectively). CoP trajectories from a representative participant are shown in Fig. 1. Although the magnitude of the horizontal GRF varied greatly across trials, the CoP trajectories measured during the three balance tasks were similar for FP and WBB systems (mean RMS: 0.31–0.74 mm; mean *r* 0.997–0.999; see Table 2).

Table 1
Dead weight noise testing for FP and WBB.

'mean CoP sway' (mm)			'CoP path velocity' (mm s^{-1})	
Weight (kg)	**FP**	**WBB**	**FP**	**WBB**
55	0.040	0.153	0.988	5.326
60	0.036	0.158	0.922	5.662
65	0.038	0.141	0.718	5.026
70	0.034	0.133	0.565	4.801
75	0.031	0.130	0.537	4.799
80	0.029	0.116	0.598	4.143

The dead weight noise measurements were conducted with synchronous CoP data collection for FP and WBB during 10 s trials, subsequently data were low-pass filtered at 12 Hz.

The WBB measurements overestimated the 'CoP path velocity' and 'mean CoP sway' averaged outcomes of FP by 3.5 to 5.3% (SD < 2%). For both balance measures, outcomes were highly correlated between FP and WBB over 140 trials per task ($r > 0.996$). Fig. 2 presents scatter plots of the two balance measures for all 420 trials (14 subjects × 10 trials × 3 tasks).

4. Discussion

In line with results of Clark et al. (2010), we found good correspondence between balance measures obtained with FP and WBB. In addition, we showed that instantaneous estimates of CoP location obtained with a WBB are very similar to those obtained with a FP. The latter adds significantly to the literature, since comparable outcomes between FP and WBB on averaged measures, such as 'CoP path velocity', do not necessarily generalize to other balance measures. Within the range of tasks evaluated, the good correspondence between CoP trajectories suggests that any balance measure based on a WBB CoP trajectory can be considered sufficiently accurate.

The present findings seem a thorough validation for further exploitation of the WBB in the design and control of tasks that train or test balance. An elegant approach was investigated by Young et al. (2011), who incorporated real-time visual feedback of CoP into virtual environments (i.e., serious gaming) in order to create custom diagnostic or training programs. By, for instance, implementing such systems in (home-based) training regimes, compliance and effectiveness of balance training might be improved. Various disciplines (e.g., sports, sports medicine and rehabilitation medicine) might benefit from large-scale implementation of the low-cost WBB.

As the WBB might be employed in the field of sports, sports medicine or rehabilitation medicine, it is of necessity to thoroughly consider the limitations of the WBB. The results of the present dead weight noise testing indicate that the 'CoP path velocity' noise of the WBB (4.1–5.3 mm s^{-1}) is considerable, which is in accordance with Pagnacco et al. (2011). The higher noise levels compared to FP (0.6–1.0 mm s^{-1}) are most likely due to hardware configuration (Pagnacco et al., 2011). Furthermore, the low sample rate might have further increased the noise as a consequence of aliasing the high frequency noise (Pagnacco et al., 1997). Estimations can be made from the result of the FP (R_{FP}) and dead weight noise measurements (N_{FP} and N_{WBB}) on how noise could have affected the WBB result (R_{WBB}). According to the equation $R_{WBB} = \sqrt{(R_{FP}^2 + N_{WBB}^2 - N_{FP}^2)}$ (Pagnacco et al., 2011) and the outcomes presented in Tables 1 and 2, it can be calculated that the present noise levels did cause an overestimation of the

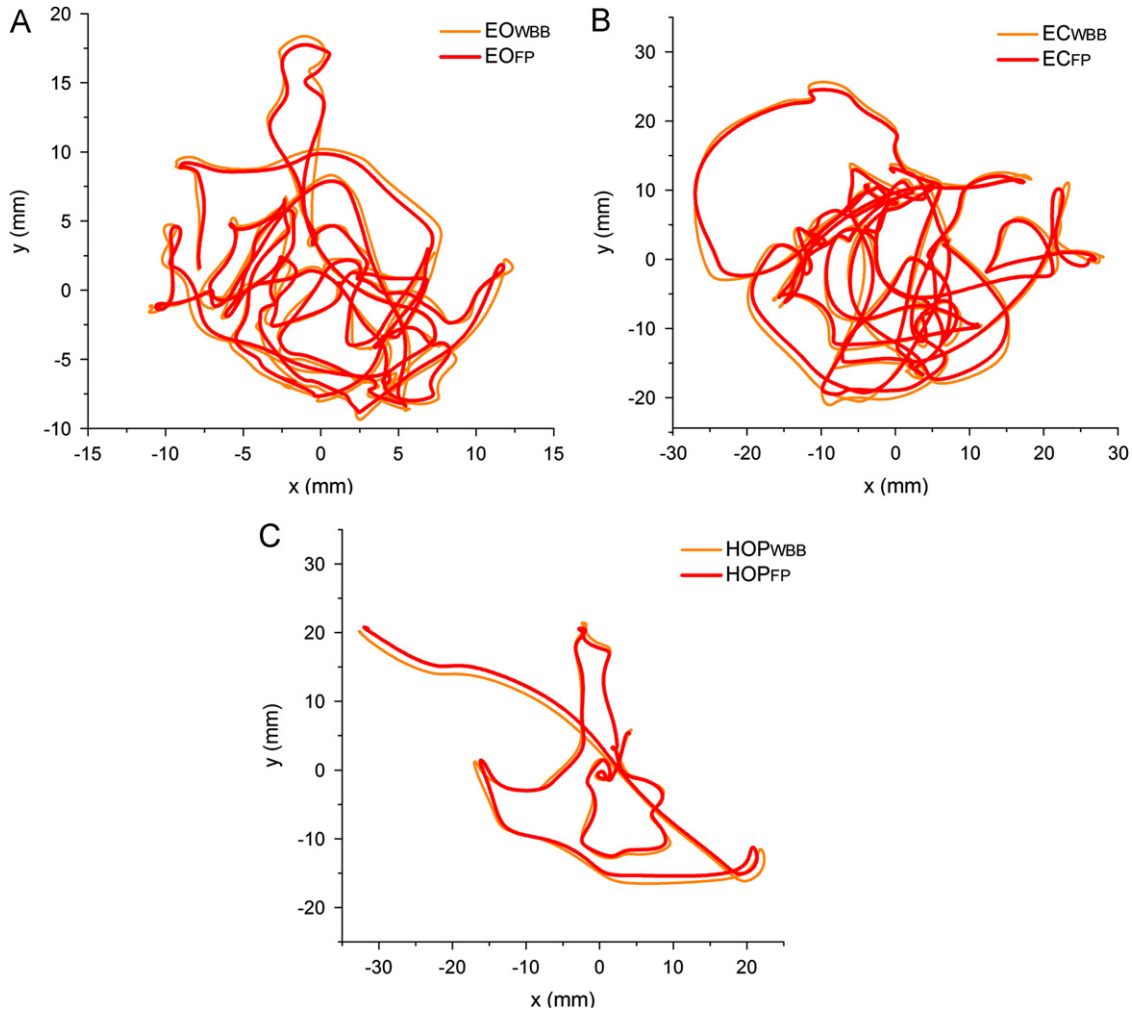

Fig. 1. Typical example of center of pressure trajectories for WBB and FP data on EO (1A), EC (1B), and HOP (1C). Presented trials were comparable with the mean outcome and mean differences of the present sample.

Table 2

	EO		EC		HOP	
Horizontal GRF (FP) [a]						
Absolute mean (|**N**|)	3.2 (0.9; 0.9)		6.9 (2.7; 1.6)		8.9 (3.3; 0.7)	
Absolute max (|**N**|)	9.3 (3.7; 3.0)		25.1 (14.5; 8.3)		39.2 (16.5; 11.8)	
CoP trajectory [a,b]						
RMS–*x* (mm)	0.33 (0.10; 0.05)		0.58 (0.17; 0.11)		0.74 (0.34; 0.14)	
RMS–*y* (mm)	0.31 (0.16; 0.06)		0.63 (0.19; 0.12)		0.74 (0.27; 0.09)	
r–*x*	0.999 (0.001; 0.000)		0.999 (0.000; 0.000)		0.999 (0.003; 0.001)	
r–*y*	0.998 (0.001; 0.001)		0.999 (0.000; 0.000)		0.997 (0.005; 0.002)	
	FP	**WBB**	**FP**	**WBB**	**FP**	**WBB**
'CoP path velocity' [c]						
Outcome (mm s^{-1}) [a]	32.3 (6.5; 5.1)	34.0 (6.5; 5.1)	75.5 (21.3; 14.7)	78.4 (22.2; 15.0)	74.1 (20.4; 10.3)	77.5 (21.4; 10.7)
Difference (%) [a]	5.3 (1.9; 1.4)		4.0 (1.4; 1.1)		4.6 (1.6; 0.8)	
r	0.997		0.999		0.998	
'mean CoP sway' [c]						
Outcome (mm) [a]	7.1 (1.7; 1.1)	7.3 (1.8; 1.1)	13.6 (3.9; 2.2)	14.1 (4.0; 2.3)	13.7 (5.4; 2.5)	14.2 (5.5; 2.6)
Difference (%) [a]	3.5 (0.7; 0.3)		3.7 (0.5; 0.3)		3.6 (1.0; 0.3)	
r	1.000		1.000		1.000	

RMS, root-mean-square error between WBB and FP data.
r, Pearson's correlation coefficient between WBB and FP data.
Diff, difference as percentage relative to FP value.

[a] Presented as mean (SD over all trials; SD over subjects after averaging over 10 trials).
[b] Within trial analysis for *x* and *y* directions.
[c] Analysis over 140 data points (14 subjects × 10 trials) per task.

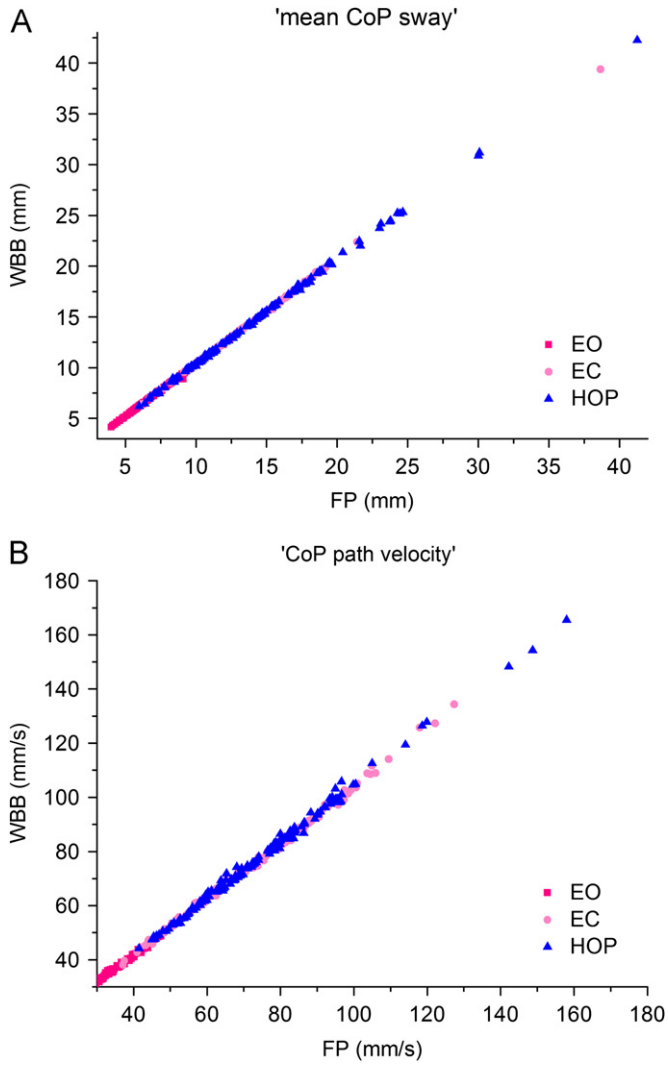

Fig. 2. Scatter plot of WBB and FP outcomes on EO, EC, and HOP concerning the balance measures 'mean CoP sway' (2A) and 'CoP path velocity' (2B) for all 420 trials ($14 \times 10 \times 3$).

WBB 'CoP path velocity' values compared to FP by 0.8 to 1.3% for EO and 0.1 to 0.3% for EC and HOP. Although this is marginal for the present study, the noise driven overestimation might become more relevant when balance testing comprises considerably lower CoP velocities. Additionally, the results presented in Table 1 suggest that the noise is dependent on weight. Consequently, there might be an increase in noise bias on the WBB when CoP velocities are low. For example, bipedal stance with the eyes open has been reported to show CoP velocities of approximately 10 mm s^{-1} (Raymakers et al., 2005). A 'CoP path velocity' of 10 mm s^{-1} measured on the FP, would equate to an overestimation on the WBB of 15% for a participant with a weight of 60 kg, but only 8% for a participant with a weight of 80 kg (see Table 1). As actual measurement conditions and inter-device variations may affect noise levels as well, we would advise to routinely perform dead weight noise testing in the actual measurement environment and to relate this to the expected 'CoP path velocity'. The latter also provides opportunities to correct 'CoP path velocity' outcomes for noise effects. It should be noted that the effect of noise on CoP position measures, such as 'mean CoP sway', is negligible.

In addition to noise effects, the overestimation of WBB 'CoP path velocity' (4.0–5.3%) is likely to be caused by a negligence of horizontal forces. The latter accounts for the overestimation of WBB 'mean CoP sway' values ($\sim$ 3.5%). Finally, the low sample rate restricts the applicability of the WBB to measurements in the relatively low frequency domain. However, this seems not to be of importance concerning standing balance assessments (Ross et al., 2009; Schmid et al., 2002).

Despite the limitations of the WBB, we found that the present balance measures showed very high Pearson's correlations and small differences in error between FP and WBB. This indicates linearity and consistency of measurement outcomes. In addition, Fig. 2 reveals that the WBB measurement error is consistent over the present ranges of 'CoP path velocity' and 'mean CoP sway'. Therefore, it is unlikely that errors of WBB estimates in single-leg balance tests will lead to false conclusions. However, this should be verified for populations and tests that differ from the ranges of the present study, especially concerning body weight and 'CoP path velocity' values.

In conclusion, the WBB is sufficiently accurate for quantifying CoP trajectory, and overall amplitude and velocity during single-leg stance balance tasks.

Conflict of interest statement

There is no conflict of interest.

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