The Validity of an Oculus Rift to Assess Postural Changes During Balance Tasks

Jonathan Marchetto and **W. Geoffrey Wright,** Temple University, Philadelphia, Pennsylvania, USA

Objective: To investigate whether shifts in head position, measured via an Oculus Rift head-mounted display (HMD), is a valid measure of whole-body postural stability.

Background: The inverted single-link pendulum model of balance suggests shifts in whole-body center of mass can be estimated from individual body segments. However, whether head position describes postural stability such as center-of-pressure (COP) remains unclear.

Method: Participants (N = 10) performed six conditions while wearing an HMD and performing a previously validated virtual reality (VR)-based balance assessment. COP was recorded with a Wii Balance Board force plate (WBB), while an HMD recorded linear and angular head displacement. Visual input was presented in the HMD (stable scene, dark scene, or dynamic scene) and somatosensory information (with or without foam) was varied across each condition. The HMD time series data were compared with the criterion-measure WBB.

Results: Significant correlations were found between COP measures (standard deviation, range, sway area, velocity) and head-centered angular and linear displacements (roll, pitch, mediolateral and anteroposterior directions).

Conclusions: The Oculus Rift HMD shows promise as a measure of postural stability without additional posturography equipment. These findings support the application of VR HMD technology for assessment of postural stability across a variety of challenging conditions.

Application: The human factors and ergonomic benefit of such an approach is in its portability, low cost, and widespread availability for clinic and home-based investigation of postural disturbances. Fall injury affects millions of people annually, so assessment of fall risk and treatment of the underlying causes has enormous public health benefit.

Keywords: virtual environments, posture, balance, fall-risk, medical devices

Address correspondence to W. Geoffrey Wright, Neuromotor Sciences Program, Department of Physical Therapy, Temple University, 1301 Cecil B. Moore Ave., Philadelphia, PA 19122, USA; e-mail: wrightw@temple.edu.

HUMAN FACTORS
Vol. 61, No. 8, December 2019, pp. 1340–1352
DOI: 10.1177/0018720819835088
Article reuse guidelines: sagepub.com/journals-permissions

BACKGROUND

Postural control in humans is a complex motor skill that involves control of the center of mass and postural orientation in space (Horak, 2006). Because such a system is inherently unstable, even minor alterations in central or peripheral nervous system function can disrupt human balance; postural deficits are well established following many neurological pathologies and injuries, including Parkinson's disease, traumatic brain injury, and vestibular dysfunction (Black, Shupert, Horak, & Nashner, 1988; Broglio & Puetz, 2008; Vaugoyeau, Viel, Assaiante, Amblard, & Azulay, 2007; Wright et al., 2017). Such deficits are associated with numerous risks, most notably an increased fall risk, which may expose patients to further injury, head trauma, and a reduced quality of life (Tinetti, Doucette, Claus, & Marottoli, 1995). Therefore, the development of methodologies to better understand the physiology and pathology of postural control with accurate and objective assessments has been a major focus in clinical posturography research.

An inverted pendulum model has been used to describe bipedal postural control during quiet stance (Winter, 1995). Quiet stance has traditionally been measured using a force plate to assess center of pressure (COP), which reflects the force vector distributed by the surface of the feet onto the ground (Winter, 1995). Although this approach is validated and widely used in research, it can be cost prohibitive and difficult to utilize outside of laboratory settings (Seimetz, Tan, Katayama, & Lockhart, 2012). This presents challenges to clinicians and on-field personnel who wish to accurately and objectively assess balance in vulnerable populations, after injury, or as a performance metric. As a result, researchers have begun exploring new means of assessing posture using accelerometers, gyroscopes, and infrared cameras to validate alternative or supplementary measures (Clark et al., 2012; Najafi et al., 2010; Seimetz et al., 2012).

Virtual reality (VR) head mounted displays (HMDs), such as the Oculus Rift, provide another possible alternative to balance assessment. HMDs traditionally have been used as a means of immersing participants in a simulated 3-dimensional environment (Desai, Desai, Ajmera, & Mehta, 2014; Wright, 2014). In order to track a participant's movement through the virtual environment and update the virtual scene accordingly, Oculus uses an internal Invensense MPU-6000 gyroscope and accelerometer, a Honeywell HMC5983 magnetometer, as well as a camera that tracks infrared LEDs on the HMD's surface (LaValle, Yershova, Katsev, & Antonov, 2014). This allows users to collect both angular rotation (roll, pitch, and yaw) and linear displacement in the *x* (medial-lateral), *y* (up-down), and *z* (anterior-posterior) planes with great fidelity and minimal time lag (Lavalle et al., 2014). Based on prior research using comparable hardware to assess posture, researchers have recently utilized Oculus to assess cervical joint kinematics (Xu, Chen, Lin, & Radwin, 2015) and head and neck joint position sense (Robins, Teodoro, & Wright, 2017). However, research has yet to establish whether the Oculus, and therefore head position, can be utilized to assess whole-body posture during traditional balance tasks.

The inverted pendulum model of balance suggests that whole-body posture can be measured by assessing movements of individual body segments (Gage, Winter, Frank, & Adkin, 2014; Winter, 1995). Although there is evidence that such a model inadequately explains postural behavior (Loram, Maganaris, & Lakie, 2009; Wright, Ivanenko, & Gurfinkel, 2012), in some conditions, this model may suffice. For example, ankle angular displacement in the sagittal plane has been shown to be highly correlated to whole-body center of mass during quiet stance (Winter, 1995). One very parsimonious model assumes human postural control can be estimated by a single link inverted pendulum model, where shifts in ankle position should be reflected through concurrent shifts in head position (Gage et al., 2014; Winter, 1995). As a result, changes in COP would be highly correlated to changes in linear and angular head position. Whether this model is sufficient to describe postural control across a variety of conditions, such as with varied visual and somatosensory input, is not well established.

Traditional measures of postural control, such as COP, have been shown to be susceptible to manipulations in visual and somatosensory input, suggesting that these systems are functionally linked in humans' control of balance (Horak, 2006; Schärli, van de Langenberg, Murer, & Müller, 2012; Wright, McDevitt, & Appiah-Kubi, 2015; Wright et al., 2017). Less is known, however, about how these measures impact head position and whether changes to COP occur independent of shifts in head position. For example, the importance of head stabilization in balance maintenance is highlighted in young children, who have a limited ability to control head position in response to gaze shifts, resulting in large body sway (Schärli et al., 2012). Other evidence shows significant reductions in postural tilt and sway when vibrotactile stimulation is utilized to inform participants about head sway position (Wall, Weinberg, Schmidt, & Krebs, 2001). Taken together, these findings underscore the close link between head position and whole-body postural. However, consensus as to the strength of this relationship, and how it is impacted by manipulations in visual and somatosensory information, is still unclear.

Therefore, a goal of this study is to investigate the relationship between traditional postural control measurements (i.e., COP from a force plate) and Oculus to assess its validity in measuring whole-body posture. This study's first aim is to assess the correlation between standard deviation and range of mediolateral (ML) and anterior-posterior (AP) linear displacement in a Wii Balance Board (WBB; Nintendo Co., Kyoto, Japan) and linear displacement in the ML and AP directions and angular displacement in the roll and pitch directions in the Oculus Rift. Secondly, this study will explore whether traditional measures used in research exploring postural control and balance, such as COP sway area and COP velocity, correlate with equivalent measures of head sway and velocity recorded by the Oculus Rift. Lastly, this study aims to establish how differing visual and somatosensory conditions in a validated balance task could affect these correlations due to changes in the multilink control of the body. If high correlations were found between all variables in the WBB and Oculus Rift across all balance conditions, this would support an inverted pendulum model of postural control. Perhaps more importantly, these findings would help establish the Oculus Rift as a valid measure of postural control, which could offer a low-cost and highly portable alternative to clinicians and field staff to assess balance without the need for additional posturography equipment.

METHOD

Subjects

Ten healthy young adults (average age: 26.1 ± 5.47 years; age range: 20–35 years; sex: 9 women and 1 man; height: 166 ± 7.8 cm) volunteered for the study. All participants were screened via verbal self-report to exclude those with visual, vestibular, or oculomotor disorders. All subjects signed a consent form approved by the institutional review board at Temple University—in accordance with the guidelines of the Helsinki Accords—before participating in this study.

Instrumentation

A custom-designed VR-based balance assessment user-interface was programmed in Unity game engine (Unity, San Francisco, CA) that integrates visual stimuli and real-time recording of kinematic time series data from an Oculus Rift Development Kit 2 (DK2) HMD (Oculus, Irvine, CA). This VR HMD-based solution (VBAL) presents an immersive visual scene in a VR setting while utilizing the Oculus Rift's internal sensors for rotational tracking of the head and its external infrared camera for positional tracking in the *x* (ML), *y* (up-down), and *z* (AP) directions. The Oculus Rift DK2 uses a 100° diagonal field of view and weighs 440g. Data is recorded with a 60 Hz sampling rate, and visual scenes are presented at a 100 Hz refresh rate. A sample scene is pictured in Figure 1.

VBAL will be compared with a previously validated forceplate-based criterion measure for balance assessment (i.e., VETS device; see Wright et al., 2015, 2017). That original postural assessment device wirelessly collecting COP data from a WBB via a Bluetooth connection while presenting simulated VR motion on a large commercially available television monitor (e.g., VIZIO E-Series Razor LED 60 inches, Irvine, CA). The WBB has been validated for posturography (Bartlett, Ting, & Bingham, 2014; Young, Ferguson, Brault, & Craig, 2010) and has been validated with our VETS user interface in previous studies (Wright et al., 2015; 2017). The WBB and HMD were synced through VETS and VBAL, respectively, using a custom code that sends a TCP/IP command to a local host, which then sends a common signal back to both programs to begin and end data collection. The delay between both systems has been found to be within a 16 ms tolerance.

Procedure

Participants performed the protocol while wearing the Oculus Rift HMD and standing on the WBB. The visual stimuli were viewed in the HMD, which presented the immersive visual scene and collected head movement data while the WBB collected COP data. The six conditions were (1) EO-Firm—eyes open with stable support surface (i.e., WBB) and earth-referenced, stable visual scene, (2) EC-Firm—eyes closed with stable support surface and dark screen, (3) DYN-Firm—eyes open with a stable support surface and rotating visual scene (40°/sec roll), (4) EO-Foam—eyes open with unstable support (Airex foam pad placed on top of the WBB) and stable visual scene, (5) EC-Foam—eyes closed with unstable foam support and dark screen, and (6) DYN-Foam—eyes open with unstable foam support and rotating scene. The protocol consisted of three 30-s trials in each condition before proceeding to the next condition, in which the trials were presented in the same order (EO-Firm, EC-Firm, DYN-Firm, EO-Foam, EC-Foam, DYN-Foam) for all participants. Between the firm and foam conditions participants were required to remove the HMD to safely step off the WBB. Approximately 1 min was taken to add the foam and reposition the participant before continuing on to the next block of trials, otherwise a 5 to 10 s break occurred between each trial. Two stance widths were tested: Half of the participants stood with feet together, the other half stood with feet at a comfortable distance approximately shoulder width apart (20–25 cm).

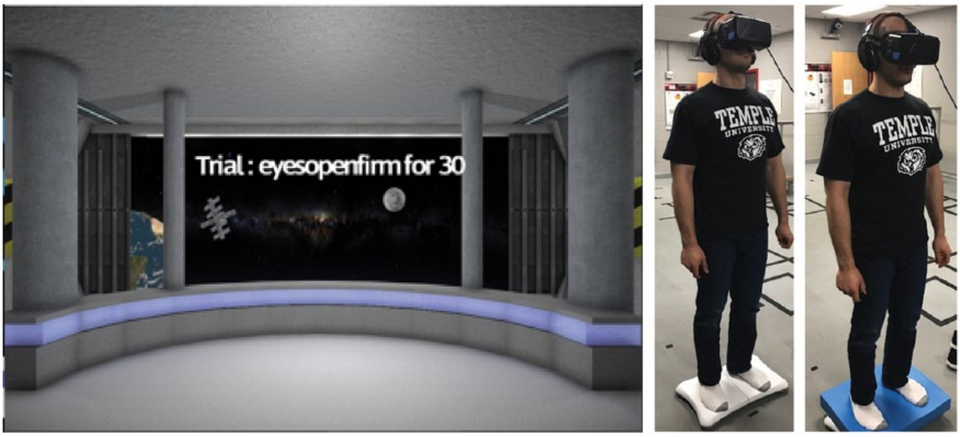

Figure 1. Experimental set-up: (left) sample scene from VR HMD-based solution (VBAL) eyes open with stable support surface (EO-Firm); (right) concurrent postural test of Wii Balance Board (WBB) and head-mounted display (HMD) on firm and foam surface.

Before testing began, the visual scene was centered within the HMD so that each participant's visual regard was oriented in the same direction. During the eyes-open and dynamic visual scene conditions, participants were instructed to maintain their gaze at a central point within the virtual environment. During the eyes-closed conditions, participants were instructed to keep their eyes closed for the duration of the trial. Participants were instructed to maintain a stable, upright posture and neutral head position in each condition to the best of their ability. An experimenter stood behind the participants during testing to guard the participants against falls.

Data Analysis

The dataset was filtered first by truncating the first and last 5 s of data, reflecting 1,000 and 750 samples for the WBB and Oculus, respectively. This is done to remove any postural adjustments in the beginning or end of the trial that reflect uninstructed changes to body position as participants adapt to the next trial. The COP data is then postprocessed to clean nonphysiological data points (e.g., not a number [NaN], data spikes) and filtered using a second order low-pass Butterworth filter with a 10 Hz cutoff frequency. The HMD has built-in complementary and predictive filters, which low-pass filter the head movement; however, exact parameters were not accessible (Lavalle et al., 2014).

A repeated-measures analysis of variance (rmANOVA) was used to compare WBB to HMD measures and validate the reliability of each device to detect change in the level of postural challenge across conditions. The effects of stance width were compared using a between-groups rmANOVA. Mauchley's test of sphericity was used to determine whether adjustments to degrees of freedom were necessary. Pearson product-moment correlation coefficients (2-tailed) were calculated in MATLAB within all 10 participants' data to establish criterion-related validity by comparing COP data to head-centered kinematics from the HMD on a trial-by-trial basis. Specifically, correlations for standard deviation were calculated between the WBB and the HMD's ability to track participants' postural variability relative to the established WBB criterion-measure. Furthermore, correlations between the range of linear and angular displacement in the WBB and Oculus were calculated to assess the relationship between the distributions of postural data in both devices. Correlations were calculated for all conditions together as well as for each condition separately to determine which conditions were more or less like a single-link inverted pendulum.

Correlations between COP velocity in the WBB and linear and angular velocity in the Oculus Rift were calculated. Additionally, correlations were calculated for COP sway area and sway area of the head measured through the Oculus Rift infrared camera. Sway area was calculated using a 2D plane formed by ML and AP linear displacement of the foot pressure for COP sway area (WBB) or head position for head sway area (HMD). Using a principal component analysis, an ellipse around the ML and AP axes was approximated using the first two eigenvectors (Wright et al., 2015). COP velocity and sway area are common measures used to measure postural control during quiet stance and have been shown to be valid and reliable measures of balance assessment (Amiridis, Hatzitaki, & Arabatzi, 2003; Doyle, Hsiao-Wecksler, Ragan, & Rosengren, 2007; Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996).

Lastly, time series correlations were calculated to assess how changes in COP and linear and angular head kinematics relate throughout the duration of a trial. This was calculated to assess whether changes in position of the head and COP are related not only in terms of the average across an entire trial, but also if there is moment-to-moment correlation. For the time-series analysis, data were not truncated at the beginning and end of trials.

RESULTS

Standard Deviation

The overall Pearson product-moment correlation coefficients between ML COP (WBB) and ML head sway (HMD) standard deviations was $r = .93$, $N = 167$, $p < .001$. Very high, positive correlations ($r = .90–.97$, $p < .001$) were also found in each condition, except EO-Foam. The overall correlation between ML COP and head roll (HMD) standard deviations was $r = .45$, $n = 167$, $p < .001$. In addition, moderate to high positive correlations were found in the EC-Firm, DYN-Firm, and DYN-Foam, $r = .41–.70$, $p \leq .03–.001$. Correlation coefficients and their significance for each condition for standard deviation of linear displacement and linear versus angular displacement are outlined in Table 1.

The overall correlation between AP COP (WBB) and AP head sway (HMD) standard deviations was $r = .88$, $N = 167$, $p < .001$. High to very high, positive correlations ($r = .81–.91$, $p < .001$) were found in each condition, except EO-Foam. A moderate correlation was found in the EO-Foam condition, $r = .47$, $p = .01$. The overall correlation between AP COP and head pitch (HMD) standard deviations was $r = .540$, $n = 167$, $p < .001$. Looking at the conditions separately, moderate correlations were found in the EO-Firm, EC-Firm, DYN-Firm, and DYN-Foam, $r = .50–.67$, $p \leq .03–.001$. Correlation coefficients and their significance for each condition for standard deviation of linear displacement and linear versus angular displacement are outlined in Table 1.

Range

Pearson product-moment correlation coefficients were high to very high when comparing the ML ranges of linear displacement between the WBB and Oculus in all conditions ($r = .76–.95$, $p < .001$), except EO-Foam. Moderate to high correlations were found in EO-Firm, DYN-Firm, and DYN-Foam for the ML range of linear displacement and roll range, $r = .43–.68$, $p \leq .018$. Correlation coefficients and their significance for each condition for range of linear displacement and linear versus angular displacement are outlined in Table 2.

Correlation coefficients for range between the WBB and Oculus in the AP direction proved to be less sensitive, as no correlations were found to be significant across all conditions. However, moderate to high correlations in range of AP and pitch were found in the EO-Firm, DYN-Firm, and DYN-Foam conditions, $r = .60–.72$, $p \leq .01–.001$. Correlation coefficients and their significance for each condition for range of linear displacement and linear versus angular displacement are outlined in Table 2.

COP Sway Velocity Versus Head Sway Velocity and Head Angular Velocity

The overall correlation between COP sway velocity and head sway velocity was $r = .91$, $N = 167$, $p < .001$. Very high correlations were found between COP velocity in the WBB and linear head velocity in the HMD in the EO-Firm, DYN-Firm, and DYN-Foam conditions, $r = .87–.95$, $p < .001$. A moderate correlation was found in the EO-Foam condition, $r = .47$, $p = .01$.

The overall correlation between COP sway velocity and head sway angular velocity was $r = .89$, $N = 167$, $p < .001$. High to very high correlations were found between COP velocity and

TABLE 1: Correlations for Standard Deviation Between WBB and HMD

		WBB ML *SD*		WBB AP *SD*	
Condition		HMD ML *SD*	HMD Roll *SD*	HMD AP *SD*	HMD Pitch *SD*
EO-Firm	*r*	.97	.32	.91	.67
	p	<.001	=.08, ns	<.001	<.001
	N	30	30	30	30
EC-Firm	*r*	.92	.41	.84	.50
	p	<.001	=.03	<.001	<.01
	N	30	30	30	30
DYN-Firm	*r*	.96	.56	.84	.64
	p	<.001	<.01	<.001	<.001
	N	29	29	29	29
EO-Foam	*r*	.19	.18	.47	−.09
	p	=.32, ns	=.36, ns	=.01	=.64, ns
	N	29	29	29	29
EC-Foam	*r*	.92	.11	.81	−.01
	p	< .001	= .56, ns	< .001	= .94, ns
	N	30	30	30	30
DYN-Foam	*r*	.90	.70	.82	.51
	p	< .001	< .01	< .001	= .03
	N	19	19	19	19

Note. WBB = Wii Balance Board; ML = mediolateral; SD = standard deviation; AP = anterior-posterior; HMD = head-mounted display; EO-Firm = eyes open with stable support surface; EC-Firm = eyes closed with stable support surface and dark screen; DYN-Firm = eyes open with a stable support surface and rotating visual scene; EO-Foam = eyes open with unstable support; EC-Foam = eyes closed with unstable foam support and dark screen; DYN-Foam = eyes open with unstable foam support and rotating scene; ns = not significant.

angular head velocity in the EO-Firm, DYN-Firm, EC-Foam, and DYN-Foam conditions, r = .68–.95, p < .001. Moderate correlations were found in the EO-Foam and EC-Foam conditions, r = .41–.60, $p \leq$ .03–.001. Correlation coefficients and their significance for each condition for COP velocity and linear and angular head velocity are outlined in Table 3.

COP and Head Sway Area

The overall correlation between COP sway area and head sway area was r = .90, N = 167, p < .001 (Figure 2a). High to very high correlations were found in all conditions, r = .82–.96, p < .001, except EO-Foam. Correlation coefficients and their significance for each condition for COP area and head sway area are outlined in Table 4. A rmANOVA showed no significant difference between the WBB and HMD measurement devices ($F_{1,9}$ = .72, p = .42, ns). Moreover, there was no interaction between measurement devices and postural conditions ($F_{2,18}$ = 1.41, p = .27, ns), which can be seen in that both COP sway area and head sway area showed a nearly identical pattern of increasing sway area across conditions (Figure 2b).

Stance Width

A comparison between stance widths showed a significant difference in sway area ($F_{1,8}$ = 13.1, p = .007), which was also found for sway velocity ($F_{1,8}$ = 8.12, p = .022). Shoulder width stance was more stable than feet together stance (i.e., smaller sway area and velocity) for both measurement devices. For shoulder width stance, COP and head sway areas were all strongly correlated (r = .85–.95), whereas feet together stance correlations, though significant, tended

TABLE 2: Correlations Between WBB and HMD for Linear and Angular Range

		WBB ML Range		WBB AP Range	
Condition		HMD ML Range	HMD Roll Range	HMD AP Range	HMD Pitch Range
EO-Firm	*r*	.95	.43	–.24	.72
	p	< .001	= .02	= .20, ns	< .001
	N	30	30	30	30
EC-Firm	*r*	.93	.31	.06	.36
	p	< .001	= .09, ns	= .76, ns	= .05
	N	30	30	30	30
DYN-Firm	*r*	.98	.68	.05	.70
	p	< .001	< .001	= .81 ns	< .001
	N	29	29	29	29
EO-Foam	*r*	.17	.18	–.10	–.17
	p	= .38, ns	= .36 ns	= .60, ns	= .38, ns
	N	29	29	29	29
EC-Foam	*r*	.84	.19	–.29	.21
	p	< .001	= .33, ns	= .12, ns	= .27, ns
	N	30	30	30	30
DYN-Foam	*r*	.76	.52	–.09	.60
	p	< .001	= .02	= .72, ns	< .01
	N	19	19	19	19

Note. WBB = Wii Balance Board; COP = center-of-pressure; HMD = head-mounted display; ML = mediolateral; AP = anterior-posterior; HMD = head-mounted display; EO-Firm = eyes open with stable support surface; EC-Firm = eyes closed with stable support surface and dark screen; DYN-Firm = eyes open with a stable support surface and rotating visual scene; EO-Foam = eyes open with unstable support; EC-Foam = eyes closed with unstable foam support and dark screen; DYN-Foam = eyes open with unstable foam support and rotating scene; ns = not significant.

to be less. The notable exception was EO-Foam, which was highly significant for shoulder width stance, but uncorrelated for feet together stance. Correlation coefficients and their significance for each condition for sway area in two stances are outlined in Table 4.

Time Series Correlation

The strength of the Pearson product-moment correlation coefficients comparing moment-to-moment changes in the WBB positional time series data to the HMD positional or angular time series data showed greater dependence on which variables were being compared than which condition was being tested (N = 3,000 in each correlation). The ML linear displacements of both systems showed moderate to very high correlations across all six conditions (r = .57–.86, $p < .001$), with 5% of trials as outliers. The AP linear displacements of both systems show high to very high correlations across all six conditions (r = .73–.86, $p < .001$), with less than 4% of trials as outliers. Although the COP ML linear displacement (WBB) relative to head roll (HMD) also included some trials with very low but significant correlations, on average none of the conditions showed significant correlations (r = .015–.105, $p > .10$, ns). Similarly, head pitch (HMD) showed some trials with low but significant correlations, on average none of the conditions showed significant correlations (r = .014–.092, $p > .10$, ns). The average correlation coefficients and their significance for each condition are outlined in Table 5.

TABLE 3: Correlations Between WBB and HMD for Sway Velocity

		WBB COP Velocity	
Condition		HMD Linear Velocity	HMD Angular Velocity
EO-Firm	*r*	.87	.76
	p	< .001	< .001
	N	30	30
EC-Firm	*r*	.12	.60
	p	= .53, ns	= .001
	N	30	30
DYN-Firm	*r*	.95	.95
	p	< .001	< .001
	N	29	29
EO-Foam	*r*	.47	.41
	p	= .01	= .03
	N	29	29
EC-Foam	*r*	.30	.68
	p	= .11, ns	< .001
	N	30	30
DYN-Foam	*r*	.89	.91
	p	< .001	< .001
	N	19	19

Note. WBB = Wii Balance Board; COP = center-of-pressure; HMD = head-mounted display; ML = mediolateral; AP = anterior-posterior; HMD = head-mounted display; EO-Firm = eyes open with stable support surface; EC-Firm = eyes closed with stable support surface and dark screen; DYN-Firm = eyes open with a stable support surface and rotating visual scene; EO-Foam = eyes open with unstable support; EC-Foam = eyes closed with unstable foam support and dark screen; DYN-Foam = eyes open with unstable foam support and rotating scene; ns = not significant.

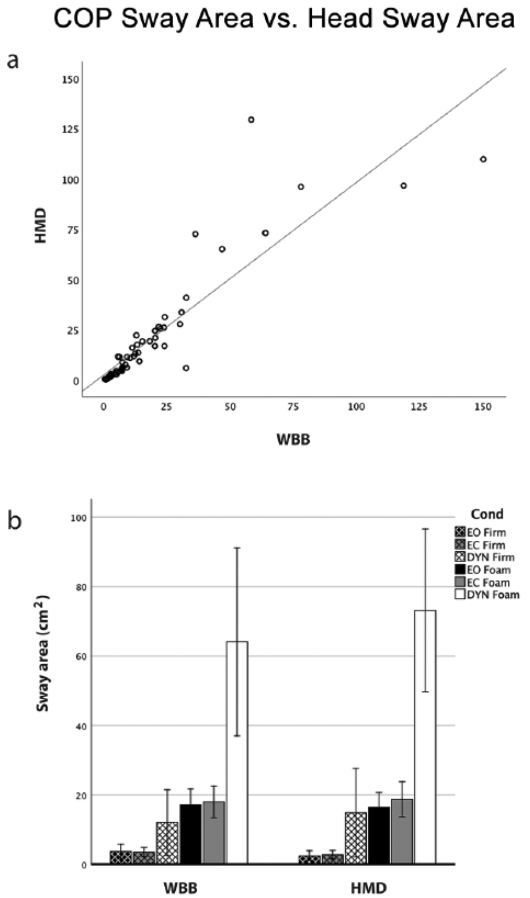

Figure 2. (a) Correlation coefficient between center-of-pressure (COP) sway area (Wii Balance Board [WBB]) and head sway area (head-mounted display [HMD]) was highly significant ($r = .90$, $p < .001$). (b) Average sway area for each participant with each measurement device. COP sway area (*left*) and head sway area (*right*) did not differ between measurement devices across six postural conditions ($p = .42$, ns). Error bars: ±2 *SE*.

DISCUSSION

This study aimed to investigate the utility of an Oculus Rift VR head mounted display to track whole body posture during a balance task by analyzing the relationship between angular and linear kinematics in the HMD to center of pressure data from a WBB. Comparing the HMD to the WBB, high to very high correlations were identified across a number of conditions and between comparisons of both linear and angular displacement. The findings suggest when COP shifts in order to maintain a stable posture, concurrent and systematically corresponding shifts in head position occur.

In order for a head-centered kinematic measurement device such as the Oculus Rift to be a suitable postural assessment metric, high correlations between the HMD and criterion-measure (i.e., COP collected on WBB) across variable conditions would be necessary. Notably, numer-

TABLE 4: Correlations Between WBB and HMD for Sway Area in Two Stances

Condition		WBB COP Sway Area		
		HMD Pooled Stances Head Sway Area	HMD Shoulder Width Head Sway Area	HMD Feet Together Head Sway Area
EO-Firm	*r*	.93	.92	.87
	p	< .001	< .001	< .001
	N	30	15	15
EC-Firm	*r*	.92	.92	.84
	p	< .001	< .001	< .001
	N	30	15	15
DYN-Firm	*r*	.96	.85	.97
	p	< .001	< .001	< .001
	N	29	15	14
EO-Foam	*r*	.30	.92	−.01
	p	= .11, ns	< .001	= .97, ns
	N	29	15	14
EC-Foam	*r*	.86	.91	.78
	p	<.001	<.001	<.001
	N	30	15	15
DYN-Foam	*r*	.82	.95	.85
	p	< .001	< .001	= .017
	N	19	12	7

Note. WBB = Wii Balance Board; COP = center-of-pressure; HMD = head-mounted display; ML = mediolateral; AP = anterior-posterior; EO-Firm = eyes open with stable support surface; EC-Firm = eyes closed with stable support surface and dark screen; DYN-Firm = eyes open with a stable support surface and rotating visual scene; EO-Foam = eyes open with unstable support; EC-Foam = eyes closed with unstable foam support and dark screen; DYN-Foam = eyes open with unstable foam support and rotating scene; ns = not significant.

ous conditions produced high to very high correlations between the WBB and HMD across multiple comparisons of standard deviation and range in the ML and AP directions. Comparisons between ML and Roll or AP and Pitch positional variables tended to be much lower. This suggests that for changes in linear displacement in the frontal and sagittal plane, COP and head position are very closely related providing some evidence for single-link inverted pendular postural responses.

The strongest evidence for criterion-related validity of the HMD for postural behavior came from the comparison of COP sway area to head sway area. Here we found significant correlations across all conditions, suggesting a strong relationship between shifts in the forces distributed through the feet and concurrent changes in head position. This relation was further informed when two stances were compared. Standing with feet at a comfortable shoulder width distance resulted in very strong correspondence between the head sway area and the COP sway area. Previous research looking at the effect of VR on balance utilizing a feet-together stance or a tandem stance found a significant increase in shoulder pitch and roll angle while wearing a HMD (Horlings et al., 2009). Although their study utilized transducers on the shoulder instead of the internal HMD gyroscope, their findings do support the current findings, which showed roll and pitch of the head tended to be

TABLE 5: Average Correlations Across Participants Comparing WBB Positional to HMD Positional and Angular Time Series Data

		WBB ML Position		WBB AP Position	
Condition		HMD ML Position	HMD Roll Angle	HMD AP Position	HMD Pitch Angle
EO-Firm	*r*	.67 ^	.09	.85	.09
	p	< .001	> .10, ns	< .001	> .10, ns
EC-Firm	*r*	.76	.09	.82	.06
	p	< .001	= .10, ns	< .001	> .10, ns
DYN-Firm	*r*	.57 ^	.04	.73 ^	.02
	p	< .001	< .01	< .001	> .10, ns
EO-Foam	*r*	.79 ^	.02	.77	− .01
	p	< .001	= .05, ns	< .001	= .08, ns
EC-Foam	*r*	.86	.11	.86	−.06
	p	< .001	> .10, ns	< .001	= .05, ns
DYN-Foam	*r*	.73	.06	.81	.03
	p	< .001	> .10, ns	< .001	= .12, ns

Note. WBB = Wii Balance Board; ML = mediolateral; AP = anterior-posterior; HMD = head-mounted display; EO-Firm = eyes open with stable support surface; EC-Firm = eyes closed with stable support surface and dark screen; DYN-Firm = eyes open with a stable support surface and rotating visual scene; EO-Foam = eyes open with unstable support; EC-Foam = eyes closed with unstable foam support and dark screen; DYN-Foam = eyes open with unstable foam support and rotating scene; ns = not significant. 10 participants per condition, 3 trials per participant, $N = 3{,}000$ per trial; ^ indicates outlier in that condition.

correlated less with COP than linear head position.

Head velocity tended to be strongly correlated with COP velocity. Comparisons showed significant correlations between COP velocity and head angular velocity in all conditions, and for head linear velocity, all but two conditions were significantly correlated. This suggests that a strong relationship exists between changes in the location of the vertical ground reaction force vector through the feet and the velocity of the head as it moves through space, both in terms of linear head position and shifts in head angle. However, this relationship may be influenced by access to visual information, as both the EC-Firm and EC-Foam conditions did not consistently show as large a correlation between WBB and HMD as in other conditions. A few possible explanations may explain why the postural behavior did not behave as a single-link inverted pendulum in these EC conditions. One explanation may be that dynamic visual information, despite producing worsened balance performance when compared to the eyes closed conditions, as has previously been found, produces large shifts in both head and COP velocity (Wall et al., 2001, Wright et al., 2015, 2017). An alternative explanation is that in the eyes-closed conditions, the head and body movements may decouple, as there is a lack of any visual reference that could help guide head position. Some evidence suggests that the hierarchy of sensory dependence during postural control is visually dominated followed by somatosensory, then vestibular dependence (Peterka, 2002). If visual input is removed, the lower body and feet may become more proprioceptively driven to include long-latency reflexes, whereas there may be a shift in reliance to the vestibular system for head stabilization (Nashner, 1976).

Reliance on the vestibular and neck reflexes should not be overlooked. The vestibulocollic, vestibulospinal, and cervicocollic reflexes can play an important role in head stabilization during the balance task utilized in our study. In low frequency movements, such as quiet stance at a frequency of <0.7 Hz, there is evidence that the head and trunk move coherently (Honegger, Van Spijker, & Allum, 2012). Pitch rotations fall at or below the sensitivity threshold for canal afferents, forcing reliance on the otolith organs, so it has been suggested that at low frequencies, locking the head to the trunk reduces reliance on these signals to maintain upright posture (Honegger et al., 2012). However, at higher frequencies (>3Hz) the vestibular and cervicocollic reflexes may be activated which dampens the antiphase motion of the body to stabilize the head. In the eyes-closed conditions, the removal of access to a visual anchor may have increased the reliance on the vestibular system for postural maintenance. As a result, if sway velocity was higher in these conditions, a shift in sensory weighting toward the vestibular system could amplify the head-stabilizing effects of the vestibular and cervicocollic reflexes while the ankles remained independently controlled. This could offer an alternative explanation to why these conditions produced significant correlations for linear and angular velocity.

These findings appear to lend some support to a single link inverted pendulum model of postural control, because we were able to estimate whole body posture through individual body segments (Gage et al., 2004; Winter, 1995). High to very high correlations between the COP shifts and changes in head position suggest that whole body posture can be predicted using head position. However, a number of the aspects of these findings suggest that a multilink inverted pendulum model could be used to account for more of the postural variability. Although average COP and head position metrics were highly correlated on a per-trial basis, the angular rotation of the head, especially in the time-series analysis, was more weakly—and in many cases, not—correlated with COP. For instance, shifts in COP in the ML direction and changes in roll angle of the head are not as strongly correlated. This may be due to the fact that the ankles generally rely on dorsiflexion and plantar flexion in the sagittal plane to reduce large shifts in posture, whereas shifts in the frontal plane may be compensated through other means such as shifts in hip position (Gage et al., 2004; Winter, 1995). It is also important to note that the validity of using a head-centered measure of posture does not need to strictly follow a single-link inverted pendulum model, which it does not always do (Loram et al., 2009; Wright et al., 2012). For example, a two-link inverted pendulum could show increases in head movement range, variability, and velocity yet be phase-shifted relative to COP; however, these measures could still be accurately reflecting decreased postural stability. Because COP has been viewed as the gold-standard for postural assessment (Hass & Burden, 2000; Palmieri, Ingersoll, Stone, & Krause, 2002), the consistently high correlations found in the various metrics used in the current study provide evidence supporting the use of Oculus Rift or perhaps any HMD for measuring posture using head position alone. To our knowledge, this is one of the first studies to directly investigate the link between COP and head position during a balance task.

Although these findings are promising for establishing the validity of the Oculus Rift for assessing whole-body posture, some limitations should be noted. First, it appears that to achieve large correlations between head position and COP, the balance task used needs to control for head position, meaning independent and/or volitional head movements that are not part of a balance adjustment strategy would decrease an HMD's validity for tracking postural stability during quiet stance. For instance, a shift in head position to intentionally look at something would corrupt the head tracking data's representativeness of postural stability. Another limitation that should be noted is that in the DYN-Foam condition, data had to be removed due to failure to complete the postural task. In these trials, participants were unable to maintain quiet stance without assistance from a spotter, therefore, these were marked as falls, and the data was removed from analysis. This limitation can be addressed, in part, by an alternative analysis that we performed. By replacing these falls with a threshold value demarcating the limits of stability, we found that correlations between postural metrics increased even more. Future embodiments of our custom-designed posturography device will incorporate automatic identification of falls to improve applicability.

APPLICATION

Our findings suggest that an Oculus Rift HMD may be used as an alternative method for assessing postural control without additional posturography equipment. Additionally, our study is perhaps the first to highlight the conditions and variables in which strong relationships between COP and the head exist. Accordingly, these findings can contribute to the development of a standalone application or device that uses only linear and angular head position to track postural control in a variety of settings and populations. Therefore, the Oculus Rift could provide a low-cost, highly portable alternative for postural testing, which has clinical applicability in assessment of numerous conditions that impact balance, such as traumatic brain injury, Parkinson's disease, and vestibular disorders.

ACKNOWLEDGMENTS

We thank the Motion-Action-Perception Laboratory members for their respective contributions. W. Geoffrey Wright, for conception and design of the study, data analysis and interpretation, critical revision of the article, acquiring funding, and final approval for publication. Jonathan Marchetto, for data collection, analysis and interpretation, first draft of the manuscript, and final approval for publication. Gregory Teodoro, for programming the user interface. IP from PCT/US18/13777 are included in this article.

KEY POINTS

- Good to excellent correlations exist when comparing COP, measured through a force plate, to linear and angular position and velocity of the head, measured through the Oculus Rift's infrared camera and internal sensors.
- This relationship seems to be affected by access to visual and somatosensory information. For example, in the eyes-closed conditions, correlations in COP velocity and head velocity were not significant for linear head velocity and smaller for angular head velocity, compared with conditions with a stable or dynamic visual scene.
- These findings suggest that head position kinematics measured by the Oculus Rift may offer a valid measure of postural control without additional posturography equipment.
- This could greatly increase the portability and decrease the cost of postural assessment devices compared with traditional measurement tools such as a force-plate, lending greater utility to clinicians and researchers who work with populations that suffer from disorders that affect postural control.
- Although average COP and head position metrics were highly correlated on a per-trial basis, the angular rotation of the head, especially in the time series analysis, was more weakly correlated with COP. This suggests at least a multilink inverted pendulum model is needed to account for these postural control results.

REFERENCES

Amiridis, I. G., Hatzitaki, V., & Arabatzi, F. (2003). Age-induced modifications of static postural control in humans. *Neuroscience Letters*, *350*(3), 137–140.

Bartlett, H. L., Ting, L. H., & Bingham, J. T. (2014). Accuracy of force and center of pressure measures of the Wii Balance Board. *Gait Posture*, *39*(1), 224–228.

Black, F. O., Shupert, C. L., Horak, F. B., & Nashner, L. M. (1988). Abnormal postural control associated with peripheral vestibular disorders. *Progress in Brain Research*, *76*, 263–275.

Broglio, S. P., & Puetz, T. W. (2008). The effect of sport concussion on neurocognitive function, self-report symptoms and postural control. *Sports Medicine*, *38*(1), 53–67.

Clark, R. A., Pua, Y. H., Fortin, K., Ritchie, C., Webster, K. E., Denehy, L., & Bryant, A. L. (2012). Validity of the Microsoft Kinect for assessment of postural control. *Gait & Posture*, *36*(3), 372–377.

Desai, P. R., Desai, P. N., Ajmera, K. D., & Mehta, K. (2014). A review paper on oculus rift-a virtual reality headset. *International Journal of Engineering Trends and Technology*, *13*, 175–179.

Doyle, R. J., Hsiao-Wecksler, E. T., Ragan, B. G., & Rosengren, K. S. (2007). Generalizability of center of pressure measures of quiet standing. *Gait & Posture*, *25*(2), 166–171.

Gage, W. H., Winter, D. A., Frank, J. S., & Adkin, A. L. (2004). Kinematic and kinetic validity of the inverted pendulum model in quiet standing. *Gait & Posture*, *19*(2), 124–132.

Haas, B. M., & Burden, A. M. (2000). Validity of weight distribution and sway measurements of the Balance Performance Monitor. *Physiotherapy Research International*, *5*(1), 19–32.

Honegger, F., Van Spijker, G. J., & Allum, J. H. J. (2012). Coordination of the head with respect to the trunk and pelvis in the roll and pitch planes during quiet stance. *Neuroscience*, *213*, 62–71.

Horak, F. B. (2006). Postural orientation and equilibrium: What do we need to know about neural control of balance to prevent falls? *Age and Ageing*, *35*(Suppl_2), ii7–ii11.

Horlings, C. G., Carpenter, M. G., Küng, U. M., Honegger, F., Wiederhold, B., & Allum, J. H. (2009). Influence of virtual reality on postural stability during movements of quiet stance. *Neuroscience Letters*, *451*(3), 227–231.

LaValle, S. M., Yershova, A., Katsev, M., & Antonov, M. (2014, May). Head tracking for the Oculus Rift. In *2014 IEEE International Conference on Robotics and Automation* (pp. 187–194). doi:10.1109/ICRA.2014.6906608

Loram, I. D., Maganaris, C. N., & Lakie, M. (2009). Paradoxical muscle movement during postural control. *Medicine and Science in Sports and Exercise*, *41*, 198–204.

Najafi, B., Horn, D., Marclay, S., Crews, R. T., Wu, S., & Wrobel, J. S. (2010). Assessing postural control and postural control strategy in diabetes patients using innovative and wearable technology. *Journal of Diabetes Science and Technology*, *4*(4), 780–791.

Nashner, L. M. (1976). Adapting reflexes controlling the human posture. *Experimental Brain Research*, *26*(1), 59–72.

Palmieri, R. M., Ingersoll, C. D., Stone, M. B., & Krause, B. A. (2002). Center-of-pressure parameters used in the assessment of postural control. *Journal of Sport Rehabilitation*, *11*(1), 51–66.

Peterka, R. J. (2002). Sensorimotor integration in human postural control. *Journal of Neurophysiology*, *88*, 1097–1118.

Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., & Myklebust, B. M. (1996). Measures of postural steadiness: Differences between healthy young and elderly adults. *IEEE Transactions on Biomedical Engineering*, *43*(9), 956–966.

Robins, R. K., Teodoro, G., & Wright, W. G. (2017). Head and neck position sense using a memory-driven joint position matching study. *IEEE International Conference on Virtual Rehabilitation 2017*, Montreal, QC, pp. 1–2. doi:10.1109/ICVR.2017.8007503

Schärli, A. M., van de Langenberg, R., Murer, K., & Müller, R. M. (2012). The influence of gaze behaviour on postural control from early childhood into adulthood. *Gait & Posture*, *36*(1), 78–84.

Seimetz, C., Tan, D., Katayama, R., & Lockhart, T. (2012). A comparison between methods of measuring postural stability: Force plates versus accelerometers. *Biomedical Sciences Instrumentation*, *48*, 386.

Tinetti, M. E., Doucette, J., Claus, E., & Marottoli, R. (1995). Risk factors for serious injury during falls by older persons in the community. *Journal of the American Geriatrics Society*, *43*(11), 1214–1221.

Vaugoyeau, M., Viel, S., Assaiante, C., Amblard, B., & Azulay, J. P. (2007). Impaired vertical postural control and proprioceptive integration deficits in Parkinson's disease. *Neuroscience*, *146*(2), 852–863.

Wall, C., Weinberg, M. S., Schmidt, P. B., & Krebs, D. E. (2001). Balance prosthesis based on micromechanical sensors using vibrotactile feedback of tilt. *IEEE Transactions on Biomedical Engineering*, *48*(10), 1153–1161.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait & Posture*, *3*(4), 193–214.

Wright, W. G. (2014). Using virtual reality to augment perception, enhance sensorimotor adaptation, and change our minds. *Frontiers in Systems Neuroscience*, *8*, 56.

Wright, W. G., Ivanenko, Y. P., & Gurfinkel, V. S. (2012). Foot anatomy specialization for postural sensation and control. *Journal of Neurophysiology*, *107*(5), 1513–1521.

Wright, W. G., McDevitt, J., & Appiah-Kubi, K. O. (2015). A portable virtual reality balance device to assess mild traumatic brain injury symptoms: A pilot validation study. *Proceedings of the 2015 International Conference on Virtual Rehabilitation* (pp. 72–79). doi:10.1109/ICVR.2015.7358591

Wright, W. G., McDevitt, J., Tierney, R., Haran, F. J., Appiah-Kubi, K. O., & Dumont, A. (2017). Assessing subacute mild traumatic brain injury with a portable virtual reality balance device. *Disability and Rehabilitation*, *39*(15), 1564–1572.

Xu, X., Chen, K. B., Lin, J. H., & Radwin, R. G. (2015). The accuracy of the Oculus Rift virtual reality head-mounted display during cervical spine mobility measurement. *Journal of Biomechanics*, *48*(4), 721–724.

Young, W., Ferguson, S., Brault, S., & Craig, C. (2010). Assessing and training standing balance in older adults: A novel approach using the 'Nintendo Wii' Balance Board. *Gait Posture*, *33*, 303–305.

Jonathan Marchetto earned an MA in exercise science from Montclair State University in 2016. He is currently a PhD student in the neuromotor sciences program at Temple University in Philadelphia, PA.

W. Geoffrey Wright is associate professor and director of the neuromotor sciences program at Temple University. He earned his PhD in cognitive neuroscience from Brandeis University in 2002.

Date received: July 16, 2018
Date accepted: February 7, 2019