Reliability and Validity of a Virtual Reality-Based System for Evaluating Postural Stability

Huey-Wen Liang, Shao-Yu Chi, Bo-Yuan Chen, and Yaw-Huei Hwang

***Abstract*— Postural stability is an important indicator of balance and is commonly evaluated in neurorehabilitation. We proposed a system based on a virtual reality (HTC Vive) system with a tracker at the lumbar area. The position data of the tracker were obtained through detection of the sensors on the tracker by the VR system. The reliability and validity of these sway parameters to measure postural stability were evaluated. Twenty healthy adults had their postural sway measured with this system and a force platform system under four stance conditions, with wide- or narrow-stance and eyes open or closed. The path data from both systems were computed to obtain the following parameters: the mean distance and the mean velocity in the medial-lateral and anterior-posterior directions and the 95% confidence ellipse area. The reliability of the Vive-based sway measures was tested with intraclass correlation coefficients (ICCs). The convergent validity was tested against the center of pressure (COP) parameters from the force platform system. Finally, the discriminative validity was tested for the above four conditions. The results indicated that the Vive-based sway parameters had moderate to high reliability (ICCs: 0.56 ∼ 0.90) across four conditions and correlated moderately to very highly with the COP parameters ($r = 0.420 \sim 0.959$). Bland-Altman plotting showed generally good agreement, with negative offset for the Vive-based sway parameters. The sway parameters obtained by the Vive-based system also discriminated well among the tasks. In conclusion, the results support this system as a simple and easy-to-use tool to evaluate postural stability with acceptable reliability and validity.**

***Index Terms*— Biomechanics, instruments, measurement, virtual reality.**

I. Introduction

TRUNK balance control refers to the production of forces to maintain the body's center of mass (COM) within the limits of the base of support. This ability is a prerequisite for the execution of daily activities and encompasses the acts of maintaining, achieving and restoring the line of gravity within the base of support [1]. The assessment of this ability is critical for patients with diseases and conditions that can influence postural control, not only for diagnosis but also for fall risk assessment and outcome evaluation.

Postural functions can be assessed by instrumental or non-instrumental methods, and the instrumental methods can be further categorized as kinetic and kinematic methods [2]. The most widely used kinetic device is a force platform, which can obtain vertical ground reaction force data to calculate the medial-lateral (ML) and anterior-posterior (AP) time series of the center of pressure (COP) during postural tests. Accordingly, several COP variables can be derived from the raw COP data to determine postural functions and analyze the mechanisms involved in postural control. COP-based measures include time-domain distance, time-domain area, time-domain hybrid and frequency-domain measures [3]. Regarding the kinematic methods, the commonly used options include video recordings, 3-D motion capture systems with markers, body-worn accelerometer-based devices, electrogoniometers and laser-displacement sensors [4]. The results can be used to calculate joint angle measurements, compute the COM and analyze the movements of a specific body landmark, such as the lumbar area. In general, a larger sway distance, higher velocity, larger path area and higher frequency reflect a poorer postural steadiness [3].

Most of these instrumental postural measurements involve technical devices that are either costly or require analytic technology. To address the needs of clinicians and researchers with limited resources, some researchers designed a swaymeter to record the displacement of the body in the horizontal plane [5]. This device provides an indirect measure of the COM movement as the device is fixed at approximately the level of the COM (pelvis) to record motion of the body in two dimensions during quiet stance. The displacement data have moderate association with that obtained from a force platform device and fair discriminative validity for different tasks (eyes open *vs*. eyes close) or populations (elderly *vs*. young persons). However, this method is nondigital and is limited to static posture evaluation only. Therefore, there is still an unmet need for a simple and feasible measurement for postural sway.

Recently, emerging technologies have provided new options to record body displacement. Virtual reality (VR) has been

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used widely in gaming, and is gaining increasing attention in health care in recent decades, especially in the fields of psychology, neuroscience and rehabilitation [6]–[8]. VR creates an enriched and immersive environment, enables real-time performance and can be incorporated into rehabilitation treatments. One of the key development of the hardware is to improve the tracking precision of the devices used for virtual environment interaction. Some examples include the controller or the trackers. Therefore, recent studies have started to explore the feasibility of collecting kinematic data with these immersive VR systems 9,10,11. For example, the HTC Vive is a fully immersive VR environment equipped with Vive trackers that can be used for object tracking. These trackers are wireless, lightweight, wearable and inexpensive and their accuracy for recording position and orientation during human movement has been validated in several studies [9], [11]–[13]. Potentially, trackers can be positioned at the lumbar area, similar to a laser displacement sensors or swaymeters [5], [14]. The trackers can then be used to record the body displacement with the VR system. Nevertheless, the reliability and validity of measuring postural stability with such a setup has not been tested.

The purpose of this study was to design Vive-based posturography as an inexpensive and simple method for evaluating postural stability. We hypothesized that this technique had acceptable reliability and good agreement with the COP parameters gained from a force platform system and could discriminate postural sway in different stance conditions. We examined the following: 1) the same-day test-retest repeatability of the sway parameters obtained from the Vive trackers; 2) the convergent validity of the Vive-based sway parameters against a force platform system; and 3) the discriminant validity of the Vive-based postural sway parameters in differentiating tasks requiring different levels of postural control.

II. Method

A. Study Design and Participants

This was a cross-sectional study with repeated measurements. The participants were 20 healthy adults who were at least 20 years old (Table I). None of the participants had any known history of visual, cognitive, cardiovascular, neurological or musculoskeletal problems and they could all walk normally. This study was approved by the Ethical Committee of the National Taiwan University Hospital (approval number: 201904129RINC, date of approval: 20/06/2019), and written informed consent was obtained prior to participation. We estimated that 20 participants were needed based on the expectation of an ICC of 0.85 as compared to the acceptable value of 0.70, with a type-1 error of 0.05 and a power of 0.80.

B. Virtual Reality System

We used the Vive Pro system (HTC, Inc. Taiwan), which included two infrared laser emitter units (lighthouses, SteamVR Base Stations V1.0) and three wireless trackers (Steam VR Tracking V1.0). The head-mounted display was not worn in the current study. The two lighthouses were positioned diagonally 3.7 meters apart and mounted at a height of 2.1 m, angled downward at an angle of 25-30°, and connected by a synchronization cable. The signal emitted from the lighthouses enables the system to synchronize the two lighthouses and to track the position and orientation of the trackers within the VR space. The tracker was sized 99.65∗42.27 mm and weighed 89 g and had sensors with surface-mounted photodiode arrays to receive signals from the lighthouses with a field of view (FOV) of 270 degrees [15]. Three trackers were docked with a standard tripod cradle head on elastic straps to secure them onto the appropriate body parts (Figure 1B). Two of the trackers were positioned on each dorsal foot and one was positioned in the lumbar area at the level of the posterior superior iliac spine (Figure 1A).

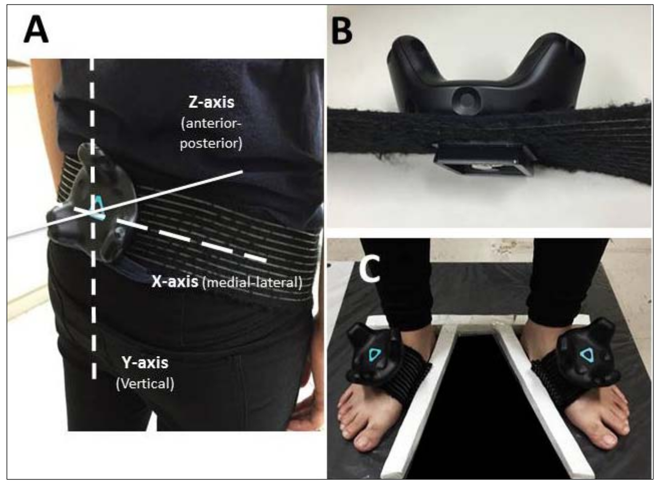

Fig. 1. The setup of trackers for body position data collection. The trackers were docked with a standard tripod cradle head on elastic straps (B) and positioned on each dorsal foot (C) and lumbar area at pelvic level (A).

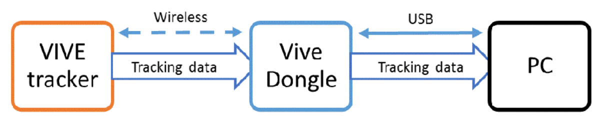

Fig. 2. The transmission of trackers' data.

The signals were then transmitted to dongles by a wireless interface in VR and then through a USB to a computer (Figure 2). A custom C# script and the SteamVR (Valve Corp, Washington, USA) plugin for Unity3D were used to provide integration with the HTC Vive to record the position and orientation of trackers with a sampling rate of 100 Hz. A local coordinate system was established according to the human body frame system with the center at the level of the lumbar tracker as the origin. First, two trackers at the feet were assigned to the height of the lumbar tracker and used to determine the X-axis (medial-lateral) by the software. Second, the Y-axis (vertical) was determined by the cross-product of two vectors, the vector of the lumbar to the right foot tracker and the vector from the lumbar to the left foot tracker. Third, the Z-axis (anterior-posterior,) was determined by the cross-product of the X and Y axes (Figure 1A). The displacements

TABLE I
DEMOGRAPHIC DATA OF ALL PARTICIPANTS

Variable	Results
Age (years, mean ± SD)	33.1 ± 7.3
Male (n, %)	10 (50%)
Body height (cm, mean ± SD)	167.9 ± 8.2
Body weight (kg, mean ± SD)	64.0 ± 12.7
BMI (kg/m^2, mean ± SD)	22.6 ± 3.2

of the lumbar tracker in the horizontal plane were recorded as a bivariate (AP and ML directions) time series to compute several time-domain sway parameters described below.

C. COP Measurements

COP was measured simultaneously with the FDM-S pressure plate (Zebris Medical GmbH, Germany), with a sample rate of 100 Hz. This system incorporates 2,560 sensors on an area of 54 by 34 cm, giving a resolution of approximately 8.7 sensors per square inch and accuracy within 5 %. The platform was connected to the WinFDM software to analyze the COP at stance. We exported the raw data of the COP positions on the X (ML directions) and Y (AP directions) as csv files to calculate the COP parameters.

D. Procedure

The assessments of postural sway were performed in a bipedal stance for a total of four conditions: wide-base with eyes open (W-EO), wide-base with eyes closed (W-EC), narrow-base with eyes open (N-EO), and narrow-base with eyes closed (N-EC). For the wide-base stance, the participants stood with their heels 15 cm apart and toes rotated outward by 25 degrees along a V-shaped separator made of heavy cardboard to standardize the posture (Figure 1C) [15]. This separator was placed at the same position on the force platform and then removed once the participant's feet were positioned. For the narrow-base stance, the participants stood with their feet side-by-side. Simultaneous recordings from the Vive and the force platform were obtained throughout each testing. The testing order of the four conditions was randomized and two tests (test 1 and test 2) of 30 seconds were conducted for each condition, yielding a total of eight tests for each participant. The participants were offered a seat and rested for one minute in between tests. For the same-day retesting, the above conditions were tested again in the same order within one hour (trial 1 and trial 2).

E. Calculation of Postural Sway and COP Parameters

The time series position data from the platform system were used to calculate COP measures and those from the lumbar tracker were used as a proxy of postural sway and calculated for the center of body displacement (COD). The bivariate distribution path defined by AP and ML coordinates according to the origin of the coordinate systems of either the force platform or the tracker system. Both sets of data (COD and COP) were passed through a fourth-order zero phase Butterworth low-pass digital filter with a 5-Hz cut-off frequency in MATLAB (MathWorks Inc, Massachusetts, USA). The following postural sway parameters were computed for the filtered COP and COD data according to the following equations [3]:

1. Mean distances in the medial-lateral (ML) direction ($MDIST_{ML}$) and anterior-posterior (AP) direction ($MDIST_{AP}$): the average ML and AP distance from the mean COP and COD;
2. Mean velocity-ML ($MVELO_{ML}$) and mean velocity-AP ($MVELO_{AP}$): the mean velocity of the COP and COD in the ML and AP direction;
3. 95% confidence ellipse area (AREA-CE): the area of the 95% bivariate confidence ellipse of COP and COD.

F. Statistical Analysis

The descriptive statistics (means and standard deviations) of five sway parameters ($MDIST_{ML}$, $MDIST_{AP}$, $MVELO_{ML}$, $MVELO_{AP}$ and AREA-CE) were calculated for the data from the Vive-based system and Zebris FDM. The reliability of the Vive-based parameters was calculated for the average data of test 1 and 2 from both trials. The intraclass correlation coefficient (ICC) estimates and their 95% confident intervals were calculated for each condition and across 5 sway parameters based on a two-way mixed-effects model for consistency ($ICC_{(3,n)}$). An ICC higher than 0.90 was considered excellent, between 0.75 and 0.9 was good, between 0.5 and 0.75 was moderate and less than 0.5 was poor [16]. We also examined the correlation and agreement between the COD and COP parameters with the Pearson product moment correlation coefficients and Bland and Altman plots for four conditions from both trials [17], [18]. The size of the correlation coefficient (r) was interpreted as being very high (0.90), high (0.7 to 0.9), moderate (0.5 to 0.7) or low (0.3 to 0.5) [19]. A general linear model (GLM) repeated measures was conducted to compare the main effect of the stance width (stance), eyes-open/closed (eye) and their interaction (stance∗eye) on the COD parameters obtained from the Vive-based system under four conditions. Statistical analyses were performed using SPSS 15.0 for Windows (SPSS Inc, Chicago, USA) with a statistical significance of $p < 0.05$.

III. RESULTS

The descriptive statistics for the sway parameters from the Vive-based and force platform systems presented in Table II and Figure 3 show an example of the COP and COD data of one participant in N-EO stance. The $MVELO_{ML}$, $MVELO_{AP}$ and AREA-EC were considerably larger for the COP compared to COD data, with a tendency towards an increase in difference as the magnitude of sway increased.

The Pearson's correlation coefficient between the COP and COD parameters in the pooled data ranged from 0.737 to 0.938 and was mostly moderate to very high among the different conditions (Table III). An exception was observed for $MVELO_{ML}$ ($r = 0.420$) in EC-W and $MVELO_{AP}$ in EO-W ($r = 0.492$). Additionally, the average difference of the measures of pooled data from the two trials ranged between

TABLE II

MEAN (STANDARD DEVIATION) OF SWAY PARAMETERS FOR FOUR CONDITIONS WITH THE VIVE-BASED AND FORCE PLATFORM SYSTEMS

Variable	Wide stance, eye open		Wide stance, eye closed		Narrow stance, eye open		Narrow stance, eye closed	
	Vive	Force platform	Vive	Force platform	Vive	Force platform	Vive	Force platform
1st trial								
mean distance- ML (mm)	0.99 (0.5)	0.96 (0.29)	1.19 (1.06)	1.14 (0.59)	2.98 (1.08)	3.38 (1.08)	3.17 (1.37)	3.60 (1.08)
mean distance- AP (mm)	2.90 (1.16)	2.81 (0.92)	2.81 (1.43)	3.16 (1.19)	3.02 (1.37)	3.10 (1.01)	3.45 (1.77)	3.77 (1.52)
mean velocity – ML (mm/s)	2.26 (0.52)	4.46 (0.97)	2.38 (0.80)	4.59 (0.87)	3.64 (0.69)	7.10 (1.87)	4.23 (1.25)	9.45 (2.81)
mean velocity- AP (mm/s)	2.90 (0.77)	5.50 (1.43)	3.34 (0.84)	6.90 (1.40)	3.32 (0.64)	6.62 (1.63)	4.11 (1.25)	9.14 (2.81)
95% conf. ellipse area (mm^2)	83.8 (69.5)	91.19 (52.18)	106.08 (141.37)	114.61 (87.29)	255.08 (173.19)	312.24 (159.07)	326.56 (268.19)	434.37 (326.11)
2nd trial								
mean distance- ML (mm)	1.08 (0.53)	1.08 (0.38)	1.20 (0.61)	1.19 (0.55)	3.27 (1.26)	3.40 (1.49)	3.37 (1.28)	3.58 (1.36)
mean distance- AP (mm)	2.88 (1.41)	2.75 (1.15)	2.88 (1.03)	3.03 (1.01)	3.64 (1.57)	4.32 (1.24)	4.20 (1. 87)	3.58 (1.36)
mean velocity – ML (mm/s)	2.16 (0.47)	4.21 (0.97)	2.24 (0.37)	4.66 (0.76)	3.95 (1.08)	7.41 (2.20)	4.49 (1.28)	9.75 (3.88)
mean velocity- AP (mm/s)	2.82 (0.63)	5.00 (1.32)	3.34 (0.82)	6.38 (1.68)	3.34 (0.84)	6.46 (1.61)	4.31 (1.13)	8.65 (2.42)
95% conf. ellipse area (mm^2)	91.93 (88.29)	88.07 (56.39)	92.18 (71.85)	105.87 (63.11)	337.54 (224.27)	354.78 (202.75)	421.00 (299.10)	477.32 (341.44)

Note: ML=medial-lateral; AP= anterior-posterior direction.

TABLE III

PEARSON CORRELATION COEFFICIENTS (r) AND BLAND-ALTMAN OFFSETS (LIMITS OF AGREEMENT) FOR THE SWAY PARAMETERS FROM THE VIVE TRACKERS AND THE FORCE PLATFORM SYSTEM

Variable	Pooled data	Wide stance		Narrow stance	
		Eyes open	Eyes closed	Eyes open	Eyes closed
mean distance- ML (mm)	*r* = 0.928 ** −0.14 (−1.26~0.99)	*r* = 0.664 * 0.02 (−0.74~0.77)	*r* = 0.674 * 0.03 (−1.21~1.26)	*r* = 0.856** −0.27 (−1.46~0.93)	*r* = 0.922 ** −0.32 (−1.43, 0.79)
mean distance- AP (mm)	*r* =0.916** -0.04 (−1.25~1.77)	*r* = 0.929** 0.11 (−0.87~0.18)	*r* =0.874** −0.25 (−1.43~0.93)	*r* = 0.947 ** 0.07 (−1.02~1.16)	*r* = 0915** −0.11 (−1.57~1.36)
mean velocity – ML (mm/s)	*r* =0.843 *** −3.29 (−7.42 ~0.84)	*r* = 0.569 ** −2.12 (−0.56~-0.369)	*r* = 0.420 * −2.32 (−3.86~0.78)	*r* =0.568** −3.46 (−0.17~-6.75)	*r* = 0.818 ** −5.24 (−10.69~0.21)
mean velocity- AP (mm/s)	*r* = 0.928** −0.14 (−1.26~0.99)	*r* = 0.664 * 0.02 (−0.74~0.77)	*r* = 0.674 * 0.03 (−1.21~1.26)	*r* = 0.856 ** −0.27 (−1.46~0.93)	*r* = 0.922 ** −0.32 (−1.43, 0.79)
95% conf. ellipse area (mm^2)	*r* =0.916 ** -0.04 (−1.25~1.77)	*r* = 0.929** 0.11 (−0.87~0.18)	*r* =0.874*** −0.25 (−1.43~0.93)	*r* = 0.947 ** 0.07 (−1.02~1.16)	*r* = 0915 ** −0.11 (−1.57~1.36)

Note: ML=medial-lateral; AP= anterior-posterior; * indicates p <0.01, ** indicates p < 0.001.

-68.3 % ($MDIST_{AP}$) and -1.3% ($MVELO_{ML}$). The mean COD-COP offsets for $MVELO_{ML}$, $MVELO_{AP}$ and AREA-CE indicated that a lower magnitude was recorded from the Vive-based system. The Bland-Altman plots indicated good agreement between the COP and COD data across the conditions, and few data points fell outside of the limits of agreement (Figure 4). Minimal COD-COP offsets were observed for $MDIST_{ML}$ for the wide stance.

ICCs were calculated for five COD parameters across four conditions (Table IV). Moderate to good repeatability was present across the conditions, with ICCs ranging from 0.522 to 0.898. The lowest ICCs were reported for $MDIST_{ML}$ for the EO-N condition and $MDIST_{AP}$ for the EC-W condition. Generally, the mean velocity had a higher ICC than the mean distance, and all the ICCs were higher than 0.80 for $MVELO_{ML}$. There was a tendency for the eyes open conditions to have higher ICCs than the eyes-closed conditions in the wide stance but not in the narrow stance.

The GLM repeated measures analysis showed a significant main effect for the stance condition for all the COD parameters ($p < 0.001$) (Table V). Additionally, the eye condition had only a significant main effect on $MVELO_{AP}$ ($p < 0.001$) and AREA-CE ($p = 0.048$). The interaction of stance and eye conditions had no significant effect on any parameters and is not presented.

IV. DISCUSSION

We evaluated the validity and reliability of a Vive-based system for evaluating postural stability in four stance conditions. This setup included a tracker positioned at the lumbar area, which was approximately the level of the center of mass (COM), and the position data of the trackers were used as a proxy for body sway in two dimensions. Our goal was not to establish a tool with high accuracy or precision as a motion capture system, but to show the feasibility of a commercially available product as an estimation of postural control with acceptable reliability and validity. Our data support this system as a reliable measure, with ICCs ranging from 0.52 to 0.90 across four conditions for repeated tests on the same day. The body displacement parameters obtained from the lumbar tracker correlated moderately to very highly with the COP parameters obtained from a force platform

TABLE IV
INTRACLASS CORRELATION COEFFICIENTS (95% CIs) FOR SWAY PARAMETERS OBTAINED FROM THE VIVE TRACKERS ACROSS FOUR CONDITIONS

	Eyes Open		Eyes Closed	
	Eyes Open	Eyes Closed	Eyes Open	Eyes Closed
mean distance- ML	0.70 (0.25~0.88)	0.87 (0.68~0.95)	0.56(-0.13~0.82)	0.71 (0.25~0.88)
mean distance- AP	0.85 (0.63~0.94)	0.52 (-0.21~0.81)	0.78 (0.43~0.91)	0.77(0.43~0.91)
mean velocity – ML	0.90 (0.74~0.96)	0.80 (0.49~0.92)	0.83 (0.57~0.93)	0.83 (0.56~0.93)
mean velocity- AP	0.76 (0.39~0.90)	0.61 (0.01~0.85)	0.84 (0.61~0.94)	0.85 (0.63~0.94)
95% conf. ellipse area	0.78 (0.46~0.91)	0.70 (0.21~0.88)	0.85 (0.62~0.94)	0.8 (0.59~0.93)

Note: ML=medial-lateral; AP= anterior-posterior.

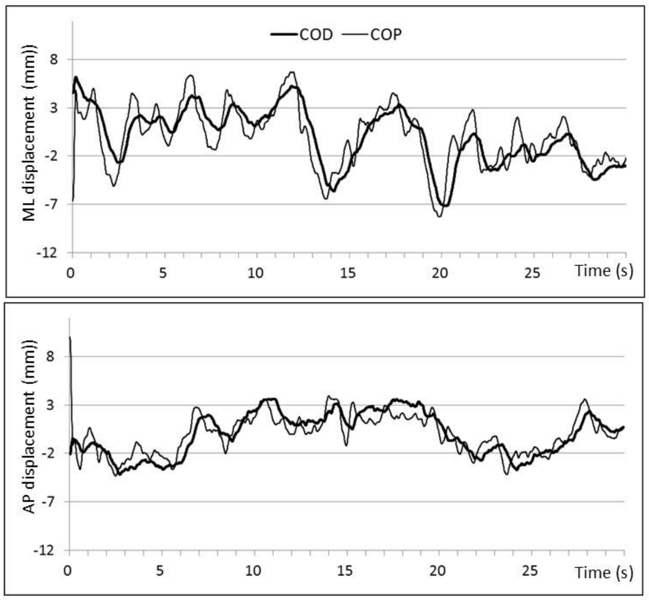

Fig. 3. An example of the Vive-based (COD) versus the force platform (COP) data at anterior-posterior and medial-lateral displacement during the eyes-closed, wide-stance (EC-W) stance.

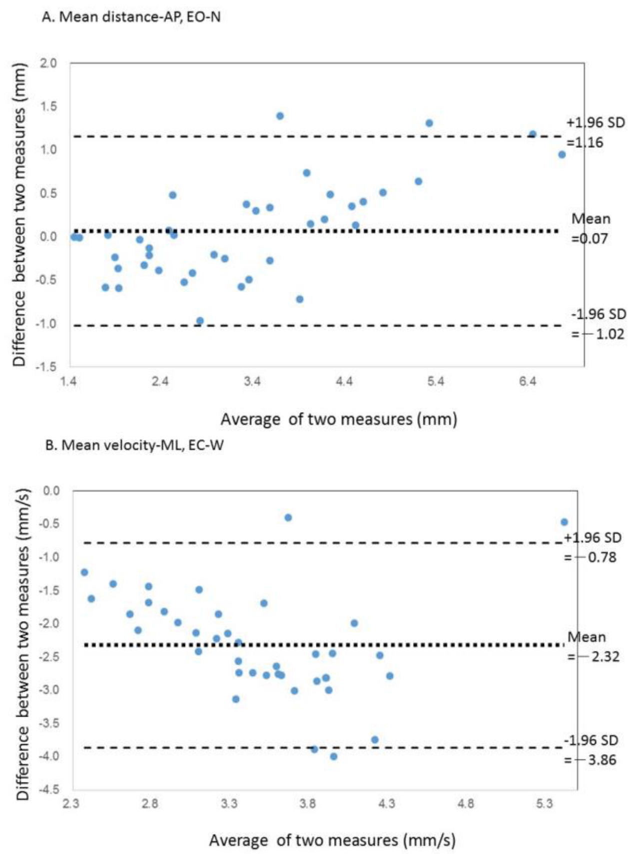

Fig. 4. Bland-Altman plots of the: a) mean anterior-posterior (AP) distance during the eyes-open, narrow stance (mm); b) mean medial-lateral (ML) velocity during the eyes-closed, wide stance (mm/s2), for the Vive-based versus force platform data.

system. The data also discriminated the four conditions well from the combination of stance width (narrow- or wide-stance) and eyes conditions (open or closed). Therefore, using this setup for postural evaluation is feasible when a more precise or expensive tool is not available.

There are several initiatives behind the design of this system for evaluating postural stability. Most available tools, such as marker-based motion capture systems or force platforms, cost more than ten thousand of US dollars. Comparatively, the virtual reality system used in our study costs no more than a thousand US dollars, is a global-commercially available system and can be set up easily. This setup provides a simple and feasible measure of postural sway, especially for effectively screening a large population, and allows measurement in more diverse postures than force platforms, for example, sitting. Moreover, this setup has the potential to be integrated into immersive scenarios created by VR to test the influence of additional environmental stimuli or simultaneous cognitive tasks on postural control. Most of the prior studies have to combine a force platform system or motion capture system for evaluation, which has high technological complexity [19], [20].

One important premise for the use of this set up is the accuracy of the position data. The end-to-end latency of the Vive system is low, 22 ms, and the noise level in the tracker output is relatively low [21]. Prior studies reported various levels of positional and rotational accuracy. An earlier study proposed that the accuracy of the Vive head-mount-display (HMD) was inadequate for scientific experiments due to tracking being lost by the lighthouses and VR space orientation

TABLE V

INTRACLASS CORRELATION COEFFICIENTS (95% CIS) FOR SWAY PARAMETERS OBTAINED FROM THE VIVE TRACKERS ACROSS FOUR CONDITIONS

	Wilks' Lamda	F[&]	p value
mean distance- ML (mm)			
Stance	0.125	132.408	<0.001
Eye	0.936	1.291	0.270
mean distance -AP (mm)			
Stance	0.544	15.940	0.001
Eye	0.921	1.638	0.216
mean velocity -ML (mm/s)			
Stance	0.083	210.462	<0.001
Eye	0.978	0.428	0.521
mean velocity -AP (mm/s)			
Stance	0.241	58.894	<0.001
Eye	0.237	61.019	<0.001
95% conf. ellipse area (mm^2)			
Stance	0.292	46.069	<0.001
Eye	0.810	4.471	0.048

Note: ML=medial-lateral; AP= anterior-posterior; [&]F (1, 19)

tilting when tracking is re-established [10]. Nonetheless, much better precision and accuracy has been documented [9], [11], [12]. For example, HTC Vive trackers agree with Vicon motion tracking with an average translational error of 0.68 ± 0.32 cm and average rotational error of 1.64 ± 0.17° [9]. A comparison between the Vive controller and tracker and a Polhemus Liberty magnetic tracking system sensor on a rigid segment shows a mean translational error below 3 mm [11]. The different levels of positional and rotational accuracy come from hardware, algorithms, device setups, or study designs. For example, tracking loss can be addressed by software and space setup [9]. In our study, tracking loss was not a problem, which may be attributed to the continuous updates provided by Steam VR and the static measurements taken in the center of the calibrated space.

Reliability is an important psychometric property for outcome measures. For postural stability, most of the previous studies showed moderate to excellent reliability of COP parameters, mainly evaluated by the ICC [22], [23]. Our results documented moderate to good same-day reliability (ICCs: 0.52 to 0.90) across four conditions, and the ICCs were mostly above 0.8 for $\mathrm{MVELO_{ML}}$, $\mathrm{MVELO_{AP}}$ and AREA-EC, but not the W-EC condition. These results are compatible with previous studies that found relative high ICCs for mean velocity, sway area and ML displacement [5], [22], [24]. The factors influencing the reliability of the COP-based measures include: task duration, task repetition, repetition timing, age, and task conditions, among others. A longer task duration, higher repetition and same-day repetition increase reliability [24]. Some authors recommended two trials for 120 seconds for the mean velocity and three trials of 120 seconds for other COP-based measurements to achieve an ICC over 0.80 [24]. Considering the possible fatigue effect for same-day retesting, our task duration was set at 30 seconds, which was not an uncommon setting in previous studies [25]. Under this suboptimal setting for reliability, our results still showed acceptable reliability. Further studies will help to explore the effect of longer task duration to achieve an improved signal stationarity and reliability of the sway parameters.

The posturography obtained by the Vive-based system is validated in two ways. First, the system was compared against the COP parameters from the force platform for convergent validity. The data showed a mostly high correlation, especially $\mathrm{MDIST_{AP}}$ ($r > 0.87$). In addition, most of the COD parameters had good agreement with the COP parameters by Bland-Altman plotting and we observed a trend of lower magnitude of the COD compared to COP. Using a tracker at the lumbar area is similar to the concept of a swaymeter or using laser displacement sensors as a proxy for COM [5], [14], and while not synonymous to, they should agree well with it due to the underlying body dynamics [26]. Similarly, a previous study using swaymeters obtained an excellent correlation for AP and ML sway displacement measures (average offset = 6 mm), with longer sway path length measures for the force platform compared to the swaymeter (average offset = 376 mm) by Bland-Altman plots [5]. This finding corresponds to the COP-COM relationship described according to the inverted pendulum model [26], and further validation through a motion caption system should be considered.

We also validated the Vive-based system by examining its discriminative validity through tasks with different levels of postural controls. We choose four stance conditions from the combination of wide or narrow stance and eyes open or closed to explore the effects of support base and visual stimuli on postural stability. Our results showed that the Vive-based postural sway parameters could effectively discriminate these conditions. The GLM repeated measures showed that all the measures were significantly larger for the narrow-base stances than the wide-base stances, similar to previous studies [23]; however, only two out of the five measures ($\mathrm{MVELO_{AP}}$ and AREA-CE) were significantly influenced by the eye conditions. These results may not be related to the validity of the measurements, but rather to the control strategies under these conditions. For example, differences between the eyes-open and eyes-closed conditions are found only in the young adult group for some distance (mainly AP direction) or area measures [3].

Limitations: This study has several limitations that should be addressed. First, the data obtained from the Vive trackers were not validated against marker-based motion capture systems, which should be further evaluated if high precision or accuracy is of interest. Second, the demographics of the current participants, mainly young adults, likely limit generalizability to the elderly or the subjects with impaired balance. For example, elderly people tend to rely on a pelvic control strategy instead of a youth preferred ankle control strategy. Therefore, a different correlation may exist between COP and COD among elderly people, who are likely to be the target population for fall risk evaluation. Further studies are warranted to help to establish the feasibility of this system with various populations.

V. Conclusion

We designed a Vive tracker-based system as a simple and easy-to-use tool for evaluating postural stability. The results showed an acceptable same-day reliability (ICC=0.56∼0.90) and good convergent validity in comparison to the COP data from a force platform system, and discriminated well among different stance conditions. This system has the potential to be a screening tool that has the advantages of low cost, high mobility and simple setup and can be integrated into simulated conditions created by a VR system to evaluate body sway in virtual environments. Further studies in various populations should be conducted to establish the feasibility of wider application.

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