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Use of mobile device accelerometry to enhance evaluation of postural instability in Parkinson's disease

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Abstract

Objective—To determine the accuracy of inertial measurement unit data from a mobile device using the mobile device relative to posturography to quantify postural stability in individuals with Parkinson's disease (PD).

Design—Criterion standard.

Setting—Motor control laboratory at Cleveland Clinic.

Participants—Fourteen mild to moderate individuals with PD and 14 healthy age-matched community dwelling controls completed the project.

Interventions—Not applicable.

Main Outcome Measures—Center of mass (COM) acceleration measures were compared between the mobile device and NeuroCom force platform to determine accuracy of mobile device measurements during performance of the Sensory Organization Test (SOT). Analyses examined test-retest reliability for both systems and sensitivity of: 1) the Equilibrium Score from the SOT and 2) COM acceleration measures from the force platform and mobile device to quantify postural stability across populations.

Results—Metrics of COM acceleration from inertial measurement unit data and NeuroCom force platform were significantly correlated across balance conditions and groups (Pearson's r ranged from 0.35 to 0.97). The SOT Equilibrium Scores failed to discriminate individuals with PD and controls. However, the multi-planar measures of COM acceleration from the mobile device exhibited good to excellent reliability across SOT conditions and were able to discriminate individuals with PD and controls in conditions with the greatest balance demands.

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Conflict of Interest: The authors have authored intellectual property related to the methods described in this manuscript.

Conclusions—Metrics employing medial-lateral movement produce a more sensitive outcome than the Equilibrium Score in identifying postural instability associated with PD. Overall, the output from the mobile device provides an accurate and reliable method of rapidly quantifying balance in individuals with PD. The portable and affordable nature of a mobile device with the application make it ideally suited to utilize biomechanical data to aid in clinical decision-making.

Keywords

Parkinson Disease; Postural Balance; Feedback, Sensory; Accelerometry; Mobile Applications

Postural instability is a disabling consequence of Parkinson's disease (PD) with approximately 70% of individuals falling at least once each year¹. Disease-related declines in balance result in hospitalization and decreased quality of life^{2,3}. Objective evaluations of postural stability in individuals with PD are critical to advance medical management and facilitate the development of interventions that may mitigate the negative effects of PD on balance.

It has been proposed that PD-related postural instability is linked to striatal dopamine loss in the basal ganglia^{4,5}. While the basal ganglia likely receive suitable afferent information from visual, vestibular, and somatosensory inputs to maintain balance, dopamine depletion limits the ability of the central nervous system to integrate and re-weight relevant postural sensory information leading to increased fall rates compared to healthy peers^{6–11}. Historically, force plate posturography measures have been used to characterize changes in balance under varying sensory conditions^{12–16}.

The Sensory Organization Test (SOT) estimates the relative contribution of visual, vestibular, and somatosensory inputs used to maintain postural stability^{17,18}. The SOT has been utilized to evaluate changes in postural stability in individuals with PD as a result of rehabilitation, medication, or disease severity^{11,13,19–22}. However, discrepancies exist across studies comparing SOT performance in individuals with mild to moderate PD and healthy controls, which questions use of the SOT and corresponding Equilibrium Score outcome measure in this patient group^{13,21,22}.

The Equilibrium Score quantifies the peak amplitude of the anterior-posterior (AP) projection of the center of mass (COM) onto the force platform (i.e. angular displacement) within a given sensory condition. The composite Equilibrium Score consists of the weighted average of the Equilibrium Scores across all six conditions¹⁸. Movement in the medial-lateral (ML) direction is not considered in the computation of the Equilibrium Score, though postural stability is achieved by regulating movement in AP and ML directions^{23,24}. The exclusion of movement in the ML direction may explain discrepancies in SOT outcomes across study findings^{25,26}.

The most prevalent model used to characterize postural stability during quiet standing is the inverted pendulum model^{24,27}, in which center of pressure (COP) and COM are utilized to evaluate balance. Briefly, the COP represents the weighted average of all forces applied perpendicular to the ground and COM is the weighted average of each body segment in space. The inverted pendulum model contends that the movement of COP varies with the

movement of COM to preserve the COM over the base of support; thus, the linkage between controlling COP about the COM allows for the maintenance of a steady posture^{24,27}.

The inverted pendulum model is reliant on activation of sensory systems and central nervous system integration to maintain postural stability²⁴. Despite impaired basal ganglia function, individuals with PD often exhibit similar levels of COM displacement compared to healthy controls²⁸. However, the frequency of adjustments necessary to maintain this COM area of displacement requires more frequent corrections of postural sway, as reflected by abnormally higher derivatives of displacement, such as acceleration^{28,29}.

The difference between the COP and COM (COP-COM) displacement measures is proportional to the acceleration of the COM, which is considered an error signal that the postural control system is sensing²⁴. Therefore, acceleration of the COM provides a more thorough quantification of balance than COP and COM displacement individually^{24,30}. Acceleration of the COM has been used to identify significant differences in postural stability between individuals with neurological disorders, older adults, and young adults, which reflect abnormal balance responses induced by aging and disease^{31–33}. In order to capture the movement of COM, which is not measured directly by a force platform, biomechanical models utilizing Newton-Euler equations of motion and proprietary software have been developed that utilize the inverted pendulum model to estimate COM during quiet standing^{34–37}.

To obtain a precise measure of postural stability that incorporates COM acceleration with ease, an application was developed to gather data from the accelerometer within the Apple iPad (Cupertino, CA)^{38–41}. Correlation between inertial sensors within the iPad and 3D motion analysis was established in a cohort of individuals with Parkinson's disease, healthy older adults, and healthy young adults^{38,40,41}. The aims of the current project were: 1) validate use of inertial sensors within the mobile device relative to ground reaction forces from the NeuroCom force platform with individuals with PD and controls, 2) estimate test-retest reliability of the mobile device and NeuroCom systems, and 3) evaluate the capability of mobile device and NeuroCom measures to detect postural instability during the SOT. The development of a validated biomechanical method, packaged within a mobile platform, to characterize postural instability in PD would provide an attractive approach to facilitate understanding and treatment of PD related postural instability.

METHODS

Participants

Fourteen individuals with mild to moderate PD (7 females; mean age 63 ± 8) and fourteen healthy age-matched controls (8 females; mean age 65 ± 9) completed the project; those with PD were on anti-parkinsonian medication (Table 1).

Severity of Parkinson's disease was evaluated by the same trained rater using the Motor Section (III) of the Movement Disorder Society (MDS)-Unified Parkinson's Disease Rating Scale (UPDRS) and the Hoehn and Yahr Scale. Healthy, age-matched (within three years)

controls were recruited. All participants provided consent to procedures approved by the institutional review board.

Data Collection

Participants completed the NeuroCom SOT using the Smart Balance Master (NeuroCom International, Clackamas, OR) while the mobile device was affixed to their waist (level with the sacrum) to approximate whole body COM during data collection. For data synchronization purposes, the mobile device received an output signal from the NeuroCom via audio jacks at the onset of each SOT trial. The SOT consisted of three 20-second trials under six different conditions: SOT1: Fixed surface, eyes open (fixed-fixed); SOT2: Fixed surface, eyes closed (fixed-absent); SOT3: Fixed surface, sway referenced vision (fixed-sway); SOT4: Sway referenced surface, eyes open (sway-fixed); SOT5: Sway referenced surface, eyes closed (sway-absent); and SOT6: Sway referenced surface, sway referenced vision (sway-sway). The conditions used in the SOT included sway referencing, which entails AP tilt of the support platform and/or of the visual surround in direct response to the individual's COP. All participants wore a harness for safety while completing the SOT.

Data Analysis

COM Acceleration—The COM acceleration was derived from the NeuroCom force platform utilizing the inverted pendulum model of postural stability and compared with mobile device COM acceleration. The NeuroCom system quantifies COP in the frontal plane (ML direction) and in the sagittal plane (AP direction). A single-link inverted pendulum model relates COP with COM, which can be shown using the Newton-Euler equations:

$$COP_x - COM_x = (I_f / m * g * h) * \ddot{COM}_x \quad (1)$$

where COM_x is the COM position with respect to the ankle joint in the ML direction, COP_x is the COP position with respect to the ankle joint in the ML direction, I_f is the total body moment of inertia about the ankle joint in the frontal plane, m is the subject's body mass, g is acceleration due to gravity, h is the height of the COM above the ankle joint, and $\ddot{COM}_x$ is the COM linear acceleration in the ML direction and

$$COP_y - COM_y = (I_{sa} / m * g * h) * \ddot{COM}_y \quad (2)$$

where y refers to displacements in the AP direction and I_{sa} is the total body moment of inertia about the ankle joint in the sagittal plane.

Using equations 1 and 2, a mathematical model was developed to compute COM position using the force plate COP positional data^{34,42}. Once the COM position was computed, ML and AP COM accelerations were calculated by determining the second order derivative of the positional data with respect to time across populations for all balance conditions and trials.

The COM acceleration was calculated from the mobile device and NeuroCom systems recorded at a 100-Hz sampling frequency and after applying a 3.5-Hz cut-off, 4th order, low-pass Butterworth filter. Four metrics that have previously been shown to be significantly correlated with 3-D motion analysis to characterize ML and AP COM acceleration were computed for each balance trial^{38,40,41}: 1) peak-to-peak (P2P), a measure of amplitude of displacement; 2) normalized path length (NPL) which normalizes the data using sampling rate and total time of each balance trial; 3) the natural log of the root mean square distance (logRMS) which quantifies the magnitude of COM displacements; and 4) ellipsoid area (95% Ellipse Area) that, with 95% of probability, contained the center of the points of sway in the ML and AP planes. The logarithmic transformation was computed for the RMS in order to ensure normality required for statistical analyses.

NeuroCom Equilibrium Scores—NeuroCom Equilibrium Scores, ranging from 0–100% (i.e. a score of 100% represents no postural sway) were quantified by the NeuroCom software for each trial. The software computes Equilibrium Score by comparing angular difference between the subject's calculated maximum AP COM displacement and the subject's theoretical maximum sway of 12.5°¹⁸. Equilibrium Scores were averaged within each condition (three trials per condition)¹⁸.

Statistical Analysis

All statistical analyses were performed with SPSS version 21.0, Chicago, IL. Differences were assumed significant when $P < 0.05$.

NeuroCom Equilibrium Scores—Differences between populations for Equilibrium Score data during each SOT condition were determined using two sample t-tests. Welch correction was applied to account for possible unequal variances.

COM Acceleration—Each metric that characterized COM acceleration (P2P, NPL, logRMS, and 95% Ellipse Area) quantified by the NeuroCom and mobile device were evaluated for goodness of fit via Bland-Altman plots. All subjects (individuals with PD and controls) were combined for the correlation analysis and each trial was treated as a separate occurrence. In this approach, the differences between each measure quantified by the NeuroCom and mobile device were subtracted from each other to quantify the measurement error (i.e. $P2P_{iPad} - P2P_{NeuroCom}$, $NPL_{iPad} - NPL_{NeuroCom}$, etc.) and were plotted against the average of each measure on a trial-by-trial basis from the two systems, which represented the best approximation of the true value for each outcome measure. The mean of the difference and the respective limits of agreement (defined as $\text{mean} \pm 1.96 \text{ SD}$) are reported for each outcome measure across SOT conditions, where a smaller gap between the limits of agreement indicates less variability and more consistency between the two systems.

Test-retest reliability was estimated for P2P, NPL, logRMS, and 95% Ellipse Area values between the three trials of each SOT condition for the NeuroCom and mobile device. Intraclass correlation coefficient (ICC) and 95% confidence intervals were used to determine the ratio of between-subject to the total variance in the NeuroCom and mobile device measures for the six SOT conditions. Each ICC was computed using two-way mixed effects

analysis of variance with a model comprised of all subjects (individuals with PD and controls) across each trial number. The test–retest reliability was categorized into four groups: poor (0–0.39), fair (0.40–0.59), good (0.60–0.74), and excellent (0.75–1.0).

Differences between populations for COM acceleration measures during each SOT condition were determined using two sample t-tests. Welch correction was applied to account for possible unequal variances.

RESULTS

NeuroCom Equilibrium Scores

No significant differences in Equilibrium Scores were found between populations during the various SOT conditions (Table 2; Fig. 1A).

COM Acceleration

The COP and COM trajectories in the AP direction recorded from the NeuroCom system for six 20-second trials of quiet standing of one representative subject are presented (Fig. 2). The figure demonstrates how closely COP and COM are in phase and how COP oscillates around COM. The COM acceleration was computed from force plate data and directly compared to COM acceleration recorded by the mobile device across balance conditions. Through visual inspection and the correlation coefficients provided, Figure 3 indicates that the COM acceleration converted from force platform COP measures correlated with the mobile device ML and AP COM acceleration measurements.

Bland-Altman plots revealed that the mean difference (bias) across all outcome metrics for each condition was close to 0 (units are in cm/s^2 and cm^2/s^4), with SOT1 (fixed-fixed) showing the smallest and SOT5 (sway-absent) and SOT6 (sway-sway) showing the largest mean difference (Table 3; Fig. 4). Inspection of the limits of agreement in the Bland-Altman plots revealed that 95% of the measurement error between the two systems was from -0.3 to 7.3 cm/s^2 for AP P2P, 3.5 to 9.4 cm/s^2 for AP NPL, 0.1 to 1.0 cm/s^2 for AP logRMS and -3.5 to $37.5 \text{ cm}^2/\text{s}^4$ for 95% Ellipse Area in the condition of best agreement (SOT1) and from -5.4 to 21.9 cm/s^2 for AP P2P, -0.3 to 20.8 cm/s^2 for AP NPL, -0.4 to 0.8 cm/s^2 for AP logRMS, and -25.1 to $116.2 \text{ cm}^2/\text{s}^4$ for 95% Ellipse Area in the condition with the poorest agreement (SOT5 or SOT6).

Reliability of the mobile device on a test-retest basis exceeded the reliability of the NeuroCom for all four metrics characterizing COM acceleration (Table 4). Test–retest reliability estimates for P2P of mobile device measurements taken across the three reiterations ranged from good to excellent (ML: 0.64 – 0.74 ; AP: 0.61 – 0.85) for SOT1–SOT4 and excellent (ML: 0.78 – 0.85 ; AP: 0.85 – 0.86) for SOT5–SOT6. The NPL, logRMS, and 95% Ellipse Area measures exhibited slightly higher ICC values, with mobile device ICC values ranging from good to excellent (0.74 – 0.90) for SOT1–SOT4 and excellent (0.88 – 0.92) for SOT5–SOT6.

Mobile device and NeuroCom systems were equally capable of differentiating postural stability across populations. Postural instability for the PD group quantified by P2P, NPL,

logRMS, and 95% Ellipse Area in ML and AP directions was significantly greater compared to controls during SOT5 (sway-absent) and SOT6 (sway-sway) with both the mobile device (SOT5 and SOT6: $P<0.02$) and NeuroCom systems (SOT5 and SOT6: $P<0.05$). Figure 1B depicts differences in 95% Ellipse Area across populations during the six SOT conditions.

DISCUSSION

The aim of this study was to determine concurrent validity of measures of accelerometry using the embedded inertial sensors within a mobile device with the NeuroCom force platform to accurately measure postural stability in individuals with PD compared to age-matched controls. The primary results indicate: 1) measures of COM acceleration (i.e. P2P, NPL, logRMS, and 95% Ellipse Area) computed from the mobile device and NeuroCom force platform were significantly correlated across sway directions, balance conditions, and populations, 2) the test-retest reliability of the mobile device measures was equal or superior to the NeuroCom across SOT conditions, and 3) COM acceleration was superior to the Equilibrium Score in detecting differences in postural stability between individuals with PD and their healthy, age-matched counterparts (Fig. 1).

The Equilibrium Scores failed to detect differences in postural instability between individuals with PD and healthy, age-matched controls, in any of the six balance conditions of the SOT. However, COM acceleration was sensitive in detecting differences between populations under the most difficult conditions, SOT5 (sway-absent) and SOT6 (sway-sway). Specifically, in SOT5, somatosensory information was inaccurate and vision was unavailable; whereas in SOT6, both somatosensory and visual information were inaccurate. Although a number of studies have detected postural instability in individuals with PD using Equilibrium Scores during SOT5 and/or SOT6^{11,13,22,43,44}, in our cohort of individuals with PD, measures incorporating COM acceleration were the only ones sensitive in detecting postural stability deficits. The majority of individuals in our study had mild PD, with median disease duration of 4.5 years and an average Hoehn and Yahr Stage of 2.4. Individuals who present at Hoehn and Yahr Stage 2 are characterized as having bilateral symptoms with no balance impairment, as defined by the pull test⁴⁵. Therefore, it is not surprising that the Equilibrium Score, which uses only the two most extreme AP positional data points of postural stability, did not discriminate across populations.

The inverted pendulum model incorporating COP and COM displacements, which are proportional to COM acceleration, provides increased sensitivity into the assessment of postural stability than COP and COM used in isolation^{46,47}. Since COM acceleration is not directly measureable from the force platform, a critical finding in the current study was demonstrating validity of the mobile device metrics to quantify postural stability during the SOT. The Bland-Altman analysis assessed the ability of the mobile device to accurately quantify the trajectories of COM acceleration within a trial as quantified by P2P, NPL, logRMS, and 95% Ellipse Area. The SOT1 had the smallest spread of differences between the 95% confidence intervals across outcome measures, whereas SOT5 and SOT6 had the largest spread between the 95% confidence intervals (Table 3). Overall, the mobile device consistently provided a measure similar to the NeuroCom system across a range of balance task difficulties.

Additionally, test-retest reliability of mobile device measures was shown to be equal or superior to the NeuroCom system across SOT conditions, indicating reproducibility and precision of mobile device outcome measures for each trial within the SOT conditions. The test-retest results are consistent with previous findings from Whitney and colleagues⁴⁸, showing good to excellent test-retest reliability of accelerometry measures comparable to those using the NeuroCom system. However, while Whitney and colleagues compared COM measures using a custom accelerometer system to COP measures using the NeuroCom force platform, the current study incorporated the inverted pendulum model, allowing for the direct comparison of COM measures across systems, thereby building upon the work of Whitney and colleagues. All outcome measures in the current study exhibited ICC values ranging from good to excellent across all subjects and SOT conditions, and specifically, the mobile device exhibited excellent reliability 79% of the time compared to the NeuroCom, which demonstrated excellent reliability 60% of the time. Our findings support use of a biomechanical balance assessment that provides a valid and reliable measure of postural instability to monitor and manage disease progression. Results from the study justify use of ML and AP COM acceleration in individuals with PD and in a healthy population to better characterize postural stability under demanding sensory conditions.

When analyzing measures of COM acceleration, individuals with PD maintained postural stability similar to controls when somatosensory and vestibular information were available with or without access to vision (SOT1-SOT3). Further, when somatosensory information was incongruent, but vision and vestibular information were available, individuals with PD maintained postural stability (SOT4) comparable to healthy controls. However, when vision was unavailable or incongruent and somatosensory information was incongruent (SOT5 and SOT6), individuals with PD had decreased postural stability compared to controls. These results do not necessarily indicate that their vestibular system is impaired⁴⁹, but rather, suggest that individuals with PD may be over-reliant on visual feedback to maintain postural stability. This visual dependence is consistent with neuroanatomical evidence of somatosensory impairment and points to largely intact visual inputs in PD. Projections from the dorsal visual stream to the basal ganglia are minor compared to the visual afferents to the cerebellum and posterior parietal circuits⁵⁰; thus, visuomotor integration may be largely spared by the disease, and could explain the increased reliance on vision by PD^{6,51}. Increased reliance on vision may be an adaptive strategy to compensate for impaired somatosensory feedback⁵¹; thus, postural instability can often be improved by utilizing external visual cues⁵²⁻⁵⁵. Additionally, individuals with PD may have difficulty re-weighting sensory information when one or more sensory modalities become unreliable or unavailable⁵⁶. When examining postural responses of individuals with PD after visual stimuli generated by discrete lateral displacements of a moveable room in which subjects stand in the upright position, investigators reported normal sway with eyes open or closed but abnormally large motor responses to room movement, which did not attenuate with stimulus repetition¹⁰. Therefore, it is plausible that individuals with PD may have experienced difficulty re-weighting sensory input from visual and somatosensory components, thus becoming unstable.

The established mobile device metrics quantifying COM acceleration provide insight into the error signal that the postural control system is experiencing during balance conditions

involving deprivation or inaccuracy of visual, vestibular, and somatosensory information. The error signal may vary based on each individual's capability of prioritizing sensory inputs to maintain equilibrium, which may aid clinicians in determining the severity of deficits in central processing capacity. Balance deficits in individuals with PD during SOT5 and SOT6 provide insight into the integration of sensory systems associated with postural stability, and were only detected using metrics quantifying the mobile device's trajectory but were lost using less sensitive measures of maximum AP displacement quantified by the NeuroCom system. Use of the proposed mobile device and application installed on a handheld mobile device may provide an accurate, easy to use, and cost-effective platform to generate quantitative scores by which clinicians and researchers may identify individuals with postural instability over time and across providers.

LIMITATIONS

Modeling postural stability as an inverted pendulum model assumes a rigid structure above the ankle joints; however, the human body is a multi-segmented structure capable of moving at all joints. Yet, because individuals with PD display significantly larger hip feedback gains with significantly smaller hip joint torques⁵⁷, we felt confident that overall postural stability resembled the inverted pendulum in response to perturbations. In applying the inverted pendulum model in healthy older adults, movement at joints other than the ankles, such as the hips, has been shown to be present during unperturbed^{58,59} and perturbed^{60,61} balance conditions; however, Gage and colleagues provided a detailed kinematic analysis of postural stability using a laboratory 3D motion capture system supporting the inverted pendulum model due to each body segment's COM synchronized to the movement of whole body COM⁶².

CONCLUSIONS

Utilizing acceleration measures from ML and AP planes increased sensitivity to discriminate between healthy and PD populations and provides a method capable of reliably detecting changes in postural stability across a range of conditions. Enhanced sensitivity of measurement provides utility in monitoring disease progression, response to treatment, potential for falling and methods of facilitating clinical hand-offs between providers.

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Abbreviations

AP	anterior-posterior
COP	center of pressure
COM	center of mass

ML	medial-lateral
PD	Parkinson's disease
SOT	Sensory Organization Test

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Figure 1.

(A) Equilibrium Scores and (B) 95% Ellipse Areas quantifying differences in postural stability between populations during each SOT condition. * Asterisk denotes p-values < 0.05; ** Asterisk denotes p-values < 0.005.

Figure 2.

Trajectories of AP COP and COM position under the six SOT conditions. Using the NeuroCom, the COP was recorded and COM position was computed using the inverted pendulum model.

Figure 3.

Typical 20-second balance trials showing how closely the mobile device's accelerometer can track the NeuroCom's force platform COM acceleration measures.

Figure 4.

Bland-Altman plots for each outcome measure (quantifying AP sway) across SOT conditions, where the difference in each outcome measure ($P2P_{iPad} - P2P_{NeuroCom}$, $NPL_{iPad} - NPL_{NeuroCom}$, etc.) between systems is plotted against the average of two values on a trial-by-trial basis. The solid lines represent the mean difference in values; dashed lines represent the upper and lower limits of agreement (mean difference $\pm$ 1.96 SD). Groups were combined for the analysis (Red plus symbols indicate PD subjects; Blue circles indicate Controls).

Table 1

Summary of characteristics for individuals with PD

Subject	Gender	Age (years)	BMI (kg/m ²)	Duration of disease (years)	MDS-UPDRS-III	Hoehn and Yahr
1	M	57	21.6	7	48	2
2	F	70	32.3	5	47	3
3	M	59	33.5	3	41	2
4	F	66	32.0	2	27	2
5	M	53	37.0	4	43	2
6	M	80	24.9	8	48	3
7	F	69	25.0	4	34	2
8	F	55	31.0	3	48	3
9	M	60	27.3	12	60	3
10	F	58	38.6	6	25	2
11	M	71	29.7	1	28	2
12	F	55	21.1	7	30	2
13	M	72	28.0	11	40	3
14	F	56	27.9	5	29	2
Mean		62.9	29.3	5.6	39.1	2.4
SD		8.3	5.2	3.2	10.5	0.5

Table 2

Mean $\pm$ SD of Equilibrium Scores with corresponding P-values for the SOT conditions using the NeuroCom measures

Condition	Individuals with PD	Control Subjects	P-values
SOT1: Fixed-fixed	94.2 $\pm$ 1.7	95.5 $\pm$ 1.7	0.315
SOT2: Fixed-absent	90.3 $\pm$ 4.7	91.4 $\pm$ 4.8	0.329
SOT3: Fixed-sway	88.4 $\pm$ 6.2	90.7 $\pm$ 3.9	0.157
SOT4: Sway-fixed	85.3 $\pm$ 7.8	87.6 $\pm$ 6.3	0.329
SOT5: Sway-absent	68.7 $\pm$ 12.3	71.2 $\pm$ 10.4	0.345
SOT6: Sway-sway	71.9 $\pm$ 10.8	73.4 $\pm$ 13.1	0.451

Table 3

Mean Bias (Lower-Upper Bound limits) from Bland-Altman Analysis across outcome measures, sway directions, and SOT conditions

SOT Condition	Peak-to-peak (cm/s ²)		Normalized path length (cm/s ²)		Log of Root mean square (cm/s ²)		95% Ellipse Area (cm ² /s ⁴)	
	ML	AP	ML	AP	ML	AP	ML/AP	
1	3.5 (0.9–6.2)	3.5 (–0.3–7.3)	6.2 (3.3–9.1)	6.5 (3.5–9.4)	1.0 (0.4–1.5)	0.5 (0.1–1.0)	17.0 (–3.5–37.5)	
2	3.6 (1.2–6.0)	3.3 (–1.6–8.2)	6.7 (3.3–10.1)	6.6 (1.2–12.0)	0.8 (0.3–1.3)	0.3 (–0.1–0.7)	23.7 (–8.8–56.3)	
3	4.6 (0.6–8.6)	4.3 (–1.3–9.9)	7.9 (3.0–12.7)	8.1 (0.1–16.0)	0.9 (0.3–1.4)	0.4 (–0.2–0.9)	31.8 (–8.6–72.2)	
4	4.8 (0.8–8.7)	4.6 (–4.7–14.0)	8.1 (3.9–12.3)	8.4 (2.5–14.3)	0.9 (0.3–1.4)	0.3 (–0.1–0.7)	36.6 (–6.0–79.1)	
5	5.8 (–0.7–12.3)	8.2 (–5.4–21.9)	10.2 (1.4–18.9)	10.2 (–0.3–20.8)	0.5 (–0.3–1.4)	0.1 (–0.5–0.7)	45.6 (–25.1–116.2)	
6	6.5 (–0.4–13.4)	6.1 (–4.4–16.6)	10.4 (3.4–17.5)	9.4 (–1.7–20.6)	0.7 (–0.1–1.5)	0.2 (–0.4–0.8)	48.3 (1.6–94.9)	

Abbreviations: ML, Medial-lateral; AP, Anterior-posterior.

Table 4

Test-retest reliability of three trials for the six SOT conditions: Intraclass Correlation Coefficients (95% confidence intervals) were calculated across outcome measures for both mobile device and Neurocom systems

SOT Condition	Peak-to-peak (cm/s ²)			Normalized path length (cm/s ²)		
	ML		AP		ML	
	Mobile device	NeuroCom	Mobile device	NeuroCom	Mobile device	NeuroCom
1	0.64 (0.32–0.82)	0.61 (0.26–0.80)	0.62 (0.28–0.81)	0.67 (0.38–0.84)	0.83 (0.68–0.91)	0.53 (0.19–0.88)
2	0.69 (0.41–0.84)	0.60 (0.25–0.80)	0.61 (0.27–0.81)	0.56 (0.21–0.86)	0.82 (0.67–0.91)	0.78 (0.58–0.89)
3	0.70 (0.44–0.85)	0.60 (0.25–0.80)	0.67 (0.38–0.84)	0.62 (0.29–0.81)	0.88 (0.78–0.94)	0.79 (0.60–0.90)
4	0.74 (0.51–0.87)	0.66 (0.35–0.83)	0.85 (0.72–0.93)	0.86 (0.74–0.93)	0.88 (0.77–0.94)	0.83 (0.68–0.91)
5	0.78 (0.59–0.89)	0.73 (0.50–0.87)	0.86 (0.73–0.93)	0.78 (0.58–0.89)	0.88 (0.78–0.94)	0.77 (0.56–0.88)
6	0.85 (0.72–0.93)	0.81 (0.65–0.91)	0.85 (0.72–0.93)	0.78 (0.60–0.89)	0.92 (0.86–0.96)	0.88 (0.77–0.94)
					0.91 (0.83–0.96)	0.88 (0.78–0.94)

SOT Condition	Log of Root mean square (cm/s ²)			95% Ellipse Area (cm ² /s ⁴)		
	ML		AP		ML/AP	
	Mobile device	NeuroCom	Mobile device	NeuroCom	Mobile device	NeuroCom
1	0.76 (0.55–0.88)	0.62 (0.29–0.81)	0.74 (0.52–0.87)	0.69 (0.42–0.85)	0.76 (0.55–0.88)	0.61 (0.26–0.80)
2	0.78 (0.60–0.89)	0.54 (0.20–0.90)	0.74 (0.51–0.87)	0.71 (0.45–0.85)	0.82 (0.66–0.91)	0.79 (0.61–0.90)
3	0.76 (0.56–0.88)	0.77 (0.57–0.89)	0.75 (0.54–0.88)	0.70 (0.44–0.85)	0.88 (0.78–0.94)	0.84 (0.70–0.92)
4	0.81 (0.64–0.91)	0.72 (0.48–0.86)	0.82 (0.67–0.91)	0.82 (0.66–0.91)	0.86 (0.74–0.93)	0.78 (0.59–0.89)
5	0.90 (0.81–0.95)	0.86 (0.73–0.93)	0.90 (0.82–0.95)	0.87 (0.75–0.93)	0.90 (0.81–0.95)	0.89 (0.80–0.95)
6	0.88 (0.78–0.94)	0.76 (0.55–0.88)	0.92 (0.85–0.96)	0.89 (0.79–0.94)	0.91 (0.84–0.96)	0.91 (0.83–0.95)

Abbreviations: ML, Medial-lateral; AP, Anterior-posterior.