Published in final edited form as:

Gait Posture. 2012 April ; 35(4): 573–578. doi:10.1016/j.gaitpost.2011.11.026.

Body-worn motion sensors detect balance and gait deficits in people with multiple sclerosis who have normal walking speed

R.I. Spain^{a,*}, St. R.J. George^b, A. Salarian^c, M. Mancini^b, J.M. Wagner^d, F.B. Horak^b, and D. Bourdette^a

R.I. Spain: spainr@ohsu.edu

^aNeurology Service and MS Center of Excellence-West, Portland VA Medical Center and Department of Neurology, Oregon Health & Science University, Portland, OR, USA

^bDepartment of Neurology, Oregon Health & Science University, Portland, OR, USA

^cBalance Disorders Lab, Oregon Health & Science University, Beaverton, OR, USA

^dDepartment of Physical Therapy and Athletic Training, Saint Louis University, St. Louis, MO, USA

Abstract

While balance and gait limitations are hallmarks of multiple sclerosis (MS), standard stopwatch-timed measures practical for use in the clinic are insensitive in minimally affected patients. This prevents early detection and intervention for mobility problems. The study sought to determine if body-worn sensors could detect differences in balance and gait between people with MS with normal walking speeds and healthy controls. Thirty-one MS and twenty-eight age- and sex-matched control subjects were tested using body-worn sensors both during quiet stance and gait (Timed Up and Go test, TUG). Results were compared to stopwatch-timed measures. Stopwatch durations of the TUG and Timed 25 Foot Walk tests were not significantly different between groups. However, during quiet stance with eyes closed, people with MS had significantly greater sway acceleration amplitude than controls ($p = 0.02$). During gait, people with MS had greater trunk angular range of motion in roll (medio-lateral flexion, $p = 0.017$) and yaw (axial rotation, $p = 0.026$) planes. Turning duration through 180° was also longer in MS ($p = 0.031$). Thus, body-worn motion sensors detected mobility differences between MS and healthy controls when traditional timed tests could not. This portable technology provides objective and quantitative mobility data previously not obtainable in the clinic, and may prove a useful outcome measure for early mobility changes in MS.

Keywords

Gait; Postural balance; Multiple sclerosis; Outcome measurement; Rehabilitation

1. Introduction

Mobility impairment is a hallmark of multiple sclerosis (MS), affecting nearly half of patients at presentation [1] and contributing to a lower quality of life [2]. Causes of balance and gait dysfunction in MS are complex and incompletely understood [3,4], and assessment is limited to specialized motion-analysis laboratories with trained personnel [5]. In the MS

*Corresponding author at: Portland VA Medical Center, Oregon Health & Science University, Mail Code: CR120 3181, S.W. Sam Jackson Park Road, Portland, Oregon 97239-3098, USA. Tel.: +1 503 346 0768; fax: +1 503 494 7289.

Conflicts of interest: There are no conflicts of interest.

clinic, balance and gait are poorly captured by subjective rating scales and traditional stopwatch-timed measures like the Timed 25 Foot Walk (T25FW). While easy to administer, the T25FW has high variability requiring at least 20% change to be considered statistically [6] or clinically [7,8] significant. High variability also creates difficulties in distinguishing deficits in minimally impaired MS [6,9]. Thus, while patients report worsening mobility, average treating neurologists without access to specialized laboratories are at a loss to objectively capture balance and gait deterioration or determine the benefit of therapeutic drug and/or rehabilitation strategies. Given that early therapy is most effective at slowing disease progression [10], treating clinicians require discriminative measures of mobility decline to justify the risks of switching disease-modifying therapies.

In contrast to stopwatch-timed measures, quantitative balance and gait measures obtained in the motion-analysis laboratory can detect early MS mobility changes [11,12]. The development of body-worn mobility-assessment technology along with automated data processing software has allowed for clinic-based data acquisition and interpretation [13,14]. The body-worn sensors used in this study identified early changes in Parkinson's disease undetected by stopwatch-timed measures [15].

We hypothesized that body-worn motion sensors would discriminate people with MS with normal walking speeds from healthy controls by capturing balance and gait abnormalities during quiet stance, gait, and postural transition tasks while stopwatch-timed tests would not. We expected significant correlations between abnormal mobility parameters and self-reported measures of balance and gait dysfunction to provide concurrent validity of these novel mobility outcome measures.

2. Patients

The Oregon Health & Science University Institutional Review Board approved the study in accordance with the Declaration of Helsinki. All subjects gave written informed consent prior to assessments.

Sample size for this pilot trial was based on prior studies in PD using these sensors [15]. Thirty-one subjects (18–70 years) with any type of MS and normal T25FW (within 2 standard deviations of the control sample, <5 s) were recruited from an MS clinic along with 28 healthy age- and sex-matched control subjects recruited from family members and MS clinic staff. MS type was determined by chart review. Subjects were excluded if they had non-MS-related causes of gait or balance problems (joint replacement, arthritis, pregnancy, etc.) or MS exacerbation in the prior 60 days.

3. Methods

3.1. Experimental protocol

Timed 25 Foot Walk [16]: subjects were asked to walk 25 feet in a hallway “as quickly and safely as possible” according to the Multiple Sclerosis Functional Composite instructions. The average of two trials recorded by stopwatch was used for data analysis.

Timed-Up-and-Go test (TUG) [17]: subjects were instructed to stand up from a chair, walk 7 m, turn, then walk back and sit down, all “as quickly and safely as possible” (Fig. 1B). Distance was increased (modified) from 3 m to 7 m to provide sufficient gait cycles for analysis [18]. Time to complete the TUG was recorded by stopwatch (modified, mTUG) and the motion sensors (instrumented TUG, iTUG) which also recorded body motion data.

Quiet standing task: wearing the lumbar motion sensor, participants stood with arms crossed and feet placed by a template block (Fig. 1A). Three 30 s trials were performed with eyes open (EO) and three with eyes closed (EC).

Self-reported balance and walking measures: subjects completed the Activities-specific Balance Confidence scale (ABC), Multiple Sclerosis Walking Scale-12 (MSWS12), and self-reported Expanded Disability Status Scale (EDSS). The ABC, a scale designed to evaluate balance in the elderly, has been shown to predict falls in the MS population [19]. The MSWS12 is a valid and responsive questionnaire reflecting the impact of MS on walking [20]. The EDSS is a standard rating scale of neurological function in MS [21]. The self-reported EDSS has been validated against the clinician-administered EDSS [22].

3.2. Equipment

The portable motion analysis system consisted of six small, body-worn sensors (Xsens, Enschede, The Netherlands www.xsens.com) each housing a 3-dimensional gyroscope and tri-axial accelerometer sampling at 50 Hz. Gyroscopes measured rotational trunk velocity in roll (medio-lateral, ML), pitch (anterior-posterior, AP) and yaw (axial rotation) planes with a $\pm 300^\circ/\text{s}$ range. Tri-axial accelerometers measured linear acceleration in vertical, lateral and sagittal directions with a $\pm 1.7\text{ g}$ range and resolution of $\pm 26\text{ }\mu\text{g}$. Sensors were attached 4 cm above each malleolus, on the dorsum of the wrists, the upper trunk 2 cm below the sternal notch, and on the lumbar trunk at L5, (approximate body center of mass, Fig. 1A). The sensing axes were oriented along anatomical AP, ML, and vertical directions. Wires from the sensors connected to a portable data-receiver on a belt that wirelessly streamed data to a laptop.

3.3. Signal processing

A MATLAB program (Matlab R2009b, The Mathworks Inc, Natick, MA) automatically calculated components of the postural sway during quiet stance using the tri-axial acceleration signals from the lumbar L5 sensor. The sensing axes were oriented along the anatomical antero-posterior (AP), medio-lateral (ML), and vertical directions. The sensor was connected via a cable to a data transmitter located on a belt around the waist. Data acceleration signals from the lumbar AP and ML directions were sampled at 50 Hz, transformed to a horizontal-vertical coordinate system [23] and filtered with a 3.5 Hz cut-off, zero phase, low-pass Butterworth filter. Four previously used measures [15] were computed from planar acceleration data to characterize postural sway: (1) sway acceleration amplitude as root mean square around the mean (RMS), (2) mean sway velocity (MV) from integration of the acceleration signal, (3) sway frequency as the centroidal frequency (CF) of sway reflecting median power of the acceleration signal, and (4) sway jerk as the derivative of the acceleration signal using established formulas [15,24]. Jerk, a measure of the rate of change in decelerations and accelerations, reflects sway smoothness and the amount of regulatory postural corrections [25]. Sway jerk was normalized (nJerk) to the peak-to-peak range of the acceleration excursion in the trial and to trial duration so that the parameter was less dependent on the amount of sway and more revealing of the smoothness; this normalization process made the parameter unit-less [26]. Each parameter was calculated in the ML and AP directions as well as the combined 2-D horizontal plane. Unless ML and AP directions showed different effects, the combined 2-D horizontal plane values were presented.

The iTUG was automatically detected and separated into its component phases: gait (lower body, upper body) and postural transitions (sit-to-stand, stand-to-sit transitions, and turning). The algorithms have been described previously [18]. Individual gait cycles were analyzed. Both temporal (cadence, double-support time) and spatial (stride length) gait parameters are

reported. Variability in leg and arm signals was recorded by the coefficient of variability. Signals from the sternal sensor were used to calculate postural transition parameters (sit-to-stand and stand-to-sit pitch velocities and durations) as well as turning parameters. Both significant differences between groups and commonly reported mobility parameters are presented.

3.4. Statistical analysis

Normality of the data was verified with the Shapiro-Wilk test before parametric analyses were performed. A two-sample *t*-test investigated differences of balance and gait parameters between the MS and control groups for each dependent variable. Corrections for multiple comparisons were not performed as the aim of this exploratory study was to investigate subject groups already known to be different (MS and controls), and doing so would have exaggerated Type II errors. For significant quiet stance parameters, a repeated measures analysis of variance (ANOVA) was performed to determine the relative changes between groups in the EO and EC conditions. The area under the Receiver Operating Characteristics (ROC) Curve (AUC) evaluated the discriminatory value of each parameter – a larger area indicating greater sensitivity and specificity. Nonparametric (Spearman) correlations investigated associations between the instrumented parameters with high discriminatory values, and associations between discriminatory parameters and the EDSS, MSWS12, ABC, and stopwatch-timed measures. The critical α level was 0.05. Statistical analyses were performed using PASW Statistics (version 18).

4. Results

4.1. Subjects

Demographics are presented in Supplemental Table 1. The groups did not differ significantly in age, sex or weight. Median EDSS was 3.0 (0–5.0). The wide range of symptom duration (0.2–33 years, median 6.8) suggests a variety of MS histories. All subjects were relapsing-remitting except three that had high-risk clinically isolated syndromes (one brainstem, one spinal cord, and one pyramidal tract symptoms with cerebral white matter changes).

4.2. Gait speed

The MS and control subjects groups were indistinguishable when tested for gait speed using the T25FW (4.15 ± 0.08 s vs 3.93 ± 0.12 s, $p = 0.073$, Fig. 2A), the stopwatch-timed mTUG (12.32 ± 0.36 s vs 11.60 ± 0.45 s, $p = 0.18$, Fig. 2B) and sensor-timed iTUG (13.76 ± 0.37 s vs 12.97 ± 0.45 s, $p = 0.18$, Fig. 2C). While a repeated measures ANOVA showed that the stopwatch recorded a small but significantly faster TUG than the sensors (mean 0.98 s, $F_{(1,49)} = 272$, $P < 0.001$), the more precise recording device did not render the TUG better able to separate MS from controls ($p = 0.18$ mTUG, $p = 0.18$ iTUG, Fig. 2B and C).

4.3. Quiet standing task parameters

Acceleration measures of sway were able to differentiate MS from control subjects during quiet stance, EC condition. In particular, sway acceleration amplitude was greater in people with MS compared to controls (0.085 vs 0.058 m/s², $p = 0.02$, Cohen's $d = 0.66$, Table 1, Fig. 3A and C). In addition, ML nJerk was lower in MS compared to controls (2.41 vs 2.82 , $p = 0.05$, Cohen's $d = -0.60$ Table 1, Fig. 3B and C). When changing from EO to EC condition, the sway acceleration amplitude increased more in MS than controls, as shown by a significant vision (EO/EC)-by-group interaction effect ($p = 0.024$). Both AUC for sway acceleration amplitude EC (0.69) and ML nJerk EC (0.65) indicated a moderate degree of discriminatory ability of the parameters (Table 1, Fig. 3C).

4.4. Walking task parameters

During the gait phase, there were no significant differences between MS and controls in temporal measures (e.g., cadence, swing, double support time) (Table 1). However, significant differences arose in aspects of gait affected by dynamic balance. The angular trunk ROM was significantly larger in MS than controls in both roll (7.5° vs 5.9° , $p = 0.017$, Cohen's $d = 0.64$) and yaw (11.1° vs 9.0° , $p = 0.026$, Cohen's $d = 0.68$) axes (Table 2). When these significant trunk parameters of ML flexion and rotation were combined via sum of squares, the resulting “total lateral trunk ROM” was the best discriminator between MS and controls (13.7° vs 10.8° , $p < 0.01$, Cohen's $d = 0.87$, AUC = 0.72, Table 2, Fig. 3D and F). The postural transition parameter turning duration was significantly longer in MS than controls (1.77 s vs 1.58 s, $p = 0.031$, Cohen's $d = 0.58$ Table 2, Fig. 3E and F). There were no significant differences between groups for sit-to-stand or stand-to-sit transitions.

4.5. Self-reported measures, correlations with significant instrumented parameters

People with MS (mean 87, 46–100) had less balance confidence on the ABC questionnaire than controls (mean 98, 88–100, $p < 0.001$). Likewise people with MS had worse self-ratings of gait on the MSWS12 (mean 16.9, 0–58) than controls (0.2, 0–5), $p < 0.001$. For people with MS, correlations were performed between the best discriminatory parameters and the ABC, MSWS12, self-rated EDSS, T25FW, and mTUG, (Table 2). The quiet stance parameter sway acceleration amplitude EC had a moderate inverse relationship with the ABC ($\rho = -0.56$, $p < 0.01$) and a lesser correlation with the MSWS12 ($\rho = 0.38$, $p = 0.04$); thus people with MS who had worse balance confidence and worse perceived walking tended to have more difficulties maintaining postural balance with eyes closed than those with MS who had better balance confidence and perceived walking abilities. None of the other parameters were significantly associated with self-reported measures of clinical disability (EDSS), functional disabilities (ABC, MSWS12), or stop-watched timed walking tests (T25FW, mTUG).

4.6. Correlations between significant instrumented parameters

With the exception of ML nJerk in the EC condition correlating with turning duration ($\rho = -0.370$, $p = 0.05$), the best discriminatory balance and gait parameters did not correlate with each other (Supplemental Table 2).

5. Discussion

Body-worn motion sensors captured significant differences in quiet stance, gait, and postural transition parameters in people with MS when standard stop-watch timed tests commonly used in clinical practice (T25FW and TUG) could not. The clinical relevance is that while patients report balance and gait problems, only the instrumented measure and not the timed measures could detect objective deficits. However, mechanisms of MS balance and gait dysfunction are as yet unknown given the few significant correlations between the distinguishing instrumented parameters and the self-reported and timed measures. Thus, motion sensors detect small changes in MS mobility and require few subjects to do so; future work will determine both the clinical meaning of individual parameters and the responsiveness of parameters to change over time.

The instrumented parameters distinguishing MS from controls appeared related to aspects of static and dynamic balance confirming prior reports of abnormal balance in early and/or minimally impaired MS [11,12]. As balance maintenance mechanisms differ depending on specific tasks, comprehensive testing of both static and dynamic balance can focus rehabilitation strategies.

During quiet stance, sway acceleration amplitude increased more in people with MS with eyes closed condition, suggesting a greater reliance on visual input presumably due to losses of other balance maintenance functions (e.g. sensory loss, vestibular dysfunction, etc.). The reduced jerk during quiet stance EC may be the direct result of visual input loss or a compensatory strategy to it.

During gait, people with MS had increased trunk ROM in roll (ML flexion), yaw (axial trunk rotation), and combined total lateral trunk ROM. The role of lateral trunk ROM as a key component of dynamic balance control during gait has been previously established [27,28]. Both, instability produced by increased lateral trunk motion and alterations in sensory inputs are compensated by varying lateral foot placement [27]. As gait ataxia is characteristic of MS, total lateral trunk ROM may prove a quantitative surrogate for gait ataxia.

Turning and postural transition parameters are seldom studied in MS. The significantly longer turning times in our study may suggest an impaired proprioceptive system. When a head turns in preparation for body turn, proprioceptive inputs generated during locomotion no longer correlate with vestibular and visual inputs. Therefore extra weighting may be placed on the proprioceptive systems for postural orientation [29]. Future studies will test sensation, proprioception, vision, and strength to determine their relative contributions to MS balance maintenance during quiet stance, gait and postural transitions.

Abnormal mobility parameters in this study differed from other studies, likely due to variations in patient selection, data acquisition protocols, and analysis. Patient selection based on normal walking speed in this study may account for our lack of detection of commonly described abnormal temporal-spatial gait parameters (e.g. velocity, step length, base of support) [5,11,12,14]. Turning and postural transition parameters have not been routinely measured in prior studies, nor have acceleration-based measures of postural stability such as jerk of trunk sway. Notably, angular motion of the trunk could previously only be calculated from motion-analysis segment rotation calculations – here we calculate it directly. Standardized testing protocols and disease-specific parameters will help make comparisons of mobility testing across centers possible.

The few subjects necessary to discriminate MS from controls using the objective mobility data captured by body-worn sensors demonstrates their potential use in clinical research, clinical practice, and rehabilitation. Clinical trials could have smaller sample sizes and shorter durations, thereby reducing costs and facilitating faster testing of novel therapies. MS clinicians could have a clinically relevant and objective gauge of their patients' mobility status unencumbered by recall bias or subjectivity [30]. Tailored rehabilitation strategies could be monitored for efficacy and ongoing modification. Ongoing data collection will determine the responsiveness of body-worn motion sensor parameters to change over time to determine their ability to measure non-relapse related disease progression.

In summary, data from body-worn motion sensors in this study discriminated MS subjects from matched controls when traditional timed tests could not. Quantitative balance and gait measures in MS have the potential to detect benefits of drug and rehabilitation interventions, document disease progression, and improve clinical trial efficiency.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

Mr. Joshua Adams is acknowledged for his assistance with data collection. Dr. Spain received grant support for this project from the National MS Society Mentor Based Post-doctoral Fellowship in Rehabilitation, and the Medical Research Foundation of Oregon Early Clinical Investigator Award. This publication was made possible with support from the Oregon Clinical and Translational Research Institute (OCTRI), grant number UL1 RR024140 from the National Center for Research Resources (NCRR), a component of the National Institutes of Health (NIH), and NIH Roadmap for Medical Research. Dr. Spain is a Career Development Award recipient from the VA Rehabilitation Research & Development service.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi: 10.1016/j.gaitpost.2011.11.026.

Fig. 1.

The experimental apparatus and protocol. A. A subject performing the quiet standing task, locations of the six body-worn sensors shown. B. Schematic of the walking task divided into gait and postural transition (sit-to-stand, stand-to-sit, turning) phases. C. An example of the motion signal from the walking task showing the horizontal angular velocity of the sternal sensor.

Fig. 2. Timed walking tests did not separate MS from healthy controls using the Timed 25 Foot Walk (T25FW, A), modified Timed Up and Go measured with a stopwatch (mTUG, B), or measured with the body-worn sensors (iTUG, sensors, C). Data points indicate each subject time, horizontal black lines show group means, and white boxes show standard deviations.

Fig.3.

Significant differences were found between MS and healthy control subjects during quiet stance, gait and postural transition phases of the instrumented mobility tasks. Cutoffs that maximize sensitivity and specificity of the significant parameters (A, B, D, E) as well as receiver operator characteristic (ROC, C, F) curves are shown. Both means (horizontal black lines) and medians (horizontal grey lines) of the parameters are shown to demonstrate that group effects are driving differences between MS and control and not outliers (A, B, D, E). AUC values are found in Table 1.

Table 1

Differences between MS and control subjects in selected quiet stance parameters divided into eyes-open and eyes-closed conditions and in walking task parameters divided into gait (upper body), gait (lower body), and postural transition phases.

	MS subjects mean $\pm$ SE	Control subjects mean $\pm$ SE	<i>p</i>	AUC
Quiet stance, eyes-open				
Sway acceleration amplitude (m/s ²)	0.062 $\pm$ 0.007	0.052 $\pm$ 0.003	0.23	0.55
Mean velocity (m/s)	0.33 $\pm$ 0.03	0.30 $\pm$ 0.03	0.52	0.51
Centriod freq. (Hz)	0.73 $\pm$ 0.03	0.76 $\pm$ 0.04	0.55	0.50
nJerk ML	2.48 $\pm$ 0.12	2.68 $\pm$ 0.13	0.27	0.57
nJerk AP	1.17 $\pm$ 0.06	1.16 $\pm$ 0.06	0.86	0.48
Quiet stance, eyes-closed				
Sway acceleration amplitude (m/s ²)	0.085 $\pm$ 0.01	0.058 $\pm$ 0.005	0.02	0.68
Mean velocity (m/s)	0.45 $\pm$ 0.06	0.33 $\pm$ 0.04	0.11	0.64
Centriod Freq. (Hz)	0.70 $\pm$ 0.03	0.72 $\pm$ 0.03	0.57	0.43
nJerk ML	2.41 $\pm$ 0.12	2.82 $\pm$ 0.14	0.05	0.65
nJerk AP	1.18 $\pm$ 0.06	1.15 $\pm$ 0.04	0.67	0.52
Gait, lower body				
Cadence (steps/min)	130.3 $\pm$ 3.0	131.4 $\pm$ 2.3	0.79	0.43
Swing (% gait cycle)	40.4 $\pm$ 0.5	40.1 $\pm$ 0.5	0.69	0.52
Double support time (% gait cycle)	19.2 $\pm$ 1.0	19.7 $\pm$ 0.7	0.69	0.48
Normalized stride length (%/ht)	89.0 $\pm$ 1.0	90.9 $\pm$ 1.0	0.19	0.39
Normalized velocity	96.5 $\pm$ 2.2	99.6 $\pm$ 2.2	0.32	0.42
Stride length variability	3.2 $\pm$ 0.2	3.2 $\pm$ 0.3	0.85	0.54
ROM shank (°)	82.6 $\pm$ 1.0	84.4 $\pm$ 1.2	0.22	0.40
Gait, upper body				
Arm swing peak speed (°/s)	220.1 $\pm$ 12.1	231.4 $\pm$ 15.8	0.57	0.47
Arm swing Speed variability (%)	15.1 $\pm$ 1.2	14.2 $\pm$ 1.1	0.60	0.53
Arm swing ROM (°)	21.9 $\pm$ 1.8	22.8 $\pm$ 2.5	0.76	0.50
Trunk ROM Roll (ML flexion,°)	7.5 $\pm$ 0.5	5.9 $\pm$ 0.35	0.02	0.67
Trunk ROM Pitch (AP flexion,°)	5.2 $\pm$ 0.3	5.1 $\pm$ 0.3	0.70	0.55
Trunk ROM Yaw (rotation,°)	11.1 $\pm$ 0.71	9.0 $\pm$ 0.48	0.03	0.65
Total lateral trunk ROM (°)	13.65 $\pm$ 0.68	10.84 $\pm$ 0.52	<0.01	0.72
Trunk acceleration (m/s ²) lateral	0.75 $\pm$ 0.04	0.74 $\pm$ 0.04	0.88	0.53
Trunk acceleration (m/s ²) sagittal	1.27 $\pm$ 0.05	1.36 $\pm$ 0.07	0.27	0.46
vertical	0.74 $\pm$ 0.04	0.72 $\pm$ 0.04	0.69	0.53
Postural transitions				
Turning duration (s)	1.77 $\pm$ 0.06	1.58 $\pm$ 0.06	0.03	0.63
Peak turning velocity (°/s)	196.2 $\pm$ 6.1	208.6 $\pm$ 8.8	0.24	0.4
Sit-to-stand duration (s)	2.35 $\pm$ 0.1	2.19 $\pm$ 0.1	0.34	0.57

When the ML and AP values both had the same level of significance, then data from combined directions is presented. For nJerk the ML and AP values are presented separately. AUC, area under curve; Freq., frequency; Hz, hertz; ML, medio-lateral; m, meter; MS, multiple sclerosis; nJerk, normalized jerk; s, second; SE, standard error.

Spearman correlations (r) and 2-tailed significance (p) among MS subjects between the most discriminatory instrumented mobility parameters captured during the test components of quiet standing, gait, and postural transitions and self-reported measures of clinical disability (EDSS), functional disabilities involving balance (ABC) and gait (MSWS12), and stop-watched timed walking tests (T25FW, mTUG).

Table 2

Instrumented mobility parameter; test component	EDSS		ABC		MSWS12		T25FW		mTUG	
	$r(p)$		$r(p)$		$r(p)$		$r(p)$		$r(p)$	
Sway RMS, EC (m/s ²); quiet stance	0.15 (0.43)		-0.56 (<0.01)		0.38 (0.04)		0.18 (0.34)		0.22 (0.25)	
ML nJerk, EC; quiet stance	0.03 (0.90)		0.21 (0.28)		-0.21 (0.29)		0.05 (0.81)		-0.01 (0.96)	
Total lateral trunk ROM (°); gait	0.051 (0.79)		-0.18 (0.36)		0.04 (0.82)		-0.37 (0.06)		-0.10 (0.59)	
Turning duration (s); postural transitions	0.04 (0.83)		-0.021 (0.28)		0.16 (0.41)		0.24 (0.20)		0.30 (0.11)	

ABC, Activities of Balance Confidence; EC, eyes closed condition; EDSS, Expanded Disability Status Scale; m, meter; MSWS12, Multiple Sclerosis Walking Scale 12; nJerk, normalized JERK; ROM, range of motion; s, second; T25FW, Timed 25 Foot Walk; mTUG, modified (7m) Timed Up and Go.