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Reliability and minimal detectable change for inertial measurement units – Derived stability, symmetry, and smoothness indexes of gait in people with multiple sclerosis

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ABSTRACT

Background: people with multiple sclerosis (PwMS) experience loss of gait stability, smoothness, and symmetry, which affects quality of life and requires accurate evaluation for effective rehabilitation. Inertial measurement units (IMUs) offer a promising approach to monitor gait quality by quantifying indices reflecting stability, symmetry, and smoothness whose minimal detectable changes (MDCs) are not defined in people with MS (PwMS).
Research question: to assess the within-day test-retest reliability and MDCs of IMUs – derived gait stability, symmetry, and smoothness metrics in PwMS during the 10 m walking test (10MWT).
Methods: 58 PwMS wore five IMUs and performed the10MWT twice with a 10 min rest between each trial. Log dimensionless jerk (LDLJ) and improved Harmonic Ratio (iHR) were calculated for each gait trial based on the signals from the pelvis – mounted IMU, normalized Root Mean Square (nRMS) were calculated also from the head and sternum-mounted IMUs. Intraclass correlation coefficient (ICC) were calculated between the results of the two 10MWT to assess test -retest reliability, and minimal detectable change scores were calculated.
Results: Reliability of the investigated parameters ranged from moderate to excellent values, with ICC ranging from 0.64 to 0.98. MDC values ranged from 0.09 to 0.53 for the nRMS, from 7.54 to 11.36 for the iHR and from 0.15 to 0.20 for the LDLJ.
Significance: This study showed moderate to excellent reliability for the investigated indexes when calculated based on 10MWT, with the LDLJ showing the highest reliability, thus providing a reliable smoothness metric in pwMS. Also, nRMS showed good reliability, but caution is warranted with iHR due to its lower reliability and higher MDCs.

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1. Introduction

Impairments in walking and balance function represent significant challenges for people with multiple sclerosis (PwMS) [1], impacting their daily activities and overall quality of life [2]. The assessment of gait and postural stability allows for the estimation of walking ability as well as the monitoring of the effectiveness of rehabilitation interventions aimed at improving it [3]. Objective gait evaluation is required in both routine clinic care and clinical research studies to improve gait and balance follow-up in PwMS. Although clinical scales remain the gold standard for assessing mobility and disability in PwMS, they may lack sufficient sensitivity to detect subtle gait alterations or early changes in disease progression. In such cases, digital biomarkers and instrumentally-derived metrics from wearable sensors may offer complementary information to enhance clinical evaluation [4].

In this context, inertial measurement units (IMUs) emerge as a promising technology to monitor gait disorders and evaluate the effectiveness of rehabilitation strategies [5]. Wearable technologies, such as IMUs, can be used to provide gait metrics outside of traditional laboratory settings as well as to improve clinical motor assessment in ambulatory outpatient facilities where more expensive gait analysis systems are not feasible. In addition to traditional spatiotemporal and joint kinematic gait parameters, a series of postural and gait indices can be calculated based on accelerations and angular velocities derived from IMUs. These metrics demonstrated to have potential as reliable digital biomarkers of mobility, thus informing about disease severity [4]. They also showed to reflect clinical improvement in terms of external responsiveness [6]. Particularly, IMUs-derived indexes of stability, symmetry, and smoothness of gait have been proposed in people with neurological conditions, including PwMS [5,7,8]. Acceleration-derived gait indexes have been shown to detect early gait and balance impairments, as well as fatigue symptoms in pwMS [9–11]. In particular, the harmonic ratio and the improved harmonic ratio (iHR) reflect the symmetry of trunk accelerative behavior during gait, although HR has also been applied in some studies as an indicator of gait smoothness [5]. Conversely, the log-dimensionless jerk score (LDLJ) is specifically designed to quantify gait smoothness using acceleration signals or angular velocities [11].". Their clinometric properties as gait symmetry metrics have been investigated in a variety of neurologic conditions, including Parkinson's disease [12], stroke [13], hereditary cerebellar ataxia [14], and multiple sclerosis, where they demonstrated the ability to detect subtle impairments even in the early stages of the disease [10]. The normalized root mean square of acceleration signals (nRMS) measures the total magnitude or intensity of body segment acceleration during gait. It is widely used to detect gait stability abnormalities in neurological conditions [13] such as multiple sclerosis [15]. The log-dimensionless jerk score (LDLJ) measures localized smoothness during gait using acceleration signals or angular velocities retrieved from IMUs [16–18], and it has been proposed to quantify gait smoothness in PwMS [5,19]. However, their minimal detectable change (MDC) scores have been rarely investigated in those populations [20,21] and are still unexplored in PwMS.

The MDC is a statistical measure that determines the smallest amount of change in a patient's performance or condition that can be considered significant beyond the margin of measurement error, thus allowing to determine whether the change in outcome score over time represents a true change, rather than just random fluctuations or inherent error associated with the measurement process [22,23]. The ability to identify subtle meaningful changes in gait characteristics becomes of paramount importance for clinicians allowing them to conduct more reliable evaluations and to assess clinical modifications along the disease history, for better management of PwMS in rehabilitative settings. Indeed, the calculation of MDC is strictly correlated with measures of outcome reliability, specifically the intraclass correlation coefficient (ICC). The more reliable a measurement instrument is, the more sensitive it is to detect true changes in the variable. When applying such IMUs – derived indices to the linear path in the steady state phase, a time series length of 20 strides or more is usually considered [12,14,24,25]. However, this condition requires a relatively long acquisition space, which is not always readily available in clinical settings, limiting the usefulness of these metrics out of laboratory contexts.

We hypothesized that IMUs-derived gait indexes might result in reliable and sensitive change scores even for shorter gait bouts. The clarification of MDCs for these indices may serve as baseline for identifying meaningful changes in walking patterns, thus supporting informed decisions in rehabilitation planning for PwMS. Therefore, the aims of this study are to assess the within – day test-retest reliability and MDCs of three IMUs-derived gait stability, symmetry, and smoothness metrics, namely the normalized Root Mean Square (nRMS), the improved Harmonic Ratio (iHR), and the log dimensionless jerk (LDLJ), respectively, in a population of persons suffering from multiple sclerosis during short bouts of linear walking, such as the 10 meters walking test (10MWT), which represents a more feasible path length also in clinical rehabilitation settings.

2. Materials and methods

2.1. Participants

This cross-sectional study was carried out at Santa Lucia Hospital (Institute for Research and Healthcare) and was approved by the Local Independent Ethics Committee with protocol number CE/ PROG.812. All procedures contributing to this work comply with the ethical standards of the relevant national and institutional guidelines on human experimentation and with the World Medical Association Declaration of Helsinki and adhere to the Strengthening the Reporting of Observational Studies in Epidemiology (STROBE) guidelines [26]. All participants gave written consent to publish the results obtained from their clinical examination and instrumental tests. A researcher who did not take part in the gait assessments determined the individual's eligibility to participate based on inclusion and exclusion criteria. Participants with a diagnosis of MS according to the McDonald Criteria [27] were recruited and enrolled based on consecutive sampling between November 2020 and October 2022. The following inclusion criteria were applied: i) diagnosis of relapsing–remitting (RR) or secondary-progressive (SP) MS diagnosed by an experienced neurologist; ii) aged between 20 and 75 years; iii) Expanded Disability Status Scale (EDSS) [28] score between 0 and 6; iv) self – reported ability to walk independently as determined by asking participants if they could walk for at least 50 m without assistance. Exclusion criteria consisted of: i) the presence of psychiatric and neurological disorders (other than MS) and other pathological conditions and/or clinical disorders that might interfere with cognitive functioning and/or the performance of motor or cognitive tasks; ii) the occurrence of a clinical relapse within three months before the enrolment; iii) steroid therapies within 30 days before the enrolment; iv) the occurrence of a lower extremity fracture within three months before the enrolment. For sample characterization, each patient was evaluated clinically using the MiniBesTest [29], Tinetti scale [30], and Modified Barthel Index [31]. Demographics and clinical characteristics of the

Table 1
Demographics and baseline characteristics of participants.

Age (years), mean (SD)	51 (10.2)
Gender (male), n (%)	20 (33.3 %)
Education (years), mean (SD)	14.2 (4.4)
Time since MS diagnosis (years), mean (SD)	13 (8.9)
EDSS, median (range)	4 (1.5–4)
Tinetti scale, mean (SD)	22.6 (6.2)
Mini-BESTest scale, mean (SD)	18.2 (6.7)
Modified Barthel Index, mean (SD)	96.8 (4.4)

SD: standard deviation; M: male; MS: multiple sclerosis; EDSS: expanded disability status scale.

included sample are described in Table 1.

2.2. Data collection

To ensure data quality, raters underwent targeted training for administration of clinical outcome measures and kinematics assessments. Inertial sensors were securely fastened to pertinent body segments using Velcro straps to minimize oscillations and to diminish motion artifacts. Additionally, two physiotherapists stayed close to participants during assessments to prevent falls and ensuring the correct test execution [8]. The IMUs - derived indices were measured during walking trials performed at a comfortable speed on a 10 m walking path with 2 m auxiliary paths at each end, for a total of 14 m straight path. The data were collected at the Santa Lucia Foundation's rehabilitation

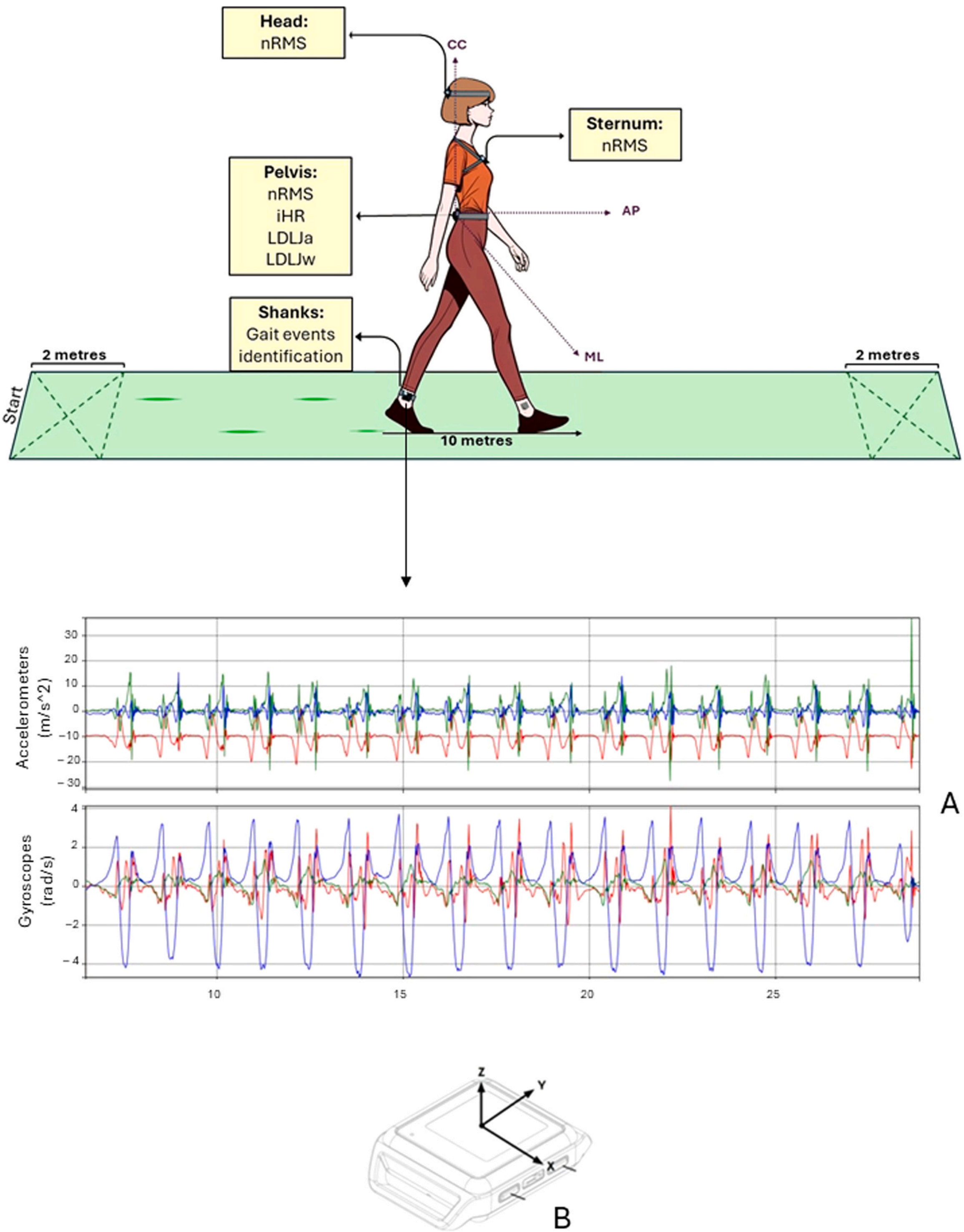

Fig. 1. Experimental setup. Subjects wore 5 mIMUs placed at head, shanks, pelvis, sternum, and head. Subjects were asked to walk along 14 m – long pathway. The first and last 2 m were excluded from the analysis. nRMS, normalized root mean square; iHR, improved harmonic ratio; LDLJa, log dimensionless jerk score based on acceleration signals; LDLJw, log dimensionless jerk score based on angular velocities signals; CC, cranio – caudal direction; AP, antero – posterior direction; ML, medio – lateral direction. (A), example of accelerations and angular velocity signals derived from the left leg of a pwMS survivor; (B), axes orientation.

facility in Rome, Italy. The corridor had no visible pavement connections or border lines, and the indirect lighting was evenly distributed along its length. This protocol was performed twice within 10 min using the same experimental setup [32,33] (Fig. 1).

The instrumental assessments were performed by two physiotherapists specifically experienced in gait analysis with IMUs. Prior to each experimental session, an ad hoc acquisition was performed to remove bias and recalibrate the IMUs [34]. During the test participants were equipped with five synchronized IMUs (128 Hz. Opal. APDM. Portland. OR. USA), acquiring samples on three-dimensional linear accelerations and angular velocities with full scale ranges of 16 g, 2000deg/s, respectively. IMUs were located on the occipital cranium bone, near the lambdoid suture of the head (H), at the center of the sternum (S), and at the L4/ L5 level just above the pelvis (P). For step and stride segmentation, one IMU was placed on each leg, just above the lateral malleoli. Initial contacts were detected using local minima in the vertical angular velocity of the shank, while final contacts were identified using peaks in the antero-posterior angular velocity of the shank. Gait speed was calculated by dividing the estimated stride length by the corresponding stride duration [35]. During the first static part of each trial, a reference system aligned with respect to the gravity vector was obtained. The time-invariant rotational matrix between each IMU and the defined reference system was calculated and applied to the dynamic part of the trials. As a result, accelerometer and gyroscope data were expressed in a reference system approximatively aligned to the antero-posterior (AP), medio-lateral (ML), and cranio-caudal (CC) anatomical axes. With this approach, the acceleration due to gravity from the CC component of the accelerometers data was removed [34]. The signals were passed through a second-order Butterworth low-pass filter (10 Hz for accelerometer data, 6 Hz for gyroscope data) before computing gait metrics, including iHR, LDLJ, and RMS. These metrics are sensitive to high-frequency noise, which can distort measurements of gait regularity, smoothness, and magnitude. The cutoff frequencies were chosen to preserve physiologically meaningful signal content associated with gait (typically less than 10 Hz) while reducing sensor-related noise. Data were processed in the Matlab® environment (MATLAB R2022b. MathWorks) for the extraction of the following acceleration-derived indexes: nRMS, iHR and LDLJ. These metrics were identified as commonly used in neurorehabilitation literature for capturing IMU-derived stability, symmetry, and gait smoothness, respectively, unless their reliability was yet to be determined.

The initial RMS was calculated based on the acceleration signals measured at the pelvis, trunk, and head levels. The RMS values of each stride acceleration were obtained for the AP, ML, and CC components. Then, to take the influence of the walking speed into account, the AP and ML components were then divided by the CC component [36,37], thus obtaining the normalized values here referred to as nRMS was calculated as follows:

$$\boldsymbol{nRMS} = \frac{\sqrt{\frac{\Sigma_i^N = \quad 1^{x^2}}{N}}}{\sqrt{\frac{\Sigma_i^N \quad = \quad 1^{y^2}}{N}}}$$

where x corresponds to the AP or ML components and y to CC one and N is the number of samples of each stride. High nRMS values were associated with higher levels of acceleration, hence decreased stability.

The iHR parameter ranges from 0 % (total asymmetry) to 100 % (total symmetry) [38] and was calculated based on the acceleration measured at the pelvis level, for each acceleration component (AP, ML and CC) as:

$$\boldsymbol{iHRj} = \frac{\Sigma_{i=1}^{K} P_I^i}{\Sigma_{i=1}^{K} \left(P_I^i + \quad P_E^i\right)} \times 100$$

where P_I^j and P_E^j are the power of the intrinsic and extrinsic j^{th} harmonic obtained considering k = 20 harmonics in this study.

The LDLJ was calculated at the pelvis level, from the linear acceleration and angular velocity signals for each spatial direction, resulting in two indexes, namely LDLJ(a) and LDLJ(w), respectively [16], that were calculated as follows:

LDLJ was calculated as follows:

$$\boldsymbol{LDLJ_s} \quad = \quad -\ln\left(\frac{(t_2 - t_1)^3}{max(\|s(t)\|_2^2)} \quad \int_{t_1}^{t_2} \| \frac{d}{dt} \quad s(t)\|_2^2 dt\right)$$

where *s(t)* represents the linear accelerations or angular velocity data and t_1 and t_2 are the starting and ending points, respectively, of the gait cycle. LDLJ values close to zero were considered as corresponding to greater smoothness.

2.3. Sample size calculation

A sample size of at least 53 PwMS was determined to be necessary to achieve a test-retest reliability ranging from moderate (ICC = 0.61) to excellent (ICC > 0.80), assuming a 95 % confidence level and 80 % power, within a design involving two assessments by the same rater [39].

2.4. Statistical analysis

Statistical analysis was performed using the IBM SPSS Statistics software (v24. IBM Corp., Armonk, NY, USA). Demographics and clinical characteristics of the sample were reported through descriptive statistics such as mean, standard deviations (SD), range and percentages. The mean and SD were also calculated for test-retest and their difference for each of the acceleration indexes. After verifying the normality of the distributions through the Shapiro-Wilk test, paired samples *t*-test or Wilcoxon test, Cohen's effect size (d) and confidence interval at 95 % were calculated to assess the significance and magnitude of the differences between the gait trials. The ICC using a two-way mixed-effects model for absolute agreement within its 95 % confidence interval (CI) was considered to assess the reliability. ICC values were interpreted as follows: less than 0.5 as poor reliability, 0.5–0.75 as moderate reliability, 0.75–0.9 as good reliability, greater than 0.90 as excellent reliability [40]. The statistical level of significance was set at alpha = 0.05. To calculate the standard error of measurement (SEM) the following formula was considered:

$$SEM = SD\sqrt{1 - ICC}$$

Based on the above-mentioned calculation, the MDC was calculated using the following formula:

$$MDC = SEM \bullet 1.96 \bullet \sqrt{2}$$

3. Results

Sixty PwMS were initially recruited, and 2 participants were excluded from the analysis due to a measuring error during the data collection. Demographics and baseline characteristics of the sample are reported in Table 1. Descriptive statistics for nRMS, iHR, and LDLJ parameters are reported in Table 2. The greatest test-retest difference was observed in AP for nRMS at head, in CC for iHR, and in both CC and AP directions for LDLJw. Reliability of the investigated parameters ranged from moderate to excellent values, with ICC ranging from 0.637 to 0.981. MDC values ranged from 0.088 to 0.529 for the nRMS, from 7.539 to 11.359 for the iHR and from 0.150 to 0.198 for both the LDLJs. All the results are detailed in Table 3.

4. Discussion

This study aimed to assess the test-retest reliability and the MDCs for the IMU-derived nRMS, iHR, and LDLJ, in PwMS performing the

Table 2
Descriptive statistics for normalized Root Mean Square (nRMS), improved harmonic ratio (iHR) and log dimensionless jerk (LDLJa and LDLJw).

		Test Mean (95 % CI)	Re - test Mean (95 % CI)	Test -retest difference Mean (95 % CI)	p -value	Cohen's d
Gait speed (m/s)		1.146 (1.004–1.288)	1.191 (1.044–1.337)	−0.061 (−0.129–0.219)	0.605	0.069
nRMS	**AP pelvis**	0.817 (0.634–0.862)	0.810 (0.727–0.873)	0.007 (−0.088–0.093)	0.960	−0.007
	ML pelvis	0.729 (0.640–0.782)	0.712 (0.651–0.792)	0.016 (−0.075–0.096)	0.805	−0.033
	AP trunk	0.630 (0.505–0.715)	0.626 (0.503–0.710)	0.004 (−0.131–0.123)	0.952	0.008
	ML trunk	0.665 (0.559–0.824)	0.642 (0.552–0.805)	0.023 (−0.155–0.130)	0.856	0.024
	AP head	0.771 (0.515–0.719)	0.695 (0.511–0.749)	0.076 (−0.131–0.156)	0.862	−0.023
	ML head	0.702 (0.525–0.754)	0.673 (0.529–0.755)	0.029 (−0.124–0.129)	0.969	−0.005
iHR	**AP**	78.627 (78.138–85.499)	80.234 (77.241–84.844)	−1.606 (−6.270–4.767)	0.786	0.036
	ML	72.196 (70.693–78.262)	72.756 (71.189–78.107)	−0.559 (−5.200–5–541)	0.949	−0.009
	CC	80.353 (80.795–87.624)	82.093 (81.407–87.950)	−1.739 (−3.928–4.866)	0.831	−0.029
LDLJa	**AP**	−5.221 (−5.309 - −5.133)	−5.229 (−5.335 - −5.124)	−0.008 (−0.149–0.133)	0.908	0.016
	ML	−5.422 (−5.514 - −5.329)	−5.383 (−5.469 - −5.298)	0.039 (−0.082–0.159)	0.521	−0.086
	CC	−5.036 (−5.125 - −4.948)	−5.050 (−5.137 - −4.963)	−0.013 (−0.128–0.101)	0.817	0.031
LDLJw	**AP**	−4.092 (−4.232 - −3.953)	−4.028 (−4.188 - −3.867)	0.065 (−0.141–0.271)	0.531	−0.084
	ML	−4.019 (−4.145 - −3.893)	−4.066 (−4.196 - −3.937)	−0.048 (−0.222–0.127)	0.586	0.073
	CC	−3.804 (−3.687 - −3.922)	−3.801 (−3.938 - −3.663)	0.004 (−0.166–0.174)	0.965	−0.006

AP: antero-posterior; CC: Cranio-Caudal; ML: medio-lateral; LDLJa: log dimensionless jerk linear acceleration; LDLJw: log dimensionless jerk angular velocity; 95 % CI: 9 % confidence interval

Table 3
Results of Interclass correlation coefficient (ICC), standard error of measurement (SEM) and minimal detectable change (MDC).

		ICC (95 % CI)	SEM (95 % CI)	MDC (95 % CI)
nRMS	**AP pelvis**	0.852 (0.752–0.912)	0.032 (0.025 0.041)	0.088 (0.068–0.115)
	ML pelvis	0.792 (0.652–0.876)	0.056 (0.043–0.072)	0.156 (0.120–0.201)
	AP trunk	0.868 (0.778–0.921)	0.060 (0.047–0.078)	0.167 (0.129–0.216)
	ML trunk	0.817 (0.693–0.891)	0.043 (0.033–0.055)	0.119 (0.092.0.154)
	AP head	0.757 (0.594–0.855)	0.191 (0.147–0.247)	0.529 (0.409–0.684)
	ML head	0.799 (0.664–0.88)	0.052 (0.040–0.067)	0.143 (0.111–0.186)
iHR	**AP**	0.683 (0.470–0.811)	3.516 (2.715–4.546)	9.747 (7.525–12.602)
	ML	0.622 (0.368–0.774)	4.098 (3.169–5.299)	11.359 (8.723–14.688)
	CC	0.815 (0.691–0.89)	2.719 (2.097–3.514)	7.539 (5.812–9.740)
LDLJ a	**CC**	0.981 (0.966–0.989)	0.054 (0.041–0.072)	0.150 (0.114–0200)
	ML	0.925 (0.867–0.958)	0.070 (0.052–0.093)	0.193 (0.145–0.258)
	AP	0.916 (0.85–0.953)	0.062 (0.046–0.083)	0.171 (0.129–0.230)
LDLJ w	**CC**	0.910 (0.839–0.949)	0.071 (0.053–0.095)	0.198 (0.148–0.263)
	ML	0.946 (0.904–0.97)	0.065 (0.048–0.087)	0.185 (0.134–0.240)
	AP	0.965 (0.937–0.98)	0.057 (0.043–0.076)	0.159 (0.119–0.212)

AP: antero-posterior; CC: Cranio-caudal; ICC: intraclass correlation coefficient; iHR: improved harmonic ratio; LDLJa: log dimensionless jerk linear acceleration; LDLJw: log dimensionless jerk angular velocity; MDC: minimal detectable change; ML: medio-lateral; nRMS: normalized root mean square; SEM: standard error of measurement

10MWT. No significant differences between two consecutive trials were found in the investigated parameters (Table 2), resulting in moderate – to – excellent reliability levels (Table 3). Particularly, LDLJ showed the highest ICC values, suggesting that this metric can be considered as reliable in PwMS even when dealing with short gait bouts, regardless of the method used for the calculation (i.e. using acceleration or angular velocity data). In this way, our results suggest that trunk smoothness of PwMS during gait can be assessed in laboratory settings. To the best of our knowledge, this is the first study that investigated test-retest reliability and MDC of LDLJ. Our findings can contribute to better assess the effectiveness of interventions on gait smoothness in terms of true change scores that are beyond measurement errors. Moreover, a recent study found a significant effect of rehabilitation on LDLJ in PwMS [8]. Our MDC findings further reinforce this result, by revealing that, when LDLJ is used as outcome measure, small changes are needed to reflect true change in LDLJ. Noteworthy, smoothness metrics rely on precise velocity data and IMU orientation, which has a direct impact on measuring movement intermittency and distinguishing between smooth and unsmooth movements [16,17]. As a result, drift in velocity may have an impact on LDLJw calculations for linear motion analysis. In this study, accelerometer and gyroscope bias removal was obtained by the ad hoc acquisition performed before each experimental session. Moreover, the consistent reference frame defined during the orthostatic portion of each trial allowed for gravity acceleration removal from accelerometer data, without relying on the IMUs orientation estimation during the dynamic portion of each trial. However, we also provided ICC and MDC values for LDLJ as calculated in the acceleration space, proving the reliability of the LDLJa and providing MDC values to refer when using a single lumbar-mounted IMU for the assessment of gait smoothness during linear walking [16].

Despite being widely used as a gait stability metric, the test-retest reliability and MDC of RMS have been rarely investigated in PwMS [15]. We found no significant differences between the two assessments, as well as substantial – to – optimal reliability values for all the nRMS indexes. This is consistent with the findings by Angelini et al. [15], which reported the root mean square ratio of pelvic accelerations during gait as a robust stability measure, regardless of the adopted gait protocol. The findings of our study revealed that also the nRMS may be considered as a reliable metric to assess gait stability during the 10MWT. However, when calculated from pelvic and trunk accelerations, nRMS slightly outperformed head acceleration-derived nRMS. Combined with the MDC values for nRMS as retrieved from head accelerations, our results suggest caution when using this IMU placement for stability calculation, particularly in the AP direction.

Despite having resulted in moderate reliability values, iHR in the AP and ML directions showed the lowest ICC values and the highest MDCs among the considered parameters. This is consistent with previous research that investigated the HR in a smaller sample of PwMS [17], confirming that this metric should be carefully managed when calculated based on short gait bouts considering few strides, such as in the 10MWT. According to our findings, improvements in iHR ranging from 7.54 % to 11.36 % are needed to identify a true change in iHR regardless of the measurement error. Although they have only been studied in people with Parkinson's disease, these MDC values are consistent with the minimal clinically important change scores for HR, where HR improvements greater than 21.5 % have been described as clinically meaningful [41–43].

Some limitations need to be acknowledged in the present study. First, we included participants with different disability levels. Despite the median score of the EDSS was 4 a few participants reported higher levels of impairment (EDSS $\leq$ 6) suggesting heterogeneity in the sample concerning disability level. However, in this study, only participants who could walk without assistance were included, reducing the bias in reliability calculations caused by the variability in gait impairment among participants. Another limitation of this study may lie in the potential heterogeneity in the subtypes of MS. However, the abnormal gait pattern of pwMS has been described to be independent of the disease phenotype [44]. Furthermore, we included people of varying ages. Because age at onset is considered an outcome predictor in pwMS, we could not rule out the possibility that this factor influenced gait metrics variability within the sample. Further studies should be implemented considering this aspect and assessing test-reliability and MDCs within subgroups of PwMS stratified by the disability level according to the EDSS. However, the use of a single gait trial and a 10-minute rest interval between assessments might have introduced variability due to learning effects or residual fatigue, which should be considered when interpreting the test–retest reliability results [45]. Furthermore, minimal clinically important differences should be evaluated after rehabilitation treatments.

Our results could be relevant for clinicians to determine the actual modifications beyond the measurement error attributable to specific balance and gait evaluation in PwMS. Furthermore, these findings suggest that nRMS, LDLJ, and, more cautiously, iHR produce stable results even if extracted from short distances, extending their applicability to ambulatory settings. However, to consider the investigated metrics as actual outcome measures for assessing the clinical effectiveness of rehabilitation interventions in PwMS, further properties of these metrics should be still assessed, such as their ability to characterize, reflect and predict the clinical characteristics of PwMS subgroups, as well as their external responsiveness to rehabilitation treatments should be calculated in terms of minimal clinically important change scores.

5. Conclusions

This study showed substantial to excellent reliability for IMU-derived postural and gait indexes when calculated based on 10MWT. Particularly, the smoothness of trunk accelerations and angular velocities during walking (as measured by LDLJ), showed the highest reliability values. The stability of acceleration signals as retrieved from inertial sensors placed at the trunk and pelvis level (measured through nRMS), also showed to be a reliable index in people suffering from multiple sclerosis. Conversely, caution is warranted when assessing the symmetry of trunk acceleration patterns during gait (as measured through iHR), due to its low reliability and high MDCs. Future research should investigate the reliability of these metrics across the disability levels of multiple sclerosis to enhance understanding and further clinical application of those parameters, as well as their responsiveness to rehabilitation treatments.

CRediT authorship contribution statement

Marco Tramontano: Writing – review & editing, Supervision, Resources, Investigation, Conceptualization. **Fulvio Dal Farra:** Writing – original draft, Formal analysis, Data curation, Conceptualization. **Mariano Serrao:** Supervision, Resources, Conceptualization. **Ugo Nocentini:** Resources, Investigation, Conceptualization. **Paolo Brasiliano:** Investigation, Formal analysis, Data curation. **Giuseppe Vannozzi:** Writing – review & editing, Investigation, Formal analysis. **Andrea Turolla:** Writing – review & editing. **Paolo Pillastrini:** Supervision, Resources, Investigation, Conceptualization. **Stefano Filippo Castiglia:** Writing – original draft, Formal analysis, Data curation, Conceptualization.

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Declaration of Competing Interest

The authors declare no conflicts of interest.

Data availability

The dataset used in this study is available from the corresponding author upon reasonable request.

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