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Gait stability at early stages of multiple sclerosis using different data sources

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ABSTRACT

Background: People at early stages of multiple sclerosis have subtle balance problems that may affect gait stability. However, differences in methods of determining stability such as sensor type and placements, may lead to different results and affect their interpretation when comparing to controls and other studies.
Questions: Do people with multiple sclerosis (PwMS) exhibit lower gait stability? Do location and type of data used to calculate stability metrics affect comparisons?
Methods: 30 PwMS with no walking impairments as clinically measured and 15 healthy controls walked on a treadmill at $1.2 \, \mathrm{ms}^{-1}$ while 3D acceleration data was obtained from sacrum, shoulder and cervical markers and from an accelerometer placed at the sacrum. The local divergence exponent was calculated for the four data sources. An ANOVA with group (multiple sclerosis and control) and data source as main factors was used to determine the effect of disease, data source and their interaction on stability metrics.
Results: PwMS walked with significantly less stability according to all sensors (no interaction). A significant effect of data source on stability was also found, indicating that the local divergence exponent derived from sacrum accelerometer was lower than that derived from the other 3 sensor locations.
Significance: PwMS with no evident gait impairments are less stable than healthy controls when walking on a treadmill. Although different data sources can be used to determine MS-related stability deterioration, a consensus about location and data source is needed. The local divergence exponent can be a useful measure of progression of gait instability at early stages of MS.

1. Introduction

Multiple sclerosis (MS) is the most common inflammatory demyelinating disease of the central nervous system causing disability in young adults [1]. Walking impairment is a frequent and disabling consequence and is given the highest priority by people with MS (PwMS) [2]. Spatiotemporal measures of gait have revealed that PwMS use strategies that may improve their stability, i.e. taking short steps [3] even when they have no overt disability [4]. These compensations are possibly employed to decrease instability that may lead to an increased risk of falling [5].

Since current clinical measures of disability have been shown to be insensitive to gait subtle changes, new markers of ambulatory function in PwMS have been recommended [6,7]. The local divergence exponent (LDE), has been proposed as a gait stability measure in neurological populations including PwMS [6,8,9]. However, there are methodological concerns about the LDE's validity derived from different sensor locations and using a different number of strides [10,11]. A good stability metric can be of high relevance as it can inform clinicians about disease progression and/or interventions effectiveness; especially at MS early stages when these interventions may be more effective.

The LDE quantifies the exponential rate of divergence and reflects the ability of the motor control system to cope with small perturbations elicited at each step [11]. Since the LDE has been derived using different methods, such as different forms of sensors, sensor locations and type of data (e.g. velocity or acceleration) so that there is still no general consensus about its clinical implementation [10]. In this study, we aimed to determine the effects different data sources on gait stability (LDE short-term) in PwMS during treadmill walking.

Table 1
Participants demographics and Expanded Disability Status Scale (EDSS) clinical scores for people with multiple sclerosis (PwMS) and spatiotemporal measures of gait for both groups. IDS: initial double support; SS: single stance; SDS: second double support.

	PwMS (n = 30)		HC (n = 15)		
	mean	sd	mean	sd	p
Age (years)	42.5	± 9.2	36.8	± 7.0	0.03
Height (cm)	167.7	± 6.5	170.8	± 12.6	0.28
Weight (kg)	70.6	± 12.0	69.8	± 14.7	0.86
Sex (f/m)	25/5	–	9/6	–	
EDSS	1.2	± 0.9	–	–	
EDSS range	[0–2.5]				
Speed (m/s)	1.2	–	1.2	–	–
Cadence (steps/min)	114.7	± 5.33	112.82	± 5.82	0.29
Step length (cm)	62.87	± 3.01	63.93	± 3.31	0.35
Stride length (cm)	125.75	± 5.89	127.87	± 6.45	0.28
Step width (cm)	9.44	± 3.29	8.27	± 2.89	0.36
Stance%	70.24	± 7.90	73.4	± 8.99	0.21
IDS%	20.39	± 7.94	23.39	± 8.99	0.25
SS%	29.33	± 8.25	26.6	± 8.99	0.34
SDS%	20.52	± 8.04	23.4	± 8.99	0.29
Swing%	29.76	± 7.90	26.6	± 8.99	0.21

2. Methods

2.1. Participants

Thirty people with relapsing-remitting MS (PwMS) were recruited. Inclusion criteria were: a) < 15 years since onset; b)aged > 18 years; c)EDSS < 4.0 (ambulation = 0). Exclusion criteria included: a)Other neurological conditions; b)Cardiovascular disease; c)Orthopaedic conditions. Healthy controls (HC, n = 15) were recruited if they had no neurological or orthopaedic conditions and were > 18 years. All demographics and clinical data are presented in Table 1. This study was approved by the Melbourne Health Ethics Committee (2015.144). All participants provided signed informed consent.

2.2. Setup

Reflective markers were placed on body landmarks following the Plug-in-Gait model. An 8-camera Vicon system (Oxford, UK) was used to collect kinematics at 200 Hz. A wireless inertial measurement unit (IMU) (Cometa, Milano, Italy) collected 3D accelerations (1000 Hz) from a sensor placed at the sacrum. Nexus 2 (Vicon, Oxford, UK) was used to record all data.

2.3. Assessments

Participants walked barefoot on a Biodex-RTM600 (Shirley, NY, USA) treadmill at $1.2 \, ms^{-1}$ without holding the rails for 5 min to obtain at least 150 gait cycles. No participant reported fatigue after walking.

2.4. Data processing

All data were processed using Matlab R2019a (Natwick, MA, USA). The LDEwas calculated for the 3D accelerations of 3 markers: mid-shoulder, average position between shoulder markers; sacrum, average position between posterior superior iliac spine (PSIS) markers and seventh cervical marker (cervical). Acceleration was also obtained from an IMU located between PSISs (sacrum$_a$). Data from 150strides was used for all participants), which were time-normalized to 150 × 100 samples. Reconstructed state-spaces were calculated using 3 embedded dimensions (9 dimensions per state-space) with a time delay of 10 samples. The 0 to 0.5 stride LDE (log(divergence)/stride), or short-term LDE, was calculated from the divergence curve for markers' acceleration and IMU data [12]. The short-term LDE reflects the ability of the motor control system to cope with step-to-step perturbations [11].
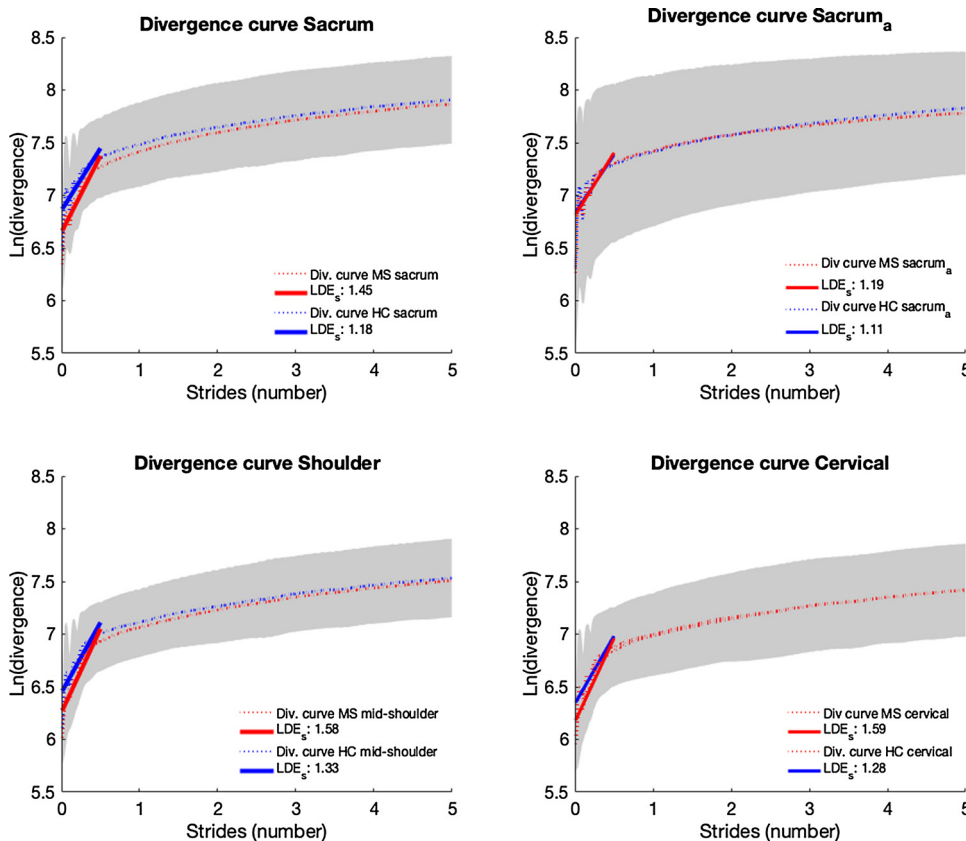
Fig. 1. Averaged divergence curves for people with MS (PwMS; dotted red line) and healthy controls (HC; dotted blue line) for the four acceleration data sources (sacrum, sacrum$_a$, mid-shoulder and cervical). The short-term local divergence exponents (LDEs ; 0-0.5 stride) are also presented for PwMS (continuous red line) and for HC (continuous blue line). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

Table 2
Descriptive statistics for the LDE measures in both groups. LDE: local divergence exponent; sd: standard deviation; CI: confidence interval; sacrum$_a$: sacrum LDE from accelerometer.

	Sacrum			Sacrum$_a$			Shoulder			Cervical			*p*-values		
	mean	sd	95 % CI	mean	sd	95 % CI	mean	sd	95 % CI	mean	sd	95 % CI	disease	location	disease*location
PwMS	1.45	± 0.29	[1.34–1.56]	1.19	± 0.13	[1.14–1.24]	1.59	± 0.29	[1.48–1.69]	1.59	± 0.33	[1.46–1.71]	< 0.01	< 0.01	0.24
Controls	1.18	± 0.24	[1.05–1.31]	1.11	± 0.14	[1.04–1.19]	1.33	± 0.27	[1.18–1.48]	1.29	± 0.30	[1.12–1.45]			

2.5. Statistical analysis

Normality of data was confirmed using Skewness and Kurtosis. An ANOVA with group as between factor (PwMS and HC) and data source (sacrum, mid-shoulder, cervical and sacrum$_a$) as within factor was used to determine disease, sensor location and their interaction effect on LDEs. A post-hoc analysis was used to determine where significant differences occurred (Bonferroni adjustments). A separate ANOVA was used to determine group differences for spatiotemporal measures. For all analyses significance was set $p < 0.05$.

3. Results

No significant between groups difference for any of the spatio-temporal measures was found (Table 1). Fig. 1 shows average divergence curves and LDEs for all data sources and both groups. PwMS walked significantly less stable than HC as shown by the significant main effect of disease ($p < 0.01$) (Table 2). A significant main effect of data source was also found ($p < 0.01$), yet no significant interaction($p = 0.24$). Post-hoc analysis on the LDE obtained from different data sources revealed that LDE calculated from the accelerometer placed on the sacrum was significantly lower than LDE calculated from the other 3 data sources ($p < 0.04$).

4. Discussion

The aim of this study was to compare gait stability (expressed as the LDE) between PwMS with no gait impairments and healthy controls when walking on a treadmill at a fixed speed using different sensor locations and data sources. Overall, the results indicate that PwMS walked less stable than HC and that LDE calculated from accelerometer data may underestimate instability compared to markers-derived LDE. On average, LDE was 18 ± 1.7 % higher in PwMS using markers acceleration yet 7 % when using sacrum accelerometer. Although it it is difficult to pinpoint a cause for this difference, it may be due to accelerometer's sensitivity, positioning, movement artifacts and/or sample rate [10]. This highlights the need for consistency within a study but also the need for a consensus about these issues in future research [10].

Previous studies using accelerometers to analyse PwMS's stability have shown that the LDE: 1) can be used in clinical settings [8,9], 2) is greater in PwMS than healthy controls [9], 3) improves after 3-weeks rehabilitation [8] and is greater in PwMS with falls history [13]. However, these studies assessed a relatively heterogeneous MS populations with EDSS scores ranging from 4.2 [6] to 5.1 [8]. PwMS with EDSS scores > 3.5 already exhibit some degree of walking disability. Also, these studies recorded acceleration over 30 s trials (~30 m walkway) [9] or between 17–60 strides [8,13], which is less than the 150 strides recommended for LDE's statistical precision [11]. In summary, patients homogeneity in terms of disability, and number of strides are of high relevance when trying to determine subtle gait changes in PwMS at early stages when gait deterioration is no evident to clinical examination.

Although treadmill walking is the most adequate alternative for large datasets acquisition, it is also known that it may impose a less challenging condition compared to overground walking [14]. Furthermore, access to a treadmill may be limited in some clinical settings. An alternative to collect large datasets may be using walking acceleration (e.g. at the sacrum as in this study) over several laps in a long corridor from which other clinical measures, e.g.6-minutes and 25-feet walking tests, can also be obtained [15].

Although all data sources showed greater LDE in PwMS than controls, the use of accelerometers is more implementable than motion tracking in clinical settings [6,8,13]. Although LDE responsiveness to the effects of rehabilitation has been shown in PwMS with evident gait impairments [8], its use as a clinical outcome measure is yet to be determined for physiotherapy and pharmacological interventions at MS early stages. Further studies should also explore the underlying mechanisms of gait stability deterioration in PwMS that may help targeting interventions.

5. Conclusion

PwMS with no evident gait impairments are less stable than healthy controls when walking on a treadmill. Different data sources can be used to determine MS-related stability deterioration. TheLDE can be a useful measure of gait disability progression at MS early stages.

Declaration of Competing Interest

None.

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