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The relationship between trunk and foot acceleration variability during walking shows minor changes in persons with multiple sclerosis

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Abstract

Background—Identifying how relationships between variability of upper and lower body segments during walking are altered in persons with multiple sclerosis may uncover specific strategies for maintaining overall stability. The purpose of this study was to examine relationships between trunk and foot acceleration variability during walking in healthy controls and in persons with multiple sclerosis.

Methods—Linear and nonlinear variability measures were calculated for 40 healthy controls and 40 persons with multiple sclerosis from the acceleration time series recorded by inertial sensors attached to the trunk and foot while subjects walked on a treadmill at self-selected preferred pace.

Findings—No main effect of group was found for any variability measures. Main effect of location was found for all variability measures, with larger magnitudes of variability at the foot compared to the trunk, and more predictable variability patterns at the foot compared to the trunk. Differences in strength of correlations between trunk and foot accelerations were found between persons with multiple sclerosis and healthy controls in the frontal and sagittal plane. Sample entropy of accelerations at the feet and at the trunk correlated significantly higher in healthy controls than in persons with multiple sclerosis.

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Interpretation—Relationships between variability of trunk and foot accelerations, which may provide a valuable comprehensive description of whole body stability during gait, showed minor changes in persons with MS compared to healthy controls.

Keywords

Gait; Accelerometers; Inertial sensors; Segmental control

1. Introduction

Multiple sclerosis (MS) is a progressive autoimmune disorder that affects an estimated 400,000 persons in North America (Evans et al., 2013). Gait deficits are reported in 80% of persons with MS (PwMS) (Souza et al., 2010), and fall risk is significantly increased with at least 50% of PwMS reporting at least one fall in a 2–6 month period (Nilsagard et al., 2009). Previous gait analysis studies have identified kinetic and kinematic differences in the gait of PwMS compared to healthy control subjects including reduced walking speed and stride length (Benedetti et al., 1999), decreased ankle and knee angular excursions (Gehlsen et al., 1986), altered trunk sway variability (Huisinga et al., 2013), altered step length and step width variability (Kaipust et al., 2012), and lower frequencies in vertical ground reaction forces (Wurdeman et al., 2011). These altered gait parameters in PwMS are influenced by decreased somatosensation (Cattaneo and Jonsdottir, 2009), altered sensorimotor responses (Huisinga et al., 2014), muscle asymmetry (Chung et al., 2008), and spasticity (Pau et al., 2015), and lead to altered gait stability and increased fall risk (Cameron et al., 2008; Huisinga et al., 2014; Nilsagard et al., 2009; Sosnoff et al., 2011).

Gait stability can be defined as the ability to maintain functional upright gait without falling (Bruijn et al., 2013). Maintaining upright stance requires a controlled interaction between the base of support (BoS) and center of mass (CoM) (Winter, 1995). The relationship between the BoS and CoM must be consistently maintained during walking through sensorimotor feedback loops controlling trunk sway and foot placement from step to step (Hurt et al., 2010; Lugade et al., 2011; Winter, 1995). However, as previous studies have shown, these sensorimotor responses are altered in PwMS (Cameron et al., 2008; Huisinga et al., 2014), which may provoke an altered relationship between the BoS and CoM.

Previous gait studies have used trunk motion to approximate CoM motion (Moe-Nilssen and Helbostad, 2002) and step length and width as indicators of dynamic balance (Maki, 1997). Wireless accelerometers can measure motion at the trunk segment and lower body segments simultaneously while also being highly portable and able to be used outside of a laboratory setting (Kavanagh and Menz, 2008). Wireless accelerometers have been used in previous studies on PwMS to measure accelerations at the trunk during walking (Huisinga et al., 2013; Solomon et al., 2015). However, these accelerometers do not allow for direct measurement of step placement or other position data. Alternatively, measuring the variability of accelerations patterns during gait can provide information about how the walking pattern is controlled in PwMS (Huisinga et al., 2013; Stergiou and Decker, 2011; Stergiou et al., 2006).

Measuring variability during gait can provide information about the sensorimotor control systems responsible for governing movement and balance during gait (Harbourne and Stergiou, 2009). Healthy persons exhibit an optimal gait pattern, and display optimal levels of variability (Stergiou et al., 2006). Compared to healthy controls, PwMS display greater variability in step width and single support time based on coefficient variation (Sosnoff et al., 2012). Linear measures of variability such as coefficient of variation or standard deviation provide information about the magnitude of variability within a time series (O'Connor et al., 2012; Terrier and Reynard, 2015). Nonlinear variability measures provide information on the temporal structure of variability within a time series (Harbourne and Stergiou, 2009; Stergiou and Decker, 2011). Nonlinear measures of variability are particularly helpful in quantifying the control systems governing time series related to movement (Stergiou and Decker, 2011; Stergiou et al., 2006). Nonlinear variability analyses showed that PwMS displayed more predictable and repeatable step length and step width during walking compared to healthy controls (Kaipust et al., 2012) while trunk acceleration displayed greater divergence in PwMS compared to healthy controls (Huisinga et al., 2013). These findings demonstrate that PwMS may be less adaptable to perturbations during gait since their foot placement is more predictable and point to a loss of complexity in their movement pattern (Kaipust et al., 2012; Stergiou and Decker, 2011; Stergiou et al., 2006).

Most studies examining variability during gait have focused on examining movement patterns of only one segment, either the trunk or the feet (Bruijn et al., 2010; Huisinga et al., 2013; Kaipust et al., 2012; Kang and Dingwell, 2009). However, stability requires coordination between upper and lower segments, so examining these segments simultaneously may provide a more comprehensive understanding of stability in a specific population. Since the coordination between segments is likely controlled in part by sensorimotor feedback loops (Hurt et al., 2010), it is possible that these segmental relationships will be altered in PwMS since we know their neural feedback loops are altered compared to healthy controls (Amadio et al., 2006; Cameron et al., 2008; Gagliardo et al., 2007; Huisinga et al., 2014).

The aim of this study was to examine the relationships between foot and trunk acceleration variability during walking in PwMS compared to the same relationships in healthy adults. Because PwMS have altered sensorimotor responses (Cameron et al., 2008; Huisinga et al., 2014) and many differences in spatial and temporal gait parameters compared to healthy controls (Benedetti et al., 1999; Gehlsen et al., 1986; Huisinga et al., 2013; Kaipust et al., 2012; Wurdeman et al., 2011), we hypothesized that (i) PwMS would exhibit different amounts and different structure of variability for foot and trunk accelerations during walking compared to healthy controls, and (ii) the strength of relationships between variability of accelerations at the foot and at the trunk during walking would be different in PwMS compared to healthy controls.

2. Methods

2.1 Participants

Forty PwMS between the ages of 20 and 60, and 40 age-matched healthy controls were recruited for this study (Table 1). Exclusion criteria for both groups were: additional

neurological or orthopedic co-morbidities possessing the potential to alter balance or gait mechanics, female subjects who were pregnant, breastfeeding, or within 3 months post-partum at the time of data collection, persons with vestibular issues, diabetes, or a pre-existing condition which could make exercising difficult (i.e. myocardial infarction, chest pain, unusual shortness of breath, congestive heart failure, etc.), unable to walk a distance of 25 feet without the assistance of a mobility aide, or currently prescribed symptom specific medication therapies (i.e. Fampridine) which can directly affect gait. Additionally, PwMS were excluded if they had a Kurtzke Expanded Disability Status Scale (EDSS) (Kurtzke, 1983) score greater than 5.5. All PwMS enrolled in the study had relapsing-remitting MS.

2.2 Protocol

Acceleration data was recorded by two APDM wireless sensors (Opal, APDM, Portland, OR, USA) secured by elastic strap to the right ankle and trunk. The trunk accelerometer was placed over the center of the sternum. The right ankle accelerometer was placed over anterior surface of the lower shank, on the distal most point of the shank, superior to the ankle joint. While the sensor was not placed directly on the foot, its position on the ankle is sufficient to approximate foot motion throughout the gait cycle, as each subsequent step is placed to maintain the subject's BoS (Bruijn et al., 2012; Savin et al., 2014). Previous studies have shown that accelerations are attenuated inferiorly to superiorly within the trunk segment (Kavanagh et al., 2004); therefore the trunk accelerometer was placed over the sternum rather than the lumbar spine in order to measure accelerations that are maximally dampened by being in the superior portion of the trunk segment. Accelerations were recorded (128Hz) while subjects walked on a motorized treadmill (Woodway Bari-Mill, Eugene, OR, USA) at self-selected comfortable pace for 3 minutes.

The raw acceleration time series were exported to Matlab (MATLAB version R2013b, The MathWorks, Inc., Natick, Massachusetts, USA) and were initially translated from local 3-dimensional Cartesian coordinates to resultant frontal and sagittal plane time series. These resultant frontal and sagittal plane time series were not aligned to the global anatomical planes, but only local to the individual sensors. All subsequent processing took place on the resultant frontal and sagittal acceleration time series. For accurate analysis of the variability and complexity within the time series, data was left unfiltered (Mees and Judd, 1993). Since there were no statistical differences in stride velocity between the groups (Table 1) we did not need to normalize the outcome variables to walking speed. A custom Matlab program was used to calculate all variability measures.

Linear measures root mean square (RMS) and range were calculated from both the frontal and sagittal plane acceleration time series. Root mean square was calculated as the square root of the mean of squares of the numbers in the time series, and was used to quantify the dispersion of the acceleration traces. Range was calculated as the difference between the maximum and minimum acceleration values within the time series.

Nonlinear measures sample entropy (SaEn) and Lyapunov exponents (LyE) were calculated from both the frontal and sagittal plane acceleration time series. A thorough explanation of the SaEn calculation can be found in previous literature (Pincus, 1991; Richman and Moorman, 2000; Yentes et al., 2013). SaEn was calculated using customized Matlab

software based on the methods of Pincus and Richman (Pincus, 1991; Richman and Moorman, 2000), using $m=3$ and $r=0.2 \times (\text{standard deviation of the time series})$. These parameters were selected after testing the algorithms on a selection of data, varying r from 0.1–0.3 in increments of 0.02. Relative consistency of results was apparent around $r=0.2$. The vector length $m=3$ was chosen after finding very inconsistent results using $m=2$ and $m=4$, regardless of r value. The LyE was calculated using customized Matlab software based on Wolf's algorithm (Wolf et al., 1985) which calculates the largest LyE and provides information about the predictability of a system. Larger LyE values relative to optimal variability indicate more divergence in the system and less predictability of the system (Stergiou, 2016). The embedding dimension (edim) was calculated separately for each group and dataset using a Global False Nearest Neighbors analysis (Stergiou, 2016). For controls, edim was chosen as 11 for sagittal trunk, 11 for frontal trunk, 8 for sagittal foot, and 8 for frontal foot based on examination of a range of constants from each group. For PwMS, edim was chosen as 9 for sagittal trunk, 10 for frontal trunk, 10 for sagittal foot, and 8 for frontal foot. The time lag (τ) was calculated for every time series based on the Average Mutual Information Algorithm (Baker and Gollub, 1996; Gomez Garcia et al., 2012; Stergiou, 2016). Time lags ranged from 5 to 30 samples, with an average time lag of 12 (SD 5) samples. SaEn and LyE calculations used each time series' specific time lag in order to appropriately capture the characteristics of each individual data set. In total, our processing resulted in four values per variability measure for each subject – trunk sagittal, trunk frontal, foot sagittal, and foot frontal.

2.3 Statistical Analysis

The assumption of data normality was accepted based on the results of Shapiro-Wilks tests. A two-way ANOVA was performed to investigate effects of Group (HC, PwMS) and Location (foot, trunk) on the results. Paired t-tests were used to analyze any significant interactions. Pearson's correlations were used to assess the relationship between sensor locations within the same plane. The strength of relationship was directly compared between groups, with correlation coefficient of 0.3–0.5 indicating a low correlation, 0.5–0.7 indicating a moderate correlation, and greater than 0.7 indicating a strong correlation (Hinkle et al., 2003). To determine if correlations were significantly different between groups, a two-tailed test using Fischer's r to z transformation was used. A significance level of 0.05 was used for all analyses, and all analyses were completed in SPSS 2013 (version 22, IBM Corp., Armonk, NY).

3. Results

3.1 ANOVA and Paired Tests

Both groups displayed larger magnitudes of variability (RMS, Range) at the feet compared to the trunk in the sagittal and frontal planes. Both groups displayed more regular accelerations (SaEn) at the trunk compared to the foot in both the sagittal and frontal planes. Both groups displayed more divergent accelerations (LyE) at the foot compared to the trunk in both the sagittal and frontal planes, and PwMS displayed more divergent accelerations at the foot and less divergent accelerations at the trunk compared to HC (Fig. 1).

3.2 Correlations

All segment correlation results are listed in Table 2. Healthy controls showed significant positive correlations between trunk and foot acceleration variability for all variables except sagittal plane Range and LyE. Of these significant correlations in HC, correlations were strong for range in the frontal plane and RMS in the sagittal plane, moderate for RMS and SaEn in the frontal plane, moderate for SaEn in the sagittal plane, and were low for LyE in the frontal plane. PwMS showed significant positive correlations between trunk and foot acceleration variability for all variables except frontal plane SaEn and sagittal plane SaEn. Of these significant correlations in PwMS, correlations were strong for RMS in the frontal plane and RMS in the sagittal plane, low for Range and LyE in the frontal plane, and low for range and LyE in the sagittal plane. Comparisons between groups using Fischer's r to z transformation showed that the SaEn correlations in the frontal and sagittal plane were significantly different between groups.

4. Discussion

The goal of this study was to determine if differences exist in the relationships between foot and trunk acceleration variability during walking in PwMS and healthy adults. Acceleration data was collected from wireless sensors and analyzed using linear and nonlinear variability measures to quantify the amount and the structure of variability within the acceleration time series. We hypothesized i) that PwMS would exhibit different amounts and different structure of variability for foot and trunk accelerations during walking compared to healthy controls and ii) the strength of relationships between variability of accelerations at the foot and at the trunk during walking would be different in PwMS compared to healthy controls. Our results partially agree with the first hypothesis and fully agree with the second hypothesis.

The magnitude of acceleration variability was higher at the foot compared to the trunk in both HC and PwMS and accelerations at the feet were more regular but less dynamically stable compared to accelerations at the trunk in both HC and PwMS. In the frontal and sagittal plane, PwMS and HC showed greater magnitudes of variability at the foot compared to the trunk which agrees with previous findings that accelerations are attenuated from inferior to superior segments of the body during walking (Kavanagh et al., 2004). Measures of entropy also showed no differences between PwMS and HC in the sagittal and frontal planes and both groups showed significantly lower values of entropy at the foot than at the trunk in the sagittal and frontal plane. These results indicate that accelerations at the feet are more regular than accelerations at the trunk in HC and PwMS in agreement with previous work (Craig et al., 2016). Other studies have found that PwMS display altered variability of kinematics at the trunk and at the lower extremities during walking and quiet standing compared to HC (Cattaneo et al., 2015; Huisinga et al., 2013; Kaipust et al., 2012). It is possible that similar differences were not identified in the current study since the PwMS were mildly affected (EDSS: 1.63 (SD 0.7)) and had a similar preferred gait speed as HC. Progression of the disease may eventually alter the ability to attenuate accelerations from inferior to superior segments in PwMS as previous studies have shown that aging populations are less capable of attenuating accelerations from lower to upper segments

during walking (Mazza et al., 2008). Future studies to examine segmental coordination would benefit from including PwMS with moderate to severe disability.

Both groups showed larger values of LyE at the foot compared to the trunk in the frontal and sagittal plane indicating less divergence or more predictable accelerations at the trunk compared to the feet in both groups. In the sagittal plane, LyE of accelerations at the feet was higher in HC compared to PwMS and LyE of accelerations at the trunk was lower in HC compared to PwMS which indicates that PwMS display greater predictability of foot accelerations in the sagittal plane compared to HC. It is possible that PwMS develop a more tightly controlled pattern of foot placement to reduce variability since their delayed sensorimotor responses may cause a slower response to any perturbations encountered during walking (Huisinga et al., 2014). In the sagittal plane, healthy controls exhibited significantly lower LyE at the trunk compared to PwMS, indicating that accelerations at the trunk are less dynamically stable for PwMS compared to HC. This finding is in agreement with previous studies which have shown that aging populations (Kang and Dingwell, 2009), populations with a known neuromuscular disorder (Hoogkamer et al., 2015; Huisinga et al., 2013), populations of fall-prone individuals (Lockhart and Liu, 2008), and PwMS all display higher LyE at the trunk compared to healthy control subjects. The present study showed that foot acceleration was less divergent in PwMS which may be a compensation for the more divergent acceleration pattern of the trunk. The combination of altered trunk and foot acceleration patterns may be indicative of fall risk in PwMS since both their CoM and BoS are altered during gait.

We hypothesized that the strength of relationships between variability of accelerations at the foot and at the trunk during walking would be different in PwMS compared to healthy controls and this was shown in our study. Correlations of acceleration variability between the feet and the trunk were similar in strength for both groups for all variables except SaEn. SaEn of accelerations at the feet and at the trunk correlated significantly higher in HC than in PwMS. This difference between the two groups indicates that more regular patterns of acceleration at the feet are related to more regular patterns of acceleration at the trunk in HC, but not in PwMS. The altered relationship between the trunk and feet in PwMS is likely reflective of disease-specific symptoms that contributed to gait deficits. PwMS experience muscle weakness (Hoang et al., 2014), strength asymmetries (Rudroff et al., 2014), and altered activation of trunk musculature (Ketelhut et al., 2015). This altered muscle functioning in PwMS, for example, may lead to an inability of the trunk to absorb accelerations via passive mechanisms, which typically attenuate higher frequency accelerations (Kavanagh et al., 2005). This change in trunk motion would affect trunk acceleration and may be represented as altered correlation between segment variability measures. These findings demonstrate the importance in studying upper and lower body motion simultaneously in order to better understand how relationships between segments contribute to walking.

During walking, motion of the frontal plane is governed by active feedback mechanisms in order to maintain proper step width to arrest and redirect the body's center of mass from step to step (Bauby and Kuo, 2000; Hurt et al., 2010; O'Connor and Kuo, 2009). Previous studies have shown that PwMS have slowed sensorimotor responses (Cameron et al., 2008;

Huisinga et al., 2014), which would directly impact these active control mechanisms in the frontal plane. In contrast to motion in the frontal plane during walking, motion in the sagittal plane is controlled by relatively passive mechanisms in healthy adults, similar to pendula motion (O'Connor and Kuo, 2009). However, this passive control may be altered in PwMS, as step length is actively constrained to increase their margin of stability from step to step (Peebles et al., 2016). As foot motion is altered, trunk motion must also be altered to compensate, leading to the observed differences in relationships between trunk and foot motion variability. In addition to previous work demonstrating relationships between trunk and foot motion, the results of the current study show that the relationship between movement variability of these segments is altered in PwMS compared to healthy adults.

A limitation of this study is that the data was collected on a motorized treadmill, which held the subjects' gait at a constant speed and confined them to walk within the boundaries of the treadmill belt. While the treadmill could have eliminated some variability in the subjects' gait, the use of the method was necessary to record a time series of sufficient length and the use of treadmill gait to assess variability of movement patterns is well established (Kaipust et al., 2012; Owings and Grabiner, 2004; Rosenblatt and Grabiner, 2010). Previous studies have also supported the validity of gait analysis using a treadmill, showing that treadmill gait is similar to over ground walking (Riley et al., 2007). Future studies should aim to expand this research to populations with other neuromuscular pathologies, which could further identify how coordination between foot and trunk motion during walking is affected by altered dynamic systems and to examine the effect of distorted visual or sensory feedback on the relationship between foot and trunk accelerations.

The current study revealed differences between upper and lower segment control during walking in HC and PwMS. Magnitude and structure of acceleration variability demonstrated differences between trunk and foot motion in both HC and PwMS, but there were few differences observed between groups. However, the relationships between trunk and foot accelerations revealed differences between HC and PwMS. These findings demonstrate the importance in studying how the relationships between upper and lower body motion contribute to walking. These differences highlight segmental control features that may contribute to increased fall occurrences in PwMS, which will be of interest in future development of fall risk assessments for clinical and home use.

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Highlights

- Variability magnitude at foot is greater than at trunk in both groups.
- Variability structure at foot is more predictable than at the trunk in both groups.
- Correlations of foot and trunk variability are altered in multiple sclerosis group.

Figure 1.

Acceleration variability results for healthy controls and persons with multiple sclerosis at the trunk and at the foot for the frontal and sagittal planes.

*Significant main effect for Location ($p < 0.05$)

†Significant Location x Group interaction, post-hoc t-test ($p < 0.05$)

Table 1

Summary statistics of subject demographics; mean (standard deviation).

	Healthy Controls (n = 40)	PwMS (n = 40)	p-value
Age (years)	44 (10)	44 (9)	0.962
Walking Speed (m/s)	1.01 (0.34)	0.88 (0.23)	0.065
EDSS	--	1.63 (0.7)	--

Correlations to measure relationship between trunk and foot acceleration variability in frontal and sagittal planes in healthy control subjects (HC) and persons with MS (PwMS), and Fischer's r to z transformation two-tailed comparison between groups; *Significant correlation, * $p < 0.05$.

Table 2

		HC <i>R-value (P-value)</i>	PwMS <i>R-value (P-value)</i>	Fischer's Transformation Z-statistic (<i>P-value</i>)
Frontal	Linear	RMS	0.565 (< 0.01) *	0.732 (< 0.01) *
		Range	0.711 (< 0.01) *	0.485 (< 0.01) *
	Nonlinear	SaEn	0.541 (< 0.01) *	0.054 (0.740)
		LyE	0.334 (0.046) *	0.396 (0.020) *
	Linear	RMS	0.810 (< 0.01) *	0.791 (< 0.01) *
		Range	0.274 (0.111)	0.372 (0.024) *
Sagittal	Nonlinear	SaEn	0.510 (< 0.01) *	2.84 (< 0.01) *
		LyE	0.259 (0.145)	-0.99 (0.322)