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The differences between overground and treadmill walking in nonlinear, entropy-based and frequency variables derived from accelerometers in young and older women – preliminary report

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Purpose: The aim of this study was to compare gait stability and variability between walking conditions and age groups. *Methods:* Twenty-six healthy younger and older females participated. Trunk acceleration in the vertical (V), medial-lateral (ML) and anterior-posterior (AP) directions during 5 minutes walking overground and 3 minutes walking on the treadmill at self-selected speed were recorded. Root mean square and standard deviations of acceleration, stride time and its variability, Lyapunov exponents (LE), multiscale entropy (MSE) and harmonic ratios (HR) were computed. *Results:* Both age groups showed significantly higher stride time variability and short-term LE in all directions during overground walking. For the older group, overground walking showed higher V and AP standard deviation. Significantly lower values for overground walking were observed for long-term LE (V and ML for the younger group, ML for the older group), HR (ML for the older group) and MSE (V for the older group). Significant age-related differences were found for V long-term LE for overground walking. *Conclusions:* The present findings suggest that both linear and advanced computational techniques for gait stability and variability assessment in older adults are sensitive to walking conditions.

Key words: *ageing, gait, stability, variability, local dynamic stability*

1. Introduction

Stability and variability of gait can be assessed using various methods that include both linear and nonlinear characteristics. It is presumed that linear characteristics seem not to include every aspect of a global complex system or movement, and hence linear parameters may not be capable of describing human gait precisely [7], [13], [28]. As a response to this problem, more sophisticated approaches have been recently implemented from theoretical mechanics and mathematics to gait analysis. In recent years, indeed nonlinear, entropy-based and frequency analyses have been successfully used to quantify stability and variability of the gait [2], [3], [7], [25], [27]. Compared to traditional variables, these approaches provide the opportunity to study inner structure, regularity, complexity and stability of the system represented by a recorded time series in a more direct way.

First, the age-related differences in gait performance have been shown in literature [2], [3], [27]. Buzzi et al. [3] found significantly higher local dynamic instability in a group of the elderly, compared to a group of young adults during treadmill walking. Terrier and Reynard [27] found increase of local dy-

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namic instability from age 40–50 and over during treadmill walking, which was significantly present in medial-lateral direction of upper trunk acceleration. Bisi et al. [2] compared toddlers, young adults and elderly using nonlinear (local and orbital dynamic stability, recurrence quantification analysis), entropy-based (multiscale entropy), frequency (harmonic ratios) variables and traditional variables derived from trunk accelerations during overground walking. Their results showed the best distinguishing ability between groups with harmonic ratios and variables derived from recurrence quantification analysis. The trend present in their work showed increased stability of the gait from toddlers to young adults, with elderly in between these two groups. Taken all together, the variables describing stability of the gait seem to provide relevant information about age-related changes in gait performance. However, there seem to be an uncertainty about the influence of conditions in which the data were collected.

Because many gait cycles are often necessary to calculate most of these variables (ranging from 10 strides for multiscale entropy to 150 strides long-term Lyapunov exponents) [24], it is common to use treadmill walking to assess gait. However, previous studies have reported that gait performance during overground and treadmill conditions could differ. First, studies concerning kinematics and kinetics of treadmill walking (TW) have reported higher hip flexion [1], [22] and a lower second peak in vertical ground reaction forces [22], compared to overground walking (OgW). These differences could suggest the difference in importance of push-off during both conditions. During TW, the active push-off is primarily not needed, but during OgW, the active push-off and the associated changes in the vertical ground reaction force are essential for forward movement. Second, differences in gait dynamics have been a concern of several studies. In a group of ten healthy, young subjects, Dingwell et al. [7] have compared TW and OgW by examining the variability of kinematic and temporal data, and local dynamic stability indicated by short- and long-term Lyapunov exponents (LE). Both linear (stride time variability and kinematic variability, indicated by standard deviation) and nonlinear (LE) measures showed significant differences between walking conditions. Terrier and Deriaz [28] have also compared the acceleration variability of TW and OgW in twenty healthy, young subjects by using standard deviation and LE. However, they found that the conditions showed no differences in the acceleration variability. During OgW, there were significantly higher short- and long-term LE values, which indicated increased local dynamic stability during TW. Despite the fact that a great deal of attention has been paid to frequency analysis end entropy-based variables for stability assessment recently, the differences of these variables between walking conditions have not been documented yet.

In recent years, inertial measurement units have been widely used for gait assessment [2], [24], [25], [27]. Compared to the optoelectronic systems more often used for gait assessment, inertial sensors are less expensive, easy to use and portable devices which can be employed in various measuring conditions and environment. As it seems, these sensors have a great potential for gait assessment [15] and also provide the opportunity to easily obtain time series which can be analyzed by sophisticated methodologies.

To our knowledge, no studies have compared harmonic ratio and entropy-based measures during different walking conditions and among different age groups. Hence, the aim of this study is to compare overground and treadmill walking trials in two different age groups by assessing local stability, multiscale entropy, harmonic ratios and linear variability measures of a trunk accelerometric signal. Even though the other indexes have been reported in literature, the aim of the present study is to have a complex insight to the movement patterns, therefore, all of these indexes will be evaluated and compared. We hypothesize that gait patterns, indicated by variability and stability measures, will be different between walking conditions and groups. Based on the results of similar studies, the five working hypotheses of the present study are as follows: 1) the local dynamic stability indicated by LE would be higher in TW; 2) traditional variability measures would show higher variability during OgW; 3) regularity and periodicity would be higher in TW indicating more stable walking pattern; 4) complexity would be higher during OgW; 5) the age-related changes would reveal more stable patterns in younger group of subjects.

2. Methods

Subjects: Twenty-six healthy females voluntarily participated in this study. Participants were divided into two groups: younger adults (n = 13, age 21.8 ± 0.9 years, height 1.70 ± 0.07 m, body mass 63 ± 9 kg) and older adults (n = 13, age 57.5 ± 4.8 years, height 1.64 ± 0.06 m, body mass 65 ± 12 kg). The participants did not have any musculoskeletal or neurological problem that could influence their balance abilities or gait patterns. The study was approved by the institutional ethics committee and all participants signed informed consent before the measurement.

Experimental protocol: Participants performed two successive gait sessions. The first session was recorded during 5 minutes of OgW in comfortable sport shoes within a 30 m long corridor. Two well-visible marks were placed on the floor, demarcating a 23 m long pathway. The participants were instructed to walk straight, maintain a preferred walking speed between those marks, and turn around immediately after crossing a mark. During this walking trial, walking speed of each participant was evaluated using two photocell gates (Fitronic, Bratislava, Slovakia) placed in the middle of the walkway, placed 1.5 m apart. Participants walked through the photocell gates 15–20 times depending on their walking speed. Preferred walking speed was estimated as an average speed recorded during the whole trial. The second session was performed during 3 minutes of walking on the treadmill (LODE Valiant, Lode, B. V. Medical Technology, Groningen, Netherlands). The 2-minute difference in the duration of the sessions was caused by the consideration of the turnaround time during the OgW trials. The participants were given a sufficient time to familiarize themselves with the treadmill and then 3 minutes to walk at the speed estimated during the OgW trial without the use of handrails. The participants wore the same shoes during both trials, and the sessions were measured one week apart. A 3D accelerometer (Trigno wireless system, Delsys Inc., Natick, MA, USA) with a sampling rate of 296.3 Hz was placed at the level of L5 vertebra and measured in the medial-lateral (ML), anterior-posterior (AP) and vertical (V) directions. The accelerometer was attached directly to the skin using a double-sided tape.

Data analysis: The first 300 samples of accelerometric signal were excluded from analysis to avoid the influence of nonstationarity [27]. The raw signal was filtered using a 2nd-order low-pass Butterworth filter with a 50 Hz cut-off frequency. In the OgW trial data, turnarounds, the last stride before a turnaround and the first stride after a turnaround were removed prior to analysis. From each trial, 140 strides were extracted and used for analysis, since that was the maximum number of strides obtainable for all of the participants and conditions. The heel strikes were identified from the AP accelerometric signal using the procedure proposed by Zijlstra and Hof [30]. For each direction, the stride time, standard deviation and coefficient of variation of stride time, root mean square and standard deviation of acceleration, short- and long-term LE, multiscale entropy (MSE) and harmonic ratios were computed as the stability and variability measures.

The stride time was computed after isolating the gait cycles as explained above. The standard deviation and coefficient of variation of stride time were computed as measures of linear variability. Root mean square was computed in each accelerometric direction for the whole walking trial, according to Menz et al. [19]. Standard deviation was computed in each of the 140 strides and then averaged to obtain one representative value. The parameters were computed from the filtered signal in Matlab (R2014a, MathWorks, Inc., Natick, MA, USA).

MSE was introduced by Costa et al. [4], [6] to assess the complexity of a system. It is based on sample entropy and uses several scales for computation. First, the coarse-grained time series is constructed by averaging an increasing number of data points in non-overlapping windows [6]. Sample entropy is then computed for each of the coarse-grained time series. The number of data points in each window is defined by a scale factor. Sample entropy indicates the similarity of consecutive data points. The computation depends on the length of consecutive data points and the similarity criterion, which is a measure of distance [6]. In the present study, MSE was computed from the filtered accelerometric signals for scales 1 to 6, as proposed by Costa et al. [6]. Entropies were computed by software available on Physionet [4], [5], [12] with a number of consecutive data points m set to 2 and a radius of 0.15 [6].

Harmonic ratios were computed from the filtered signals, after decomposition, using fast Fourier transform to the frequency domain. The harmonic ratios for the AP and V directions were computed by dividing the sum of the amplitudes of the first ten even harmonics, by the sum of the amplitudes of the first ten odd harmonics. The harmonic ratio for the ML direction was computed as the inverse ratio. The harmonic ratios were computed using custom-written Matlab scripts.

LE represents local dynamic stability [8]. They quantify the ability of the system to respond to small, local perturbations [8] and denote the mean exponential rate of divergence among initially neighboring points in the state space [28]. LE is computed in practice from the slope of a linear fit to the average logarithmic divergence plot. In the present study, the filtered accelerometric signal was normalized to 14,000 points to obtain approximately 100 data points per stride. The time delay was assessed by the first minimum of the average mutual information function. There were delays of 10 samples in the V direction, 7 samples in the

ML direction and 9 samples in the AP direction. An embedding dimension of 6 was used as computed by global false nearest neighbor analysis, and according to the existing literature. An algorithm proposed by Rosenstein et al. [26] was used to compute the short-term (over one step) and long-term LE (over the fourth to tenth stride). The computations were performed by a custom-written Matlab algorithm.

Statistical analysis: A Kolmogorov–Smirnov test was used to verify the normality of the computed variables. The data were normally distributed in all cases. A two-way repeated measure analysis of variance with Bonferroni *post-hoc* test was used to determine differences between walking conditions and groups. The level of significance was set to $p = .05$. Statistical analysis was performed in Statistica (version 12, StatSoft, Inc., Tulsa, OK, USA).

3. Results

The results are shown in Tables 1–3. There was no significant difference in walking speed between groups (younger: 4.95 ± 0.57 $km \cdot h^{-1}$, older: 5.14 ± 0.39 $km \cdot h^{-1}$).

Age-related differences were found only for the long-term LE in the V direction during OgW ($p = .021$), with higher values obtained for the older group.

In the younger group, we found significant differences between TW and OgW for short-term LE in all directions (V: $p = .018$, ML: $p = .016$, AP: $p = .001$), the values in the OgW trials being higher. Compared to TW, the long-term LE for OgW were significantly lower in the V ($p = .001$) and ML ($p < .001$) direc-

Table 1. Results of linear measures stated as mean (standard deviation)

Variable	Direction	Younger (n = 13)		Older (n = 13)	
		Treadmill	Overground	Treadmill	Overground
Stride time [s]		1.07 (0.08)	1.07 (0.09)	0.99 (0.08)	1.00 (0.08)
SD stride time [s]		0.021 (0.013)	0.030 (0.010)*	0.018 (0.015)	0.031 (0.012)*
CV stride time [%]		1.94 (1.04)	2.81 (0.72)*	1.77 (1.26)	3.01 (0.98)*
SD [g]	V	0.22 (0.06)	0.24 (0.06)	0.23 (0.04)	0.27 (0.04)*
	ML	0.16 (0.04)	0.16 (0.04)	0.18 (0.04)	0.18 (0.04)
	AP	0.19 (0.03)	0.20 (0.04)	0.19 (0.02)	0.21 (0.02)*
RMS [g]	V	0.86 (0.06)	0.87 (0.05)	0.86 (0.07)	0.90 (0.06)
	ML	0.19 (0.04)	0.18 (0.04)	0.22 (0.04)	0.22 (0.05)
	AP	0.70 (0.09)	0.70 (0.08)	0.64 (0.17)	0.59 (0.15)

n – number of participants included in group, SD – standard deviation, CV – coefficient of variation, V – vertical, ML – medial-lateral, AP – anterior-posterior.

* $p < .05$ for effect of conditions in groups – younger treadmill vs. overground and elder treadmill vs. overground.

Table 2. Harmonic ratios and Lyapunov exponents stated as mean (standard deviation)

Variable	Direction	Younger (n = 13)		Older (n = 13)	
		Treadmill	Overground	Treadmill	Overground
HR	V	6.2 (2.0)	5.3 (1.8)	4.9 (1.3)	4.6 (1.7)
	ML	3.3 (0.8)	2.7 (0.7)	3.7 (1.0)	2.7 (0.9)*
	AP	7.4 (2.4)	6.2 (2.2)	5.9 (1.3)	4.6 (1.5)
stLE	V	0.58 (0.13)	0.73 (0.17)*	0.60 (0.16)	0.76 (0.18)*
	ML	0.76 (0.17)	0.94 (0.23)*	0.89 (0.23)	1.06 (0.26)*
	AP	0.60 (0.15)	0.79 (0.17)*	0.75 (0.19)	0.93 (0.15)*
ltLE	V	0.040 (0.009)	0.024 (0.010)*	0.037 (0.011)	0.037 (0.012)$
	ML	0.023 (0.004)	0.009 (0.004)*	0.028 (0.010)	0.014 (0.006)*
	AP	0.046 (0.012)	0.037 (0.012)	0.044 (0.008)	0.033 (0.014)

n – number of participants included in group, HR – harmonic ratio, stLE – short-term Lyapunov exponent, ltLE – long-term Lyapunov exponent, V – vertical, ML – medial-lateral, AP – anterior--posterior.

* $p < .05$ for effect of conditions in groups – younger treadmill vs. overground and older treadmill vs. overground.

$ $p < .05$ for effect of age during overground walking – overground younger vs. older.

Table 3. Multiscale entropy results stated as mean (standard deviation)

Variable	Direction	Younger (n = 13)		Older (n = 13)	
		Treadmill	Overground	Treadmill	Overground
MSE1	V	0.38 (0.07)	0.40 (0.10)	0.43 (0.10)	0.35 (0.06)*
	ML	0.50 (0.07)	0.47 (0.09)	0.54 (0.10)	0.54 (0.10)
	AP	0.28 (0.04)	0.27 (0.05)	0.28 (0.09)	0.26 (0.06)
MSE2	V	0.55 (0.09)	0.58 (0.14)	0.66 (0.17)	0.55 (0.10)*
	ML	0.76 (0.13)	0.71 (0.14)	0.82 (0.18)	0.83 (0.19)
	AP	0.41 (0.06)	0.40 (0.08)	0.45 (0.13)	0.41 (0.09)
MSE3	V	0.67 (0.14)	0.73 (0.19)	0.83 (0.21)	0.69 (0.13)*
	ML	0.97 (0.17)	0.90 (0.21)	1.04 (0.23)	1.04 (0.24)
	AP	0.52 (0.08)	0.49 (0.10)	0.57 (0.17)	0.51 (0.12)
MSE4	V	0.79 (0.18)	0.86 (0.23)	0.96 (0.23)	0.79 (0.15)*
	ML	1.15 (0.19)	1.06 (0.25)	1.19 (0.25)	1.19 (0.25)
	AP	0.60 (0.11)	0.57 (0.12)	0.66 (0.19)	0.58 (0.13)
MSE5	V	0.89 (0.22)	0.96 (0.26)	1.05 (0.23)	0.86 (0.16)*
	ML	1.30 (0.20)	1.20 (0.28)	1.30 (0.25)	1.30 (0.25)
	AP	0.67 (0.14)	0.63 (0.14)	0.71 (0.20)	0.62 (0.14)
MSE6	V	0.99 (0.24)	1.05 (0.27)	1.12 (0.22)	0.93 (0.16)*
	ML	1.43 (0.21)	1.32 (0.29)	1.38 (0.23)	1.38 (0.23)
	AP	0.72 (0.17)	0.68 (0.15)	0.76 (0.21)	0.66 (0.15)

n – number of participants included in group, MSE1–6 – multiscale entropy for scales 1 to 6, V – vertical, ML – medial-lateral, AP – anterior-posterior.

* $p < .05$ for effect of conditions in groups – younger treadmill vs. overground and older treadmill vs. overground.

tions. The standard deviation and coefficient of variation of stride time showed significantly higher values for OgW (standard deviation: $p = .003$, coefficient of variation: $p = .004$), compared to TW.

When comparing walking conditions in the older group, significant differences were found for all of the short-term LE (V: $p = .020$, ML: $p = .014$, AP: $p = .001$), the values obtained for the OgW trials being higher than the TW trials. The opposite situation was found for long-term LE in the ML direction such that the OgW trials were lower than the TW trials ($p < .001$). In the older group, OgW showed significantly lower harmonic ratio in the ML ($p = .031$) compared to TW. The MSE in the V direction showed lower values during OgW for all scales ($p = .008$–$.028$). The standard deviation in the V and AP directions was higher during OgW (V: $p < .001$, AP: $p = .006$) than in TW. The standard deviation and coefficient of variation of stride time showed significantly higher values for OgW (both $p < .001$).

4. Discussion

TW is a walking condition that is frequently used during clinical sessions among patients with neurological problems to increase gait symmetry, step length, step width, rhythmicity and posture [9], [11], [14]. With the increasing trend to use more advanced computational techniques for data analysis, which usually require a relatively high number of gait cycles, treadmills have become more often used in research area. Unlike OgW, TW allows for stable gait speed and the opportunity to record a high number of gait cycles in a small laboratory room. However, gait performance during TW is less natural, because of a fear of falling and a fear of the continuously moving belt underneath a patient's feet. The purpose of this study was to compare gait stability and variability during OgW and TW among two different age groups. We formed five hypotheses expecting TW and younger adults to show more stable and less variable walking pattern. The hypotheses were supported partially.

Previous studies involving OgW and TW researched mostly spatial, temporal, kinetic, kinematic and EMG data [1], [17], [20], [22], [23], [29]. For instance, Riley et al. [23] assessed spatial-temporal, kinematic and kinetic variables in a group of young, healthy adults. Although they found significant differences in most of the variables when comparing OgW and TW, the absolute differences were small enough to be considered natural variability for kinematic assessment (differences less than 2°) and for all of the moments and powers from kinetic analysis except for peek knee extension moment (differences smaller than values of coefficients of repeatability evaluated by authors). The authors concluded that the gait in both of the walking conditions are quantitatively and qualitatively similar. Other authors also confirmed these results for elderly adults. Lee and Hidler [17] studied young and older adults and reported similar results for kinematic and temporal gait parameters. On the other hand, their results for joint moment, joint powers and muscle activity suggest that motor control is different between walking conditions. The results of these studies seem to be inconsistent. Some authors claim that OgW and TW are similar in terms of kinematic analysis [22], [23], while the other reported distinct differences between walking conditions [17], [20], [29].

Advanced computational techniques have been implemented to assess gait patterns during both walking conditions. In a group of young, healthy adults, Dingwell et al. [7] assessed spatial-temporal characteristics, their variability and local dynamic stability with data obtained by kinematic analysis of lower limb movement and trunk acceleration. They found significantly lower short-term LE for both the trunk accelerations and lower limb kinematics during the TW trial compared to the OgW trial. However, the results for long-term LE did not show any significant differences in trunk acceleration, although for the lower limb kinematics, the long-term LE during TW was lower than OgW, similar to short-term LE. Our results for LE also showed lower stLE for TW in both age groups, which can be explained by the compulsory regular movement and the need to respond immediately to the treadmill belt to successfully walk. On the contrary, we observed higher long-term LE during the TW trial in both age groups. Our results could be expected due to the visual imagination of the movement during OgW. As Terrier and Deriaz [28] showed in their study, the differences between walking conditions could be induced by different proprioceptive and visual information. During OgW, a subject usually knows where he is going, and the aim of that movement is not the walking itself. One is not primarily interested in small perturbations it is a natural movement, and thus a type of automatic subcortical movement. In contrast, small perturbations during TW could cause a disturbance because it is a less natural, learned cortical process.

In the older group, harmonic ratio in the ML direction was significantly higher during TW, compared to OgW. This result suggests that there is better harmonicity and periodicity during TW in the ML direction. In this group, higher TW MSE in the V direction for all scales has also been found. Higher values of MSE imply more complex movement. Kang et al. [16] showed that lower complexity implies frailty in an elderly population, which is connected to higher fall risk and, therefore, instability. Our results could suggest better stability during TW in the V direction.

The linear measures evaluated in the present study were also able to distinguish between walking conditions. In both groups, we found significantly higher stride time variability for OgW, compared to TW. These results agree with those of Dingwell et al. [7], who observed decreased stride time variability during TW trials. Also, our results for the standard deviation of acceleration were similar to those reported in their study. They found significantly higher standard deviation for the OgW trial in the AP direction, with similar trends in other directions. Taken together, these results suggest that advanced computational techniques are not the only ones that can be used to differentiate between walking conditions. However, they provide a different insight into locomotor control.

We found age-related differences during the OgW trials only for the long-term LE in the V direction. A possible reason for this result could be the physical condition of our older participants, who were fit and active despite their age. A similar study by Lee and Hidler [17] intended to assess differences connected to age, however, they did not detect any significant differences. Their study assessed kinematic, kinetic and EMG characteristics, and therefore, our data are not comparable. On the other hand, it is possible that when older participants are healthy and fit, differences in gait performance, compared to a younger group, vanish in treadmill walking which do not provide such an open variability and possibilities for movement compared to OgW. As for variables used in the present study, results of previous works also support the relationship between age, stability and variability measures [27]. Our results, however, were not in agreement with results of their study. As mentioned above, our results could have been affected by the physical condition of our participants.

There are several limitations of this study. One of the main complications when computing linear and nonlinear characteristics is the use of these methods for non-continuous walking intervals. It was proven that the local stability could be computed from one long trial or multiple shorter walking episodes aligned one after another. This confirmation was not available for other approaches used in this study, i.e., harmonic ratios and entropy. Moreover, filtering plays a very important role in the computational process. Several authors have claimed that linear filtering before nonlinear analyses is undesirable [8], [21]. In spite of that, other authors applied low-pass filtering before computation [10]. We believe that frequencies higher than 50 Hz do not need to be considered in the investigated time series when studying gait. Another limitation could be the choice of walking speed. Preferred walking speeds during treadmill and overground walking could differ [18]. To ensure the influence of the walking speed on computed variables was minimal, we decided to use the same speed – the preferred walking speed during the overground walking trial – for both conditions. However, it remains a possibility that gait performance is slightly altered during treadmill walking. Lastly, the study was conducted on a relatively small group of participants (n = 26). Further investigation with larger groups of various ages is then needed to generalize the presented results.

5. Conclusions

This study compared the gait stability and variability of trunk accelerations during overground and treadmill walking in two age groups. According to the results of this study, only the Lyapunov exponents were sensitive to the change of walking conditions in younger participants. In the older group, Lyapunov exponents, harmonic ratio in the medial-lateral direction, standard deviation in vertical and anterior-posterior directions and multiscale entropy in vertical direction were distinguishable between walking conditions. We found age-related changes in gait performance only for the long-term Lyapunov exponents in the vertical direction during the overground walking trial. It can be assumed that both linear and advanced computational techniques for gait stability and variability assessment in the older population are sensitive to walking conditions. Researchers should take these differences into account when interpreting their results because as it seems, the change of the experimental conditions induces changes in stability and variability of the gait performance.

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