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Effect of age on the variability and stability of gait: A cross-sectional treadmill study in healthy individuals between 20 and 69 years of age

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ABSTRACT

Falls during walking are a major health issue in the elderly population. Older individuals are usually more cautious, walk more slowly, take shorter steps, and exhibit increased step-to-step variability. They often have impaired dynamic balance, which explains their increased falling risk. Those locomotor characteristics might be the result of the neurological/musculoskeletal degenerative processes typical of advanced age or of a decline that began earlier in life. In order to help determine between the two possibilities, we analyzed the relationship between age and gait features among 100 individuals aged 20–69. Trunk acceleration was measured during a 5-min treadmill session using a 3D accelerometer. The following dependent variables were assessed: preferred walking speed, walk ratio (step length normalized by step frequency), gait instability (local dynamic stability, Lyapunov exponent method), and acceleration variability (root mean square [RMS]). Using age as a predictor, linear regressions were performed for each dependent variable. The results indicated that walking speed, walk ratio and trunk acceleration variability were not dependent on age ($R^2 < 2\%$). However, there was a significant quadratic association between age and gait instability in the mediolateral direction ($R^2 = 15\%$). We concluded that most of the typical gait features of older age do not result from a slow evolution over the life course. On the other hand, gait instability likely begins to increase at an accelerated rate as early as age 40–50. This finding supports the premise that local dynamic stability is likely a relevant early indicator of falling risk.

1. Introduction

Falls during walking are a major health issue in older adults. Elderly individuals exhibit more conservative gait patterns characterized by slower preferred walking speeds (PWS) and reduced step lengths [1], which are indications of greater cautiousness [2]. Musculoskeletal weakness is strongly associated with falls [3]. The decline of cognitive function is correlated with fall risk [4] and is specifically associated with reduced walking speed [5].

Many different methods have been proposed to describe gait characteristics in the older population to determine the causes of falls. Besides basic spatiotemporal gait features that are modified in older, healthy adults compared to their younger counterparts [1], it is also important to assess the variability of the gait pattern, which is caused by the decreased ability to optimally control gait from one stride to the next [6]. In this context, the root mean square (RMS) of trunk acceleration is often used as a measure of gait variability [7]. Optimal dynamic balance results in smooth trunk acceleration during walking; therefore, a low RMS value is considered evidence of a healthy gait. Another popular method is the estimation of local dynamic stability (LDS), which is derived from chaos theory (maximal Lyapunov exponent [8]). This method takes the nonlinear features of human movement into account more effectively than classical variability estimates (RMS, standard deviation, coefficient of variation). It is assumed that motor control ensures a dynamically stable gait (high LDS) if the divergence remains low between trajectories in a reconstructed state space that reflects gait dynamics. The usefulness of gait LDS to assess gait stability and falling risk has been shown in theoretical [9], experimental [10], and clinical [11] studies [12].

Although the abovementioned parameters have already been proposed to characterize gait in elderly individuals [2,13,14], there is insufficient information regarding the changes in these parameters with age. Most studies have compared older adults

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to matched young controls. However, some aspects of cognitive capabilities decline as early as the second or third decades of life [15]. Similarly, significant strength loss in the lower extremities begins between ages 40 and 50 [16]. Because musculoskeletal and cognitive status are key factors in the etiology of falls in the elderly, gait features in middle-aged adults (40–60 years) demand further investigation [7]. In other words, it is unclear whether the idiosyncrasy of gait in elderly individuals is the result of musculoskeletal/neurological degenerative processes that occur with advanced age, or whether it is the result of a slower evolution throughout the life course.

The objective of the present cross-sectional study, therefore, was to document the effect of age on gait features in 100 healthy individuals aged 20–69. In addition to basic spatiotemporal measures (PWS, step length), gait variability (RMS) and gait stability (LDS) were analyzed while each participant walked on a treadmill. More generally, we sought to assess the extent to which the typical gait characteristics observed in older adults were already present in middle-aged individuals.

2. Methods

2.1. Subjects

The study included 100 healthy subjects (50 males, 50 females) without neurological or orthopedic conditions. There were 10 males and 10 females for each decade between the ages of 20 and 69. Their anthropometric features are presented in Table 1. All participants were accustomed to treadmill walking. A subset (95/100) of the subjects was analyzed in a parallel study about LDS reliability [17]. The study was approved by the regional medical ethics committee (Commission Cantonale Valaisanne d'Ethique Médicale, Sion, Switzerland).

2.2. Experimental procedure and data pre-processing

The subjects wore a tri-axial accelerometer (Physilog® System, Gaitup, Lausanne, Switzerland) fixed with a belt at the anterior upper trunk level, 5 cm under the sternal notch. The accelerometer measured trunk acceleration along 3 axes: mediolateral (ML), vertical, (V), and anteroposterior (AP). Each participant walked barefoot on a treadmill (Venus model, h/p/cosmos®, Traunstein, Germany) while wearing a safety harness that did not impede movement of the arms and legs. PWS was assessed using the method described by Dingwell and Marin [18]. Trunk accelerations were recorded for 5 min while the subjects walked at PWS. Because acceleration data had already been used in the above-mentioned study [17], we employed an identical method for consistency. The data analysis was performed with MATLAB R2013a (MathWorks, Natick, MA). To lower the effect of sensor misplacement, the 3D-acceleration signals were reoriented according to the procedure proposed by Moe-Nilssen [17,19]. To avoid starting effects, the first 5 s were discarded. The raw 200-Hz signals were then down-sampled to 50 Hz to facilitate the subsequent analyses. Step frequency (SF) was assessed using the fast Fourier transform of the vertical acceleration signal. In the frequency domain, the SF was defined as the higher peak in the 0.5–2.5 Hz band. A duration corresponding to 175 strides was then selected for further analysis (i.e., 152–235 s, depending on individual walking speed and cadence). This length was chosen because it provided sufficient reliability for estimating the LDS and RMS [17,20].

2.3. Walk ratio

The average step length (SL) of the 175 strides was computed from the average treadmill speed (SL = PWS/SF). The walk ratio (WR) is the SL normalized by SF (WR = SL/SF): WR represents what would be SL assuming a SF of 1 step/s. This method is an appropriate means of characterizing gait pattern [21] and takes advantage of the invariant relationship between SL and SF, regardless of walking speed [21].

2.4. Gait variability (RMS_{RATIO})

Because acceleration RMS is highly correlated with walking speed [7], the normalization method recently introduced by Sekine [22] was employed. To compute the RMS ratio of the trunk acceleration (RMS_{RATIO}) the vector norm (L) of the 3D acceleration (x, y, z) for each sample n was first computed ($L_n = \sqrt{x_n^2 + y_n^2 + z_n^2}$). The RMS of the vector norm was $L_{RMS} = \sqrt{1/N \sum_{n=1}^{N} (L_n)^2}$. The same procedure was applied to the ML acceleration signal to compute ML_{RMS}. The RMS_{RATIO}, $RMS_{RATIO} = ML_{RMS}/L_{RMS}$, quantified the proportion of trunk acceleration variability that occurred in the ML direction compared to the total acceleration variability.

2.5. Gait instability (local dynamic stability)

The LDS quantification was based on the maximal Lyapunov exponent method using Rosenstein's algorithm. (The reader interested in a full theoretical background may refer to our recently published articles [8,17] that include a more detailed methodology.) The acceleration signals were time-normalized to a uniform length of 10,000 samples to thwart the trend toward a

Table 1
Descriptive statistics and ANOVAs. The 100 participants were classified into 5 age categories and walked for 5 min on a treadmill at preferred walking speed. Trunk accelerations in the mediolateral (ML), vertical (V), and anteroposterior (AP) directions were recorded by a 3D accelerometer. Walk ratio was defined as step length divided by step frequency. Gait variability is the RMS of the lateral acceleration normalized (RMS_{RATIO}) to attenuate the influence of speed (see Section 2). Gait instability (local dynamic stability, LDS) was computed using the maximal finite-time Lyapunov exponent method. Mean (SD) is shown for each age category. One-way ANOVAs were used to assess the differences among age categories. F and p-values are shown, as well as ω^2, which is an unbiased equivalent of η^2; 95% CIs of ω^2 are shown parenthetically. The bold value indicates a significant result.

N = 100	Global mean	20–29 years (N = 20)	30–39 years (N = 20)	40–49 years (N = 20)	50–59 years (N = 20)	60–69 years (N = 20)	$F_{(4,95)}$	p	ω^2
Age (y)	44.2 (14.1)	24.7 (2.8)	34.6 (2.8)	43.9 (2.9)	54.8 (2.7)	63.3 (3.2)	–	–	–
Body weight (kg)	70.2 (14.6)	68.4 (11.9)	65.4 (12.8)	74.2 (15.6)	71.1 (14.4)	72.0 (17.2)	1.10	0.36	0.00 (0–0.15)
Body height (m)	1.72 (0.07)	1.74 (0.06)	1.70 (0.08)	1.74 (0.06)	1.71 (0.08)	1.69 (0.06)	1.94	0.11	0.04 (0–0.19)
Preferred walking speed ($m\,s^{-1}$)	1.09 (0.18)	1.10 (0.15)	1.13 (0.13)	1.11 (0.17)	1.04 (0.24)	1.06 (0.17)	0.95	0.44	0 (0–0.18)
Walk ratio ($m\,Hz^{-1}$)	0.31 (0.04)	0.32 (0.04)	0.32 (0.04)	0.32 (0.04)	0.30 (0.05)	0.30 (0.04)	0.78	0.54	0 (0–0.12
Gait variability (RMS_{RATIO})	0.48 (0.07)	0.49 (0.06)	0.47 (0.05)	0.45 (0.07)	0.47 (0.06)	0.51 (0.10)	1.79	0.14	0.03 (0–0.20)
ML gait instability (LDS, λ)	0.86 (0.06)	0.85 (0.06)	0.84 (0.06)	0.86 (0.04)	0.88 (0.07)	0.90 (0.06)	3.23	**0.02**	0.08 (0.01–0.28)
V gait instability (LDS, λ)	1.23 (0.15)	1.22 (0.18)	1.19 (0.14)	1.22 (0.16)	1.24 (0.13)	1.26 (0.16)	0.52	0.72	0 (0–0.12)
AP gait instability (LDS, λ)	1.09 (0.12)	1.08 (0.10)	1.06 (0.11)	1.12 (0.16)	1.07 (0.12)	1.12 (0.12)	1.12	0.35	0.01 (0–0.16)