Differences in gait stability, trunk and foot accelerations between healthy young and older women

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Abstract

Background: Gait stability has been shown to be affected by age-related mobility problems, but exercise habits may reduce decline in gait stability. Our aim was to evaluate the variability and stability of feet and trunk between older healthy females and young females using inertial sensors.

Method: 20 older females (OF; mean age 68.4, SD 4.1 years) and 18 young females (YF; mean age 22.3, SD 1.7 years) were asked to walk at their preferred speed, while kinematics were measured using inertial sensors on heels and lower back. Spatiotemporal parameters, acceleration characteristics and their variability, as well as trunk stability as assessed using the local divergence exponent (LDE), were calculated and compared between age groups with two-way ANOVA analyses.

Results: Trunk-foot vertical acceleration attenuation, foot vertical acceleration maximum and amplitude, as well as their variability were significantly smaller in OF than in YF. In contrast, for trunk mediolateral acceleration amplitude, vertical acceleration maximum and amplitude, as well as their variability were significantly larger in OF than in YF. Moreover, OF showed lower stability (i.e. higher LDE values) in ML acceleration, ML and VT angular velocity on the trunk.

Conclusion: These findings suggest that healthy older females had a lower maximum toe clearance so that were more likely to trip. Moreover, the acceleration of trunk was sensitive to the difference between healthy older and young females, both in variability and stability. Combined, although older adults had exercise habits, our metrics indicate that they were less stable, which may increase the risk of tripping and balance loss.

Keywords: walking, aging, wearable system, motor control, balance

1. Introduction

Due to poorer physiological function and balance control ability [1], the incidence of falls increase with age, with a 35 to 40% increase in falls in people over 60 years of age, and the figure is higher for females [2]. In addition, falls among older adults becomes the leading indirect causes of disability and death, which brings a big burden to family and society [3]. According to an epidemiological analysis in the community, 53% of the older adults fall while walking in China [4]. Therefore, it is particularly important to pay attention to the gait performance of older adults.

There are several ways to evaluate gait, such as clinical function tests, questionnaires and measurements in a biomechanics laboratory [5]. Questionnaires and clinical tests cannot reflect the comprehensive gait ability, and sometimes have poor objectivity [6]. Gait assessment in a biomechanical laboratory is accurate but costly, time-consuming and limited to the space and time [7]. Therefore, an ambulatory measurement system that can test and collect gait data is particularly important. Inertial sensors have become widespread due to their advantages of being low cost, portable and having low power consumption [8]. Therefore, inertial sensors show great potential to identify stability problems.

Gait stability reflects the ability to keep walking in the face of perturbations [9, 10]. Gait stability can be divided into two categories, global stability and local stability [11].Global stability is the stability of human body against external perturbations, such as slippery ground, sudden obstacles and so on [12]. Local stability refers to the ability to remain stable in the presence of infinitely small disturbances that come from the nervous system or muscle control, when people walk without obvious external perturbations [12].

Some methods adapted from dynamical systems and non-linear time series analysis also have been used to evaluate gait stability, as these are appropriate to quantify the complex and chaotic characteristics of the human body [13]. The most popular stability metric is the local divergence exponent (LDE) [14, 15], which measures the rate of kinematic expansion of initially nearby trajectories [16]. The local divergence exponent quantifies the average exponential rate of divergence of neighboring trajectories in state space, and thus provide a direct measure of the sensitivity of a system to local perturbations [17].

Internal perturbations of the human body cause variability and randomness in periodic gait [18]. If gait is within a stable range, people would not need to correct this variability. Increased variability likely reflects a less automatic gait pattern, instability and susceptibility to falls [19]. Studies also confirmed that variability in some gait parameters (such as stride length, stride width, stride time) were highly related to the risk of falling [20, 21]. However, some studies suggested that variability is not equal to stability, as the level of variability was not necessarily negatively related to the level of stability [12, 22].

Gait stability and variability measures have been shown to differ between young adults and older adults with increased fall risk or mobility problems [23-25]. However, several reviews have indicated the effectiveness of exercises on gait and balance ability of older adults, such as Tai Chi, therapeutic exercise, strengthening and aerobic exercise [26]. It has been found that older persons who actively engaged in exercise performed better in gait such as self-selected walking speed compared with matched groups of older non-exercisers [27]. So far, it is unclear whether gait stability is maintained in older adults who exercise regularly.

Because older females were more likely to fall than older men over 60 years of age [2], we focused on the gait of females in our study. We aimed to compare spatiotemporal measures, variability measures (e.g., standard deviation and coefficient of variation) and gait stability (e.g., Local Divergence Exponent) between older females (OF) with exercises habits and young females (YF), based on inertial sensor data. We hypothesized that healthy older females with exercise habits still have a less stable gait pattern (i.e. higher LDE values, higher variability) compared with younger females.

2. The Materials and Methods

2.1. Participants

A total of 20 healthy OY and 18 YF were recruited among the community and university in this study (Table 1). These OF had the habit of morning exercises over many years, such as dance or tai chi clubs. None of our participants had any orthopedic or neurological disorders, acute pain or other complaints that might have affected gait and they were all able to walk at least 20 m. Participants were informed about the research procedures before they gave consent in accordance with the ethical standards.

Table 1. Participant characteristics

Groups	YF	OF	p-Value (T-test)
Age(years)	22.3 (1.7)	68.4 (4.1)	<0.001
Height (m)	1.65 (0.04)	1.59 (0.05)	<0.001
Body mass (kg)	54.66 (3.93)	63.2 (7.95)	<0.001
BMI (kg/m^2)	20.19 (1.53)	24.96 (2.60)	<0.001
Leg length (cm)	88.21 (3.53)	87.28 (3.19)	0.29

2.2. Data Acquisition

All participants were equipped with three inertial sensors (Xsens MTw Awinda, the Netherland) on both heels and the trunk at the level of L3-L4, sampling at 100 Hz. All participants wore the same model of shoes to walk twice on a 100-meter straight running track at a self-selected speed, since gait variability is expected minimal at this speed for healthy people [28].

2.3. Temporal gait parameters

We discarded the first and last step to ensure that measurements did not reflect accelerations and decelerations associated with starting and stopping. MATLAB (R2019b, MathWorks Inc, Natick, MA, USA) was used to analyze data.

We used angular velocity around the x-axis of the heel sensors to identify gait events[23], with one gait cycle includes the time of heel-strike (T_{heel_strike}), toe-off (T_{toe_off}) and foot-flat (T_{foot_flat}).

In our study, stride time was defined as the duration between two successive T_{heel_strike} of right foot. Combined with the gait events of both feet, we got the Initial double support period (IDS) and the terminal double support period (TDS).

2.4. Spatial gait parameters

We integrated the acceleration of foot between every consecutive $T_{foot\text{-}flat}$ when the foot was steady on the ground [29]. We used the zero-velocity-update method to distribute the drift of velocity into every moment of the cycle, assuming linearity of the drift [29]. After that, we integrated the velocity again and got the displacement of every gait cycle, which could be added to get a displacement time-series.

All data of the feet sensor were realigned with a coordinate system which had the walking direction as first X axis, and the vertical direction as Z axis, with the Y-axis pointing mediolaterally and to the left.

2.5. Acceleration Measures and variability

The raw data from the trunk sensor was extracted and realigned with anatomical axes, based on the accelerometer's orientation with respect to the gravity and optimization of left-right symmetry [30].

For both feet and the trunk sensors, we calculated the maximum vertical acceleration, which reflects the intensity of ground contact [31]. We calculated the trunk-foot vertical acceleration attenuation by the difference of vertical acceleration maximum between trunk and foot, in order to assess the impact absorption of the lower limbs. Moreover, to get an idea of the movement intensity, for each sensor (both feet and trunk) and direction (AP, ML and VT) we calculated the acceleration amplitude as the maximum minus minimum acceleration during a stride.

For above parameters of each person, after getting the mean and standard deviation (SD) over all cycles, we obtained coefficient of variation (CV) by dividing the standard deviation by the mean [32].

2.6. Local Divergence Exponent (LDE)

In this study, we calculated the LDE of the signals of three dimensions separately for the accelerometer data and gyroscope data, leading to 6 LDE on trunk sensor. The time series of 50 gait cycles was normalized into 5000 samples, with the average of 100 samples per cycle.

From these data, state spaces were reconstructed using the method of correlation integral (C-C method), which not only can determine both embedding dimension and delay time, but also has a good robustness to the noise in small amount of data [33] (see supplementary tables 1 and 2 for dimension and delay values). LDE was expressed as the mean logarithmic rate of divergence per stride using Rosenstein's method [34]. Higher values of the LDE indicate a lower local stability.

2.7. Statistical Analysis

Normality was assessed using the Kolmogorov–Smirnov test. For all spatiotemporal, variability and stability parameters of the left and right feet, differences were test using two-way ANOVAs, with within-subject factors Foot (left and right) and between-subject factor Group (YF and OF). Post hoc Tukey tests tallied differences between the factors' modalities. For other parameters, we used unpaired Student's t-tests. For LDE, which appeared not distributed normally, we compared between groups using the Mann–Whitney test. For all parameters, $p<0.05$ was considered as a significant effect.

3. Results

3.1. Mean gait parameters

Table 2 shows the average values for all measures. We found no interaction between Foot and Group for any of the outcome measures. For spatial-temporal gait measures, we found no statistically significant differences between OF and YF.

For vertical acceleration maximum, values of OF were significantly smaller than for YF in both left and right feet, whereas for the trunk, they were significantly higher in the OF. Also, the trunk-foot vertical acceleration attenuation of OF was significantly smaller than those of YF.

For acceleration amplitude, the vertical value of OF were significantly smaller than of YF, in both left and right feet, whereas the ML acceleration amplitude, VT acceleration amplitude on trunk were significantly larger in OF than in YF.

The LDE calculated from ML acceleration of trunk, ML angular velocity of trunk and VT angular velocity of trunk were significantly larger (less stable) for OF than for YF.

Table 2. Average (SD) outcome measures

parameters	OF	YF	p-value
Spatial-temporal measures			
Stride time(s)	1.04 (0.07)	1.07(0.05)	0.06
IDS %	14.30(1.70)	14.00(1.20)	0.50
TDS %	14.10(1.60)	14.30(1.40)	0.76
Swing %	35.80(1.70)	35.90(1.2)	0.78
Velocity of left foot (m/s)	1.35(0.17)	1.37(0.15)	0.88
Velocity of right foot (m/s)	1.36(0.16)	1.36(0.13)	0.88
Stride length of left foot (m)	1.28(0.11)	1.35(0.12)	0.06
Stride length of right foot (m)	1.28(0.10)	1.35(0.11)	0.08
VT Acceleration Maximum (m/s^2)			
VT acceleration maximum of left foot	25.24(4.75)	35.24(9.05)	<0.001**
VT acceleration maximum of right foot	24.38(3.89)	31.81(8.51)	0.001**
VT acceleration maximum of trunk	18.04(2.75)	16.43(1.09)	0.011*
Trunk-foot vertical acceleration attenuation(m/s^2)	17.00(3.73)	28.62(8.61)	0.023*
Acceleration Amplitude (m/s^2)			
AP acceleration amplitude of left foot	84.38(15.50)	92.47(12.32)	0.08
AP acceleration amplitude of right foot	88.69(13.21)	95.75(14.58)	0.13
ML acceleration amplitude of left foot	31.59(6.24)	27.96(4.61)	0.07

ML acceleration amplitude of right foot	30.90(7.77)	28.78(4.45)	0.29
VT acceleration amplitude of left foot	59.53(9.92)	68.82(9.52)	0.007**
VT acceleration amplitude of right foot	60.69(10.75)	67.78(9.95)	0.036*
AP acceleration amplitude of trunk	8.97(2.69)	7.35(1.69)	0.07
ML acceleration amplitude of trunk	9.11(2.89)	8.52(1.28)	0.010*
VT acceleration amplitude of trunk	11.94(3.88)	9.65(1.39)	0.010*
Local Divergence Exponent (LDE)			
AP acceleration of trunk	1.15(0.46)	1.00(0.37)	0.28
ML acceleration of trunk	0.84(0.18)	0.68(0.12)	0.005**
VT acceleration of trunk	0.92(0.26)	0.87(0.21)	0.89
AP angular velocity of trunk	0.80(0.31)	0.69(0.30)	0.06
ML angular velocity of trunk	0.97(0.23)	0.69(0.11)	<0.001**
VT angular velocity of trunk	0.73(0.32)	0.56(0.12)	0.048*

Notes: AP = anteroposterior direction; ML = mediolateral direction; VT = vertical direction. *p < 0.05, **p < 0.01.

3.2. Variability measures

Table 3 and Table 4 show the SD and CV of all measures. There were no significant differences in either SD or CV of spatial-temporal gait parameters between YF and OF.

The SD and CV of vertical acceleration maximum were significantly smaller in OF than in YF for both feet, while the SD of vertical acceleration maximum of the trunk was significantly higher in OF than in YF. As for trunk-foot vertical acceleration attenuation, the SD was smaller in OF than in YF. For acceleration amplitude, SD of OF in three directions were significantly smaller than of YF in both feet. Except CV of AP acceleration amplitude of left foot, which was not different between groups, CV of OF in three directions were also smaller than in YF in both feet. However, the SD of OF in ML and VT acceleration amplitude of trunk were larger significantly than of YF.

Table 3. SD(SD) of all measures

parameters	OF	YF	p-value
Spatial-temporal Gait parameters			
Stride time(s)	0.02(0.006)	0.02(0.005)	0.64
IDS %	0.74(0.11)	0.71(0.11)	0.48
TDS%	0.80(0.20)	0.79(0.15)	0.93
Swing %	0.75(0.16)	0.75(0.19)	0.97
Velocity of left foot (m/s)	0.06(0.03)	0.08(0.04)	0.05
Velocity of right foot (m/s)	0.06(0.03)	0.07(0.04)	0.05
Stride length of left foot (m)	0.02(0.01)	0.05(0.05)	0.13
Stride length of right foot (m)	0.03(0.01)	0.04(0.05)	0.49
VT Acceleration Maximum (m/s^2)			
VT acceleration maximum of left foot	4.08(1.22)	7.98(2.94)	<0.001**

VT acceleration maximum of right foot	4.02(1.56)	7.82(3.58)	<0.001**
VT acceleration maximum of trunk	1.02(0.44)	0.78(0.13)	0.007**
Trunk-foot vertical acceleration attenuation(m/s^2)	4.02(1.09)	7.99(2.91)	0.008**
Acceleration Amplitude (m/s^2)			
AP acceleration amplitude of left foot	5.68(1.59)	7.07(1.53)	0.026*
AP acceleration amplitude of right foot	6.25(1.94)	7.97(2.29)	0.006**
ML acceleration amplitude of left foot	4.64(1.23)	5.68(1.66)	0.045*
ML acceleration amplitude of right foot	4.64(1.16)	5.99(2.09)	0.010*
VT acceleration amplitude of left foot	4.84(1.24)	9.67(2.90)	<0.001**
VT acceleration amplitude of right foot	4.99(1.51)	9.70(2.53)	<0.001**
AP acceleration amplitude of trunk	1.26(0.45)	1.05(0.31)	0.35
ML acceleration amplitude of trunk	0.98(0.48)	0.82(0.21)	0.017*
VT acceleration amplitude of trunk	1.31(0.67)	0.08(0.15)	<0.001**

Notes: AP = anteroposterior direction; ML = mediolateral direction; VT = vertical direction. *$p < 0.05$, **$p < 0.01$.

Table 4. CV(SD) of all measures

parameters	OF	YF	p-value
Spatial-temporal Gait parameters			
Stride time	0.02(0.005)	0.02(0.004)	0.86
IDS %	5.28(0.90)	5.11(0.84)	0.56
TDS %	5.67(1.70)	5.58(0.86)	0.83
Swing %	2.10(0.42)	2.11(0.57)	0.94
Velocity of left foot	0.06(0.03)	0.08(0.04)	0.11
Velocity of right foot	0.06(0.03)	0.07(0.04)	0.09
Stride length of left foot	0.02(0.01)	0.03(0.03)	0.16
Stride length of right foot	0.02(0.01)	0.03(0.03)	0.66
VT Acceleration Maximum (m/s^2)			
VT acceleration maximum of left foot	0.16(0.03)	0.22(0.05)	0.002*
VT acceleration maximum of right foot	0.16(0.05)	0.24(0.08)	0.002*
VT acceleration maximum of trunk	0.06(0.02)	0.05(0.01)	0.13
Trunk-foot vertical acceleration attenuation(m/s^2)	0.24(0.05)	0.27(0.05)	0.94
Acceleration Amplitude (m/s^2)			
AP acceleration amplitude of left foot	0.07(0.01)	0.08(0.02)	0.09
AP acceleration amplitude of right foot	0.07(0.02)	0.08(0.02)	0.017*
ML acceleration amplitude of left foot	0.15(0.03)	0.21(0.07)	<0.001**
ML acceleration amplitude of right foot	0.15(0.03)	0.21(0.05)	0.001**
VT acceleration amplitude of left foot	0.08(0.02)	0.14(0.03)	<0.001**
VT acceleration amplitude of right foot	0.08(0.02)	0.14(0.03)	<0.001**
AP acceleration amplitude of trunk	0.14(0.03)	0.15(0.05)	0.10
ML acceleration amplitude of trunk	0.10(0.03)	0.10(0.02)	0.11
VT acceleration amplitude of trunk	0.11(0.04)	0.09(0.02)	0.23

Notes: AP = anteroposterior direction; ML = mediolateral direction; VT = vertical direction. *$p < 0.05$, **$p < 0.01$

4. Discussion

We compared spatiotemporal parameters, gait stability (LDE) and variability measures between healthy older females and young females based on inertial sensor data.

4.1. Mean gait parameters

In this study, we obtained spatiotemporal gait parameters, including gait speed, stride length and stride time. Older females walked with the same preferred walking speed and stride length and time as the younger females. Previous studies suggest that older adults walk slower due to physical limitations like muscle weakness or loss of flexibility [35]. Possibly, the habitual exercise of the older females in our study delayed their muscle loss and therefore the decline of gait speed and stride length and time [27].

The average of VT acceleration maximum and amplitude for the foot sensors were significantly smaller for OF. Based on the similar stride time and percentage of swing, this would suggest that OF had lower maximum toe clearance (MTC) than YF. Considering controlling the MTC was essential for walking without tripping [35], OF could be more likely to trip. However, calculating MTC from accelerometer data is hard, which is why we opted to only calculate peak vertical acceleration, as a proxy. In addition，in this study, OF showed significantly smaller trunk-foot vertical acceleration attenuation than YF, which suggests decreased skeletal muscle cushioning to protect the head's stability[36].

Reduced stability, strength and ROM seem to occur in healthy older adults before slower gait, which is a major clinical predictor of decreased motor function [37]. For stability, the LDE of trunk time-series data can better reflect gait stability difference of age compared to data from other segments [38]. Therefore, our study implied that although older adults had the habit of exercise, the ability to maintain stability of the trunk was still weaker than younger females. Specifically, our OF showed significantly decreased local dynamic instability in ML acceleration, ML and VT angular velocity. Many studies suggested that ML direction needs more control during gait [39, 40], because higher postural sway measures were shown in fallers in the ML direction than non-fallers, such as larger area and excursion of the center of pressure[41, 42]. Considering we found an effect in ML of our older healthy females, LDE in ML direction could hence be an early indicator of decreased gait stability.

4.2. Variability measures

All participants in this study walked under the same environmental conditions. Thus, any between-subject differences in stochastic inputs arose from differences in (internal) neuromotor noise and not (external) environmental noise. No difference was found in the variability of spatiotemporal measures, which was consistent with a previous study showing that temporal gait variability of older non-fallers was not significantly different from young adults in terms of standard deviation and coefficient of variation [28].

Our results showed that the variability of ML and VT acceleration amplitude of the trunk was larger for the OF, which could suggest OF are at a higher risk of balance loss and falling [32].

Our OF walked with similar variability of spatiotemporal gait parameters and vertical acceleration maximum of feet variability compared to YF (Tables 3 and 4). This suggests that stability of the trunk might be more sensitive indicators of locomotor impairment and potential future risk of falls than changes in variability [37].

4.3. Limitations

All tests in our study were aimed at testing the same hypothesis, that is, OF are less stable than YF, hence, we did not use a correction for multiple testing. Nonetheless, not correcting may lead to Type I errors, and thus, some caution is warranted.

Furthermore, our study did not contain an older control group without exercise habits and we have limited information about the exercise habits of our participants, or other qualifications about their muscle mass.

Limited literature exists on the effects of life-long exercise on gait stability in elderly people. Considering the importance of gait stability to predict fall risks, and the importance of mobility to elderly people, further research is warranted.

5. Conclusions

Although healthy older females had similar stride time and percentage of swing as the young females during steady state walking, they showed lower vertical foot accelerations, suggesting a lower minimum toe clearance (MTC), which would make them more likely to trip. The accelerations of the trunk were sensitive to discriminate between healthy older females and young females, both in gait variability and stability. Despite the older females' exercise habits, they still showed less gait stability and hence, were more likely to fall.

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