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Biomechanical characteristics of high- and low-frequency trunk acceleration upon gait initiation related to balance ability in community-dwelling elderly people: a cross-sectional study

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ABSTRACT

Although trunk acceleration during walking is widely used as a measure of stability, few studies have focused on sensitive postural control in community-dwelling elderly people to detect components related to balance ability during gait initiation. This study aimed to clarify the biomechanical characteristics of movement and sensitive postural control related to balance ability, focusing on high- and low-frequency components of trunk acceleration during gait initiation. Healthy older participants were divided into two groups (high-performance older people [Older(H)], n = 11; age, 76.2 ± 3.3 years, and low-performance older people [Older(L)], n = 17; age, 75.8 ± 3.2 years) based on the Timed Up and Go Test time related to balance ability while walking at their chosen speed. Trunk acceleration data were obtained from an accelerometer on the L3-4 level spinous process. The gait velocity was measured at the first step using a motion capture system. The acceleration data were separated into high- and low-frequency components, and the root mean square was calculated. The level of significance was set at 5%. For the high-frequency component, the root mean square of acceleration in Older(L) was significantly lower than that of Older(H) in the mediolateral direction ($p = 0.019$) and correlated with gait velocity ($r = 0.415$; $p < 0.001$). For the low-frequency component, the root mean square of acceleration in Older(L) was significantly lower than that of Older(H) in the vertical ($p = 0.034$) and anteroposterior direction ($p = 0.039$). The results suggest that low- and high-frequency components of trunk acceleration can reveal biomechanical characteristics in community-dwelling elderly people.

Keywords: gait initiation, frequency component, trunk acceleration analysis

Abbreviations:
AP: anteroposterior
COM: center of mass
FES-I: Fall Efficacy Scale International
ML: mediolateral
MMSE: Mini-Mental State Examination
Older(H): high-performance older people
Older(L): low-performance older people
RMS: root mean square
RMS-acc: root mean square of trunk acceleration

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TUG: Timed Up and Go Test
V: vertical

INTRODUCTION

Gait initiation involves the transition from a state of standing static stability to a dynamic state through voluntary postural adjustments eventually leading to a rhythmic gait pattern. Because gait initiation requires greater control and complexity than steady-state gait, postural control during this phase has been associated with potential fall risks.[1,2] When considering postural control during gait initiation, it is important to examine both the shift in the center of mass (COM) and sensitive postural control. A COM shift is necessary to move forward and increase gait velocity within the base of the support, whereas sensitive postural control involves anticipatory postural adjustments and correction of trajectory errors in the COM shift. These control adjustments are highly voluntary in nature. Consequently, many studies have examined gait initiation stability, a causal factor of falling incidents, from the perspective of balance related to postural stability.[3-5]

Both gait function and balance are closely related to walking ability, encompassing motor, sensory, cognitive, and emotional aspects.[6,7] Among the prevalent risk factors, falls are often attributed to lower limb dysfunction in older people, particularly those with health conditions or on medication.[8] Specifically, falls during walking are frequently linked to inadequate trunk–lower limb postural control due to improper shifts in the COM, insufficient lower limb support, and stumbling, which account for over 70% of falls.[9]

In recent years, research interest in trunk acceleration analysis using small accelerometers that can precisely measure the behavior of the COM has continued to increase.[10] The above study reported that trunk acceleration data can reflect general instability. However, trunk acceleration may not accurately indicate specific gait abnormalities, such as step variability and non-steady-state gait phase, which are critical for determining fall risks in older adults.[11] Root mean square (RMS) analysis of trunk acceleration during gait has emerged as a valuable method for quantitatively assessing balance control. RMS values reflect the variability and magnitude of trunk movements and have been reported to correlate with postural stability and fall risk.[12] For instance, higher RMS values in the mediolateral (ML) direction are associated with increased instability, whereas lower RMS values in the vertical (V) direction indicate better postural control.[13] Similarly, RMS measurements can effectively differentiate older adults with different levels of mobility and balance performance.[14] On the other hand, RMS values are difficult to evaluate quantitatively for non-steady-state motion such as gait initiation. Therefore, a more detailed RMS analysis is necessary to properly assess a subject's physical function.

Another aspect of trunk acceleration is that it contains some components derived from principal component analysis.[15] Trunk acceleration may be derived from two major components. First, the basic component generated by regular stepping within the 1–2-Hz frequency band represents the COM shift due to the basic localization activity of the vestibular and postural control system loop activity. Second, the supplemental component generated by sensitive postural control related to balance disorder over the 3-Hz frequency band represents sensitive postural control due to the spatial localization activity for subtle manipulation of the body during task execution. According to a previous study, the characteristics of the frequency components during level walking in young people, community-dwelling elderly people, and stroke patients fundamentally differ in terms of gait cycle and postural control.[16] Therefore, we hypothesized that the meaning of trunk acceleration RMS may differ depending on the two parameters mentioned above.

To capture participants' sensitive control during gait initiation, in a non-steady-state gait, it is important to analyze the frequency component of trunk acceleration quantitatively, which has so far been little studied. Clarifying the meaning of trunk acceleration RMS in terms of frequency components will result in a more precise evaluation, enhancing its potential application in motion tasks that require postural control ability. Therefore, this study aimed to clarify the biomechanical relationship of COM movement and sensitive postural control with differences in characteristics between high- and low-frequency components of trunk acceleration during gait initiation in older people.

METHODS

Study participants

Twenty-eight healthy older individuals participated in the study. The selection criteria were being registered at the staffing center for community-dwelling elderly people aged 70 years or older and giving consent. Participants with neuromuscular, orthopedic, or cognitive disorders verified by a qualified physical therapist during interview and based guidelines prior to the experiment, having experienced events that could be regarded as an accident, and having orthopedic conditions that may affect their daily lives were excluded.

A Timed Up and Go Test (TUG) time of 13.5 s is typically used as a clinical threshold for assessing fall risk. However, this study focused on community-dwelling adults, and a TUG time of 9.0 s was used to reflect their balance abilities, as indicated by two previous studies.[17,18] Using this cutoff time, the group with faster TUG times was classified as high-performance older people (Older(H)) for high performance, whereas the group with slower TUG times was classified as low-performance older people (Older(L)) for low performance at the timing of analysis. This study adhered to the principles outlined in the Declaration of Helsinki. All participants provided informed consent to participate in the study, which was approved by the Institutional Review Board of Nagoya University (23-503).

Experimental protocol

The participants walked a 12-m path five times at a self-selected speed when requested by the examiner to start walking. The acceleration data in the tri-axis directions were measured using a gyrometer combined with an accelerometer (MVP-RF8, Microstone Inc, Nagano, Japan) positioned at the level of the L3-4 spinous process.[19] Trunk acceleration was determined by analyzing both the acceleration and angular velocity components, and gravitational acceleration was corrected at 200 Hz in accordance with the Euler angle principle. A motion capture system (Optitrack Trio, Acuity Inc, Tokyo, Japan) was used to measure gait velocity from a reflective marker on an accelerometer on the trunk part at 120 Hz. Because the sampling frequencies were different, a preliminary movement was performed from a quiet standing position, and synchronization was performed by down-sampling the acceleration data after aligning the starting points.

RMS data split into high- and low- frequency components

The acceleration data in each direction were processed using a low-pass filter (4 dim, cutoff frequency: 3.0 Hz) applied to the low-frequency component based on the gait cycle and a band-pass filter (4 dim, cutoff frequency: 3.0–6.0 Hz) applied to the high-frequency component, based on a previous study.[16,20] The data were then used to calculate the RMS of the period from the starting motion to the first step, which was defined as the root mean square of trunk acceleration in the ML direction (RMS-acc-ML), root mean square of trunk acceleration in the

V direction (RMS-acc-V), and root mean square of trunk acceleration in the anteroposterior direction (RMS-acc-AP) for each direction. The data were used in the frequency analysis for all trials without averaging for each participant, because gait initiation is a non-stationary movement and the differences among trials are meaningful. We used MATLAB 2022b (MathWorks Inc, Natick, MA, USA) to perform this process.

Gait velocity

We detected the velocity at the time of the first step using a reflective marker because the first step is the most effective point of speed change for gait initiation.

Physical, mental, and cognitive factors

To assess the participants' performance levels, we conducted several tests, namely the 10-m walking test under comfortable conditions to reflect walking ability, the Mini-Mental State Examination (MMSE) to evaluate cognitive function in older participants, the Fall Efficacy Scale International (FES-I) to screen for subjective fear of falling, and the Short Physical Performance Battery (SPPB) to assess overall physical function.

Statistical analysis

Statistical analyses were performed using EZR software[21] (ver 1.55). The Kolmogorov–Smirnov test was first conducted to examine the normality of the data distribution in all groups, and the Bartlett test was then conducted to examine the variability of each indicator. To evaluate the MMSE and SPPB, we performed comparisons between groups using the Mann–Whitney U test because the data did not follow a normal distribution. To compare the other indicators between the two groups, we used a t-test for group comparison. The Pearson correlation coefficient was evaluated for the RMS-acc and gait velocity using correlation analysis. The level of significance was set at $p < 0.05$.

RESULTS

Patient characteristics

There were no significant differences in the fundamental characteristics among all participants or in age between the groups (Table 1).

Table 1 Participant data

	Older(H) (n = 11)	**Older(L) (n =17)**	**p-value**
Age (year)	76.2 ± 3.3	75.8 ± 3.2	0.779
Sex (Male/Female)	5/6	11/6	—
Height (cm)	158.5 ± 7.1	160.5 ± 8.2	0.518
Weight (kg)	53.3 ± 6.3	57.9 ± 8.7	0.141
TUG time (s)	8.60 [8.08–8.79]	10.38 [9.55–11.23]	—

The values for each item, except for TUG time, are shown as the mean ± standard deviation. TUG time values are presented as the median with the range from the first to the third quartile.
Older(H): high-performance older people
Older(L): low-performance older people
TUG time: Timed Up and Go Test time under comfortable speed conditions

Physical, mental, and cognitive performance

Older(L) exhibited significantly lower MMSE (p = 0.025) and FES-I (p < 0.001) than Older(H). In contrast, no significant differences were observed between Older(H) and Older(L) in the 10-m gait time and SPPB (Table 2).

Table 2 Performance tests

	Older(H) (n = 11)	Older(L) (n =17)	p-value
10m-gait time (s)	7.01 ± 0.83	7.47 ± 1.03	0.229
MMSE	29.0 [27–29]	26.0 [24–28]	0.025
FES-I	19.4 ± 3.3	27.9 ± 6.5	<0.001
SPPB	11.5 [11.0–12.0]	11.0 [9.0–11.0]	0.118

To compare the two groups, we conducted a t-test for the 10-m gait time and FES, and the Mann–Whitney U test for the MMSE and SPPB.
The 10-m gait time and FES-I values are shown as the mean ± standard deviation. The MMSE and SPPB values are presented as the median with a range from the first to the third quartile.
10-m gait time: 10-m walking test time under comfortable speed conditions
Older(H): high-performance older people
Older(L): low-performance older people
MMSE: Mini Mental State Examination
FES-I: Falling Efficacy Scale International
SPPB: Short Physical Performance Battery

RMS-acc in each direction for high- and low-frequency components

For the high-frequency component, Older(L) had a significantly lower RMS-acc-ML than Older(H) (Older(L), 0.085 ± 0.040; Older(H), 0.103 ± 0.039; p = 0.019; Fig. 1a). For the low-frequency component, Older(L) had a significantly lower RMS-acc-V and RMS-acc-AP

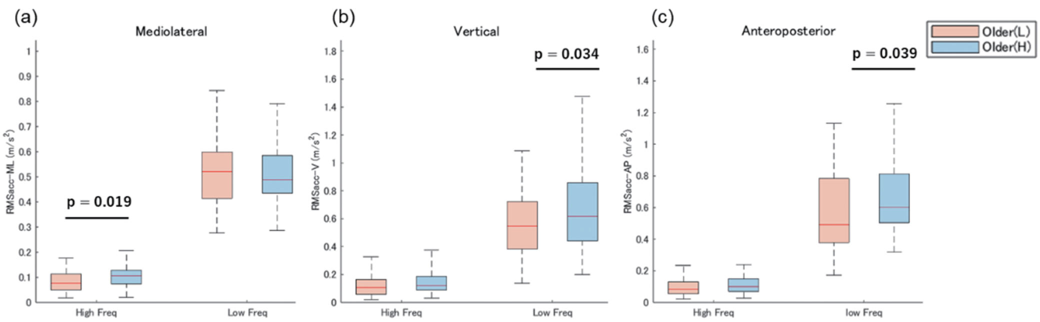

Fig. 1 High and low-frequency components for RMS-acc in each direction
Older(H): high-performance older people
Older(L): low-performance older people
RMS-acc-ML: root mean square of trunk acceleration in the mediolateral direction
RMS-acc-V: root mean square of trunk acceleration in the vertical direction
RMS-acc-AP: root mean square of trunk acceleration in the anteroposterior direction
High freq: high-frequency component of trunk acceleration
Low freq: low-frequency component of trunk acceleration

than Older(H) (V: Older(L), 0.570 ± 0.236; Older(H), 0.684 ± 0.303; p = 0.034; Fig. 1b, AP: Older(L), 0.575 ± 0.249; Older(H), 0.660 ± 0.220; p = 0.039; Fig. 1c).

In the subgroup analysis, the high-frequency component in the ML direction showed a significant correlation with MMSE in the overall participant data (r = 0.487; p = 0.009), and a significant correlation with FES-I in the Older(L) group (r = –0.572; p = 0.016).

Correlation analysis of the time-series data of RMS-acc and gait velocity for high- and low-frequency components

For the high-frequency component, we observed a significant correlation in the ML direction (r = 0.415; 95%CI, 0.248–0.558; p < 0.001; Fig. 2a), whereas for the low-frequency components, we observed a significant correlation in the V direction (r = 0.543; 95%CI, 0.397–0.662; p < 0.001; Fig. 2b) and AP direction (r = 0.262; 95%CI, 0.080–0.428; p = 0.005; Fig. 2b).

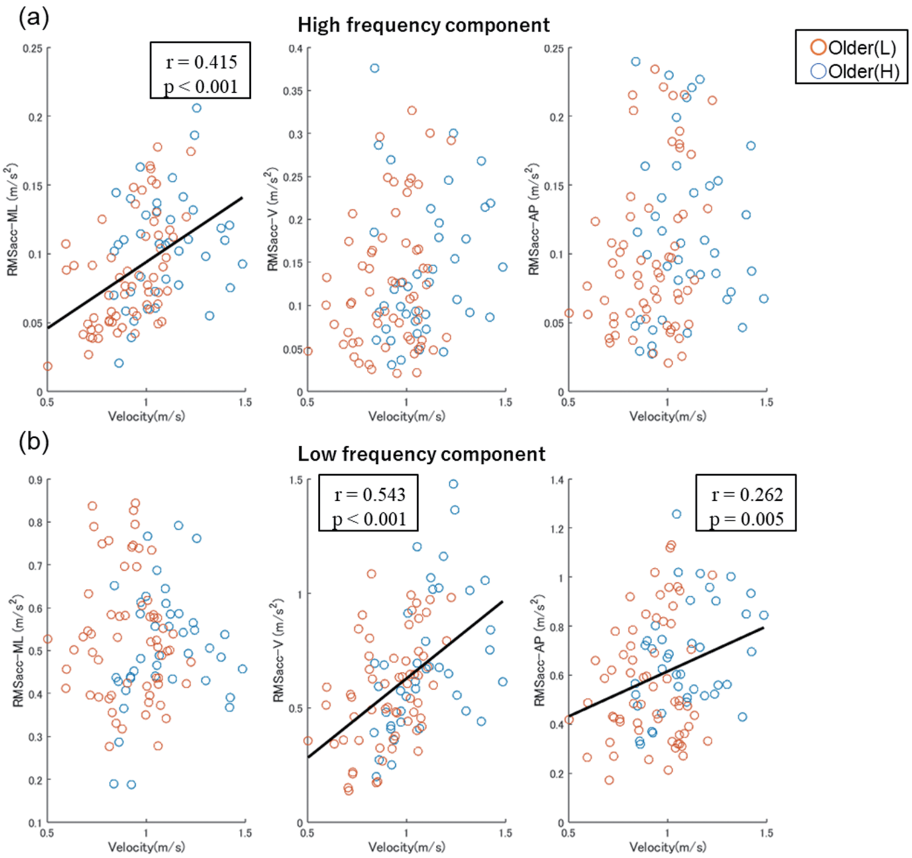

Fig. 2 Trend differences between high and low-frequency components of the relationship between RMS-acc and gait velocity in each direction

Older(H): high-performance older people
Older(L): low-performance older people
RMS-acc-ML: root mean square of trunk acceleration in the mediolateral direction
RMS-acc-V: root mean square of trunk acceleration in the vertical direction
RMS-acc-AP: root mean square of trunk acceleration in the anteroposterior direction

DISCUSSION

This study examined the characteristics of the high- and low-frequency components of the RMS of trunk acceleration in each direction and their trends in relation to walking speed. The results of this study suggested that the trunk acceleration frequency component can be quantitatively evaluated in terms of balance ability during gait initiation, a non-steady-state gait.

Regarding the significant differences between the V and AP directions of the trunk acceleration RMS of the low-frequency component, postural control based on body function is required for forward movement of the COM, which will not deviate from the base of the support.[22] One study has found differences in the propulsion ability of young and older participants related to trunk-lower extremity coordination in the AP direction.[23] Therefore, our results may indicate whether forward propulsion during gait initiation can be performed efficiently. The high-frequency component in the ML direction, which indicated that the RMS value was higher in the group of older adults with a higher balance function because of both components being affected during gait initiation, included highly sensitive postural control, such as anticipatory postural control and COM trajectory error modification. The process was performed for approximately 1.0 s by complex trunk-lower limb coordination and may reflect the extent to which a supportive component is at work in the transition from a static to a dynamic state.[24] Accordingly, the interpretation of RMS values varies greatly depending on whether the participant has a prevalent disease or is an older person living in the community. Specifically, in children with cerebral palsy, RMS in the ML direction increases with increased instability,[25] whereas in community-dwelling elderly people, it shows no difference or a decrease.[12,26] The participants in our study were older people living in a community with a relatively high balance ability, and no significant difference was found in the low-frequency component, but only in the high-frequency component related to the fine postural control component. This observation is likely due to the different target population. Regarding the TUG time, the average time for the participants in this study was 10.4 s compared to a cutoff value of 13.5 s, which is the golden standard for fall risk indicators. Therefore, when considering the general older population as the control population, our study population can be regarded as a population with a relatively higher balance ability. Previous research suggests that differences in trunk-lower limb coordination between older people at high and low risk of falling in the ML direction are a main causal factor in the sensitive postural control in this case.[23] In addition, participants with ACL injuries tend to have lower mean frequencies of high-frequency components than healthy participants, which may also reflect poor coordination due to lower extremity dysfunction.[27] These results are consistent with those reported in a previous study that focused on the frequency components of each participant's gait and helped improve the accuracy of gait analysis.[16]

In terms of the relationship between low-frequency components and gait velocity, the correlations were in good agreement in the directions in which significant differences were observed in the results for each frequency component. Previous research on young people found that the RMS values varied with walking speed in all directions.[15] In contrast, our results showed different characteristics for each frequency component. This indicated that the frequency component can reflect postural control and help evaluate trunk sway caused by a participant's balance ability, which is a fundamental factor for walking speed.

In this study, the older group with lower balance ability had lower cognitive function and lower self-efficacy related to falls compared to the other group, based on the MMSE and FES-I results. It is known that participants with cognitive decline have increased trunk sway in the ML direction and decreased balance ability.[28] Another study suggested that the same phenomenon occurs for participants with low self-efficacy for falling.[29] In particular, the two groups differed for each FES-I item, namely walking on slippery surfaces and on uneven terrain. These tasks require a higher level of postural control and may best capture the characteristics of the participants in this study. In our study, significant differences in the ML direction were found only in the high-frequency component, which reflects the fine postural control component related to cognitive function and self-efficacy. In addition, in the subgroup analysis, the high-frequency component in the ML direction showed a significant correlation with MMSE in the overall participant data, and a significant correlation with FES-I in the Older(L) group. In other words, a decrease in the high-frequency component led to an increase in trunk sway and a decrease in postural control ability.

This study has some limitations. First, the sum of each component separated into low- and high-frequency components does not reflect all the information for the original acceleration component. Time delay and the components around the cutoff frequency were considered to not be significantly affected; however, there was a certain degree of error in these areas. Second, even if we divide the acceleration into high- and low-frequency components, trunk acceleration of low-frequency components that are highly dependent on the movement itself tend to be more represented. Future studies should investigate this protocol among community-dwelling older adults with a higher risk of falls or prevalent disease.

CONCLUSION

This study performed a valid quantification of the frequency components of the RMS data of trunk acceleration to clarify its characteristics during gait initiation. We found that the high-frequency component can detect differences in the balance ability of community-dwelling older adults living with relatively high levels of physical function and could help to indicate a potential loss of balance ability.

CONFLICT OF INTEREST DECLARATION

All authors have no competing interests.

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