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Original Article

Changes in acceleration and deceleration factors associated with active gait speed adjustment

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Abstract. [Purpose] The ability to actively adjust walking speed is fundamental and the factors enabling it should be assessed. The present study aimed to demonstrate how active gait speed is kinematically adjusted. [Participants and Methods] Walking acceleration and deceleration were evaluated in 16 healthy adults using three-axis accelerometers and surface electromyographs. The root mean square (RMS) of each axis in the center-of-gravity acceleration was calculated as an index of gait stability. Electron myograph data were obtained from images captured of the right lower muscles, and the integral value of total muscle activity per gait cycle was calculated. [Results] The RMS of each axis increased during acceleration and decreased during deceleration. The integral values of total activity of the gastrocnemius, biceps femoris, and tibialis anterior muscles increased in acceleration. In contrast, the values increased in the biceps femoris but decreased in other muscles during deceleration. [Conclusion] These results suggest that the specific kinematic mechanisms of each factor regulate the acceleration and deceleration of walking. In addition, these mechanisms and factors indicate how exercise therapy may be used in rehabilitation to improve the ability to adjust walking speed in daily life.
Key words: Accelerometer, Electromyograph, Kinematics

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INTRODUCTION

As walking speed decreases with aging, the risk of falling increases[1–4]. Lower walking speed reportedly correlates with a higher fall risk[5–8]. However, other studies have reported that a decrease in walking speed is a coping mechanism for physical function decline[9–11]. In older individuals, comfortable walking speed is dependent on the surrounding environment. Older individuals with a slower walking speed are more likely to fall indoors, while those with faster walking speeds are prone to falling in outdoor environments; nevertheless, such relationships are not linear[12]. Therefore, we believe that walking speed is not a simple indicator of fall risk and that gait stability should be evaluated in situations closer to those of daily life.

Daily life often involves varying walking speeds, which is more challenging than maintaining a constant speed. Furthermore, it has been reported that approximately 35–50% of all walking in a typical daily routine involves turning[13] and that walking speed is markedly reduced when approaching a stationary or moving obstacle[14]. Therefore, the ability to control walking speed should be regarded as a goal for rehabilitation considering the purpose and surrounding environment, and walking tests should approximate the daily life environment. In acceleration and deceleration control, changes in ankle joint moments produce braking of walking speed in the early stance phase[15]. Lower limb function in acceleration and decelera-

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tion depends on changes in joint angles and is not related to joint moments[16]. Although these studies have explored the biomechanics of walking speed acceleration/deceleration control, the main experimental equipment used was a 3D motion capture system, which can analyze kinematic parameters, including joint moments, in real-time when combined with a floor reaction force meter. However, several studies have focused on the periods immediately before and after the change in walking speed and in a small number of steps[6, 10, 15]. Since approximately 40% of all walking in daily life consists of less than 12 consecutive steps[17], it is important to capture changes in kinematic parameters that follow changes in walking speed over longer periods.

The 3-axis accelerometer is simple, has high repeatability[18] in various motion analyses, and a useful tool to evaluate gait stability[19]. Also, surface electromyography (EMG) is widely used clinically for gait analysis. Studies exist on the relationship between gait speed and fall risk[20] and on changes in lower limb muscle activity during acceleration/deceleration control[21]. Further analysis of COG acceleration values obtained with a 3-axis accelerometer is expected to improve gait analysis, and we believe that results are important as a complementary element to COG acceleration changes. Therefore, this study used a 3-axis accelerometer and a surface EMG system to estimate specific parameters of acceleration/deceleration control of walking speed. This study aimed to show how active gait speed is if adjusted kinematically.

PARTICIPANTS AND METHODS

This was a cross-sectional study. Students attending Kanagawa University of Human Services who provided informed consent were publicly recruited. This study included 16 healthy adults (4 male and 12 female adults, mean age 21.4 years, standard deviation [SD] 1.0, mean height 164.0 ± 7.0 cm). Participants had no history of orthopedic or neuromuscular disease that would interfere with the performance of daily activities or gait. All participants provided written informed consent. This study was approved by the Office of Research Ethics at Kanagawa University of Human Services (Approval No. 7-20-67) and conformed with the Declaration of Helsinki.

We used a 3-axis accelerometer (Ultium EMG (EM-U810M8), Noraxon, Scottsdale, AZ, USA) and a surface EMG system (Ultium EMG (EM-U810M8), Noraxon) for measuring and evaluating gait stability[22, 23], and a footswitch that identifies heel contact (HC) via a pressure sensor. Participants walked along a straight walking path of approximately 50 m in a straight 200-m corridor. All participants walked the entire distance. There were two task conditions: acceleration and deceleration. Acceleration required a rapid shift from a comfortable walking speed to the maximum walking speed following a sound cue. Deceleration required a rapid shift from the maximum walking speed to a comfortable walking speed following the cue. The sound cue was controlled via a software program (LabView 2015, National Instruments, Austin, TX, USA) approximately 5–10 s after the initiation of the experiments.

For each participant, a 3-axis accelerometer was attached to the third lumbar spinous process, which reflects most movement of the COG[24, 25]. Muscle activity during walking was recorded using surface EMG for the vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA), and medial head of the gastrocnemius (GAS) muscles on both lower limbs. EMG recordings were performed by bipolar derivation with 10-mm spaced electrodes on the belly of each muscle, as recommended by a previous study[26]. Additionally, footswitches were fixed to the central portion of the right and left heels to identify the left and right HC. The footswitches were attached with tape to prevent them from falling during walking or obstructing walking. Participants walked barefoot. They performed three trials of the 10 meters walking tests at comfortable and maximum walking speeds. They were instructed to “walk as fast as you feel comfortable walking” at the comfortable walking speed and “walk as fast as you can without running” at the maximum walking speed, and the average speed of each trial in the comfortable and maximum conditions was used as the representative speed for each walking condition. Subsequently, acceleration and deceleration tasks were performed for 10 trials (20 trials in total). The condition order was randomly determined. Participants were instructed to “start walking at a comfortable walking speed and adjust to reach the maximum walking speed as soon as possible after hearing the sound” in the acceleration condition and “start walking at the maximum walking speed and adjusting to approaching the comfortable walking speed as soon as possible after hearing the sound” in the deceleration condition. To avoid the effects of fatigue, participants rested sufficiently between tasks. Signals from the 3-axis accelerometer, surface EMG, and footswitch were A/D converted in real-time using LabChart7 (ADInstruments, Bella Vista, NSW, Australia). The 3-axis accelerometer data were recorded at a 200 Hz sampling frequency. The surface EMG and footswitch data were recorded at a 2-kHz sampling frequency. The data were analyzed offline.

Figure 1 shows the schematic of the walking tasks. The 3-axis accelerometer results recorded in LabChart7 were extracted according to the timing of the right HC. The extracted data comprised two gait cycles before the cue and nine gait cycles after the cue (six gait cycles for EMG), with one gait cycle from the time of right HC to the next right HC. The right foot was used as the reference since it was the dominant foot in all the participants. These two gait cycles before the cue were used for comparing each of the gait cycles after the cue. The gait cycles were used as a landmark that showed gait speed changed significantly.

We used the myoMOTION software (Noraxon, Scottsdale, AZ, USA) for our analysis. The root mean square (RMS) was calculated for the left/right, front/rear, and vertical (x, y, and z, respectively) axes of the 3-axis accelerometer data. The RMS represents the dynamics of the acceleration component for each axis in walking and functions as an index of walking stability[27, 28]. The RMS of each axis was used to quantitatively compare the dynamics of the acceleration components in the

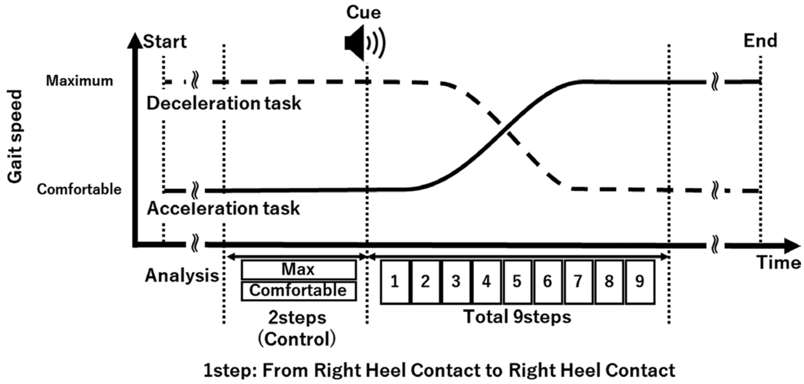

Fig. 1. Schematic of the walking tasks performed.
For the two walking tasks, the vertical axis represents the walking speed, while the horizontal axis represents the time course. The acceleration task required a shift from comfortable walking speed to maximum walking speed as quickly as possible upon receiving the cue. The deceleration task required shifting from maximum to comfortable walking speed immediately after receiving the cue. The data from 3-axis accelerometer and the integrated value of total muscle electromyographic activity (%IEMG) were extracted nine steps after the cue, according to the timing of right heel contact. Additionally, the two gait cycles before the cue were used as a control in both tasks.

x, y, and z planes (ax_RMS, ay_RMS, and az_RMS). RMS values were averaged at 50-ms intervals. Using equation (1), we calculated a value integrating the RMS of x, y, and z ("composite RMS") and used it as an index reflecting the total change in body dynamics. The axial (ax_RMS, ay_RMS, az_RMS) and the composite RMS were compared for each walking speed condition, with the values averaged over the respective number of walking trials for each participant as representative values.

$$compositeRMS = \sqrt{RMS_x^2 + RMS_y^2 + RMS_z^2} \cdots A \quad (1)$$

The EMG signals were filtered using myoMOTION software and rectified along with band-pass filtering (10–500 Hz)[29]. For each muscle, the integrated value of the total muscle EMG activity (%IEMG) was calculated for each gait cycle. The %IEMG was standardized to the gait cycle in which the maximum muscle activity was recorded in all trials. This was used as an index of total muscle activity for each muscle per gait cycle. %IEMG values were averaged over the respective numbers of walking trials for each participant and compared for each condition.

SPSS 26.0 (IBM, Armonk, NY, USA) was used for statistical analysis. Multiple comparisons were performed using a paired t-test with Holm's method adjustment on the composite and axial RMS data from the 3-axis accelerometer, and the %IEMG averaged over all participants for each walking speed condition. The significance level for all tests was set at 5%.

RESULTS

The COG acceleration was analyzed for nine gait cycles, which characteristically reflected changes in walking speed following the sound cues. Figure 2 shows the results from the multiple comparisons of the composite RMS. In the acceleration condition, the composite RMS acceleration significantly increased from the value at comfortable walking speed in the first gait cycle ($p<0.05$), and after the fourth gait cycle, no difference was detected from the value at maximum walking speed. In the deceleration condition, composite RMS decreased significantly from the value at maximum walking speed in the fifth gait cycle ($p<0.05$), and after the sixth gait cycle, no difference was detected from the value at comfortable walking speed.

Figure 3a–3f show the results from the multiple comparisons of the axial RMS. In the acceleration condition, ax_RMS significantly increased from the value at comfortable walking speed in the second gait cycle ($p<0.05$), and after the fifth gait cycle, no difference was detected from the value at maximum walking speed. In the deceleration condition, ax_RMS decreased significantly from the value at maximum walking speed in the sixth gait cycle, and after the seventh gait cycle, no difference was detected from the value at comfortable walking speed ($p<0.05$). Furthermore, In the acceleration condition, ay_RMS significantly increased from the value at comfortable walking speed in the second gait cycle ($p<0.05$), and after the fourth gait cycle, no difference was detected from the value at maximum walking speed. In the deceleration condition, ay_RMS decreased significantly from the value at maximum walking speed in the sixth gait cycle, and after the fifth gait cycle, no difference was detected from the value at comfortable walking speed ($p<0.05$). In addition, In the acceleration condition, az_RMS significantly increased from the value at comfortable walking speed in the first gait cycle ($p<0.05$), and after the third gait cycle, no difference was detected from the value at maximum walking speed. In the deceleration condition, az_RMS decreased from the value at maximum walking speed within the seventh gait cycle ($p<0.05$), and after the fourth gait cycle, no difference was detected from the value at comfortable walking speed.

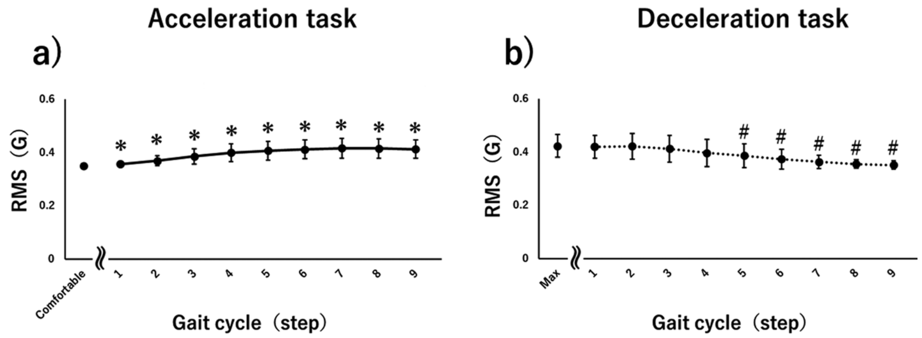

Fig. 2. Comparison of walking speed conditions using composite root mean square (RMS).
The composite RMS for each of the acceleration; a) and deceleration; b) tasks are shown (N=16, error bar=standard deviation).
*: Values that differed significantly from those obtained during comfortable walking.
#: Values that differed significantly from those obtained during maximum walking.

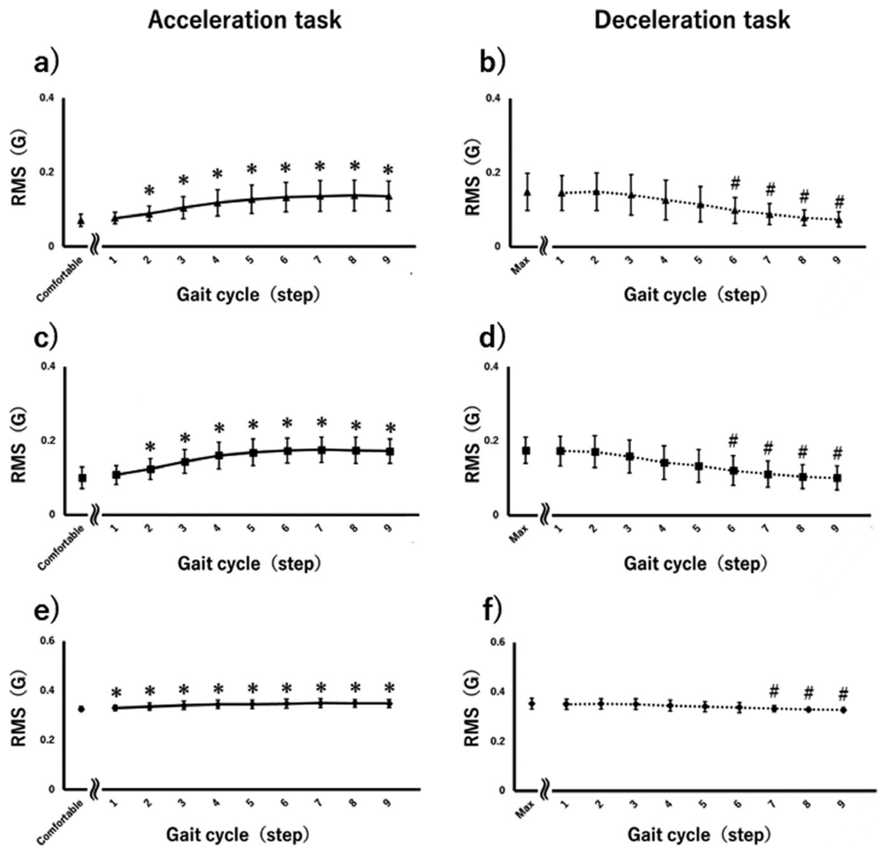

Fig. 3. Comparison of walking speed conditions using root mean square (RMS) of the three axes.
a), b) ax_RMS (medio-lateral); c), d) ay_RMS; e), f) az_RMS (vertical). In addition, a), c) and e) represent acceleration tasks, and b), d) and f) represent deceleration tasks (N=16, error bar=standard deviation)
*: Values that differed significantly from those obtained during comfortable walking.
#: Values that differed significantly from those obtained during maximum walking.

Lower limb muscle activity was analyzed for six gait cycles, which characteristically reflected changes in gait speed from sound cues. Due to incomplete data, only 15 participants were included in this analysis. Figures 4 and 5 show the results of the multiple comparisons of %IEMG.

In the acceleration condition, VM activity significantly increased from the value at comfortable walking speed in the second gait cycle ($p<0.05$), and after the second gait cycle, no difference was detected from the value at maximum walking

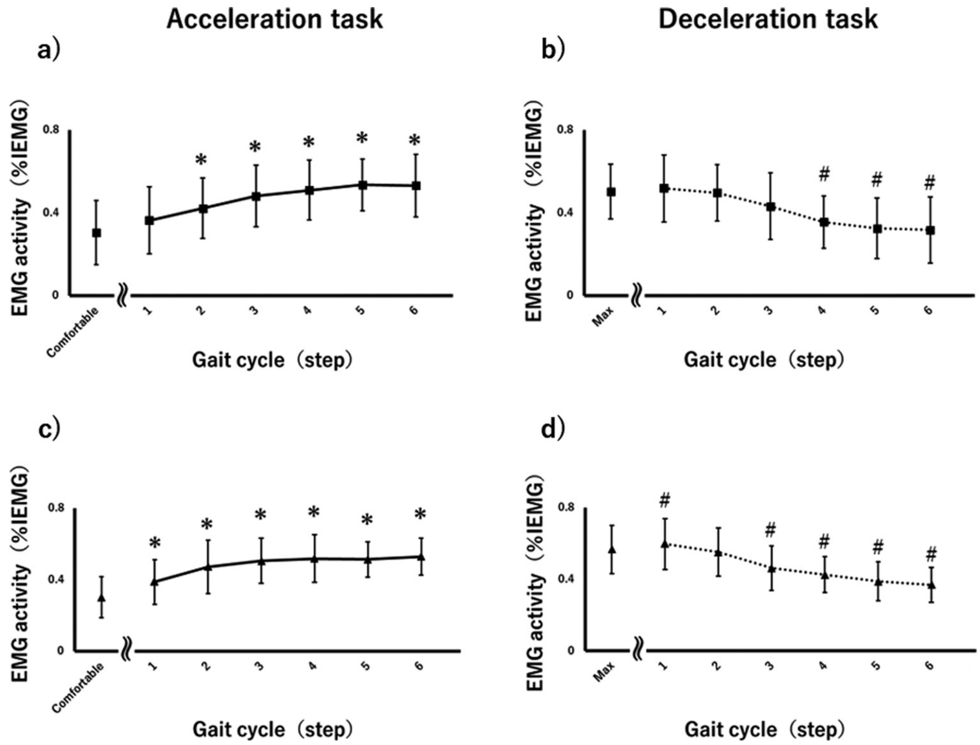

Fig. 4. Comparison of integral values of total muscle activity (%IEMG) between walking speed conditions.
a) and c) represent acceleration tasks, and b) and d) represent deceleration tasks (N=15, error bar=standard deviation)
*: Values that differed significantly from those obtained during comfortable walking.
#: Values that differed significantly from those obtained during maximum walking.
EMG: electromyography.

speed. In the deceleration condition, VM activity decreased from the value at maximum walking speed within the fourth gait cycle ($p<0.05$), and no difference was detected from the value at comfortable walking speed. In the acceleration condition, BF activity significantly increased from the value at comfortable walking speed in the first gait cycle ($p<0.05$), and after the second gait cycle, no difference was detected from the value at maximum walking speed. In the deceleration condition, BF activity increased significantly in the first gait cycle after the cue ($p<0.05$). Furthermore, BF decreased from the third gait cycle ($p<0.05$), and after the sixth gait cycle, no difference was detected from the value at comfortable walking speed. In the acceleration condition, TA activity significantly increased from the value at comfortable walking speed between the cue and the first gait cycle ($p<0.05$), and no difference was detected from the value at maximum walking speed. In the deceleration condition, TA activity decreased from the value at maximum walking speed in the third gait cycle ($p<0.05$), and after the fourth gait cycle, no difference was detected from the value at comfortable walking speed. In the acceleration condition, GAS activity significantly increased from the value at comfortable walking speed between the cue and the first gait cycle ($p<0.05$), and no difference was detected from the value at maximum walking speed. In the deceleration condition, GAS activity decreased from the value at maximum walking speed within the second gait cycle ($p<0.05$), and no difference was detected from the value at comfortable walking speed.

DISCUSSION

It is known that during walking on flat ground, GAS increases muscle activity from mid-stance to pre-stance, contributing to forefoot rocker function[30]. The GAS not only modulates the time to swing the contralateral lower limb by slowing the drop of the COG but also helps in producing forward acceleration by promoting push-off force. In the acceleration condition, the task was to rapidly increase gait speed to maximum. Therefore, the GAS muscle activity increased significantly from the beginning of the acceleration. In a previous study, the GAS was shown to be involved in increasing gait speed[31], which is consistent with this study's findings. The forward propulsive force is generated mainly by storing and releasing energy within the Achilles tendon[31]. Conversely, accelerated gait is associated with increased positive torque around the ankle joint[16, 32] and decreased ankle plantar flexor activity for shock absorption[33–36], which have been associated with reduced ankle plantar flexor activity for shock absorption[33]. Reducing the shock absorption during the early stance phase contributes to an increase

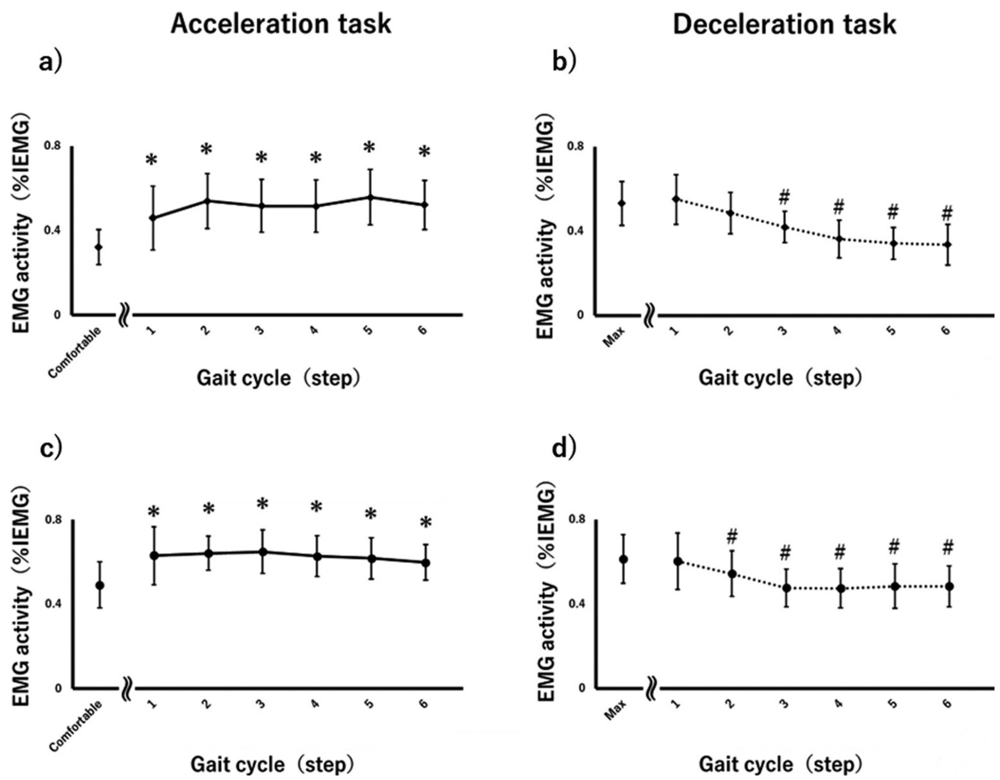

Fig. 5. Comparison of integral values of total muscle activity (%IEMG) between walking speed conditions.
a) and c) represent acceleration tasks, and b) and d) represent deceleration tasks (N=15, error bar=standard deviation)
*: Values that differed significantly from those obtained during comfortable walking.
#: Values that differed significantly from those obtained during maximum walking.
EMG: electromyography.

in gait speed[33]. Moreover, an increase in gait speed to maximum velocity results from increased ankle plantar flexor activity and timing adjustments in ankle angle changes as gait speed increases[37]. Based on these factors, it was possible that the TA and GAS muscle activities occurred at the earliest possible time. Additionally, BF and VM are responsible for controlling COG downward fall by generating the hip flexion moment via eccentric contraction during the initial ground-load response phase[30]. These muscle activities tilt the lower leg forward, with the heel as the axis of rotation, drawing the femur forward and producing a forward propulsive force. In the previous study, as gait speed increases, the joint angles of the hip and ankle joints change, which is associated with energy absorption and production[16]. Our findings suggest that the decrease in shock absorption by the ankle joints immediately after the sound cue may lead to a rapid increase in gait velocity and cause an increase in the VM and BF muscle activities that contribute to shock absorption. Temporal specificity is thereby generated, resulting in a smooth and gradual increase in acceleration.

In the RMS results, only az_RMS showed a significant increase in the first gait cycle of comfortable walking, indicating that vertical COG sway was the earliest to increase. Vertical sway fluctuation may occur earliest during acceleration since GAS increases stride length. BF and TA muscle activities adjust the external joint moment generated by increasing stride length, thereby gradually increasing gait speed. It has been reported that, RMS depends on gait speed since COG sway increases with increasing speed[38, 39]. Furthermore, stride length and cadence increase linearly with the increase of gait speed[40]. In these factors, we speculate that under the acceleration condition, as in the previous study, the COG trajectory was linear because stride length and cadence increased with the increase in gait speed. Our study showed significant increases in ax_RMS and ay_RMS, indicating that an increase in COG movement velocity occurs in the left and right directions for a more linear COG trajectory. Moreover, it took the same time for the forward COG acceleration to increase significantly. These results suggest that an active increase in gait speed may not involve a uniform change in each axis, but rather an axis-specific time-series change.

In deceleration adjustment, previous studies have analyzed only a small number of steps. In this study, together with the previous knowledge of muscle activity, allowed us to analyze the changes in COG sway with increasing gait speed over a larger number of steps and capture the specific changes in this area. Previous studies have analyzed muscle activity when walking is stopped in response to a cue, although the gait speed differed[41–43]. Hase et al. used a task in which healthy adults were given a cue during comfortable gait that instructed them to stop suddenly[42]; when gait speed was rapidly reduced, a rapid increase in the activity of the hip extensors, knee extensors, and ankle plantar flexors in the early stance phase caused braking of the hip flexion moment and forward acceleration force. Additionally, eccentric contraction of the ankle dorsiflexors during the late stance phase inhibited push-off. These muscle activities decrease step length in the contralateral lower extremity, resulting in rapid gait stopping[42]. Unlike previous studies, in our study, the task was to continue walking rapidly from maximum gait speed to a comfortable speed, i.e., to walk while slowing down and adjusting the speed. Consequently, we found that VM, TA, and GAS activities did not increase significantly after the cue compared with those in the control but rather decreased gradually with the progress of gait cycles. However, BF activity increased significantly in the first gait cycle immediately after the cue, similar to that reported previously[42]. Our results suggest that, in contrast to the stopping motion, during continuous adjustment of gait speed while decelerating, adjusting the hip flexion moment in the early stance phase may decrease the forward propulsion of COG. In the knee and ankle joints, there was no increase in muscle activity greater than that observed during maximal walking, suggesting that progressive adjustments to decreased muscle activity reduce the forward propulsive force generated by the braking of the hip flexion moment in the early stance phase. The ax_RMS and ay_RMS significantly decreased first, indicating that BF activity first causes an adjustment from maximal gait to rapidly decreasing forward acceleration. Furthermore, a decrease in the speed of COG movement occurred from the left and right directions, and finally, it is speculated that vertical COG movement decreased as stride length decreased. These results suggest that the kinematic control of walking while decreasing speed may differ from that of stopping the walking motion.

This study aimed to analyze kinematic factors during active gait speed change and clarify EMG changes in lower limb muscles during COG acceleration. This study revealed factors necessary for gait speed adjustment based on changes in COG acceleration and muscle activity during active acceleration and deceleration. Our findings suggest that acceleration and deceleration of walking are regulated by factor-specific kinematic processes. Based on our results, we were able to capture changes in new kinematic parameters that were not previously analyzed using an instrument that can be easily applied in clinical practice over a longer time from the onset of gait velocity change. This study contributes to the development of rehabilitation assessment and physical exercise therapy strategies related to the gait speed adjustment required in daily living, which have not been elucidated by previous methods.

In this study, we used a 3-axis accelerometer and surface EMG systems to conduct a more clinically relevant analysis of gait speed acceleration/deceleration control. However, we were unable to measure real-time changes in gait speed; thus, it is challenging to confirm what kind of change in gait speed actually caused the changes in COG acceleration observed in this study. Furthermore, although we used linear gait speed acceleration and deceleration tasks here, there are many situations in daily life including walking in a straight line, turning[13], and avoiding obstacles, in which gait speed must be adapted depending on the surrounding environment. Therefore, these situations should be analyzed in combination with changes in COG sway. To further validate the changes in kinematic parameters obtained in this study, comparative studies should include young, healthy, and older adults, as well as other participants whose walking ability is expected to worsen.

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Conflict of interest

None declared.

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