ORIGINAL

Characteristics of the stand-to-sit motion in healthy older women : Evaluation of sitting impact by measurement of ground reaction forces

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Abstract : Objectives : The aims of this study were to examine the biomechanics of StandTS movements in older adults and to identify their optimal StandTS motion by measuring sitting impact forces. Methods : Healthy older women (n = 17) and healthy young women (n = 18) were asked to perform SitTS and StandTS motions at a natural speed using a chair. We measured the ground reaction forces from the participants' feet and the chair, the angle of the trunk and ankle, vertical velocity, and postural muscle activities using a force plate, motion analyzer, and electromyography, respectively. Results : Sitting impact force was significantly greater in the older women than in the young women during the StandTS motion. There was a significant difference between the trunk angle and the ankle angle during the StandTS motion and sitting impact force had a significant negative correlation with the ankle joint motion in the older women. Conclusions : The ankle joint strategy was characterized by body sway resembling a single-segment-inverted pendulum and suggests that this response is less developed in the older adult. These results indicate that the ankle joint strategy may be an important factor involved in the sitting impact force. J. Med. Invest. 69 : 278-286, August, 2022

Keywords **:** *older adults, stand-to-sit, kinetics, sitting impact, postural configurationg*

INTRODUCTION

Sit-to-stand (SitTS) and stand-to-sit (StandTS) are representative motions in daily life. The StandTS motion has been reported to be frequently performed by community-dwelling older adults living in Europe, on average 39 to 71 times per day (1-3). The StandTS motion may become unbalanced in older adults due to loss of lower limb muscle strength and poor balance. A decline in the standing and sitting abilities of older adults leads to higher risk of falls (4) and the need for long-term care (5), so maintaining these abilities seems to be important for continued independent living.

The StandTS motion requires postural control accompanied by backward movement of the center of mass (COM). The COM displacement range of this backward movement is narrower in older adults than in young adults (6). Thus, older adults are more likely to lose their balance when performing the StandTS motion. Because the StandTS motion is performed with the assistance of gravity, failure to properly carry out this motion could cause serious consequences. Loss of balance may lead to poorly controlled acceleration, which would result in a larger sitting impact to the spine and pelvis. This could cause osteoporosis-related kyphosis, especially in frail older adults (7).

Many studies have investigated the kinematics and kinetics of the SitTS motion in older adults (8-11) because the biomechanical demands of this motion are larger than those of the StandTS motion and the ability to carry out the SitTS motion could be easily lost with aging. Several studies have reported that body trunk flexion angle (12) and angular velocity (13, 14) are smaller during the StandTS motion in older adults than in young adults. However, few studies have investigated the effect of different modes of operation on the sitting impact force when performing the StandTS motion. The impact force on the buttocks when falling backwards has been reported to be 4–6 kN (15, 16), which is enough force to deform a vertebral body (17, 18). However, many older adults are unaware of vertebral fracture (19), so it is necessary to investigate minor impacts that occur in daily life.

In this study, we aimed to examine the influence of biomechanics on the sitting impact force on the buttock during the StandTS motion and to identify the optimal StandTS motion for decreasing the associated health risks in older adults. To achieve these aims, we performed motion analyses of the trunk and legs and compared the kinetics and kinematics between older adults and young adults.

MATERIALS AND METHODS

Participants

We recruited healthy older women (n = 17) and healthy young women (n = 18). The older women were aged 60 years or over and comprised the older group ; they were participants of exercise classes organized by a local public foundation. The young women were aged 24-35 years and comprised the young group ; they were medical staff and medical students from the authors' institutions and volunteered to participate in the study. The exclusion criteria were a history of back pain or spinal problems requiring medical treatment ; a history of hip, knee or ankle joint pain ; or a reported pregnancy. Participant characteristics are shown in Table 1. All participants provided written informed consent. This study was approved by the Medical Ethics Committee of Tokushima University, Japan (reference number : 3355).

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Table 1. Participant characteristics

	Young (n = 18)	Older (n = 17)	P-value
Age (years)	25.2 ± 3.6	72.9 ± 6.2	< 0.001
Height (cm)	159.5 ± 4.5	152.9 ± 6.6	0.002
Weight (kg)	53.6 ± 7.2	51.6 ± 4.1	0.327
BMI (kg/m^2)	21.1 ± 3.2	22.2 ± 2.7	0.274
Muscle strength			
Grip (kg)	27.7 ± 4.7	24.8 ± 3.0	0.037
IKEF/weight (%)	60.7 ± 13.1	48.3 ± 16.5	0.020
ROM (degrees)			
Knee extension	0.0 ± 0	1.2 ± 2.8	0.104
Ankle dorsiflexion	37.9 ± 8.7	33.8 ± 7.0	0.128

BMI : body mass index, *IKEF* : isometric knee extension force, *ROM* : range of motion
Values are means ± standard deviation

Experimental settings and task

Participants sat on an adjustable chair with a backrest and no armrests. We adjusted the chair's height so that the participants could sit with their thighs horizontal, lower legs vertical, and feet flat on the ground. The average seat height was 38.4 ± 1.1 cm in the older group and 39.0 ± 1.2 cm in the young group. The chair frame was made of aluminum and the seat surface was polyethylene coated with ethylene-vinyl acetate. The participants were asked to stand up to a vertical position then sit back down until their back touched the backrest with their arms folded. They were instructed to perform these motions 5 times at a natural speed.

Data collection

We measured height, body weight, muscle strength (hand grip and isometric knee extension force), and range of motion (knee and ankle joint), which affect ground reaction forces and motion analysis data. Hand grip was recorded using a hand-held dynamometer (Grip-D ; TAKEI, Niigata, Japan). Isometric knee extension force was recorded using a hand-held dynamometer (μ-TAS F-1 ; ANIMA, Tokyo, Japan). Range of motion of the knee and ankle joints was measured using a goniometer.

Ground reaction forces were recorded at 1000 Hz using 4 embedded force plates (model OR-06 ; Advanced Mechanical Technology, Inc., Watertown, MA). The chair was placed on force plates 3 and 4 and the participant placed their feet on force plates 1 and 2 (Fig. 1a). The standing width was defined as the width of the shoulders. The sum of the force values from plates 1 and 2 was defined as the foot floor reaction force. Two ground reaction force curves were calculated from the foot floor reaction force : anteroposterior (Fy feet) and vertical ground reaction force (Fz feet) curves (Fig. 1b). The sum of the force values from plates 3 and 4 was defined as the chair ground reaction force. The magnitude of the chair ground reaction force reflected the impact of the seat of the chair with the buttocks at the time of sitting (sitting impact ; Fig. 1c). Sagittal and vertical chair ground

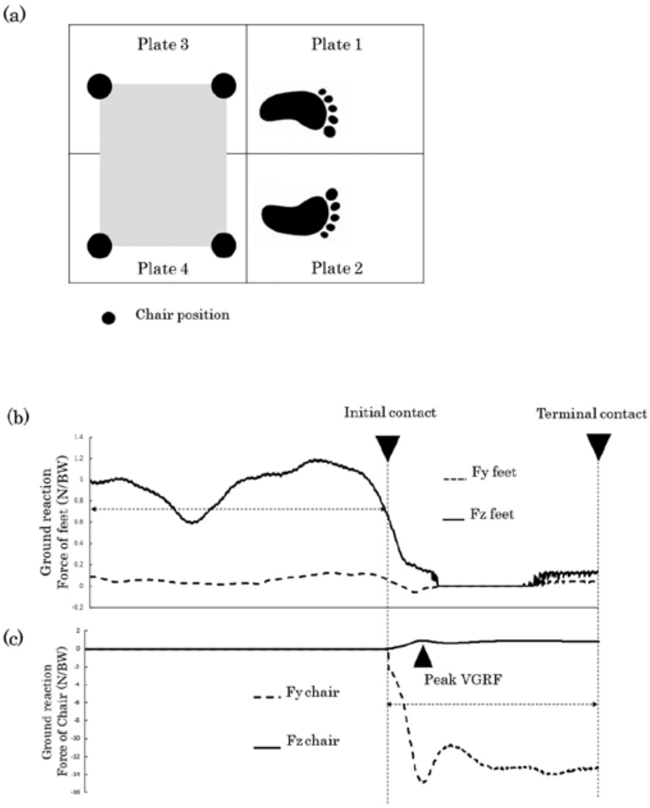

Fig 1. Experimental setting, ground reaction force and synchronization between the stand-to-sit parameters
(a) Feet and chair position on the force plates.
(b) Typical example of a time curve for the stand-to-sit vertical and anteroposterior forces recorded using force plates 1 and 2 from the start of movement to initial contact with the chair. Fy feet is the anteroposterior ground reaction force curve of the feet, Fz feet is the vertical ground reaction force curve of the feet. BW : body weight.
(c) Typical example of a time curve for sitting impact force recorded by force plates 3 and 4 from the instant the buttocks contacted the chair to the instant the trunk was maximally extended. Fy chair is the anteroposterior ground reaction force curve of the chair, Fz chair is the vertical ground reaction force curve of the chair. The time point when the value exceeded 1 N was taken as initial contact.

reaction force (Fy chair and Fz chair, respectively) curves were calculated from when the buttocks contacted the chair (initial contact) to when the trunk was maximally extended and the back was in contact with the backrest (terminal contact). The descent time was defined as the period from the start of movement to initial contact with the chair.

Motion analysis was performed during the StandTS motion using the Vicon MX system® (Vicon Motion Systems, Oxford, United Kingdom). Kinematic data were collected at 250 Hz using a passive 8-camera system (Vicon MX T20 ; Vicon Motion Systems). Reflective markers (diameter, 14 mm) were placed on the participant's left side at the acromion, lower sacrum, greater trochanter, fibula head, and lateral malleolus (Fig. 2a) (12). The 8 MX cameras each captured the motions of the markers and Nexus 1.4 (Vicon Motion Systems) processed the motion data as stick images based on marker positions in three dimensions. The trunk angle was defined as the angle between the vertical and the line connecting the markers of the acromion and the greater trochanter (Fig. 2a). The shank angle, which reflects ankle joint motion (20), was defined as the angle between the vertical and the line connecting the fibula head and the lateral malleolus (Fig. 2a). Two segmental angle curves from start to initial contact were calculated from the motion data : trunk angle and shank angle curves (Fig. 2b). The vertical velocity of the marker attached to the lower sacrum was defined as the descent velocity (Fig. 2c). We derived the vertical deceleration rate (Δvelocity) during descent by using the following formula :

$$\Delta velocity = \frac{\text{(peak vertical descent velocity-vertical descent velocity at impact)}}{\text{(peak vertical descent velocity)}}$$

Muscle activity was measured on the right side of the lumbar paravertebral muscles (PVM, Fig. 3a), vastus lateralis (VL, Fig 3b), gastrocnemius muscle (GC, Fig 3c), and tibialis anterior muscle (TA, Fig 3d) using surface electromyography (EMG). The electrodes used in this study were 1-cm-diameter metal plates and were placed with 3 cm between the centers of the two electrodes. The skin where the electrodes were placed was shaved and cleaned with alcohol. Surface EMG signals were recorded using the Telemyo 2400T system (Noraxon USA, Scottsdale, AZ) and stored on a PC for further processing and analysis. The recording frequency was 1000 Hz for all channels. An integrated bandpass filter of 10–500 Hz was applied to the EMG signals to avoid noise artifacts. The EMG and Vicon system were synchronized using an external 5-V square-wave trigger voltage. Surface EMG of each muscle was recorded from the start of movement to initial contact with the chair.

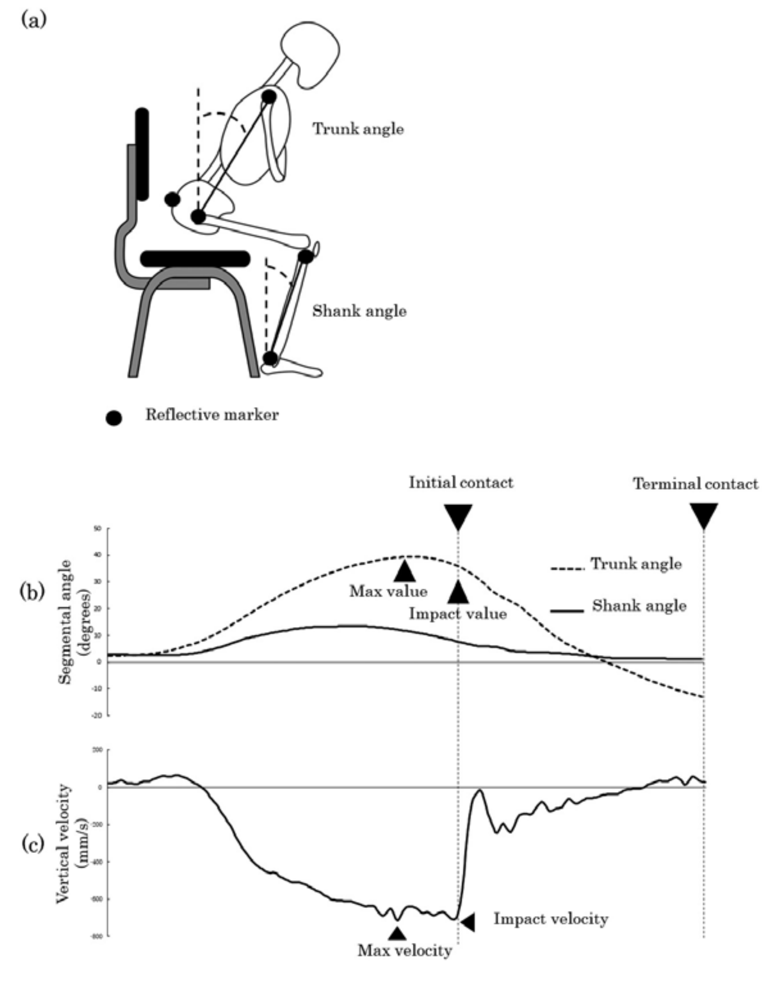

Fig 2. Marker positions, and synchronization between the stand-to-sit parameters
(a) Reflective marker position and segmental angle.
(b) Angle of the body segment calculated based on the position of markers from the start of movement to initial contact with the chair. The maximum bending angle and the angle at initial contact were extracted for the trunk and shank.
(c) Vertical descent rate was recorded using the marker on the lower sacrum. The maximum descent speed and the speed at the initial contact were extracted from the recorded curve from the start of movement to initial contact with the chair.

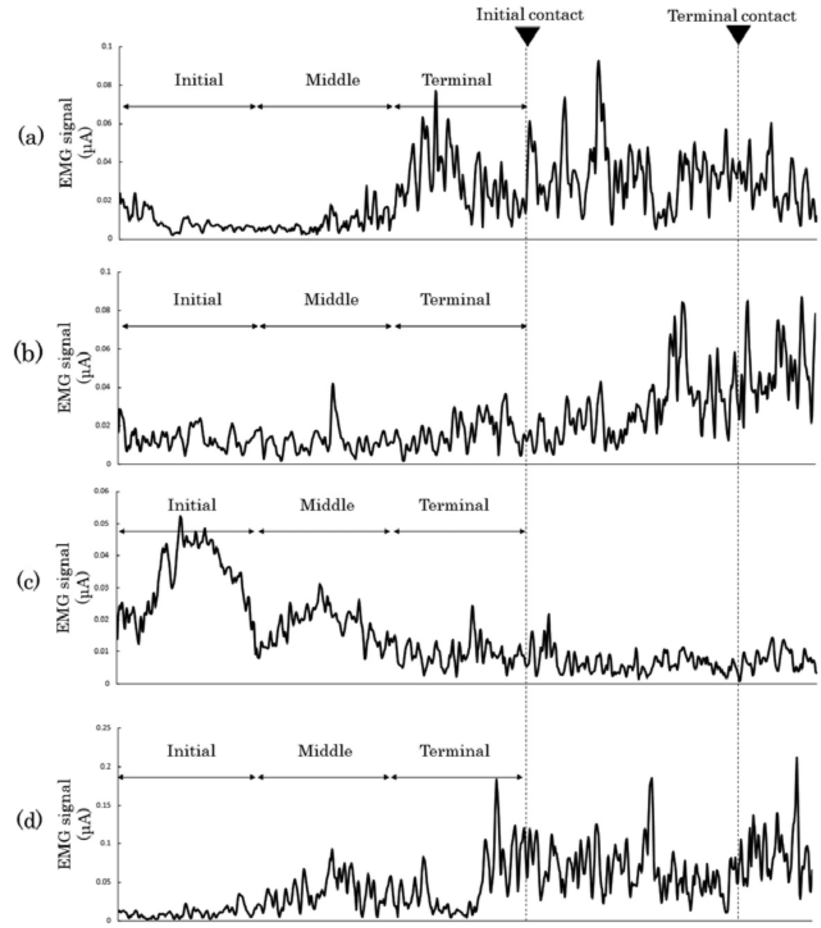

Fig 3. Synchronization between the typical surface electromyography recordings and platform data
Muscle activity of the (a) lumbar paravertebral muscles, (b) vastus lateralis, (c) gastrocnemius muscle, and (d) tibialis anterior muscle using surface electromyography. The period during the stand-to-sit motion was divided equally into initial, middle, and terminal phases.

Data analysis

To account for the learning of the movement and fatigue, we analyzed the data from the second to fourth motions. The ground reaction force data used for examination were normalized by body weight. Data on Fy chair and Fz chair were normalized by the time from initial contact with the chair to terminal contact ; other data from the ground reaction force and motion analysis were normalized by the time from the start of movement to initial contact. The same period was divided into three equal parts in the EMG data, and for each of these parts, the integrated EMG (iEMG) was calculated and normalized by the value of maximum voluntary contraction.

Statistical analysis

Data are presented as the mean ± standard deviation. Characteristic data, kinematic and kinetic data were compared between the two groups using Welch's t-test. Correlations between peak chair ground reaction force per body weight (Fy chair and Fz chair) and all kinematic parameters in the older group was examined using Pearson's correlation coefficient. All statistical analyses were performed using EZR version 1.37 (Saitama Medical Center, Jichi Medical University, Saitama, Japan), which is a graphical user interface for R (The R Foundation for Statistical Computing, Vienna, Austria). Statistical significance was set at $p < 0.05$.

RESULTS

Ground reaction force parameters

The chair and feet ground reaction forces per body weight are plotted against normalized StandTS duration in Fig. 2. There was no significant difference in peak Fy feet (Fig. 4a) between the two groups ($p = 0.237$, Table 2), but there was a significant difference in peak Fz feet (Fig. 4b, $p = 0.014$, Table 2). The peak of the Fy chair (Fig. 4c) curve was significantly higher in the older group than in the young group ($p = 0.046$, Table 2). Similarly, the peak of the Fz chair (Fig. 4d) curve was significantly higher in the older group than in the young group ($p = 0.020$, Table 2).

Motion analysis parameters

The initial trunk angle was approximately the same in both groups and increased further with movement in the older group than in the young group. Thus, the trunk angle at maximum and at initial contact was significantly higher in the older group than in the young group (at maximum, $p < 0.001$; at initial contact, $p = 0.012$, table 2). In contrast, the shank angle increased further with movement in the young group than in the older group. Thus, the shank angle at maximum and at initial contact was significantly higher in the young group than in the older group (at maximum, $p = 0.015$; at initial contact, $p = 0.001$, table 2). The vertical descent velocity did not differ significantly between the two groups (Table 2). However, the shapes of the

Table 2. VGRF of the chair, and reflective marker and EMG data

	Young (n = 18)	Older (n = 17)	P-value
Peak VGRF			
Fy feet	0.14 ± 0.03	0.13 ± 0.03	0.237
Fz feet	1.20 ± 0.09	1.29 ± 0.11	0.014
Fy chair	−7.73 ± 7.67	−16.21 ± 14.98	0.046
Fz chair	1.10 ± 0.19	1.33 ± 0.26	0.020
Descent time (s)	0.66 ± 0.15	0.72 ± 0.21	0.360
Segmental angle			
Trunk, max (°)	33.2 ± 5.6	41.9 ± 7.6	<0.001
Trunk, at sitting impact (°)	30.2 ± 5.5	36.5 ± 8.1	0.012
Shank, max (°)	13.7 ± 4.1	10.6 ± 2.9	0.015
Shank, at sitting impact (°)	11.0 ± 4.2	6.1 ± 3.8	0.001
Vertical descent velocity			
Peak (mm/s)	−823.8 ± 175.0	−733.3 ± 139.3	0.106
Impact (mm/s)	−738.3 ± 137.9	−694.8 ± 157.3	0.398
Δvelocity (%)	9.3 ± 9.8	5.7 ± 7.3	0.240
EMG in initial phase (%)			
PVM	13.3 ± 10.0	19.1 ± 9.4	0.153
VL	11.4 ± 13.3	17.0 ± 11.1	0.237
GC	18.9 ± 10.4	16.6 ± 11.8	0.609
TA	9.9 ± 6.8	17.4 ± 9.3	0.010
EMG in middle phase (%)			
PVM	31.1 ± 22.7	29.7 ± 11.8	0.725
VL	13.1 ± 8.5	20.0 ± 8.1	0.010
GC	13.8 ± 6.9	11.4 ± 4.2	0.135
TA	17.5 ± 6.8	24.7 ± 11.5	0.028
EMG in terminal phase (%)			
PVM	51.3 ± 40.3	30.2 ± 10.4	0.049
VL	29.6 ± 12.2	29.4 ± 6.7	0.849
GC	14.4 ± 9.5	12.0 ± 6.1	0.323
TA	27.4 ± 11.1	26.5 ± 10.9	0.710

VGRF : vertical ground reaction force ; *EMG* : electromyography ; *PVM* : paravertebral muscles ; *VL* : vastus lateralis gastrocnemius muscle ; *GC* : gastrocnemius muscle ; *TA* : tibialis anterior muscle
Values are means ± standard deviation

curves were different and the young group more quickly reached the maximum descent velocity. Furthermore, the young group slightly decelerated just before sitting, whereas the older group did not. Fig. 5a–c shows data on segmental angles and vertical descent velocity corrected for the StandTS duration.

Electromyography parameters

The average values of iEMG (%) during the initial, middle, and terminal phases were calculated. During the initial phase, TA activity was significantly higher in the older group than in the young group ($p = 0.010$, Table 2). During the middle phase, VL and TA activities were significantly higher in the older group than in the young group (VL, $p = 0.010$; TA, $p = 0.028$, Table 2). During the terminal phase, PVM activity was significantly higher in the young group than in the older group ($p = 0.049$, Table 2).

Correlation between peak sitting impact force and each parameter in the older group

Table 3 shows the correlations between peak sitting impact force (Fz chair) and each kinematic and kinetic parameter. Only the shank angle at sitting impact was significantly associated with sitting impact force in the older group (r = -0.534, $p = 0.027$).

Table 3. Correlation between the ground reaction force and each parameter in the older group

	Older group (n = 17)	
	r	p-value
Descent time (s)	0.113	0.667
IKEF/BW (%)	−0.356	0.16
Segmental angle		
Trunk, max (°)	−0.262	0.31
Trunk, at sitting impact (°)	−0.254	0.325
Shank, max (°)	−0.465	0.059
Shank, at sitting impact (°)	−0.534	0.027

IKEF : isometric knee extension force ; *BW* : body weight

DISCUSSION

We compared sitting impact forces and movements during the StandTS motion between older women and young women to investigate the factors that influence the sitting impact force and to promote safe sitting in older adults. To our knowledge, this is the first study to show differences in the sitting impact force and movement parameters during the StandTS motion between healthy older adults and healthy young adults.

We observed several differences in the StandTS motion between the younger and older adults in this study. Compared with the StandTS motion in young adults, this motion in older adults was characterized by a larger sitting impact force, and a higher trunk angle and smaller shank angle at maximum during StandTS motion and at initial contact with the chair. There were also differences in muscle activity, with the older group showing higher TA activity during the initial phase, higher VL and TA activities during the middle phase, and lower PVM activity during the terminal phase. The only significant association observed between sitting impact force and these electromyography parameters was between the shank angle at sitting impact and the sitting impact force in the older group.

This study showed that the sitting impact force was approximately 20% greater in the older group than in the young group. Chen reported that the sitting impact was 0.7–0.8 times body weight in stroke patients (21). In the present study, the sitting impact was approximately 1.3 times body weight. We assume that stroke patients take care in sitting down because they are

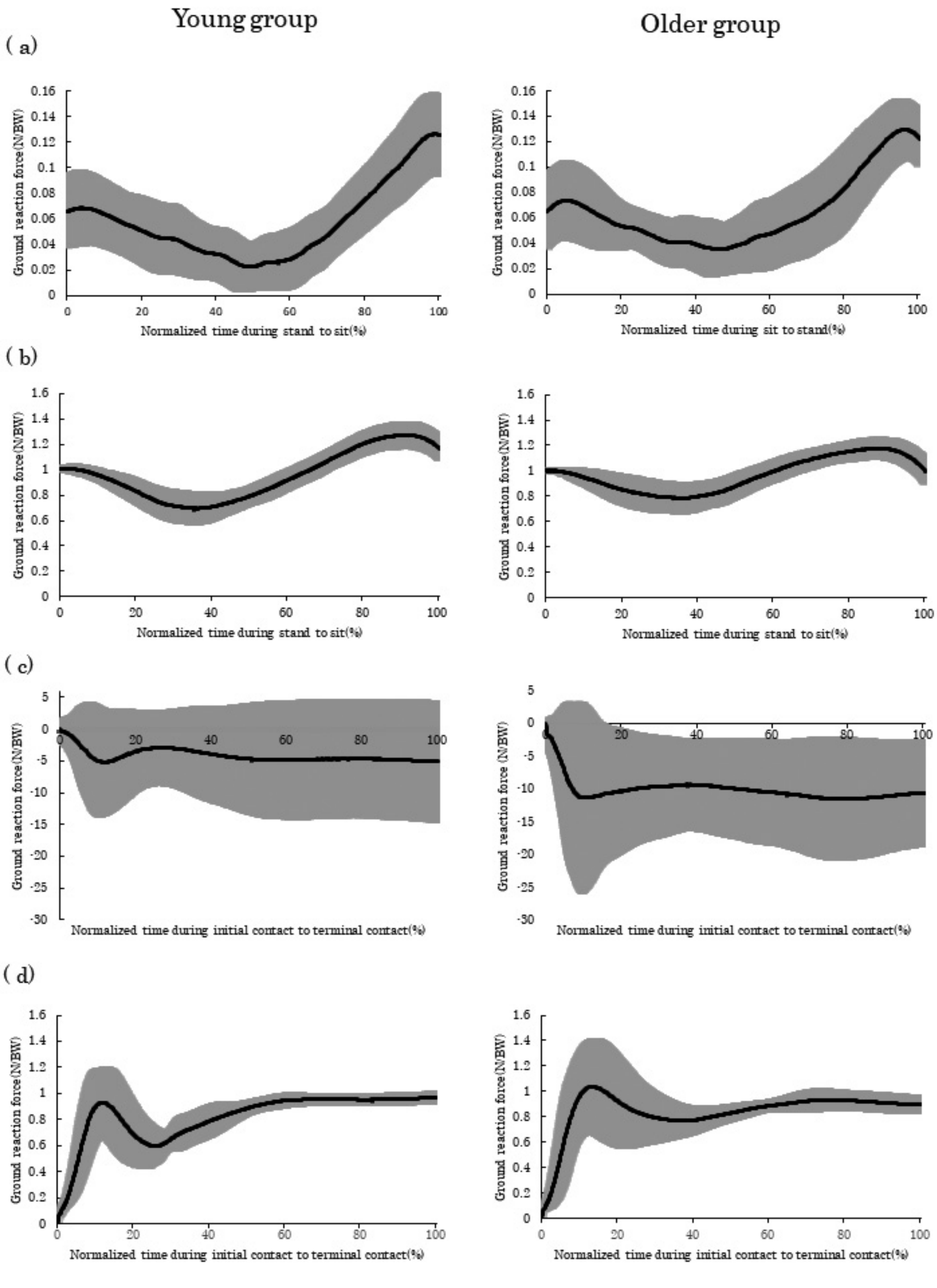

Fig 4. Ground reaction force stand-to-sit parameters
(a) Means ± standard deviation (SD) of normalized Fy feet data. (b) Means ± SD of normalized Fz feet data. (c) Means ± SD of normalized Fy chair data. (d) Means ± SD of normalized Fz chair data. BW : body weight.

aware that their legs are weak and their balance is poor. In addition, in the present study, we asked the participants to sit in a chair so that their back touched the backrest, such that the center of gravity would shift further backward, causing a greater sitting impact because of poorer control.

Some studies have reported that trunk anteversion during the StandTS motion results in a sitting impact pressure of approximately 1.5 kN (22) and that the sustained pressure causing deformation of the vertebral body is 2.3–2.7 kN (17, 18). A sitting impact force of 1.3 times body weight as observed in this study could be transmitted from the buttocks to the vertebral column, which could cause vertebral fracture in older adults with severe osteoporosis.

Dubost *et al*. reported that the trunk anteversion angle is smaller and the lower leg anteversion angle is larger in older adults than in younger adults (12). Their results were opposite to those in the present study. In our experimental setting, the sitting movement was defined by bringing the back into contact with the backrest. Therefore, it is necessary to compare the COM in backward movement with that in normal sitting movement. The range of the COM in backward movement is narrower in the standing posture in older adults than in young adults (6). Furthermore, it has been reported that the contribution range of the ankle strategy is narrow in the dynamic stability of older adults and that the range may be extended by control via the hip strategy. In contrast, the range that can be controlled via the ankle strategy is wider in young adults than in older adults (23, 24). We assume that in this experimental setting, the COM had

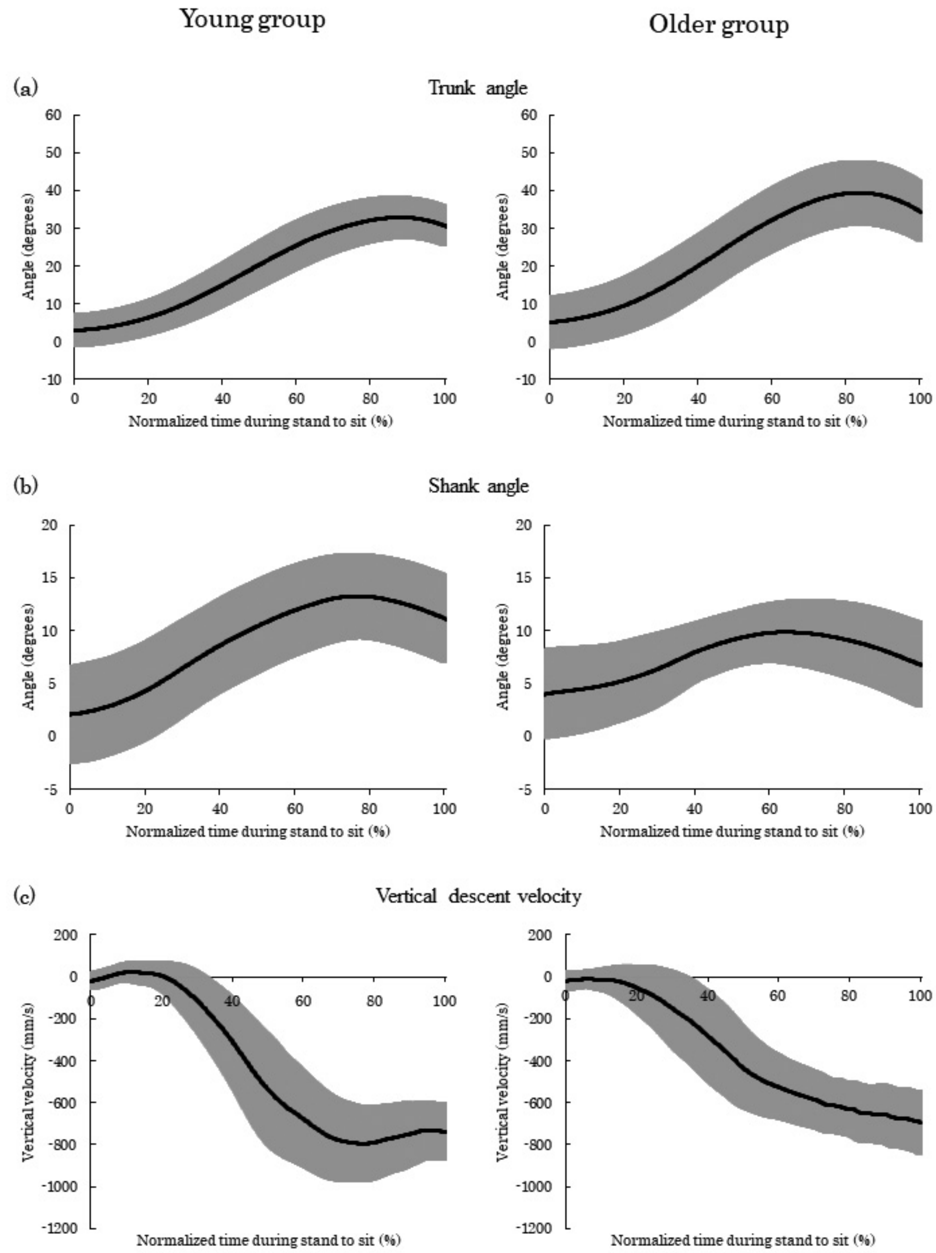

Fig 5. Segmental angles and vertical descent velocity of the stand-to-sit parameters
(a) Means ± standard deviation (SD) of normalized trunk angle data. (b) Means ± SD of normalized shank angle data. (c) Means ± SD of normalized vertical descent velocity data.

to be moved backward beyond the controllable range of the ankle strategy in the older group. As a result, the trunk anteversion angle increased in the older group and decreased in the young group, which was caused by the use of the hip strategy and the use of the ankle strategy, respectively. The results of electromyographic analyses may support this assumption. The muscle activities during the initial and middle phases of the StandTS motion were reflected in the movement of the segmental angles. Because the older group used the hip strategy to move their center of gravity backward, the TA and VL muscles might have generated larger forces to counteract the backward motion of the center of gravity. On the other hand, PVM activity may have become larger at the end of the StandTS motion in the young group because their motions were led by the ankle strategy and they used the hip strategy during the terminal phase. In addition, although the maximum speed of vertical descent was reached earlier in the young group than in the older group, the movement might have been performed mainly by the ankle strategy. By lowering the center of gravity early, even if it deviated from the supporting basal plane, the sitting impact would be lower in the young group than in the older group because the height of descent was lower.

Correlations were evaluated between the peak sitting impact force and various parameters of the older group and only ankle joint dorsiflexion angle at sitting impact was found to have a significant correlation. This result suggested that the StandTS motion was performed more easily and safely in the older adults who used the ankle joint strategy and kept their center of gravity within the support base. Therefore, motion instruction using an ankle strategy is considered important to lessen the sitting impact force in older adults.

There are some limitations in this study. First, all the participants were women. Differences in muscle strength and flexibility between the sexes might affect the center of gravity position as well as the kinematics and kinetics of the StandTS motion. These differences may not be negligible even among same-sex participants, and their influence on the StandTS motion should be examined in the future. Second, we could not measure the position of the COM in the participants in this study because the markers were attached to only one side of the body. In the future, the movement of the COM during the StandTS motion should be analyzed while considering the participant's posture. Third, there may have been surface EMG crosstalk due to the action of antagonist muscles. Therefore, because the soleus and TA are antagonist muscles in the StandTS motion, future studies should consider the possibility of EMG contamination.

CONCLUSION

The ankle joint strategy was characterized by body sway resembling a single-segment-inverted pendulum and suggests that this response is less developed in the older adult. The sitting impact force on the buttocks during the StandTS motion was significantly greater in older adults than in young adults. The ankle joint strategy was an important factor affecting the sitting impact force, and by improving this strategy, it may be possible to mitigate the sitting impact force.

DECLARATION OF COMPETING INTEREST

The authors have no conflicts of interest to declare.

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