



Full length article

Muscle activity and balance control during sit-to-stand across symmetric and asymmetric initial foot positions in healthy adults

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ABSTRACT

Background: Rising from a sit to a stand has biomechanical factors that are dependent on initial foot position. Little is known about the effect of initial foot position on leg muscle activation patterns during a sit-to-stand and balance maintenance of stance after a sit-to-stand.

Research question: What are the effects of different symmetric and asymmetric initial foot positions on leg muscle activation patterns and balance during and after a sit-to-stand?

Methods: Three symmetric (neutral; both ankles positioned under the knees at a 90° flexion; one-third; and two-thirds foot length posterior to neutral) and three asymmetric (neutral non-dominant leg with one-third back dominant leg, neutral non-dominant with two-thirds back dominant leg, and one-third back non-dominant leg with two-thirds back dominant leg) initial foot positions were tested. EMG of the lower extremity muscles and sagittal plane kinematic data were measured along with balance assessments in the anterior-posterior and medial-lateral axes.

Results: In the symmetric initial foot positions, a faster forward velocity of the body's center of mass was required for more anterior initial foot positions. Even though the hip extensors activated earlier to decelerate the forward velocity of the body's center of mass before rising, the greater forward velocity caused postural sway following completion of upright stance. In the asymmetric initial foot positions, the posterior leg supported more weight during the sit-to-stand, resulting in balance perturbations in the posterior leg. In the one-third back non-dominant leg with two-thirds back dominant leg asymmetric initial foot position, however, the weight-bearing symmetry was not different from the symmetric initial foot positions during the sit-to-stand. Postural stability after completion of uprising was also improved in this asymmetric initial foot position, showing greater but delayed onset of the tibialis anterior in the anterior leg during the momentum transfer phase.

Significance: With a neutral symmetric initial foot position, earlier onset of the hip extensors during eccentric lengthening contributed to decelerating the forward velocity of the body's center of mass for balance control during a sit-to-stand. With asymmetric initial foot positions, the weight distribution during a sit-to-stand can be increased by positioning both feet posterior to neutral foot position. Performing a sit-to-stand with this asymmetric initial foot position can improve postural stability after uprising. Thus, this foot position could be used in designing rehabilitation interventions for clinical populations and the frail elderly.

1. Introduction

Rising to a stand from a seated position is a ubiquitous task and an essential functional activity in daily life [1]. The sit-to-stand (STS) is a vital prerequisite to regaining mobility in many clinical and elderly populations. STS exercise is one of the most effective resistance training modalities to improve lower limb strength, especially knee extensor strength, which declines after 50 years of age at approximately 2–4% per year [2,3]. However, lifting off from a seat and maintaining balanced standing is a physically demanding task, due to the required

control and coordination of the lower extremities and simultaneous smooth movement of the upper body.

Many studies have suggested several strategies to improve STS performance. These strategies include adjusting chair height, increasing trunk flexion, and changing initial foot position (IFP) [4,5]. A symmetric posterior IFP has kinetic advantages over an anterior IFP due to the shorter distance between the center of mass (CoM) and the acting point of the ground reaction force (GRF) [6]. Additionally, the functional linkage between hip and knee extension from the biarticular muscles (rectus femoris and gastrocnemius) in a posterior IFP

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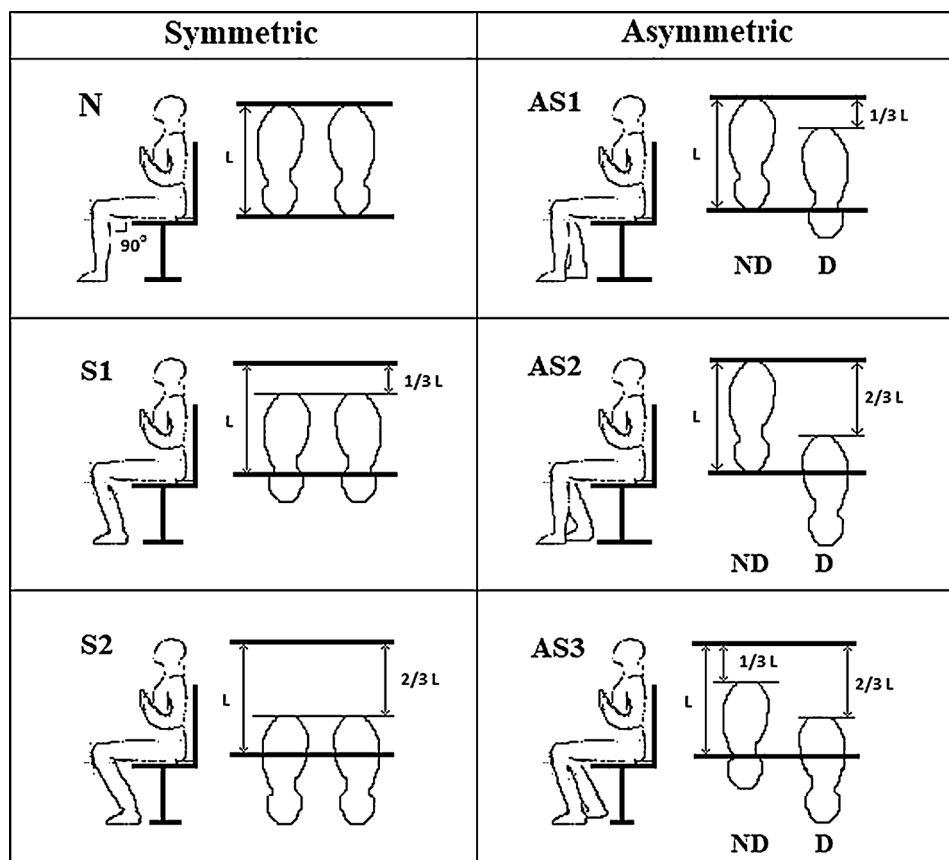


Fig. 1. Six different symmetric and asymmetric initial foot positions (IFPs) ND: Non-Dominant; D: Dominant).

contributes to transferring the vertical force from one joint to another efficiently to lift the body [7,8].

Although a symmetric posterior IFP provides these advantages, allowing structurally and functionally equivalent contribution of the lower limbs, an asymmetrical IFP is commonly used in daily life in healthy adults, particularly when transitioning from a stand to a walk. In an asymmetric IFP, the neuromuscular control system supports and corrects the initial unlevelled and asymmetrical body components against gravity [9]. In addition to the initial strategies for performing the STS, maintaining balance with control after rising is also a critical determinant of a successful STS [10]. A symmetric anterior IFP increases hip extension torque, whereas an asymmetric IFP increased the ankle plantarflexion and knee extension torques of the posterior leg while reducing the anterior leg joint extension torques [12]. Thus, positioning the affected foot posteriorly can reduce the vertical reaction force asymmetry during STS in patients with hemiparesis following stroke [11–13]. Lower limb muscle activity and onset times are modified by different symmetric IFPs, and there resulting postural adjustments [5,7]. However, differences between symmetric and asymmetric IFPs and their effects on leg muscle activation patterns and balance have not been compared together in one study in healthy individuals.

The purpose of this study was to investigate changes in leg muscle activation patterns during a STS across six different symmetric and asymmetric IFPs and to examine the effects of IFP on balance during the stabilization phase after STS completion. We hypothesized that trunk forward tilt angle from the vertical axis would be greater for anterior IFPs and that EMG activity of the hip extensors during STS would be higher with the IFP in a neutral compared to a posterior IFP because greater trunk flexion is required to move the body's CoM over the base of support requiring greater hip extension torque [14,15]. We also hypothesized that balance during the stabilization phase would improve with a more posterior IFP because of a shorter required

displacement of the body's CoM. The majority of CoM displacement during STS termination occurs in the sagittal plane [16]. Thus, we hypothesized that asymmetric IFPs would improve anterior-posterior (A–P) balance during the stabilization phase by increasing the A–P boundary of the base of support and allowing involvement of the anterior leg to correct A–P postural sway.

2. Methods

2.1. Participants

Twenty healthy individuals (10 males and 10 females, 24 ± 3 years) participated in this study. All procedures were approved by the University of Texas at Austin's Institutional Review Board and were in accord with the Helsinki Declaration of 1975.

2.2. Data collection

A Bagnoli EMG System (Delsys, Inc) was used for acquisition of EMG signals. Adhesive pre-gelled Ag/AgCl surface EMG electrodes (Delsys Inc., Boston, MA) were placed bilaterally on five lower limb muscles: rectus femoris (RF), biceps femoris (BF), gluteus maximus (GM), tibialis anterior (TA), and soleus (Sol). The positioning of the electrodes was in accord with Rainoldi et al [17].

After electrodes placement, the volunteers performed three maximum voluntary isometric contractions (MVIC) for each muscle. For the Sol and TA, MVICs were performed against a strap secured over the foot (ankle joint 90° ; knee joint 90°). For RF, knee extension MVIC was elicited with the participant seated on a chair (hip joint 90° ; knee joint 90°) with a cuff around his/her ankle. Maximum knee extension was produced against external resistance. For BF, the participant lay on a massage table in prone position and performed MVIC knee flexion at

45° knee flexion against manual resistance applied to the ankle. For GM, the participant was in prone position with legs extended and toes-pointed (hip joint 0°; knee joint 0°). They performed maximal hip extension against manual resistance at the ankle joints.

After the MVIC testing, 39 reflective markers were placed on the body according to the Vicon Full-Body Plug-In Gait Model. A 10-camera Vicon motion capture system (VICON Motion Systems Ltd, UK) was used to record body kinematics.

During the STS, the participants sat upright with their hands on their chest. They sat on an armless, backless height-adjustable bench with the back of the knees not touching the bench. The seat height for each participant was set as the distance from the center of the knee joint to the floor. Each foot was placed on a separate force plate (Bertec Corporation, Columbus, OH, USA) with the feet shoulder-width apart. A pressure-sensitive pad (Microgate, NY, USA) was placed on the seat to determine when the participant lifted off. Participants performed the STS at their preferred speed after the visual light cue was turned on. They performed three non-consecutive trials for each of the six IFPs (Fig. 1) in random order. The order was also randomized across participants. Following STS completion, they maintained standing for five seconds.

Symmetric Initial Foot Positions: Neutral position (N) was set with both ankles positioned under the knees at 90° flexion. Two baselines were set parallel to each other at the center of the heel and toes. The other two symmetric positions tested were with the feet positioned symmetrically at one-third (S1), and two-thirds (S2) of the participant's foot length posterior to the baselines.

Asymmetric Initial Foot Positions: For all asymmetric IFPs, the dominant leg (leg preferred for kicking a ball) was placed posteriorly. In asymmetric IFPs 1 and 2, the non-dominant leg was placed in N. The dominant leg was placed one-third (AS1) and two-thirds (AS2) foot length behind the baselines. In AS3, the anterior leg was placed one-third and the posterior leg was placed two-thirds foot length behind the baselines.

2.3. Data processing

The average of three trials for each IFP was calculated for all data analysis.

2.3.1. Kinematics

Real-time Vicon data during the STS was processed by Plug-In Gait Nexus 1.8.5 Software (Vicon, Oxford Metrics, UK) to estimate the body's CoM trajectory, lower limb joint positions, and trunk flexion angle. Moments of hip and knee joints were calculated using standard inverse dynamics by the Plug-In Gait Dynamic pipeline. All moments were normalized to body mass. The time point of full extension was defined as the time when hip angular velocity first reached 0°/sec [18].

2.3.2. Weight-bearing symmetry

Weight-bearing symmetry for all IFPs during STS were calculated using vertical ground reaction forces as shown by the following equation (adapted from Talis et al. [19]):

$$\text{Symmetry (\%)} = 100 - \frac{|\text{Dominant max} - \text{Non dominant max}| \cdot 100}{\text{Dominant max} + \text{Non dominant max}}$$

2.3.3. EMG during sit-to-stand

Surface EMG was filtered with a 5–500 Hz band-pass filter. A moving root mean square with a 100 ms window was applied as a digital smoothing algorithm. Then, the EMG signals and the area under the rectified curves were normalized to the MVIC EMG for each muscle. The onset time of activation for each muscle during the STS was determined as the time at which the EMG exceeded three standard deviations (SD) of the initial mean baseline of EMG activity [20].

2.3.4. Kinetic (force plate) measurements

The force plate recorded center of pressure (CoP) and ground reaction force (GRF) on the anterior-posterior (A–P), medial-lateral (M–L), and the vertical axes. Measures of the standard deviation (SD) of the body's CoM acceleration, SD (mm/s²), on the A–P and M–L axes were used to quantify postural sway during the stabilization phase before entering quiet standing. The peak GRF during STS was normalized to body mass (kg) [21]. The stabilization phase was defined as the time period between completing full leg extension and the beginning of the quiet standing phase. To find the starting point of quiet standing, a sliding window composed of 0.5 s intervals was used with 0.1 s increments. The starting point of the quiet standing phase was set as the time when the difference between the locations of the CoP and center of gravity was steady within the range of ± 0.5 cm [22]. The A–P displacement of CoM after completing the STS was used as the center of gravity trajectory.

To measure the sway area, the 95% confidence ellipse area enclosed by the points of the CoP path during the stabilization phase was calculated first, then the sway area was normalized to the time interval of the stabilization phase (mm²/sec). For the frequency spectrum of CoP, the resultant distance was obtained from the following equation (adapted from Prieto et al. [23]).

Resultant Distance [n] = $[\text{AP}[n]^2 + \text{ML}[n]^2]^{1/2}$ (n = 1,2,3,...,N; N is the number of data points included in the analysis, AP[n] = AP_o[n] - $\bar{\text{AP}}$, ML[n] = ML_o[n] - $\bar{\text{ML}}$)

AP_o and ML_o = The AP and ML CoP path relative to the origin of the force plate coordinate system, $\bar{\text{AP}}$ and $\bar{\text{ML}}$ = The mean CoP positions of the AP_o and ML_o; $\bar{\text{AP}} = 1/N \sum \text{AP}_o[n]$, $\bar{\text{ML}} = 1/N \sum \text{ML}_o[n]$. Resultant distance time series was the vector distance from the mean CoP, where AP and ML were the time series of the displacement of the CoP in the A–P and M–L axes respectively.

A Fourier transform was applied to the resultant distance to calculate the frequency spectrum. The frequencies in the 0.5–2 Hz range represent reactive control for balance corrections in response to transient body oscillations. Frequencies below 0.5 Hz are in the range of anticipatory control mechanisms during standing [24,25]. Frequencies above 2 Hz are due to random postural adjustments [26]. Thus, total spectral energy during the stabilization phase was divided into three frequency domains (0 – 0.5 Hz, 0.5–2 Hz and > 2 Hz) and the percentage of total for each domain was calculated.

EMG, force plate, and pressure pad data were sampled at 1200 Hz and kinematic data was sampled at 120 Hz. Matlab 9.3 (Matworks Inc., Natick, MA, USA) was used to filter and analyze the EMG, kinematics and balance data.

2.4. Statistical analysis

A one-way ANOVA was used to examine whether there are significant differences between the means of the variance (SD) of the three trials within each IFP and the variance (SD) between six IFPs for all parameters across all participants. A one-way repeated measures ANOVA with Tukey's post hoc analysis was used to compare kinematic parameters and EMG for all muscles during the STS, and balance parameters during the stabilization phase for both legs in all conditions. Paired t-tests were used to compare differences in trunk flexion angle and anterior displacement and velocity of body's CoM during the STS between all symmetric and asymmetric IFPs. Pearson's correlation coefficients (r) were used to estimate the correlation between displacement of the IFP and 1) anterior displacement of the body's CoM 2) trunk forward tilt-angle 3) forward velocity of the CoM, and between the forward velocity of the CoM and the onset time of the hip extensors. SPSS (Chicago, IL) was used for all statistical analysis with an alpha level of significance of p < 0.05 set A–Priori. All data are presented as mean ± SD in the text and tables.

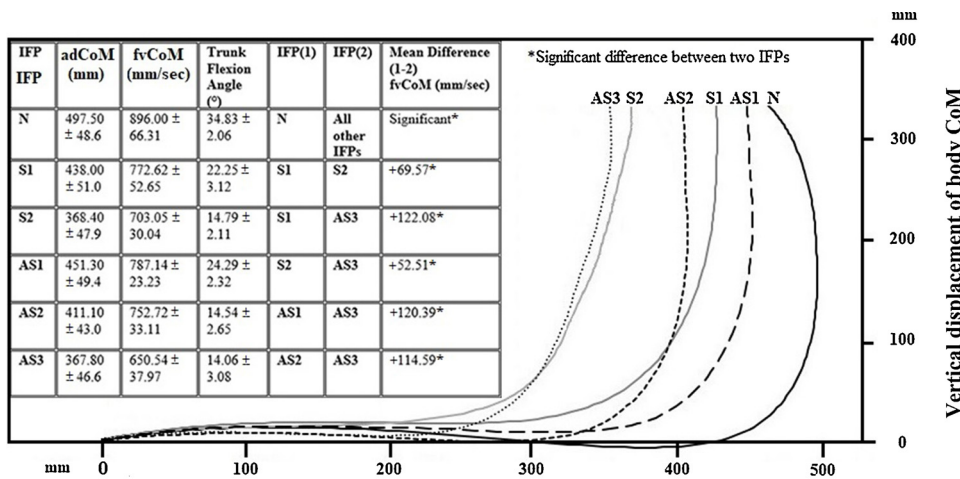


Fig. 2. Anterior displacement and forward velocity of the body’s center of mass, and forward trunk tilt angle during sit-to-stand between six symmetric and asymmetric initial foot positions across all participants. Mean values (± standard deviation) of these variables and statistics of the differences in forward velocity of the body’s center of mass between initial foot positions are shown in the table on this figure. Asterisks indicate a statistically significant difference. The origin of the coordinate system is the location of body’s center of mass in seated posture.

Anterior - posterior displacement of body CoM during sit-to-stand in each initial foot position

3. Results

For all parameters tested, the average of the variance (SD) of the three trials within each individual IFP across all participants was significantly smaller than the average of the variance (SD) between six IFPs ($p < 0.01$).

The average speed of STS performance in S2 and AS3 were higher than other IFPs (1057.04 ± 30.61 mm/sec and 1033.15 ± 23.15 mm/sec, respectively, $p < 0.05$) due to the shorter required anterior displacement of body’s CoM. There was no difference in the average speed of STS performance across all other IFPs ($N = 970.26 \pm 45.75$ mm/sec, $S1 = 996.91 \pm 32.95$ mm/sec, $AS1 = 973.90 \pm 47.97$ mm/sec, $AS2 = 989.21 \pm 45.32$ mm/sec).

3.1. Kinetics and kinematics

3.1.1. Displacement and velocity of center of mass / Trunk flexion / Distance between center of mass and base of support

Fig. 2 displays all the data for the anterior displacement of the CoM, forward velocity of the CoM and maximum trunk flexions angle. The anterior displacement of the body’s CoM and trunk forward tilt angle from the vertical axis were greater for anterior IFPs ($r = 0.70$, $p < 0.01$ and $r = 0.87$, $p < 0.01$ respectively). There was also a significant positive correlation between anterior positioning of the IFP and forward velocity of the body’s CoM ($r = 0.59$, $p < 0.01$).

There was a difference in the forward velocity of the CoM ($F = 4.128$, $p = 0.047$) and the trunk forward tilt angle ($F = 8.318$, $p < 0.01$) between symmetric and asymmetric IFPs. This is likely because N required greater trunk flexion than the other IFPs (Fig. 2).

There was a main effect for the distance between the CoM and the base of support across all IFPs ($p < 0.01$). It was longest in N (973.07 ± 7.09 mm) and shortest in S2 and AS3 (925.34 ± 10.73 mm, 932.04 ± 9.80 mm). The distance between CoM and the base of support in S1, AS1, AS2 (949.14 ± 9.86 mm, 953.39 ± 9.59 mm, 948.28 ± 6.33 mm, respectively) were not significantly different.

3.1.2. Weight-bearing asymmetry

Weight bearing symmetry between the anterior and posterior legs was significantly lower in AS1 ($91.18 \pm 2.7\%$) and AS2 ($84.09 \pm 5.3\%$) compared to all other IFPs. The weight bearing symmetry for AS3 ($98.30 \pm 1.6\%$) was not different from the symmetric IFPs ($N = 97.6 \pm 1.3\%$, $S1 = 98.7 \pm 1.0\%$, $S2 = 99.3 \pm 0.8\%$).

3.1.3. Ground reaction force

The vertical GRF during the STS was largest for the posterior leg of AS2. In AS1 and AS2, there was a significant difference between the anterior and posterior legs, whereas there was no difference between two legs in AS3. The A–P GRF to decelerate body’s forward momentum during the STS was largest in N and smallest in the anterior leg of AS1 and AS2 (Table 1). There was no difference in M-L GRF across all IFPs.

3.1.4. Balance during the stabilization phase

On the A–P axis, the SD of the acceleration of the body’s CoM, and the normalized sway area of CoP during the stabilization phase were largest in N and smallest in AS3 (Table 1). In AS1 and AS2, they were smaller in the anterior leg than the posterior leg. However, there was no difference in AS3 between the two legs. On the M-L axis, there was no difference across all IFPs.

In all IFPs, more than 90% of the total spectral energy (power) was accumulated in the 0 – 0.5 Hz domain. When comparing the percentage of total spectral energy in the 0 – 0.5 Hz domain across all IFPs, the anterior leg of AS2 and AS3 showed the highest percentage, whereas the posterior leg of AS2 and N had a smaller percentage (Table 1). In the 0.5–2 Hz domain, the percentage for the posterior leg of AS2 and N was more than 5% ($6.99 \pm 0.33\%$ and $6.70 \pm 0.44\%$, respectively), whereas power was less than 5% in all other IFPs. The posterior leg of AS2 and N also had more than 1% of the total power in the > 2 Hz domain ($1.59 \pm 0.45\%$ and $1.65 \pm 0.34\%$, respectively). Whereas, power over 2 Hz was less than 1% in all other IFPs.

3.2. EMG activity

3.2.1. Onset time

Fig. 3 displays the onset times for all muscles in all IFPs. The sequence of EMG onset times of muscle activity showed the same pattern across all IFPs. The TA activated first, prior to seat-off, followed by the leg extensor (RF), and then the hip extensors (BF and GM). The Sol was activated with the hip extensors and persisted for approximately one second into the stabilization phase. In N, the hip extensors (BF and GM) displayed two significantly different EMG peaks ($p < 0.01$). The first peak started close to seat-off and ended before hip extension began (time from “Go” signal to first peak = 0.13 ± 0.05 s). The second peak started at the beginning of hip joint extension (time from “Go” signal to second peak = 0.34 ± 0.09 s).

In AS1 and AS2, the onset time of the RF in the posterior leg was earlier than in the anterior leg, whereas the offset time of RF in the posterior leg was not different from the anterior leg, resulting in a longer activation time of the RF in the posterior leg compared to the

Table 1
GRF and maximum joint torque during sit-to-stand and balance in the stabilization phase.

IFP	Sit to stand phase				Stabilization phase		
	GRF(%BW, Fz)	GRF(%BW, Fx)	Knee extension torque (Nm/Kg)	Hip extension torque (Nm/Kg)	SD of CoM Acceleration (mm/s ² , Fx/m)	Normalized CoP Sway Area (mm ² /s)	% Spectral Energy (< 0.5 Hz)
N	134.21 ± 4.27 ^c	22.13 ± 2.32 ^{b,c}	0.80 ± 0.18 ^c	1.34 ± 0.36 ^c	74.38 ± 15.00 ^{b,c}	23.49 ± 7.50 ^{b,c}	91.00 ± 2.06 ^c
S1	129.34 ± 2.94	18.70 ± 1.08 ^a	0.91 ± 0.28	1.06 ± 0.36	48.95 ± 3.77 ^{a,c}	15.31 ± 6.19 ^{a,b}	92.05 ± 1.18 ^c
S2	126.64 ± 5.70 ^a	18.64 ± 0.43 ^a	1.13 ± 0.22 ^a	0.92 ± 0.15 ^a	36.28 ± 2.26 ^{a,b}	9.29 ± 2.62 ^{a,b}	94.75 ± 1.37 ^{a,b}
AS1 (A)	103.85 ± 3.37 ^{a,b,c,d}	14.96 ± 2.01 ^{a,b,c,d}	0.88 ± 0.32	1.02 ± 0.29	41.49 ± 2.86 ^{a,b,c,d}	9.94 ± 1.92 ^{a,b}	93.35 ± 1.46 ^a
AS1 (P)	135.45 ± 2.15 ^{b,c}	20.37 ± 2.62	0.98 ± 0.28	1.01 ± 0.20	51.58 ± 5.58 ^{a,c}	12.35 ± 3.58 ^{a,c}	93.26 ± 1.87 ^a
AS2 (A)	102.43 ± 1.90 ^{a,b,c,d}	16.94 ± 2.67 ^{a,d}	0.68 ± 0.15 ^{c,d}	0.91 ± 0.09 ^a	31.09 ± 1.54 ^{a,b,c,d}	7.12 ± 1.07 ^{a,b,d}	97.82 ± 0.69 ^{a,b,c,d}
AS2 (P)	143.48 ± 4.14 ^{a,b,c}	20.70 ± 2.75	1.23 ± 0.27 ^{a,b}	0.87 ± 0.13 ^a	75.98 ± 4.06 ^{b,c}	26.91 ± 6.64 ^{b,c}	91.34 ± 1.60 ^c
AS3 (A)	122.19 ± 6.26 ^a	20.70 ± 1.17 ^c	1.00 ± 0.15 ^a	0.92 ± 0.14 ^a	30.32 ± 2.26 ^{a,b,c}	6.21 ± 1.39 ^{a,b,c}	96.71 ± 1.14 ^{a,b,c}
AS3 (P)	127.33 ± 5.29 ^a	17.62 ± 2.44 ^a	1.02 ± 0.13 ^a	0.93 ± 0.22 ^a	31.54 ± 1.28 ^{a,b,c}	6.47 ± 1.82 ^{a,b,c}	97.08 ± 1.15 ^{a,b,c}

GRF (ground reaction force). F_z (vertical GRF), F_x (anterior-posterior GRF). %BW (percent of body weight). SD (standard deviation). m (mass), CoM (center of mass), CoP (center of pressure). A, anterior non-dominant leg; P, posterior dominant leg. There was a main effect for initial foot position.

- ^a Significantly different than N (p < 0.05).
- ^b Significantly different than S1 (p < 0.05).
- ^c Significantly different than S2 (p < 0.05).
- ^d Significantly different than the posterior leg (p < 0.05).

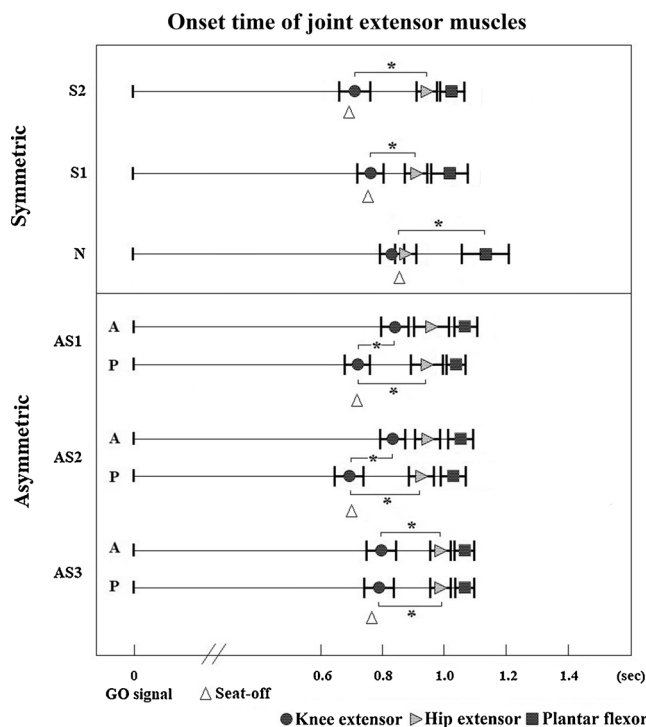


Fig. 3. Onset times of joint extensor muscles. Asterisks represent a statistically significant difference (p < 0.05) between EMG onset times. The left and right error bars display the standard deviation of the muscle onset times.

anterior leg (duration of the RF activation: AS1 (posterior 0.91 ± 0.2 s, anterior 0.75 ± 0.24 s, p < 0.01), AS2 (posterior 0.86 ± 0.33 s, anterior 0.59 ± 0.23 s, p = 0.02). However, AS3 showed no difference between the anterior and posterior legs. For symmetric IFPs, the onset time of the hip extensors was delayed as the forward velocity of the CoM for STS decreased (r = 0.83, p < .01).

3.2.2. EMG peak amplitude and area

All EMG peak amplitude and area data for all IFPs are shown in Table 2. There was no significant difference in EMG area or peak amplitude between the dominant and non-dominant legs for the symmetric IFPs (N, S1, and S2) (Table 2). In symmetric IFPs, EMG peak amplitude of the TA was largest in N (p < 0.05) and was relatively smaller in S1 and S2, whereas the Sol maintained 15–25% MVIC EMGmax

throughout the STS for all IFPs. BF and GM showed similar EMG patterns and their activity was greater with more anterior IFPs. RF activity, however, decreased with anterior foot positioning.

In AS1 and AS2, the EMG peak amplitude and area in the posterior leg were greater than in the anterior leg. In AS3, however, the EMG peak amplitude of the TA in the anterior leg (which occurred just after ‘seat-off’) was higher than the TA EMG peak that of the posterior leg.

4. Discussion

A successful STS depends upon the individual’s neuromuscular control strategy [27], lower limb muscle strength, and balance maintenance [28]. The purpose of this study was to investigate changes in lower limb muscle activation patterns and balance measures during and after a STS performed with various symmetric and asymmetric IFPs.

4.1. Muscle activation patterns

Trunk flexion in the sagittal plane during the STS produces a forward driving force to accelerate the body’s CoM anteriorly while the knee extensors are used to accelerate the body upward with the assistance of the hip extensors [29,30]. Trunk, leg, and hip extensors are the prime movers for the STS maneuver and their onset time sequence always shows the same pattern; the activation of trunk extensors is followed by knee extensors, then hip extensors during the rising phase of the STS, indicating that this is a centrally programmed sequence for preparatory postural control [5]. The present study demonstrates that different IFPs require different levels of forward velocity of the CoM before uprising and the onset times and activation levels of hip and knee extensors depends on the requirement for forward velocity of the CoM.

Compensatory postural adjustments are movement strategies regulated by sensory feedback to recover postural balance during perturbation [9]. For example, in the present study, when a small trunk flexion was enough to move the body CoM anteriorly over the base of support in S2 and less forward velocity of the CoM was required. Thus, there was a delayed onset and relatively small activation of the hip extensors (GM and BF). The hip extensors activated after seat-off following the knee extensor (RF), to assist with accelerating the body upward. The knee extensor activation in S2, however, was earlier and greater during uprising. The small trunk flexion in S2 increased the moment arm between the rotation point of the knee and the point on the GRF vector line at the beginning of seat-off [31]. Thus, the increased torque of the knee extensors in S2 was due to greater RF

Table 2
Highest peak (% MVIC EMG) and area (% MVIC EMG) of EMG.

IFP		TA		RF		BF		GM	
		Peak %	Area %	Peak %	Area %	Peak %	Area %	Peak %	Area %
N		42.86(5.12) ^{b,c}	44.77(9.83) ^{b,c}	30.29(2.11) ^{b,c}	16.93(7.14) ^c	24.06(3.81) ^{b,c}	22.12(2.10)	19.03(3.11) ^{b,c}	25.11(2.11) ^c
S1		34.22(6.33) ^a	28.78(7.78) ^a	33.40(1.08) ^{a,c}	18.12(5.64) ^c	18.63(2.32) ^a	22.01(1.12)	15.01(2.31) ^a	24.15(1.21) ^c
S2		32.11(4.11) ^a	26.78(7.99) ^a	35.22(1.72) ^{a,b}	27.54(4.32) ^{a,b}	16.88(1.99) ^a	20.21(1.28)	14.80(1.98) ^a	21.10(1.73) ^{a,b}
AS1	A	32.34(3.23) ^a	11.25(5.22) ^{a,b,c,d}	30.34(1.98) ^d	25.32(2.28)	23.11(3.02) ^{c,d}	23.56(2.31) ^d	16.13(0.92)	20.56(2.49) ^d
	P	33.11(4.32) ^a	25.11(6.88) ^a	35.32(3.21) ^a	30.02(4.44)	20.52(1.08) ^a	17.50(1.21) ^{a,b}	15.23(1.33)	14.50(2.14) ^a
AS2	A	32.12(5.10) ^{a,d}	11.22(3.21) ^{a,b,c,d}	32.01(1.52) ^{c,d}	18.99(3.15) ^{c,d}	30.06(3.98) ^{b,c,d}	25.96(3.22) ^{b,c,d}	19.02(2.09) ^{c,d}	25.76(3.69) ^{c,d}
	P	39.01(4.58)	43.27(8.21)	47.11(3.41) ^{a,b,c}	34.99(5.42) ^{a,b}	22.23(1.99)	16.32(2.34) ^a	16.33(0.88)	15.82(3.41) ^a
AS3	A	39.21(4.08) ^d	15.61(3.44) ^{a,b,d}	32.52(1.22)	24.36(2.88)	20.06(2.51)	17.82(2.89)	16.36(1.23)	19.82(2.51) ^{a,b}
	P	32.02(3.31) ^a	24.12(5.42) ^a	34.21(1.74) ^a	30.52(3.52) ^{a,b}	19.63(2.55)	16.37(2.77)	15.89(1.01) ^a	18.37(3.41) ^{a,b}

Data represent the mean \pm SD. A, anterior non-dominant leg; P, posterior dominant leg, Main effect: initial foot position.

^a Significantly different than N.

^b Significantly different than S1.

^c Significantly different than S2.

^d Significantly different than the posterior leg.

activity.

When a large trunk flexion was required to move the body CoM farther over the base of support in N, a greater forward velocity of the CoM was required. Due to the long anterior displacement of the CoM, trunk flexion was not complete even after seat-off. The hip extensors in N displayed the earliest onset with two clear EMG peaks; the early first peak started around seat-off and was maintained until shortly before rising. This suggests that the initial activity of the hip extensors was an eccentric contraction in response to active trunk flexion because they were activated before hip joint extension began.

An active eccentric lengthening is used to dissipate mechanical energy for decelerating the body [32]. Thus, our findings indicate that the initial eccentric contraction of the hip extensors provided a control mechanism to reduce forward velocity of the CoM after seat-off to prepare for the change of force direction from forward to upward. The second EMG peak of the hip extensors started right after the completion of trunk flexion. A greater trunk flexion before uprising requires a larger hip extension moment [31]. Thus, the second peak of the hip extensors was greatest in N compared to the other IFPs.

In AS2, the trunk forward flexion angle was small thereby placing the body CoM only over the posterior leg during the initial momentum transfer. This caused a greater vertical GRF in the posterior leg, resulting in greater and earlier EMG peak amplitude and area of all muscles in the posterior leg than in the anterior leg.

Asymmetric IFPs are often used for rehabilitation in clinical populations. Patients with hemiplegia after stroke rely on a compensatory strategy of recruiting the muscles on the unaffected side earlier and with greater activity [33]; Thus, the AS2 IFP would be best to help rehabilitate weak and delayed responses of the hemiplegic side by placing the affected side behind the unaffected side.

4.2. Balance in the stabilization phase

To our knowledge this is the first study to examine the effects of changes in muscle activation patterns across different IFPs on balance during the stabilization phase before entering quiet standing following a STS. In symmetric IFPs, the farther the body's CoM moved forward, the faster the forward velocity of the CoM was required. Although eccentric lengthening of the hip extensors in N decreased the propulsive force, additional ankle joint stabilization was required for upward momentum transfer. This was demonstrated by a greater activation of the TA that persisted until after seat-off. To obtain standing balance stability, the Sol was also activated after achieving the STS and persisted throughout the stabilization phase.

In asymmetric IFPs, we hypothesized that a greater base of support would allow the anterior leg to become more involved in stabilizing

sway after completing the STS. Our results, however, showed that the weight-bearing asymmetry between the anterior and posterior legs impaired balance on the posterior side. Schultz et al. found that pushing more vertically against the ground during a STS moves the body CoM backward quickly [34]. In AS2, the excessive weight bearing on the posterior leg caused a greater vertical GRF on the posterior leg during whole-body extension. This explains why all balance parameters on the posterior foot showed large perturbations during the stabilization phase.

In AS3, however, weight-bearing symmetry was not different from the symmetric IFPs. It is likely that AS3 is the most common position used during daily activity. This may be because IFPs that bring both feet backward from N improve weight distribution between both legs. Furthermore, the posterior foot position of both legs in AS3 reduced the anterior displacement of the CoM with trunk flexion before rising. This is similar to what occurred in S2. Interestingly, despite no differences in the anterior displacement of the CoM and trunk flexion between S2 and AS3, the forward velocity of CoM was approximately 7% slower for AS3 than S2. This might be explained by greater braking of A–P GRF required in the anterior leg of AS3 compared to S2. In the other asymmetric IFPs (AS1 and AS2), heavy asymmetrical loading on the posterior leg caused greater GRFs for the posterior leg on both vertical and A–P axes during the STS. In AS3, however, the anterior leg tends to have greater A–P braking GRF than the posterior leg because weight symmetry was not different from the symmetric IFPs.

In symmetric IFPs, the TA activates first before seat-off to move the shank forward to prepare for the anterior displacement of the CoM [7]. Thus, the highest TA EMG amplitude is normally just before seat-off. We found the same activation pattern of the TA in the posterior leg of AS3, which reflects the typical role of the TA. However, the anterior leg of AS3 showed a delayed TA onset, which activated after seat-off throughout the momentum transfer phase and its highest EMG peak was greater than that of the posterior leg.

Dorsiflexion by the TA holds CoP under the feet to maintain the CoP position close to the body CoM for uprising [35]. Thus, greater, but delayed onset of the TA in the anterior leg of AS3 might have helped to improve ankle stabilization on the anterior side when the propulsive force was arrested, providing stabilization for upward momentum transfer. This explains why AS3 had the highest percentage of CoP spectral energy in the 0 – 0.5 Hz domain with smaller CoP excursions, whereas S2 showed more front-back sway during the stabilization phase. Thus, performing a STS with an AS3 IFP could be used as an intervention for rehabilitation exercises for mobility-limited adults or frail elderly who have decreased flexibility and weakened muscles, allowing for relatively diminished A–P perturbations after completion of uprising.

5. Conclusions

In symmetric IFPs, the anterior displacement of the body's CoM and trunk flexion was increased with anterior translation of IFP. Since an anterior symmetric IFP (N) required larger trunk flexion to move the body CoM farther over the base of support in a short time, a greater forward velocity of the CoM was required. Thus, the initial eccentric activation of the hip extensors shown in the present study plays an important role in decelerating the forward velocity of the CoM as a postural compensatory mechanism when the IFP requires a relatively long anterior displacement of the CoM. The greater A–P sway in N IFP, however, requires additional ankle stabilization for full extension, causing balance deterioration in the stabilization phase.

In asymmetric IFPs, placing both feet posterior to neutral position (AS3) greatly improves weight-bearing symmetry. AS3 also provides greater postural stability after completion of uprising, controlling TA activation patterns differently between the anterior and posterior legs. These findings are important for designing STS exercise interventions for clinical populations such as the frail elderly to improve dynamic balance control during and after STS transfer.

Conflict of interest

The authors have no conflict of interest.

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