

Research article

Postural Stability and Muscle Activation Onset during Double- to Single-Leg Stance Transition in Flat-Footed Individuals

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Abstract

The effects of foot posture on postural stability and on muscular activation pattern for postural control remain unclear. This study aimed to investigate postural stability and muscular activation onset during the transition task from double- to single-leg stance in individuals with different foot postures. Twenty-seven healthy men (age: 21.5 ± 1.5 years) were divided into 3 groups using the Foot Posture Index: neutral foot ($n = 10$); flatfoot ($n = 8$); and high-arched foot ($n = 9$). Center of pressure (COP) data and muscle activation onset times of the tibialis anterior, peroneus longus, gastrocnemius medialis, and soleus during the transition task with eyes closed were compared among groups using one-way analysis of variance and a post-hoc Tukey honestly significant difference test ($p < 0.05$) when the data were normally distributed and the Kruskal-Wallis test and a post-hoc Mann-Whitney U-test with Bonferroni correction ($p < 0.0167$) when the data were not normally distributed. The COP displacements in the mediolateral and anteroposterior directions and the resultant COP displacement during the first 3 s after a stability time point, as determined by sequential estimation during the single-leg stance phase, differed significantly among the three groups ($p < 0.05$). Post-hoc tests showed that the displacements were significantly greater in the flatfoot group than in the neutral and high-arched foot groups ($p < 0.05$), and the effect sizes for these results were large. No muscular activation onset times showed significant intergroup differences. Postural stability was significantly decreased only in the flatfoot group, while muscle activation onsets did not differ significantly by foot posture during the transition task. Decreased postural stability may be one mechanism underlying the link between flatfoot and risk of lower limb injury, and foot posture represents a potential confounder for measuring postural stability during the transition task.

Key words: Foot alignment, pes planus, pes cavus, pronated foot, postural control, balance.

Introduction

Foot posture is commonly categorized into neutral foot, flatfoot, and high-arched foot. Flat and high-arched foot postures are associated with increased risks of lower limb injuries compared with the neutral foot posture (Tong and Kong, 2013). Specifically, a flatfoot posture is a risk factor for medial tibial stress syndrome and patellofemoral pain syndrome (Neal et al., 2014), while high-arched foot has been associated with a high risk for overuse injuries of the foot or ankle (Cain et al., 2007). Previous reports have also

suggested associations of foot posture with lateral ankle sprain (Morrison and Kaminski, 2007) and anterior cruciate ligament injury (Loudon et al., 1996). Although the mechanisms underlying links between foot posture and increased risk of lower limb injury remain unclear, changes in lower limb biomechanics (Hollander et al., 2019), neuromuscular controls (Murley et al., 2009), and postural stability (Cote et al., 2005; Kim et al., 2015) are considered to be among the mechanisms.

Deficits or changes in postural stability are identified as intrinsic risk factors for sport-related injuries, such as overuse injuries of the lower limb (De Blaiser et al., 2019), lateral ankle sprain, and anterior cruciate ligament (Dingenen et al., 2016). Postural stability requires athletes to maintain a stable center of gravity over the base of support, especially under single-leg stance conditions, and would be an essential component for avoiding falls and high-risk postures for lower limb injuries (Dingenen et al., 2016). A few studies have reported that static postural stability was significantly affected by foot postures (Cote et al., 2005; Kim et al., 2015; Hertel et al., 2002; Huang et al., 2019; Tsai et al., 2006), while deficits in dynamic postural stability have been shown to not differ between flat and high-arched feet (Kim et al., 2015; Tsai et al., 2006). Flatfoot showed poor static postural stability (Huang et al., 2019; Kim et al., 2015; Tsai et al., 2006) or no difference in stability (Cote et al., 2005; Hertel et al., 2002) during single-leg stance on a force plate or balance system, compared with neutral foot. On the other hand, previous studies have reported that high-arched foot showed poor static stability (Hertel et al., 2002; Tsai et al., 2006) or no differences in stability (Cote et al., 2005) compared with neutral foot. Based on these findings, no consensus has yet been reached regarding associations between foot posture and static postural stability.

Static postural stability has previously been assessed using the simple single-leg stance task (Cote et al., 2005; Kim et al., 2015; Hertel et al., 2002; Tsai et al., 2006). This task has some limitations, including a lack of definition of the start times for push-off by the non-stance limb and for single-leg stance phase, and the unstandardized elevation speed of the non-stance limb, which affects variables of postural stability measured using a force plate (Dingenen et al., 2013). A study using a standardized single-leg stance task is thus needed to clarify the association of foot posture with static postural stability.

The transition task from double-leg to single-leg stance was developed to overcome the above-mentioned limits as a standardized method for single-leg stance (Dingenen et al., 2013) and has been used to assess postural stability and muscular activation onset times for postural control in individuals with anterior cruciate ligament injury (Dingenen et al., 2015a; 2015b) or chronic ankle instability (Dingenen et al., 2013; 2015c). The task has been useful for detecting postural stability deficits after reaching a stability point during the single-leg stance phase and delayed muscular onset times of lower limb during the transition in the knee or ankle ligament injured individuals compared with non-injured individuals (Dingenen et al., 2013; 2015a; 2015b). In addition, displacement of the center of pressure (COP) after reaching a stability point during the task could predict increased risks for noncontact knee and ankle ligament injuries (Dingenen et al., 2016). Since the transition task was shown to be useful for assessing postural stability and muscular activity onset to reveal lower limb dysfunction, the differences in those values due to different foot postures may be discriminated by this task. However, no studies have investigated this.

Postural stability requires adequate timing and a coordinated pattern of muscular activity to maintain the stability of joints, segments, and center of mass. Previous studies have shown that muscular activity and reaction times around the ankle are altered by foot postures (Murley et al., 2009; Denyer et al., 2013; Hunt and Smith, 2004). For example, muscular activity of the tibialis anterior and tibialis posterior was increased and activity of the peroneus longus was decreased in a flatfoot group during walking (Murley et al., 2009; Hunt and Smith, 2004), and reaction time of peroneus longus was delayed with simulated ankle sprain (Denyer et al., 2013). The reasons for these altered muscular activity patterns may be stability of the foot due to differences in foot mobility and/or mechanical disadvantage of muscles around the ankle joint; for example, flatfoot would result in decreased foot stability by the hypermobile foot and/or a shortened peroneus longus by increased flattening of the foot (Tiberio, 1988). However, to the best of our knowledge, no studies have investigated the effect of foot posture on muscular activity patterns during a single-leg balance task. Clarifying the muscular activity pattern during the balance task should help to understand functional deficits and define compensatory strategies for foot posture. In addition, identifying functional differences by foot posture during balance task may contribute to clarifying mechanisms associated with the development of sport injuries due to foot posture.

The objective of this study was to investigate the effects of foot posture (neutral foot, flatfoot or high-arched foot) on both postural stability and muscular activity pattern at the ankle during the transition task from double-leg to single-leg stance. We hypothesized that individuals with flatfoot or high-arched foot would have poorer postural stability and delayed onset times of muscular activities compared with the individuals showing a neutral foot posture.

Methods

Participants

Participants comprised 27 healthy male university students (age: 21.5 ± 1.5 years; height: 1.71 ± 0.06 m; weight: 63.9 ± 9.6 kg) who engaged in recreational or competitive sport activities (e.g., baseball, lacrosse, track and field, and soccer). All participants were informed about this study and provided voluntary consent to participate prior to enrolment. This study was approved by the institutional review board of our university (approval number: 17-54). We included participants with no history of surgery or fracture of the lower limb, and no history of musculoskeletal injury in the lower limb within the 6 months before participating this study.

The foot posture of participants was screened by the Foot Posture Index (FPI-6). The FPI-6 is widely used to assess static foot posture and has shown high intra-rater and moderate inter-rater reliability (Redmond et al., 2006; Cornwall et al., 2008). The categorization of foot posture by this method has been associated with lower limb injuries (Tong and Kong, 2013). Participants were divided into three groups: neutral foot group ($n = 10$), flatfoot group ($n = 8$), and high-arched foot group ($n = 9$) (Table 1). In our pilot study regarding the reliability of measuring the FPI-6, intra-rater reliability on two different days by the same tester was good (intraclass correlation coefficient_{1,1} = 0.87; 95% confidence interval: 0.69–0.95) in 10 participants, and inter-rater reliability by two different testers was also good (intraclass correlation coefficient_{2,1} = 0.89; 95% confidence interval: 0.74–0.95) in 9 participants. A single examiner assessed the dominant foot (defined as the preferred foot for kicking a ball) with the participant in a relaxed standing posture with bare feet based on the standard protocol reported in a previous study (Redmond et al., 2006). Scores were made on a 5-point scale (-2 to +2) for each of the 6 items in the FPI-6 (talar head palpation, curves above and below the lateral malleolus, inversion/eversion of the calcaneus, bulge in the region of the talonavicular joint, congruence of the medial longitudinal arch, and

Table 1. Demographic data. All values are mean \pm standard deviation.

	Neutral (n = 10)	Flatfoot (n = 8)	High-arch (n = 9)
Age (years)	21.6 \pm 0.8	21.3 \pm 1.5	21.6 \pm 2.1
Body height (m)	1.74 \pm 0.07	1.68 \pm 0.06	1.71 \pm 0.03
Body weight (kg)	65.0 \pm 13.3	62.9 \pm 7.9	63.7 \pm 6.4
Foot length (cm)	25.6 \pm 1.3	25.3 \pm 1.8	25.3 \pm 1.1
Foot width (cm)	10.0 \pm 0.5	9.8 \pm 0.5	10.0 \pm 0.3
Foot Posture Index (FPI-6)	2.7 \pm 1.6	7.2 \pm 1.3*	-3.0 \pm 1.6†

* Significant differences compared with neutral and high-arch groups on post-hoc Tukey honestly significant difference tests ($p < 0.05$). † Significant differences compared with neutral and flatfoot groups on post-hoc Tukey honestly significant difference tests ($p < 0.05$).

abduction/adduction of the forefoot on the rearfoot), and categorized foot posture using the total scores as follows: high-arched foot, -12 to 0; neutral foot, 0 to +5; and flat-foot, +6 to +12 (Redmond et al., 2006; Cornwall et al., 2008).

Procedure

Each participant performed the transition task from double-leg stance to single-leg stance, based on previous studies (Dingenen et al., 2013; 2015a; 2015b; 2015c; 2016). The reason why this task was used was that the standardized single-leg stance task was a useful task to assess postural stability and onset of muscular activity in lower limb dysfunction. The participant stood on a force plate barefoot, with feet hip-width apart, arms hanging at the sides. The location of the feet was marked by color tapes on the force plate to perform tests using the same location between trials. Participants maintained the double-leg stance for 13 s, then elevated the non-dominant leg within 1 s to 60° of hip flexion when a beep sound was played. Next, the single-leg stance was maintained for 13 s. The task in the eyes-closed condition was recorded and analyzed, because this condition can detect differences in postural stability due to lower limb injuries and predict the risk of injuries (Dingenen et al., 2013; 2015b; 2016). All participants were allowed to practice such that the non-dominant leg was controlled to maintain hip flexion of 60° as measured by a goniometer before the task was recorded. A trial was excluded when the elevated foot touched the force plate, the standing foot deviated from the start position, or the participant opened their eyes. Three success trials were recorded for each participant.

Data collection

A force plate (Type 9286; Kistler, Winterthur, Switzerland) was used to collect the data for ground reaction forces and moments at a sampling rate of 1000 Hz. Electromyography (EMG) for tibialis anterior (TA), peroneus longus (PL), gastrocnemius medialis (GM), and soleus (SOL) were measured using a wireless surface EMG system

(WEB-1000; Nihon Kohden Corporation, Tokyo, Japan) with surface-type electrode telemeters at a sampling rate of 1000 Hz. Electrodes for all muscles were positioned according to the recommendations of the SENIAM project (Surface ElectroMyography for the Non-Invasive Assessment of Muscles. Recommendations for sensor locations on individual muscles. <http://www.seniam.org>). Data were synchronously collected using Cortex-64 software (5.5.0.1579 version; Motion Analysis Corporation, Santa Rosa, CA).

Data analysis

We analyzed force and EMG data using a custom MATLAB program (Mathworks, Natick, MA) according to the analysis method used in previous studies (Dingenen et al., 2013; 2015a; 2015b; 2015c; 2016). Ground reaction force data were filtered by a low-pass filter with a cut-off frequency of 5 Hz. Anteroposterior (AP) and mediolateral (ML) displacements of the COP were calculated from moments of force and vertical ground reaction force. The transition task from double-leg to single-leg stance consists of the following phases: the double-leg stance phase, the contralateral push-off phase during the double-leg stance, the transition phase, and the single-leg stance phase (Figure 1a). Three time points during the task were identified using time-series COP data: start point of contralateral push-off, crossing point, and time to the new stability point (TNSP) (Figure 1a). The start point was defined as the last time point when the ML COP crossed the mean ML COP for 10 s during the double-leg stance before contralateral push-off (Figure 1a). The crossing point was defined as the time point when the ML COP again crossed the mean ML COP after contralateral push-off (Figure 1a). The TNSP was determined using the sequential estimation technique, which incorporates the algorithm to calculate a cumulative average of COP data points after the crossing point in a series by successively adding 1 point at a time (Colby et al., 1999). TNSP was defined as the time point when the sequential average of the series reached the mean of overall COP data points minus 0.25 of the standard deviation (Colby et al., 1999) (Figure 1b).

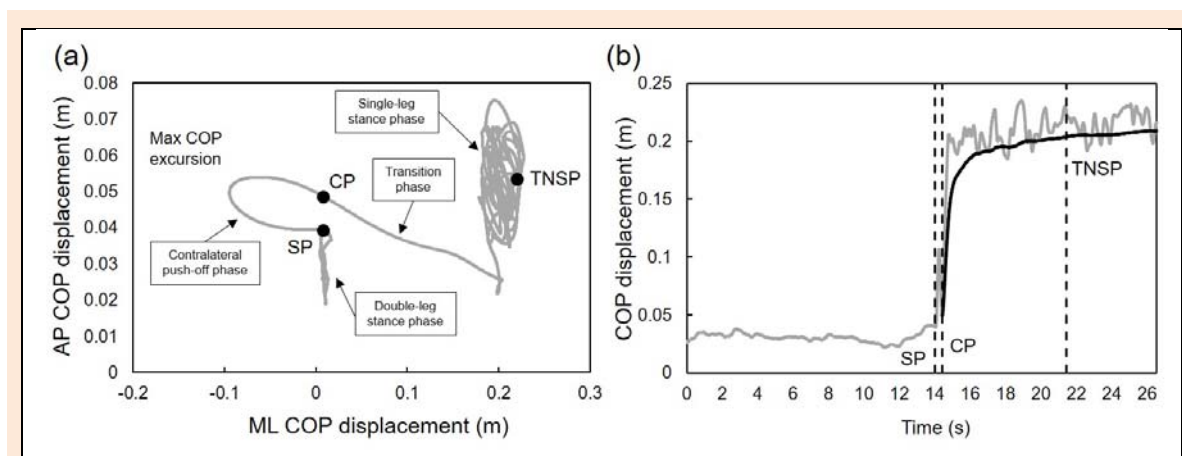


Figure 1. (a) Representative example of center of pressure (COP) displacement data (gray solid line) during the transition task from double-leg to single-leg stance. Black circles represent the time points of starting point (SP), crossing point (CP), and time to new stability point (TNSP). (b) Representative examples of sequential estimation (black solid line) and COP displacement data (gray solid line). Dotted vertical lines represent the time points of SP, CP, and TNSP. AP COP: anteroposterior COP; ML COP: mediolateral COP; Max COP excursion: maximum COP excursion during anticipatory postural adjustments.

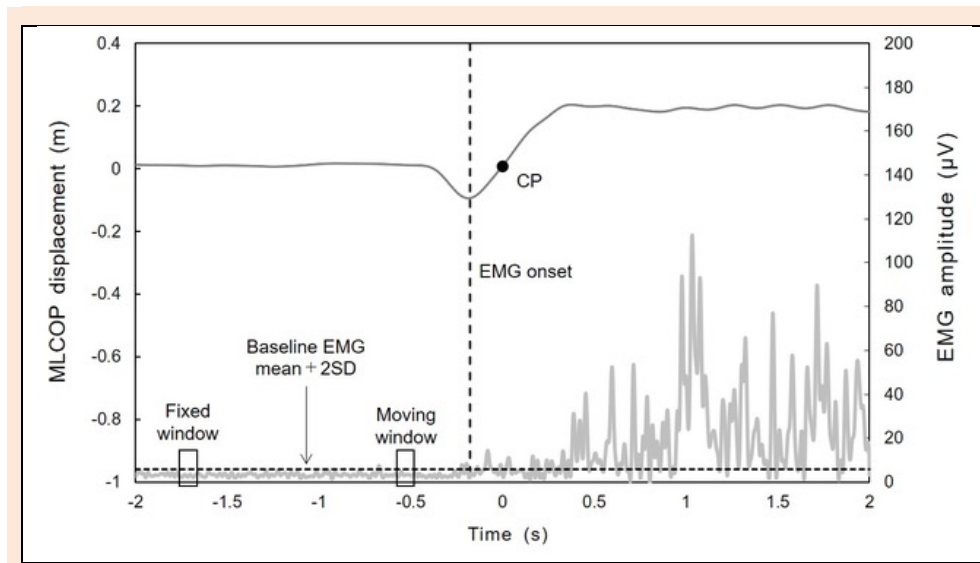


Figure 2. Representative example of mediolateral center of pressure (ML COP) displacement data (black solid line) and EMG amplitude of the tibialis anterior (gray solid line). Left axis represents ML COP displacement (m) and right axis represents the EMG amplitude of tibialis anterior (μV). Dotted vertical line represents EMG onset and dotted horizontal line represents baseline EMG mean + 2 standard deviations (SDs). Crossing point (CP) was considered as time 0 for EMG onset.

Raw EMG signals were rectified and filtered using a 6th-order Butterworth low-pass filter with a cut-off frequency of 45 Hz. Muscle activation onset was then defined as the time point at which EMG signals exceeded the threshold value (2 standard deviations beyond baseline EMG signal) and determined using a sliding window (100-ms sample width) (Hodges and Bui, 1996) (Figure 2). The baseline of EMG signals was determined as the mean of data in the fixed-window (100-ms sample width) during the double-leg stance (Dingenen et al., 2015a; 2015c). The time point at the crossing point determined by ML COP during the transition phase after contralateral push-off was considered as time zero for muscle activation onset times (Dingenen et al., 2015a; 2015c; Van Deun et al., 2007). A muscle onset time was indicated as a negative value when muscular activity occurred before the crossing point.

Statistical analysis

We used the average of 3 trials of the following variables for statistical analyses: 1) demographic data; 2) time during preprogrammed anticipatory postural adjustments from starting point to crossing points as the time could affect postural stability outcomes (Dingenen et al., 2013); 3) COP displacement and maximum excursion during anticipatory postural adjustments; 4) time until TNSP from crossing point; 5) ML and AP COP displacements and the resultant COP displacement during the first 3 s after TNSP, as displacement was a significant discriminative and predictive parameter for lower limb injuries (Dingenen et al., 2013; 2015b; 2016); and 6) muscle activation onset times for TA, PL, GM, and SOL. EMG data for TA were statistically analyzed for 26 participants because the data failed to record in 1 participant with a high-arched foot. The normality of data was assessed using the Shapiro-Wilk test. Since all demographic data except for age, COP displacement during anticipatory postural adjustments, and the first 3 s after

TNSP and onset time of SOL were normally distributed for all groups, one-way analysis of variance was used to determine the main effect of foot posture on these variables. A post-hoc Tukey honestly significant difference test was performed when a significant effect was identified ($P < 0.05$). On the other hand, age, time and COP excursion during anticipatory postural adjustments, time until TNSP, and onset times of TA, PL, and GM were not normally distributed. For these variables, the Kruskal-Wallis test ($p < 0.05$) and a post-hoc Mann-Whitney U-test with Bonferroni correction ($p < 0.0167$) were used to assess differences among three groups. These statistical analyses were performed using SPSS Statistics version 17.0 (IBM, Armonk, NY). The effect sizes were also calculated as Cohen's d when the data were normally distributed and r when the data were not normally distributed in order to indicate the magnitude of the group differences. Effect sizes were defined as small ($d = 0.2$ or $r = 0.1$), moderate ($d = 0.5$ or $r = 0.3$), or large ($d = 0.8$ or $r = 0.5$) (Cohen, 1988). In addition, post-hoc power analyses were performed by G*Power 3.1.9.7 (University of Dusseldorf, Germany).

Results

Demographic data

No significant differences in demographic data were identified among the three groups except for FPI-6 (Table 1).

Postural stability

For postural stability data, COP displacements in the ML and AP directions and the resultant COP displacement during the first 3 s after TNSP differed significantly among the three groups (Table 2). The statistical power for the resultant COP displacement was 84%, while the powers for the COP displacements in the ML and AP directions were 76% and 78%, respectively. Post-hoc Tukey honestly signifi-

cant difference test revealed significantly increased COP displacements in ML and AP directions and resultant COP displacement during the first 3 s after the TNSP in the flat-foot group compared with the neutral and high-arched foot groups (Table 2). In addition, these effect sizes were large (Table 2).

No significant differences were identified for other postural stability outcomes among the three groups (Table 2). These statistical powers were poor (< 49%), and the

effect sizes other than the COP displacement during anticipatory postural adjustments were small (Table 3).

Muscular activation onset

No significant differences in any muscular activation onset times were identified among the three groups (Table 3). The statistical powers for the muscular activation onset times were poor (< 40%), and the effect sizes except for soleus were small (Table 3).

Table 2. Postural stability data for the 3 groups. All values are mean \pm standard deviation.

	Neutral	Flatfoot	High-arch	P values	Effect sizes ^b		
					Neutral vs Flatfoot	Neutral vs High-arch	Flatfoot vs High-arch
Time during APA (s) ^a	1.13 \pm 1.40	0.96 \pm 1.21	0.66 \pm 0.37	0.744	0.08	0.00	0.23
COP displacement during APA (m)	0.21 \pm 0.03	0.18 \pm 0.07	0.16 \pm 0.04	0.110	0.57	1.50	0.38
COP excursion during APA (m) ^a	0.10 \pm 0.01	0.08 \pm 0.03	0.08 \pm 0.02	0.060	0.27	0.54	0.26
Time to new stability point (s) ^a	4.95 \pm 2.04	3.87 \pm 2.30	4.70 \pm 1.42	0.627	0.23	0.07	0.16
AP COP displacement during the first 3 s after TNSP (m)	0.15 \pm 0.03	0.21 \pm 0.05*	0.15 \pm 0.05	0.020	1.37	0.16	1.28
ML COP displacement during the first 3 s after TNSP (m)	0.17 \pm 0.04	0.23 \pm 0.06*	0.17 \pm 0.04	0.023	1.24	0.03	1.25
Resultant COP displacement during the first 3 s after TNSP (m)	0.25 \pm 0.05	0.34 \pm 0.08*	0.25 \pm 0.07	0.012	1.40	0.11	1.37

APA: anticipatory postural adjustments; COP: center of pressure; TNSP: time to new stability point; AP: anteroposterior; ML: mediolateral.^a The data were not normally distributed. ^b Cohen's *d* and *r* were calculated when the data were normally distributed and not normally distributed, respectively. * Significant differences compared with neutral and high-arch groups on post-hoc Tukey honestly significant difference tests ($p < 0.05$).

Table 3. Muscle activation onset times for the 3 groups. All values are mean \pm standard deviation.

	Neutral	Flatfoot	High-arch	P values	Effect sizes ^b		
					Neutral vs Flatfoot	Neutral vs High-arch	Flatfoot vs High-arch
Tibialis anterior (s) ^a	-0.16 \pm 0.17	-0.16 \pm 0.17	-0.18 \pm 0.14	0.998	0.00	0.02	0.00
Peroneal longus (s) ^a	0.03 \pm 0.08	0.08 \pm 0.24	0.01 \pm 0.08	0.937	0.04	0.09	0.00
Gastrocnemius medialis (s) ^a	0.06 \pm 0.11	0.04 \pm 0.14	0.03 \pm 0.11	0.973	0.00	0.06	0.04
Soleus (s)	0.10 \pm 0.14	0.17 \pm 0.24	0.01 \pm 0.11	0.164	0.35	0.81	0.92

^a The data were not normally distributed. ^b Cohen's *d* and *r* were calculated when the data were normally distributed and not normally distributed, respectively.

Discussion

This is the first study to investigate simultaneously the postural stability and shank muscle activation onset times during transition tasks from double-leg to single-leg stance in participants with different foot postures. We found that participants with flatfoot had significantly decreased postural stability, but no difference in muscle activation onset compared with those with neutral or high-arched foot. As for participants with high-arched foot, postural stability and muscle onset time did not differ compared to those with neutral. These findings partially supported our hypothesis that participants with flatfoot or high-arched foot would exhibit poor postural stability, although our hypothesis regarding muscle activation onset was not supported.

Postural stability

Postural stability during the first 3 s after TNSP in the transition task was decreased in participants with flatfoot compared to those with neutral or high-arched foot, and the effect sizes were large. Decreased postural stability for maintaining a single-leg stance in individuals with flatfoot has also been shown in previous studies: flatfoot individuals demonstrated increases in COP displacements (Tsai et al., 2006; Huang et al., 2019), COP speed (Kim et al., 2015), and standard deviation of COP (Tsai et al., 2006). The

present study supported the findings of those studies. As discussed in previous studies, the poor postural stability may be attributable to decreased passive stability within the foot (Kim et al., 2015; Tsai et al., 2006). The transition tasks used in the present study induced an internal perturbation by weight-shift from double- to single-leg stance, so COP displacement during the first 3 s after TNSP provides an indication of the participant's ability to maintain a steady stance position against perturbation (Dingenen et al., 2013; 2015b; 2016). Motions within foot joints may be increased to control postural stability against the internal perturbation, which in turn might result in increased displacement of the COP. The joint axis of the subtalar joint, which mainly controls the pronation and supination of the foot, is directed obliquely from posterior-lateral-plantar to anterior-medial-dorsal (Kirby, 2001). The axis direction can rotate internally and translate medially in the pronated position (Kirby, 2001) and subtalar joint motion in sagittal plane may be increased. Foot motion control around the medially deviated subtalar joint axis could lead to increases in both ML COP and AP COP displacements during the first 3 s after TNSP in participants with flatfoot.

For a high-arched foot posture, postural stability did not differ significantly from that reported by Cote et al. (2005), who found no effects of foot posture on center of balance and postural sway during single-leg stance. On the

other hand, other studies have reported that individuals with a high-arched foot have poorer postural stability as assessed by COP parameters, compared to those with neutral foot (Hertel et al., 2002; Tsai et al., 2006). These discrepancies may be mainly attributable to differences in the methods of categorizing foot posture and the balance task. We used the FPI-6 to categorize three foot postures, whereas previous studies have used the medial longitudinal arch angle and rearfoot angle under weight-bearing conditions (Tsai et al., 2006), or forefoot and rearfoot angles under non-weight-bearing conditions (Hertel et al., 2002). The balance task in previous studies used a single-leg stance (Cote et al., 2005; Hertel et al., 2002; Tsai et al., 2006). A high-arched foot should be more rigid within the foot compared with neutral and flat feet, and the increased passive stability of the foot could overcome the internal perturbation during the transition from double- to single-leg stance in the present study. Postural stability in participants with a high-arched foot thus showed no significant change compared to those with neutral foot type.

Muscular activation onset

Counter to our hypothesis, muscle activation onset times did not differ significantly by foot posture. Although we cannot directly compare our findings with those of previous reports, because the tasks differed, individuals with flatfoot showed delayed reaction time of PL to a simulated ankle sprain (Denyer et al., 2013) and altered muscular activities of the TA, PL, and tibialis posterior during walking (Murley et al., 2009; Hunt and Smith, 2004). The transition task in the present study, unlike the dynamic tasks used in previous studies (Murley et al., 2009; Denyer et al., 2013; Hunt and Smith, 2004), may not have placed adequate demands to detect changes in muscular activation onset of the shank by foot posture. The flatfoot posture would have decreased passive stability due to dysfunction of the static stabilizers of the foot (e.g., spring ligament, plantar fascia, or joint capsule), but muscular control in dynamic stabilizers was unchanged in the present study. We assume that suitable muscular controls for the flatfoot posture were required to control postural stability, because structure and mobility differed from those of the neutral foot. No changes in muscular activation patterns for individuals with flatfoot would not be beneficial for maintaining a steady stance position and ultimately might lead to decreased postural stability. However, the findings for muscle activation onset times are considered underpowered. Future studies should recruit more participants to investigate the effect of foot posture on muscular activity patterns during a single-leg balance task.

Clinical implications

Concerning clinical relevance in the present study, decreased postural stability during the transition task was associated with increased risk of sport-related lower limb injuries in a prospective cohort study (Dingenen et al., 2016). Flatfoot posture has also been associated with increased lower limb injuries compared with neutral foot (Tong and Kong, 2013; Neal et al., 2014; Loudon et al., 1996). Based on these previous findings and our own results, decreased

postural stability may be one mechanism underlying the link between flatfoot and increased risk of lower limb injury. Foot orthoses or taping interventions may increase passive stability within the foot and may be beneficial for improving postural stability in individuals with flatfoot. Our findings also suggest that foot posture should be considered as a potential confounder in the measurement of postural stability to assess the risk of sports injuries, as indicated by Hertel et al. (2002). Researchers, trainers, and clinicians should take into account the foot posture of individuals when measuring balance capability during transition tasks. In addition, the transition task was shown to be useful for detecting deficits in postural stability, not only in individuals with ligament injury as reported by previous studies (Dingenen et al., 2015a; 2015b), but also in uninjured individuals with flatfoot. Future studies should investigate the effects of interventions with foot orthoses or taping on postural stability in individuals with flatfoot. In addition, prospective research is also needed to clarify whether the decreased postural stability may be one of the mechanisms underlying the links between flatfoot and increased risk of sports injuries.

Limitations

Some limitations to the present study need to be considered. First, we did not collect EMG data for the intrinsic muscles of the foot or the tibialis posterior. Muscular activity of the abductor hallucis is reportedly lower during single-leg stance and is associated with increased difficulty of the balance task for the flatfoot posture compared with the neutral foot (Huang et al., 2019). The tibialis posterior is also a dynamic stabilizer against flattening of the foot (Kohls-Gatzoulis et al., 2004). Activation patterns for these muscles should be assessed during balance tasks in individuals with different foot posture in the future. Second, the present study investigated the transition task only under barefoot conditions. The findings demonstrated in the present study may differ from those under shod conditions. Lastly, kinematic data were not collected in the present study. The kinematics of the lower limb joints differ between foot postures and this may affect the postural stability (Hollander et al., 2019). Kinematics should therefore be simultaneously assessed during balance tasks to gain further insights into postural control in flat and high-arched feet.

Conclusion

The present study showed that individuals with flatfoot had significant decreased postural stability during the transition task from double-leg to single-leg stance compared to those with neutral or high-arched feet. Individuals with a high-arched foot did not differ in postural stability from those with a neutral foot posture. In addition, muscular activation onset times for shank muscles did not differ significantly by foot posture. The findings suggest that decreased postural stability may be one of the mechanisms underlying links between flatfoot and increased risk of lower limb injury, so foot posture can be a potential confounding

factor for measurement of postural stability during transition tasks.

Acknowledgements

This work was supported by JSPS KAKENHI Grant Number JP19K19823. The experiments comply with the current laws of the country in which they were performed. The authors have no conflict of interest to declare.

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Key points

- We compared postural stability and muscular activation onset during double- to single-leg stance transition among participants with neutral foot, flatfoot, and high-arched foot.
- The center of pressure displacements in the mediolateral and anteroposterior directions were significantly increased in the flatfoot group.
- No muscular activation onset times showed significant intergroup differences.
- Decreased postural stability may be one of the mechanisms underlying links between flatfoot and increased risk of lower limb injury.

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