The Foot 38 (2019) 19–23

Contents lists available at ScienceDirect

The Foot

journal homepage: www.elsevier.com/locate/foot

Original Article

The effect of additional activation of the plantar intrinsic foot muscles on foot kinematics in flat-footed subjects

Kazunori Okamura[a,b,*], Shusaku Kanai[b], Kengo Fukuda[c], Satoshi Tanaka[b], Takeya Ono[b], Sadaaki Oki[b]

[a] *Graduate School of Comprehensive Scientific Research, Prefectural University of Hiroshima, 1-1 Gakuen-cho, Mihara-shi, Hiroshima 723-0053, Japan*
[b] *Department of Physical Therapy, Faculty of Health and Welfare, Prefectural University of Hiroshima, 1-1 Gakuen-cho, Mihara-shi, Hiroshima 723-0053, Japan*
[c] *Department of Rehabilitation, Innoshima Ishikai Hospital, 1962 Innoshima nakanosyo-cho, Onomichi-shi, Hiroshima 722-2211, Japan*

ARTICLE INFO

Keywords:
Flat-foot alignment
Foot kinematics
Plantar intrinsic foot muscle
Medial longitudinal arch

ABSTRACT

Background: Strengthening exercises of the plantar intrinsic foot muscles (PIFMs) are often prescribed to flat-footed subjects because of the capacity of the PIFMs to support the medial longitudinal arch (MLA). However, it is unclear whether the capacity of the PIFMs to support the MLA is enough to change the foot kinematics in flat-footed subjects. To confirm this, the current study examined changes in foot kinematics in flat-footed subjects during standing and gait accompanied by changes in the activity of the PIFMs.

Methods: Eighteen flat-footed subjects were randomly assigned to an electrical stimulation group (ESG) or a control group (CG). In the ESG, electrical stimulation to the PIFMs was applied during standing and gait to simulate reinforcement of the PIFMs. Then, foot kinematics were measured using 3D motion analysis, and the amount of change from baseline (when no electrical stimulation was applied) was compared between the groups.

Results: In the gait analysis, the time at which the MLA height reached its minimum value was significantly later in the ESG, with no reduction in the MLA height at that time. Moreover, forefoot inversion angle and tibial external rotation angle were significantly increased in the ESG at that time. In the standing analysis, there were no significant differences between the groups.

Conclusion: The results revealed that in flat-footed subjects, the PIFMs have the capacity to support the MLA enough to change foot kinematics during gait. Strengthening these muscles may be effective in preventing or treating lower extremity overuse injuries related to flat-foot alignment.

1. Introduction

Flat-foot alignment in which the medial longitudinal arch (MLA) height is decreased has been described as a significant etiologic factor in several lower extremity overuse injuries, including plantar fasciitis [1,2] and medial tibial stress syndrome [3,4]. Although the plantar fascia [5,6] and the posterior tibialis [7,8] are principal components that prevent the decrease in MLA height, many previous studies have focused on the capacity of the plantar intrinsic foot muscles (PIFMs) to support the MLA as well [9–11]. Kelly et al. [9] reported that reinforcement of the PIFMs simulated by intramuscular electrical stimulation counteracts MLA compression under load. In addition to this, previous studies have reported that weakening of the PIFMs simulated by a tibial nerve block [10] and fatigue-inducing exercise [11] leads to a significant decrease in MLA height, as assessed by navicular drop. Furthermore, because these studies focused on the capacity of the PIFMs to support the MLA only under static conditions, in a recent experiment in our laboratory [12], an attempt to confirm the capacity of the PIFMs to support the MLA during gait was conducted. The results revealed that additional electrical activation of the PIFMs during gait caused a delay in the timing of the MLA height to reach the minimum value without a reduction in MLA height, indicating that additional activation of the PIFMs slowed the MLA deformation during the stance phase of gait. These results suggest that the PIFMs have the capacity to support the MLA enough to change the foot kinematics not only under static conditions but also during gait.

Based on these findings, in the fields of rehabilitation and sports medicine, strengthening exercises of the PIFMs, such as short foot exercises, have been prescribed to correct foot alignment [13,14] and prevent and/or treat lower extremity overuse injuries related to flat-foot alignment [15]. However, previous studies confirmed that the PIFMs have the capacity to support the MLA only in normal subjects

* Corresponding author. Present address: 1-1 Gakuen-cho, Mihara-shi, Hiroshima 723-0053, Japan.
E-mail address: k-okamura@pu-hiroshima.ac.jp (K. Okamura).

https://doi.org/10.1016/j.foot.2018.11.002
Received 19 July 2018; Received in revised form 9 October 2018; Accepted 5 November 2018
0958-2592/

[9–12], although strengthening exercises of the PIFMs are often prescribed to flat-footed subjects. In flat-footed subjects, not only the PIFMs but also other MLA support structures, such as ligaments, exhibit hypofunction [16,17]. Therefore, it is unclear whether the capacity of the PIFMs to support the MLA is enough to change the foot kinematics in flat-footed subjects. To confirm this research question would be valuable for developing strengthening exercises of the PIFMs to flat-footed subjects.

The current study sought to examine the effects of simulated reinforcement of the PIFMs via electrical stimulation on foot kinematics during standing and gait, and to confirm whether the capacity of the PIFMs to support the MLA is enough to change the foot kinematics in flat-footed subjects. It is our hypothesize that, in flat-footed subjects, simulated reinforcement of the PIFMs would cause changes in foot kinematics during standing and gait associated with an enhanced ability to support the MLA.

2. Material and methods

2.1. Subjects

A sample of 169 volunteers were screened at the authors' institution to identify individuals with flat-foot alignment. The navicular drop test and 6-item foot posture index were used to select flat-footed subjects. Although foot posture index is subjective, this non-invasive test has been shown to be reliable and boundaries for different foot types have been developed [18]. Flat-foot alignment was defined as a navicular drop of > 10 mm [19,20] and a foot posture index score of > 5 points [21]. The exclusion criteria included a history of a lower extremity injury up to 6 months before participation. Moreover, subjects who had preexisting conditions such as seizures and demand pacemakers were also excluded, following the specifications of the electrical stimulation device.

Eighteen subjects met all of the inclusion criteria. Subjects were randomly allocated to 1 of 2 groups: an electrical stimulation group (ESG) and a control group (CG). After randomization, both groups included 9 subjects (CG: age = 20.1 ± 1.4 years, height = 156.4 ± 5.4 cm, mass = 50.6 ± 6.0 kg, ESG: age = 20.3 ± 1.3 years, height = 161.5 ± 9.5 cm, weight = 55.6 ± 9.4 kg). This study was approved by the Ethics Committee of the Prefectural University of Hiroshima, and written informed consent was obtained from all subjects.

2.2. Instruments

In this study, similar to the previous study [12], electrical stimulation was delivered to the PIFMs using the WalkAide device (Innovative Neurotronics Inc., Austin, Texas, USA). Foot kinematics and kinetics were measured using a VICON system (Oxford Metrics, Oxford, UK). The VICON system was comprised of 12 MX-T20S cameras running at 100 Hz and 6 force plates running at 1000 Hz (2 Kistler and 4 AMTI). Reflective markers (diameter: 9.5 mm and 14 mm) were attached to the subjects with double-sided tape. The VICON NEXUS was used to visualize and process 3-dimensional motion.

2.3. Electrical stimulation of the PIFMs

To strengthen the contractile force of the PIFMs, electrical stimulation to the PIFMs using a method described in a previous study was applied [12]. The electrical stimulation (250 µs pulse width, 20 Hz frequency) applied to the PIFMs was provided from surface electrodes attached on the muscle belly of the abductor hallucis (ABH) muscles. The timing of electrical stimulation was controlled using the hand switch by the same tester, and checked using the VICON system synchronized with the WalkAide device. The stimulation intensity was determined to be the threshold that each subject could endure without feeling pain.

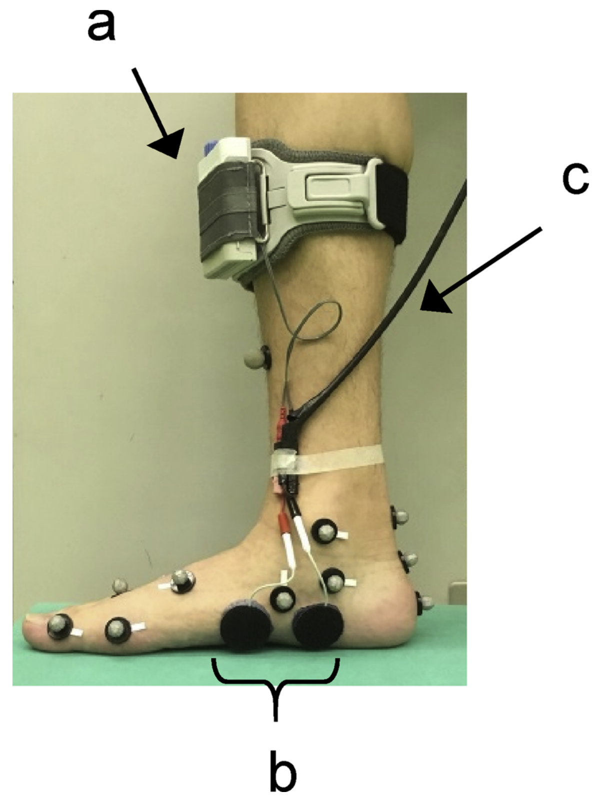

Fig 1. WalkAide and marker placements.
a: The main unit of the WalkAide device.
b: The surface electrodes to deliver electrical stimulation to the abductor hallucis muscles.
c: The cable to synchronize the WalkAide device with the VICON system.

2.4. Data collection protocol

The 29 reflective markers (14 mm or 9.5 mm) were placed following the Plug-in-Gait lower model and Oxford Foot Model specifications. Similar to our previous study [12], the Plug-in-Gait lower model was used to calculate kinetic parameters to prevent an avoidance response from a sensation such as pain from electrical stimulation. The Oxford Foot Model markers and an additional navicular tuberosity marker were applied to the side of the lower extremity that achieved the highest score on the foot posture index (Fig. 1). The WalkAide device was attached to the same side of the lower extremity (Fig. 1). After calibration with the subjects in an anatomical neutral position, the medial malleolus and posterior calcaneus proximal markers were removed, according to the protocol of the Oxford Foot Model [22,23]. The marker of the distal 1st metatarsal, recorded as one of the removable markers on the protocol, was not removed because it was necessary to calculate the MLA height. The MLA height during gait was represented by the navicular height. The navicular height during gait was defined as the perpendicular distance of the navicular tuberosity marker above the plantar plane of the foot bisecting the markers on the distal 1st metatarsal, distal 5th metatarsal, and heel [12].

Following these procedures, the gait trials and static standing trials were conducted with 28 markers. The details of the protocol for the gait and standing trials were described in our previous study [12]. In brief, all subjects were asked to walk along an 8-m walkway at their preferred normal speed for the baseline measurement, and 5 gait trials were recorded. The stimulation intensity was then determined for each subject. The determination of stimulation intensity was performed not only for the ESG, but also for the CG. A static standing trial was then conducted.

The subjects were required to place approximately 90% of their body weight on the observed lower extremity, and were allowed to stabilize their balance with a rod held in the opposite hand and be toe-touch contact to the floor with the opposite lower extremity [13]. After the force plate data determined that the weight bearing on the observed lower extremity was approximately 90% of body weight, recording was conducted for 10 seconds. In the ESG, electrical stimulation was delivered to the PIFMs for the last 5 seconds. Upon completion of a static standing trial, gait trials were conducted again. At this time, in the ESG, electrical stimulation to the PIFMs was delivered from mid-stance to pre-swing, at which time the PIFMs are known to be active [24,25]. Regardless of whether electrical stimulation was provided or not, all gait trials in both groups were performed with the tester walking diagonally behind the subject to operate the hand switch.

2.5. Statistical analysis

In the gait trials, gait velocity and the following parameters during the stance phase of the observed limb were selected for statistical analysis: stance phase duration, minimum navicular height, time at which the navicular height reached the minimum value, forefoot angle relative to the rear foot and rear foot angle relative to the tibia at that time, internal ankle moment (plantar flexion, eversion, abduction) at that time, and maximum ground reaction force (anterior, medial, and vertical directions) in the second half of the stance phase. All parameters were averaged across all 5 trials.

The navicular height from the floor was used for statistical analysis only in the standing trials, compared with previous studies [12,13]. The navicular height from the floor of the intervals excluding the first and last second of the first and second half of the 10-s recording periods were averaged.

The Shapiro–Wilk test was used to determine the normality of the data. For the baseline value of all of the parameters and the amount of change from the baseline of all parameters, independent sample *t*-tests or Mann–Whitney U tests were conducted to evaluate the differences between the 2 groups. Differences were considered significant at the $p < 0.05$ level. All statistical tests were conducted using SPSS 20.0 for Windows (SPSS Inc., Chicago, Illinois, USA).

3. Results

There were no statistically significant differences in the baseline values of the parameters between the 2 groups (Table 1). Similar to previous studies [12,26], in both groups, the time at which the navicular height reached the minimum value was equivalent to the phase of terminal stance.

Table 2 shows the amount of change from baseline. In the gait analysis, the electrical stimulation to the PIFMs in the ESG was delivered from 20.2 ± 3.6% to 57.7 ± 2.2% of the gait cycle corresponding to the phase from mid-stance to pre-swing. Regardless of whether electrical stimulation to the PIFMs was provided or not, gait velocity and minimum navicular height were consistent. In contrast, slightly extended right stance phase duration was observed in the ESG. Moreover, the time at which the navicular height reached the minimum value was significantly later in the ESG compared with the CG. At that time, the significantly increased reduction in forefoot eversion (frontal plane plane) angle relative to the rear foot, and rear foot abduction (transverse plane) angle relative to the tibia, were seen in the ESG. There were no significant differences between groups in the change in kinetic parameters, such as internal ankle moment and ground reaction force.

In the standing analysis, there were no significant group differences in the change in navicular height from the floor.

Table 1
The baseline value in each group. NH = navicular height, FF = forefoot, RF = rear foot, TB = tibia, DF = dorsiflexion, EV = eversion, AB = abduction, PF = plantarflexion, BW = body weight. Stance phase duration, RF DF angle relative to TB, and internal ankle PF moment were analyzed using Mann–Whitney U tests.

Variable		CG (N = 9)		ESG (N = 9)		Mean diff	p
		Mean	SD	Mean	SD		
Gait							
Speed (m/s)		1.2	0.2	1.2	0.2	0.1	0.38
Stance phase duration (ms)		634.4	68.6	646.7	60.6	−12.2	0.86
NH from plantar plane							
Minimum value (mm)		6.5	4.8	3.8	5.9	2.7	0.29
Timing (% stance phase)		75.8	5.3	77.8	3.8	−2.0	0.21
Foot kinematics							
FF angle relative to RF (°)	DF	8.0	4.4	8.5	4.9	−0.6	0.80
	EV	−0.1	7.2	−2.7	2.6	2.6	0.34
	AB	11.0	7.5	9.4	3.8	1.6	0.58
RF angle relative to TB (°)	DF	8.3	5.3	12.2	3.6	−3.8	0.11
	EV	−2.8	5.7	−3.8	5.8	−1.0	0.80
	AB	−27.8	11.8	−23.2	11.5	−4.6	0.42
Internal ankle moment (N mm/kg)	PF	1114.5	164.5	1221.1	153.2	−106.6	0.30
	EV	43.3	44.8	67.1	40.3	−23.8	0.25
	AB	74.6	72.0	64.3	47.9	10.2	0.73
Ground reaction force (% BW)							
Anterior direction		21.7	3.7	20.6	3.9	1.1	0.55
Medial direction		6.2	1.5	6.7	1.6	−0.5	0.52
Vertical direction		106.4	6.2	107.1	4.1	−0.8	0.76
Standing							
NH from floor (mm)		35.7	4.6	35.3	6.6	0.4	0.90

4. Discussion

In the current study, the authors examined changes in foot kinematics during standing and gait accompanied by changes in PIFM activity to evaluate whether the capacity of the PIFMs to support the MLA is enough to change the foot kinematics in flat-footed subjects.

The main finding of this study was that the reinforcement of the PIFMs during the stance phase of gait caused a delay in the time required for the navicular height to reach the minimum value, without any reduction of navicular height at that time. Because similar changes have been reported in subjects with normal feet [12], the current finding suggests that, in not only normal subjects but also flat-footed subjects, the PIFMs have the capacity to slow MLA deformation during the stance phase of gait. In addition to this, although the minimum navicular height during the stance phase of gait did not increase with reinforcement of the PIFMs, significant changes in foot segment kinematics associated with increased MLA height were observed. In the ESG, electrical reinforcement of the PIFMs caused a significantly increased forefoot inversion angle relative to the rear foot and an increased rear foot adduction angle relative to the tibia, which is identical to the increased tibial external rotation angle relative to the rear foot. These movements are caused by ABH contraction [27], and accompanied with increased MLA height. Therefore, the PIFMs may have attempted to elevate MLA height in the ESG. However, we were likely unable to detect the change in MLA height because the changes between foot segments were relatively small. Thus, it remains possible that a more powerful contractile force of the PIFMs would increase the minimum navicular height during the stance phase. Therefore, the results of this study suggest that, in flat-footed subjects, the PIFMs have the capacity to support the MLA enough to change foot kinematics during gait.

Table 2
The amount of change from baseline in each group. NH = navicular height, FF = forefoot, RF = rear foot, TB = tibia, DF = dorsiflexion, EV = eversion, AB = abduction, PF = plantarflexion, BW = body weight. Stance phase duration, FF DF angle relative to RF, internal ankle AB moment, and NH from floor at standing, were analyzed using Mann–Whitney U tests.

Variable		CG (N = 9)		ESG (N = 9)		Mean diff	p
		Mean	SD	Mean	SD		
Gait							
Speed (m/s)		−0.01	0.02	−0.04	0.06	0.03	0.12
Stance phase duration (ms)		−1.1	12.7	25.6	26.5	−26.7	0.01
NH from plantar plane							
Minimum value (mm)		−0.3	0.5	−0.3	1.1	−0.01	0.97
Timing (% stance phase)		−0.1	2.4	2.2	1.6	−2.3	0.03
Foot kinematics							
FF angle relative to RF (°)	DF	−0.9	1.6	0.5	1.0	−1.4	0.05
	EV	0.6	1.0	−0.8	1.4	1.4	0.02
	AB	0.3	0.8	0.03	1.2	0.2	0.63
RF angle relative to TB (°)	DF	1.0	1.4	−0.1	1.6	1.1	0.14
	EV	−0.6	0.9	0.3	2.1	−1.0	0.22
	AB	0.2	1.0	−1.4	1.4	1.6	0.01
Internal ankle moment (N mm/kg)	PF	−7.0	45.7	−2.3	60.6	−4.7	0.85
	EV	−4.6	26.1	2.0	26.1	−6.6	0.60
	AB	14.3	24.6	1.4	18.7	12.9	0.34
Ground reaction force (% BW)							
Anterior direction		0.04	0.8	−0.3	1.4	0.3	0.67
Medial direction		0.2	0.3	−0.3	0.9	0.5	0.11
Vertical direction		1.1	2.1	0.2	3.4	0.9	0.53
Standing							
NH from floor (mm)		−0.2	0.3	0.4	0.6	−0.6	0.06

These results are valuable for clinicians who treat lower extremity overuse injuries related to flat-foot alignment. Flat-footed patients have been found to exhibit smaller cross-sectional areas of the PIFMs [16]. Atrophy and hypofunction of the PIFMs may be some of the etiologic factors for lower extremity overuse injuries related to flat-foot alignment. In fact, patients with plantar fasciitis, a typical overuse injury related to flat-foot alignment [2], have been reported to exhibit reduced volume in the PIFMs [28,29]. Because plantar fascia tension during gait is strongly correlated with foot alignment during gait [30], to strengthen the capacity of the PIFMs as dynamic supporters of the MLA would contribute to reduction in the stress on the plantar fascia, and may be helpful to prevent and/or treat plantar fasciitis.

In the standing analysis, the navicular height from the floor tended to increase with reinforcement of the PIFMs. However, this change was small (0.4 mm) and did not achieve statistical significance. In subjects with normal feet, simulated reinforcement of the PIFMs using the protocol applied in the present study was previously found to induce a 1.6-mm increase in navicular height in the standing position, and this change was significant [12]. This difference would be caused by hypofunction of the MLA support structures other than PIFMs in flat-footed subjects. Therefore we should be aware that strengthening exercises of the PIFMs may be less effective in flat-footed subjects at changing static and dynamic foot alignment compared with subjects with a normal foot.

This study had several limitations that should be considered, in addition to the limitations of our experimental protocol discussed in our previous study [12]. First, we used the navicular drop test and the 6-item foot posture index to select flat-footed subjects. These non-invasive tests are common in clinical settings, and have been widely used in previous studies [16,19,20]. However, the validity and reliability of these measures is inferior compared with invasive X-ray examination. Second, the randomized research design and the use of 2 types of tests to select flat-footed subjects required relatively small sample sizes in each group. Although our experimental design was useful for improving the quality of the study, the small sample size may have made it difficult to detect statistically significant differences. Finally, this study was limited to evaluations during standing and gait. For example, plantar fasciitis is also common in runners [1,2]. Moreover, the activities of the PIFMs increase with increasing gait velocity [31]. Therefore, future studies should be performed to confirm in flat-footed subjects that the capacity of the PIFMs to support the MLA is enough to change the foot kinematics during running.

5. Conclusion

The simulated reinforcement of the PIFMs by electrical stimulation caused changes in foot kinematics during gait associated with an enhanced ability to support the MLA. This finding suggests that, in flat-footed subjects, the PIFMs have the capacity to support the MLA enough to change foot kinematics during gait. Therefore, strengthening this capacity may be effective in preventing and/or treating lower extremity overuse injuries related to flat-foot alignment.

Funding

This work was supported by the JSPS KAKENHI [Grant Number JP17K01524 and JP18H06441].

References

[1] Pohl MB, Hamill J, Davis IS. Biomechanical and anatomic factors associated with a history of plantar fasciitis in female runners. Clin J Sport Med 2009;19:372–6.
[2] Schwartz EN, Su J. Plantar fasciitis: a concise review. Perm J 2014;18:e105–7.
[3] Newman P, Witchalls J, Waddington G, Adams R. Risk factors associated with medial tibial stress syndrome in runners: a systematic review and meta-analysis. Open Access J Sports Med 2013;4:229–41.
[4] Reshef N, Guelich DR. Medial tibial stress syndrome. Clin Sports Med 2012;31:273–90.
[5] Huang CK, Kitaoka HB, An KN, Chao EY. Biomechanical evaluation of longitudinal arch stability. Foot Ankle 1993;14:353–7.
[6] Iaquinto JM, Wayne JS. Computational model of the lower leg and foot/ankle complex: application to arch stability. J Biomech Eng 2010;132:021009.
[7] Imhauser CW, Siegler S, Abidi NA, Frankel DZ. The effect of posterior tibialis tendon dysfunction on the plantar pressure characteristics and the kinematics of the arch and the hindfoot. Clin Biomech (Bristol, Avon) 2004;19:161–9.
[8] Kokubo T, Hashimoto T, Nagura T, Nakamura T, Suda Y, Matsumoto H, et al. Effect of the posterior tibial and peroneal longus on the mechanical properties of the foot arch. Foot Ankle Int 2012;33:320–5.
[9] Kelly LA, Cresswell AG, Racinais S, Whiteley R, Lichtwark G. Intrinsic foot muscles have the capacity to control deformation of the longitudinal arch. J R Soc Interface 2014;11:20131188.
[10] Fiolkowski P, Brunt D, Bishop M, Woo R, Horodyski M. Intrinsic pedal musculature support of the medial longitudinal arch: an electromyography study. J Foot Ankle Surg 2003;42:327–33.
[11] Headlee DL, Leonard JL, Hart JM, Ingersoll CD, Hertel J. Fatigue of the plantar intrinsic foot muscles increases navicular drop. J Electromyogr Kinesiol 2008;18:420–5.
[12] Okamura K, Kanai S, Hasegawa M, Otsuka A, Oki S. The effect of additional activation of the plantar intrinsic foot muscles on foot dynamics during gait. Foot (Edinb) 2018;34:1–5.
[13] Mulligan EP, Cook PG. Effect of plantar intrinsic muscle training on medial longitudinal arch morphology and dynamic function. Man Ther 2013;18:425–30.
[14] Sulowska I, Oleksy L, Mika A, Bylina D, Soltan J. The influence of plantar short foot muscle exercises on foot posture and fundamental movement patterns in long-distance runners, a non-randomized, non-blinded clinical trial. PLoS One 2016;11:e0157917.
[15] Kamonseki DH, Goncalves GA, Yi LC, Junior IL. Effect of stretching with and without muscle strengthening exercises for the foot and hip in patients with plantar fasciitis: a randomized controlled single-blind clinical trial. Man Ther 2016;23:76–82.
[16] Angin S, Crofts G, Mickle KJ, Nester CJ. Ultrasound evaluation of foot muscles and plantar fascia in pes planus. Gait Posture 2014;40:48–52.
[17] Mansour R, Teh J, Sharp RJ, Ostlere S. Ultrasound assessment of the spring ligament complex. Eur Radiol 2008;18:2670–5.
[18] Redmond AC, Crane YZ, Menz HB. Normative values for the Foot Posture Index. J Foot Ankle Res 2008;1:6.
[19] Mueller MJ, Host JV, Norton BJ. Navicular drop as a composite measure of excessive pronation. J Am Podiatr Med Assoc 1993;83:198–202.

[20] O'Sullivan K, Kennedy N, O'Neill E, Ni Mhainin U. The effect of low-dye taping on rearfoot motion and plantar pressure during the stance phase of gait. BMC Musculoskelet Disord 2008;9:111.
[21] Redmond AC, Crosbie J, Ouvrier RA. Development and validation of a novel rating system for scoring standing foot posture: the Foot Posture Index. Clin Biomech (Bristol, Avon) 2006;21:89–98.
[22] Carson MC, Harrington ME, Thompson N, O'Connor JJ, Theologis TN. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. J Biomech 2001;34:1299–307.
[23] de Vos JvHS, Verbruggen Jpam WPHE. Repeatability of the Oxford foot model for kinematic gait analysis of the foot and ankle. Clin Res Foot Ankle 2015:03.
[24] Mann R, Inman VT. Phasic activity of intrinsic muscles of the foot. J Bone Joint Surg Am 1964;46:469–81.
[25] Gray EG, Basmajian JV. Electromyography and cinematography of leg and foot (normal and flat) during walking. Anat Rec 1968;161:1–15.
[26] Prachgosin T, Chong DY, Leelasamran W, Smithmaitrie P, Chatpun S. Medial longitudinal arch biomechanics evaluation during gait in subjects with flexible flatfoot. Acta Bioeng Biomech 2015;17:121–30.
[27] Wong YS. Influence of the abductor hallucis muscle on the medial arch of the foot: a kinematic and anatomical cadaver study. Foot Ankle Int 2007;28:617–20.
[28] Chang R, Kent-Braun JA, Hamill J. Use of MRI for volume estimation of tibialis posterior and plantar intrinsic foot muscles in healthy and chronic plantar fasciitis limbs. Clin Biomech (Bristol, Avon) 2012;27:500–5.
[29] Cheung RT, Sze LK, Mok NW, Ng GY. Intrinsic foot muscle volume in experienced runners with and without chronic plantar fasciitis. J Sci Med Sport 2016;19:713–5.
[30] Lee SY, Hertel J, Lee SC. Rearfoot eversion has indirect effects on plantar fascia tension by changing the amount of arch collapse. Foot (Edinburgh, Scotland) 2010;20:64–70.
[31] Kelly LA, Lichtwark G, Cresswell AG. Active regulation of longitudinal arch compression and recoil during walking and running. J R Soc Interface 2015;12:20141076.