Journal of Biomechanics 93 (2019) 118–125

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Effects of a foot orthosis inspired by the concept of a twisted osteoligamentous plate on the kinematics of foot-ankle complex during walking: A proof of concept

Vanessa L. Araújo [a], Thales R. Souza [a], Fabricio A. Magalhães [a], Thiago R.T. Santos [a], Kenneth G. Holt [b], Sergio T. Fonseca [a,*]

[a] Department of Physical Therapy, School of Physical Education, Physical Therapy and Occupational Therapy, Universidade Federal de Minas Gerais, Belo Horizonte, MG, Brazil
[b] Department of Physical Therapy and Athletic Training, College of Health and Rehabilitation Sciences: Sargent College, Boston University, Boston, MA, USA

ARTICLE INFO

Article history:
Accepted 22 June 2019

Keywords:
Biomechanics
Foot-ankle complex
Orthoses
Twisted osteoligamentous plate
Locomotion

ABSTRACT

It has been suggested that the foot acts as a twisted osteoligamentous plate to control pronation and facilitate supination during walking. The aim of this study was to investigate the effect of an orthosis inspired by the concept of a foot's twisted osteoligamentous plate on the kinematics of foot-ankle complex. Thirty-five subjects underwent a kinematic assessment of the foot-ankle complex during walking using three different orthoses: (1) Twisted Plate Spring (TPS) orthosis: inspired by the concept of a twisted osteoligamentous plate shape and made with a spring-like material (carbon fiber); (2) Flat orthosis: control orthosis made of a non-elastic material with a non-inclined surface; and (3) Rigid orthosis: control orthosis made of a non-elastic material, with the same shape of the TPS. Repeated measures analyses of variance demonstrated that the TPS reduced the duration and magnitude of rearfoot eversion ($p \leq 0.03$), increased rearfoot inversion relative to shank ($p < 0.01$), increased forefoot eversion relative to rearfoot ($p < 0.01$), and increased peak of plantar flexion of forefoot relative to rearfoot during the propulsive phase ($p = 0.01$) compared to Flat orthosis. The effects of the TPS were different from the Rigid orthosis, demonstrating that, alongside shape, material properties were a determinant factor for the obtained results. The findings of this study help clarify the role of a mechanism similar to a twisted osteoligamentous plate on controlling foot pronation and facilitating supination during the stance phase of walking.

1. Introduction

The foot structure has been described as a twisted osteoligamentous plate (Fig. 1A e B) (Eslami et al., 2007; Leardini et al., 2014; Sarrafian, 1987). The anterior part of this plate corresponds to the metatarsal heads, which are horizontally oriented, and the posterior part of this plate corresponds to the calcaneus, which is vertically oriented. This twisted plate determines the shape of the longitudinal and transverse arches of the foot (Sarrafian, 1987). In this view, many soft tissues are responsible for maintaining this twisted plate and establishing its stiffness, such as fascia, ligaments and muscles (Flanagan, 2013).

During loading phase of walking, weight bearing tends to untwist the plate which corresponds to the occurrence of foot pronation. This motion can be visualized as rearfoot eversion and longitudinal arch flattening (i.e. movement of dorsiflexion of forefoot relative to rearfoot) (Fig. 1C e D) (Flanagan, 2013; Sarrafian, 1987). As the rearfoot everts and forefoot dorsiflexes relative to rearfoot, the foot soft tissues elongate and store elastic energy (Ker et al., 1987; Sanchis-Sales et al., 2018; Stearne et al., 2016). The deformation-related increased tension at these tissues may help decelerate foot pronation. During the propulsive phase of walking, when weight bearing is transferred from rearfoot to forefoot during heel rise, the same tissue tension inverts the rearfoot and returns the foot longitudinal arch to its original shape (i.e. plantar flexion of forefoot relative to rearfoot) (Flanagan, 2013; Stearne et al., 2016).

We suggest, therefore, that one of the mechanisms that control foot pronation and facilitate its supination is the torsional spring behavior of the twisted plate resisting both rearfoot eversion and

* Corresponding author at: Graduate Program in Rehabilitation Science, Department of Physical Therapy, Universidade Federal de Minas Gerais, Av. Presidente Antônio Carlos, 6627 Campus Pampulha, CEP 31270-901 Belo Horizonte, MG, Brazil.
E-mail address: sfonseca@ufmg.br (S.T. Fonseca).

https://doi.org/10.1016/j.jbiomech.2019.06.020
0021-9290/

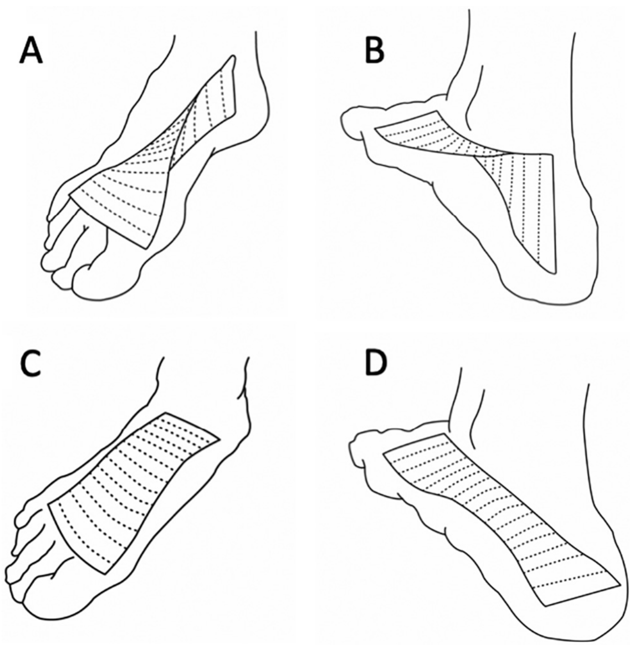


Fig. 1. The concept of a twisted foot osteoligamentous plate, in which the anterior part represents the metatarsal heads and the posterior part represents the calcaneus. (A) and (B): foot in the neutral position of pronation/supination with the osteoligamentous plate twisted (i.e. posterior part of the plate is vertical). (C) and (D): pronated foot in closed kinematic chain with the osteoligamentous plate untwisted (i.e. posterior part of the plate is less vertical).

longitudinal arch flattening (Gomes et al., 2019; Souza et al., 2014b). The 'foot spring' would be able to store energy during its deformation into pronation and, then, return to its original shape by means of elastic energy release (Riddick and Kuo, 2016; Willwacher et al., 2013), which facilitates foot supination as the foot plate restores to neutral. Foot supination converts the foot into a rigid lever, which is necessary to carry out effective propulsion (Blackwood et al., 2005; Kirby, 2000).

Despite the existence of a theoretical framework for the role of the foot's twisted osteoligamentous plate on foot pronation/-supination, this mechanism is not considered when building foot orthoses for controlling excessive pronation. It has been suggested that orthotics with rigid medial wedges reduce rear-foot eversion (Cheung et al., 2011; Desmyttere et al., 2018). These orthoses function as hard physical constraints to motion, and are not inspired by a physiological dynamic mechanism of pronation control. Moreover, the use of rigid medial profile orthotics may reduce the magnitude of pronation, but it seems they do not reduce its duration (Bishop et al., 2016; Brown et al., 1995). An earlier pronation end may favor foot supination and effective propulsion (Blackwood et al., 2005). These rigid orthoses also have the potential to block pronation and arch flattening, impairing the energy absorption and the elastic storage/return mechanism (Stearne et al., 2016; Takahashi et al., 2016).

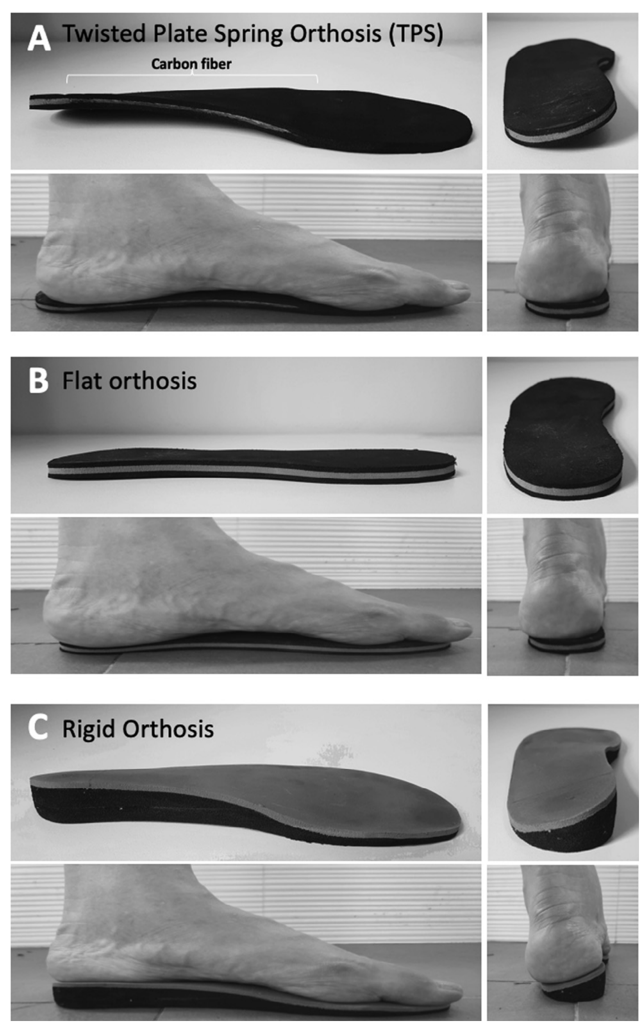
Thus, we propose a biologically inspired orthosis that has the characteristics of the foot's osteoligamentous plate. This orthosis would have the shape of a twisted plate (i.e forefoot horizontally oriented, and rearfoot more vertically oriented), positioning the forefoot more everted than the rearfoot when no weight is supported. On landing, as the body weight compresses the orthosis (loading phase), it would untwist. In addition, as the rearfoot starts to unload (heel rise), the orthosis would provide elastic energy as it returns back to its twisted position.

Hence, the aim of the present proof of concept study was to investigate the effects of an orthosis inspired by the foot's twisted osteoligamentous plate on kinematics of the foot-ankle complex during the stance phase of walking. We hypothesized that the orthosis inspired by the Twisted Plate Spring (TPS) would reduce the magnitude and duration of pronation during loading response through midstance, and increase supination during propulsion compared to a Flat orthosis that simulate the natural behavior of the foot. We also compared the effects of the TPS orthosis with a medial wedge rigid orthosis, and we hypothesized that the Rigid orthosis would decrease foot pronation magnitude, but would not decrease pronation duration neither increase supination.

2. Methods

2.1. Participants

Thirty-five subjects (17 men and 18 women, aged 22.7 ± 3.1 years, body mass of 62.7 ± 10.5 kg, height of 1.68 ± 0.07 m) participated in this study. The eligibility criteria were: (1) age between 18 and 40 years; (2) having Body Mass Index (BMI) less than 30 kg/m^2; (3) wearing a shoe size of 7.5″, for women, or a shoe size of 8.5″, for men, according to United States system; (4) experiencing no pain or injury in the lower limbs or trunk during the week before testing; (5) having no history of



orthopedic surgery or fracture in the lower limbs or trunk; (6) being free of neurological or rheumatologic pathologies; and (7) not currently using foot orthoses. The exclusion criterion was the inability to perform any assessed task. The number of participants was determined based on the effect size achieved in a pilot study with 10 participants and the same study design ($d_z = 0.5$), a statistical power of 80% and a probability of type I error of 0.05. This work was approved by the institution's Ethics in Research Committee.

2.2. Foot orthoses and shoes

This study used three prefabricated foot orthoses (Fig. 2). The TPS orthosis was inspired by the foot's twisted osteoligamentous plate, had a shape similar to this plate and was made of a spring-like material. The other two orthoses worked as control to the effects of the TPS. A Flat orthosis had a non-inclined surface to simulate the foot's natural behavior, and worked as a control for the effect of the shape and material of the TPS. The last orthosis, the Rigid, had a medial wedge on rearfoot, and as a consequence had its upper surface with the same shape of the TPS but, with no spring-like behavior. The Rigid orthosis was designed to control for the effect of the shape of the TPS upper surface, but not for its springiness. The orthoses were available for women's shoe size of 7.5″ and men's shoe size of 8.5″. The orthoses were inserted into neutral sneakers (Futsal Topper Dominator III, Topper, São Paulo, Brazil) (Fig. 3). More details of each orthosis are described in Fig. 2. The TPS orthosis' torsion torque *versus* angle curve is showed in Fig. 4. Supplementary material show the equipment used to perform this measure.

Fig. 2. Foot orthoses: (A) Medial and posterior views of the Twisted Plate Spring (TPS) orthosis with and without a person stepping on it. The TPS orthosis has a twisted shape – its anterior part was flat, keeping the metatarsal heads horizontal, and its posterior part inclined at 17° relative to the ground, simulating the twisted shape of the foot osteoligamentous plate. Under weight, this orthosis untwists so its posterior part goes flat (i.e. decreases its inclination), as we can observe in the picture with a person stepping on it. The orthosis stiffness causes it to twist as soon as the weight reduces, returning the posterior part to 17° inclination. This orthosis had, initially, its prototype manufactured in a 3D printer (uPrint SE Plus, Stratasys, Eden Prairie, USA). The prototype allowed the construction of a model, which was used to create the orthosis, with carbon fiber to provide a spring-like behavior. The carbon fiber is a stiff material with high elastic modulus (230 gigapascals GPa) and high tensile strength (4900 megapascal MPa). This carbon fiber (thickness of 0.2 cm) covered from posterior part of the calcaneus up to the metatarsal heads, and ethyl vinyl acetate (EVA) (thickness of 0.2 cm) covered the phalanges and the posterior part of calcaneus. This EVA was medium soft with a hardness of 40, according to Shore A scale (0–100). The carbon fiber did not cover the metatarsophalangeal joints to prevent limiting their dorsiflexion during the propulsive phase of gait. Furthermore, the carbon fiber did not cover the least posterior part of the calcaneus (approximately 1.5 cm) in order to allow comfortable heel contact during walking (i.e. help to cushioning the calcaneus for initial contact) and prevent a large external plantar flexion torque after initial contact. Finally, the surface of this orthosis was superiorly and inferiorly covered with a thin layer (thickness of 0.2 cm) of low-density EVA to provide comfort. This EVA was soft with a hardness of 25, according to Shore A scale (0 to 100). The total thickness of this orthosis was 0.6 cm. As this orthosis had a twisted shape and its posterior part was inclined, the distance between its medial posterior part and the ground was of 2.2 cm. (B) Medial and posterior views of the Flat orthosis with and without a person stepping on it. This orthosis had a flat surface and was made of medium soft EVA (thickness of 0.3 cm) with a hardness of 40, according to Shore A scale (0–100). This orthosis was covered with a thin layer (thickness of 0.2 cm) of soft EVA (hardness of 25 Shore A) superiorly and inferiorly to provide comfort. The total thickness of this orthosis was 0.7 cm. (C) Medial and posterior views of the Rigid orthosis with and without a person stepping on it. This orthosis was posted medially in the rearfoot using medium soft EVA with a hardness of 40, according to Shore A scale (0–100). This orthosis was also covered with a thin layer (thickness of 0.2 cm) of soft EVA (hardness of 25 Shore A). The amount of posting produced the same inclination and shape of the TPS orthosis' upper surface (i.e. 17°). However, the Rigid orthosis' cannot untwist during weight bearing situations as happen with the TPS orthosis. The total thickness of the TPR orthosis was 0.7 cm in the anterior and lateral parts, whereas the posterior medial part had 2.2 cm thickness.

2.3. Procedures

After signing the written informed consent, volunteers had their body mass and height measured. An optoelectronic system with eight cameras (ProReflex, Qualisys AB, Gothenburg, Sweden) was used to collect the kinematic data at 120 Hz. To determine the gait events, kinetic data were acquired by two force plates (AMTI OR6-7-1000, Watertown, USA) at 1200 Hz. Passive reflexive markers were attached to the participant's left foot and shank as shown in Fig. 5. Firstly, a static trial of five seconds was carried out with the participant in the relaxed barefoot standing position. Then, three additional static trials of five seconds in barefoot standing position were conducted with the subtalar joint positioned in neutral by an experienced examiner (Houck et al., 2008). The intra-examiner reliability for the subtalar joint placement in neutral was performed in a pilot study with 15 subjects. The measure was carried out in two distinct days, with an interval of four to seven days. The Intraclass Correlation Coefficient ($ICC_{3,3}$) was 0.93 for the rearfoot-shank angle in the frontal plane, which represents an excellent intra-examiner reliability (Fleiss, 1986).

The following steps were repeated once for each orthosis condition. First, the participants put on the sneakers with one of the three conditions: (a) TPS, (b) Flat orthosis, or (c) Rigid orthosis. The order of these three conditions was block randomized for each volunteer. Additional markers were placed over the right shoe and over the sacrum to determine gait events and speed (Fig. 5). The participant walked 5 min at self-selected speed to familiarize with each orthosis. Subjects walked on an 8 m walkway at their self-selected speed, until six valid trials were obtained for each condition. As soon as the valid trials were obtained, the participants filled a comfort visual analog scale (VAS) (Mills et al., 2010).

2.4. Data processing and reduction

The kinematic and kinetic data were processed using the Visual 3D™ (C-motion Inc., Germantown, USA) software. The forefoot, rearfoot and shank kinematic model (Souza et al., 2014a) were implemented. Markers trajectories and ground reaction forces were filtered using low-pass Butterworth with a cut-off frequency of 6 Hz and 10 Hz, respectively. The relative angles between forefoot and rearfoot as well as between rearfoot and shank were calculated using the following Cardan sequence: medio-lateral, antero-posterior and longitudinal axes. The neutral positions (0°) for all angles were defined as the mean joint position of the static trials with the subtalar joint in neutral position. Initial contact and toe-off of the left foot were determined using vertical ground reaction force (threshold of 10 N) and the distance between the foot's center of mass and the platform's center of pressure (threshold of 0.20 m). Right foot's toe-off was determined based on the minimum value of anterior-posterior displacement of the fifth metatarsal head marker relative to sacral marker (Zeni et al., 2008). Left foot's heel rise was determined based on vertical velocity of the heel marker (Ghoussayni et al., 2004).

A MATALB® routine was written to extract the variables. All variables were measured for each stride, and the average across all measured strides in each condition was used for analysis. The variables chosen to be analyzed were: (1) mean rearfoot-shank angle in the frontal plane—foot pronation/supination indexing (Chuter, 2010); (2) mean forefoot-rearfoot angle in the frontal plane—amount of twist of the osteoligamentous plate indexing; and (3) mean forefoot-rearfoot angle in the sagittal plane—longitudinal arch flattening and raising indexing (Wilken et al., 2011). These mean angles values were calculated during loading response

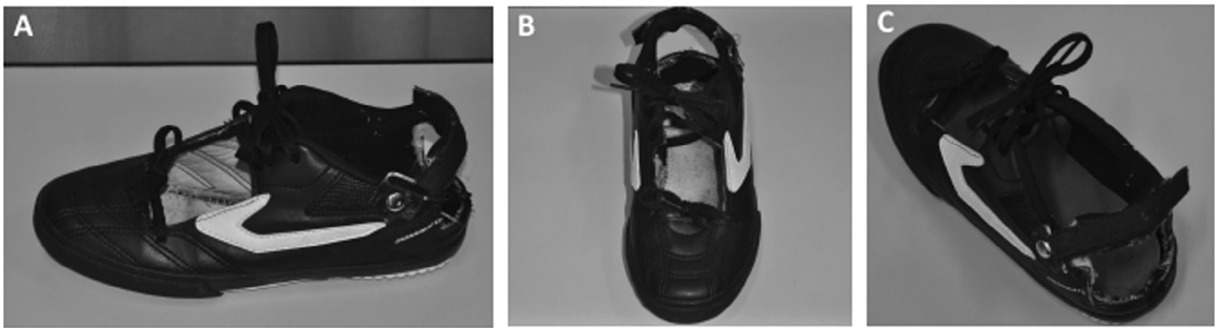


Fig. 3. Lateral (A), superior (B) and posterior (C) views of the adapted sneaker with anterior and posterior custom made openings. The sneakers' insole was removed, and an anterior and posterior opening was made in the sneakers to allow direct placement of forefoot and rearfoot clusters.

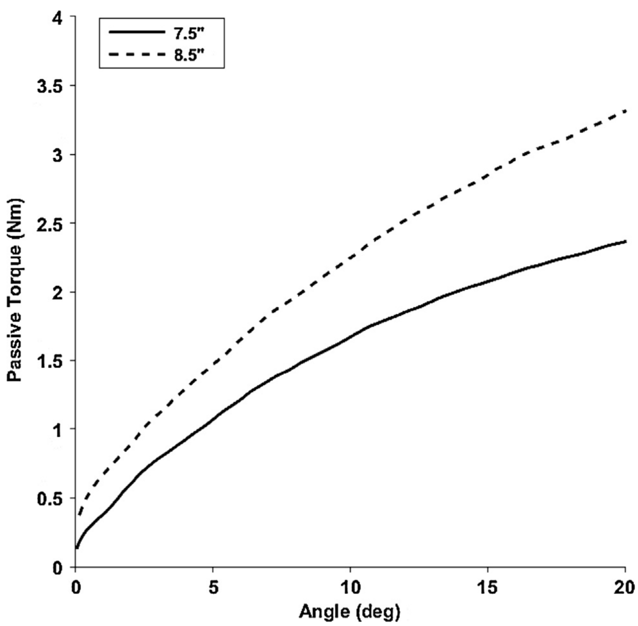


Fig. 4. Torque-angle curves of both Twisted Plate Spring (TPS) orthoses. Their slopes represent the passive stiffness, which is slightly higher for the larger orthosis than for the smaller one. Although both insoles were made of the same material, the larger orthosis (i.e. Orthosis men's size 8.5″) has more carbon fiber due to its size. Thus, the increased amount of resisting material resulted in a slightly higher stiffness of the orthosis size 8.5″ compared to the size 7.5″.

(between left initial contact and right toe-off), midstance (between right toe-off and left heel rise) and propulsive (between left heel rise and left toe-off) phases. Moreover, rearfoot-shank eversion peak, time elapsed until rearfoot-shank eversion peak, and rearfoot-shank inversion peak during propulsive phase were calculated. Finally, forefoot-rearfoot dorsiflexion peak, time elapsed until forefoot-rearfoot dorsiflexion peak, and forefoot-rearfoot plantar flexion peak were also calculated during the propulsive phase.

Gait speed (ms^{-1}) was calculated as the antero-posterior distance covered by the sacrum marker during the stance phase divided by the time spent. The volunteers' comfort for each orthosis was measured as the length (cm) in the VAS.

The TPS orthosis inclination was measured with a digital inclinometer - Mini Digital Protactor (Cyntrax, Rio de Janeiro, Brazil). The maximum torque of the TPS orthosis was calculated as the maximum torque achieved during the orthosis untwisting movement, and the stiffness as the slope of the torque-angle time-series during the orthosis untwisting movement.

2.5. Statistical analysis

Repeated measures analyses of variance (ANOVA) were conducted to verify if the dependent variables changed among the three conditions. When main effects were significant, two least-significant difference contrasts were conducted: (a) TPS *versus* Flat orthoses, and (b) TPS *versus* Rigid orthoses. The probability of type I error was set at 0.05.

3. Results

Mean and standard deviation of the dependent variables as well as the results of the ANOVA are presented in Table 2. Mean curves of rearfoot-shank and forefoot-rearfoot motion are plotted in Fig. 6.

There was a significant main effect of the foot orthoses on all variables related to rearfoot-shank in the frontal plane. Contrasts showed that the TPS orthosis condition reduced rearfoot eversion peak, increased rearfoot inversion peak and decreased mean eversion angles of rearfoot-shank during all walking phases compared to the Flat orthosis. On the other hand, when compared to Rigid orthosis, the TPS orthosis condition presented greater eversion peak angle, smaller inversion peak and greater mean eversion angles in all the walking phases. Moreover, TPS orthosis condition decreased time to reach eversion peak in comparison with both Flat and Rigid orthoses.

Regarding the forefoot-rearfoot in the frontal plane, there was a significant main effect of the foot orthoses on all variables. Contrasts demonstrated that the TPS orthosis condition increased mean eversion angle during all walking phases compared to the Flat orthosis. In addition, the TPS orthosis condition had reduced eversion angles during all phases compared to the Rigid orthosis.

In relation to the forefoot-rearfoot in the sagittal plane, no significant main effect was observed for mean angles during the loading response phase. However, there was a significant main effect of the foot orthoses on the following variables: dorsiflexion and plantar flexion peaks; time to achieve dorsiflexion peak; and mean angle during the midstance and propulsive phases. Contrasts showed that TPS orthosis condition was not different from Flat orthosis for dorsiflexion peak, time to dorsiflexion peak and mean angle during midstance and propulsive phase. However, contrasts demonstrated that the TPS orthosis reduced dorsiflexion peak, time to achieve dorsiflexion peak and mean dorsiflexion angle during midstance and propulsive phases when compared to the Rigid orthosis. Finally, TPS orthosis condition showed greater plantar flexion peak during the propulsive phase than both Flat and Rigid orthoses.

The results for gait speed and orthoses comfort are presented in Table 3. The ANOVA showed that speed did not change across conditions. Furthermore, ANOVA revealed significant main effect for

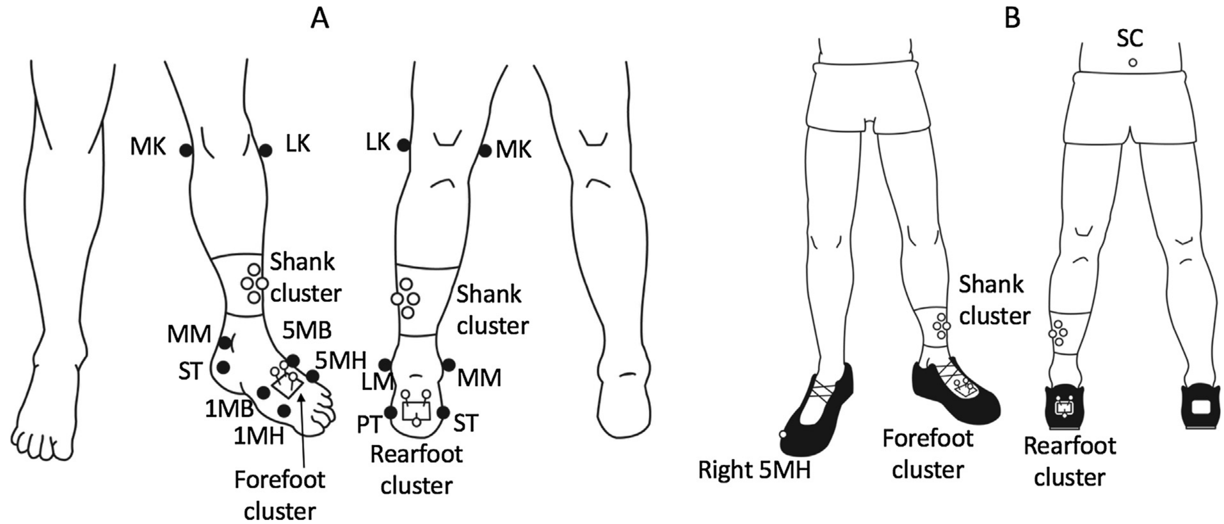


Fig. 5. (A) Markers and clusters used to collect kinematic data of the foot-ankle complex during static trial of five seconds with the participant in the relaxed standing position. (B) Markers and clusters used to collect kinematic data of the foot-ankle complex during walking trials. Anatomical markers are displayed in black, whereas tracking markers are displayed in white. MK = medial epicondyle of the femur; LK = lateral epicondyle of the femur; MM = medial malleolus; LM = lateral malleolus; 1 MB = first metatarsal base; 1MH = first metatarsal head; 5 MB = fifth metatarsal base; 5MH = fifth metatarsal head; PT = peroneal tuberosity; ST = sustentaculum tali; SC = sacrum marker placed between posterior superior iliac spines.

Table 1
Descriptive statistics of Twisted Plate Spring (TPS) orthosis characteristics.

	Orthosis women's size 7.5″			Orthosis men's size 8.5″		
	Measure 1	Measure 2	Measure 3	Measure 1	Measure 2	Measure 3
Inclination (°)	17.00	17.50	16.20	16.40	17.00	16.60
Passive torque peak (Nm)	2.37	2.51	3.02	3.31	3.47	3.87
Passive stiffness (Nm/°)	0.11	0.12	0.14	0.15	0.17	0.17

Measure 1 = in the beginning of data collection (day 1). Measure 2 = in the middle of data collection (day 15). Measure 3 = in the end of data collection (day 30).

Table 2
Descriptive and inferential statistics of the dependent variables measured during walking trials in the three test conditions: (a) Twisted Plate Spring (TPS) orthosis, (b) Flat orthosis, and (c) Rigid orthosis.

Variables			Mean (SD)			ANOVA			Contrasts			
			TPS	Flat	Rigid	p	F	η^2p	TPS × Flat		TPS × Rigid	
									p	d_z	p	d_z
Frontal plane	Rearfoot-shank	Loading resp. phase (°)	−4.23 (3.63)	−4.77 (3.52)	−0.12 (3.47)	<0.01*	109.98	0.76	0.03*	0.38	<0.01*	1.80
		Midstance phase (°)	−6.78 (3.43)	−7.71 (3.59)	−1.94 (3.60)	<0.01*	216.83	0.86	<0.01*	0.58	<0.01*	2.98
		Propulsive phase (°)	−2.82 (4.00)	−4.24 (3.98)	−0.22 (4.08)	<0.01*	87.39	0.72	<0.01*	0.72	<0.01*	1.57
		EV peak (°)	−8.48 (3.70)	−9.44 (3.77)	−4.02 (3.88)	<0.01*	202.94	0.86	<0.01*	0.63	<0.01*	2.82
		Time to EV peak (%)	44.33 (15.17)	49.17 (14.71)	58.03 (19.49)	<0.01*	9.43	0.22	0.01*	0.47	<0.01*	0.60
		INV peak (°)	1.72 (4.28)	0.34 (4.67)	3.32 (4.38)	<0.01*	39.28	0.54	<0.01*	0.67	<0.01*	0.87
	Forefoot-rearfoot	Loading resp. phase (°)	−1.86 (3.51)	1.10 (3.58)	−3.46 (3.01)	<0.01*	181.07	0.84	<0.01*	0.52	<0.01*	0.91
		Midstance phase (°)	−3.06 (2.99)	−1.24 (2.97)	−5.77 (2.84)	<0.01*	134.76	0.80	<0.01*	1.26	<0.01*	1.67
		Propulsive phase (°)	−3.89 (3.12)	−2.01 (3.25)	−5.74 (3.18)	<0.01*	96.83	0.74	<0.01*	1.26	<0.01*	1.08
Sagittal plane	Forefoot-rearfoot	Loading resp. phase (°)	1.14 (2.32)	1.69 (2.72)	1.79 (3.19)	0.16	1.97	0.06	–	–	–	–
		Midstance phase (°)	3.65 (2.28)	3.99 (3.04)	5.10 (3.15)	<0.01*	10.15	0.23	0.25	0.20	<0.01*	0.67
		Propulsive phase (°)	0.23 (2.79)	0.46 (3.30)	1.96 (3.51)	<0.01*	17.17	0.34	0.46	0.12	<0.01*	0.92
		DF peak (°)	5.24 (2.53)	5.51 (3.02)	6.66 (3.38)	<0.01*	8.97	0.21	0.40	0.15	<0.01*	0.64
		Time to DF peak (%)	46.83 (16.53)	42.99 (16.42)	55.20 (14.53)	<0.01*	16.40	0.33	0.09	0.30	<0.01*	0.71
		PF peak (°)	−7.34 (3.56)	−6.24 (3.67)	−5.11 (4.18)	<0.01*	15.81	0.32	0.01*	0.50	<0.01*	0.89

SD = standard deviation; resp: response; EV = eversion; INV = inversion; DF = dorsiflexion; PF = plantar flexion; η^2p = partial eta squared; d_z = Cohen's effect size for paired sample; * = significant effect.
Negative values of angles indicate eversion or plantar flexion, and positive values indicate inversion or dorsiflexion.
The angles during loading response phase, midstance and propulsive phase were calculated as the mean angle of each phase. The INV and PF peaks were extracted as the peaks of the angle curves during the propulsive phase.

orthoses comfort. The contrasts demonstrated no difference between Flat and TPS orthoses, and TPS orthosis was more comfortable than Rigid orthosis.

The TPS orthosis characteristics did not have large variation during the study. The values of orthosis inclination, maximum torque and stiffness are shown in Table 1.

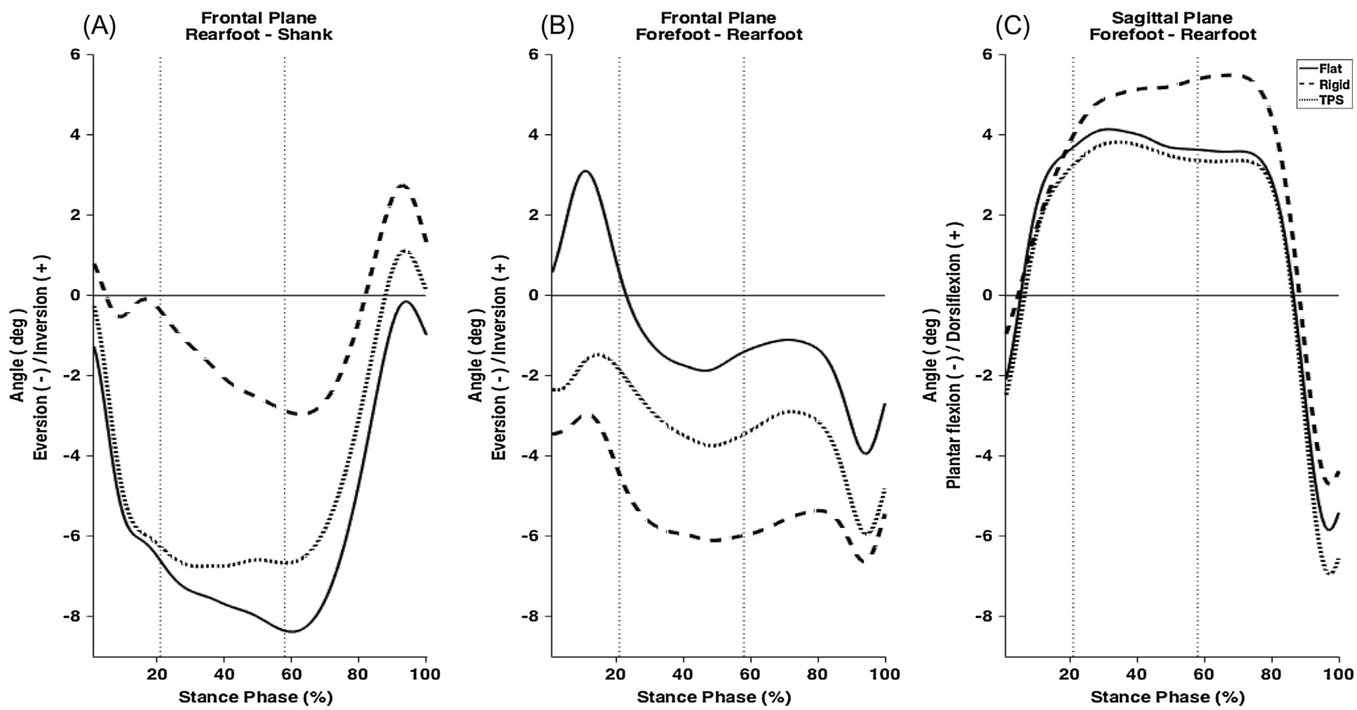


Fig. 6. Curves of the mean values of all subjects (n = 35) for each orthosis condition for: (A) rearfoot-shank in the frontal plane, (B) forefoot-rearfoot in the frontal plane, and (C) forefoot-rearfoot in the sagittal plane during stance phase of gait. Vertical lines are for right foot's toe-off and left foot's heel rise, and divide the stance phase into three sub phases: loading response phase (0 to ±20%), midstance (±20% to ±60%) and propulsive phase (±60% to 100%).

Table 3
Descriptive and inferential statistics for gait speed and orthoses comfort measured during walking trials in the three test conditions: (a) Twisted Plate Spring (TPS) orthosis, (b) Flat orthosis, and (c) Rigid orthosis.

Variables	Mean (SD)			ANOVA			Contrasts			
	TPS	Flat	Rigid	p	F	η^2p	TPS × Flat		TPS × Rigid	
							p	d_z	p	d_z
Gait speed (m/s)	1.26 (0.13)	1.25 (0.15)	1.26 (0.15)	0.86	0.16	0.01	–	–	–	–
Orthoses comfort (cm)	6.97 (1.92)	7.08 (1.67)	3.51 (1.90)	<0.01*	60.28	0.64	0.71	0.06	<0.01*	1.42

SD = standard deviation; * = significant effect.
For orthoses comfort, the greater is the value, better is the comfort.

4. Discussion

The findings of the present study support the hypothesis that the TPS orthosis reduced the duration and magnitude of rearfoot eversion and facilitated rearfoot inversion relative to shank during the stance phase of walking compared to the Flat orthosis. The TPS orthosis, as expected, also increased forefoot eversion relative to rearfoot during the entire stance phase, and increased peak of plantar flexion of forefoot relative to rearfoot during the propulsive phase in comparison with the Flat orthosis. These findings suggest that the TPS orthosis increased the twisting of the osteoligamentous plate during stance, and raised the longitudinal arch during propulsion. Moreover, the effects of TPS orthosis were different from the Rigid orthosis, demonstrating that the orthosis shape was not the only determinant for the observed effects, but also the orthosis' material. Therefore, these results reinforce the hypotheses that the TPS orthosis can be an important mechanism for controlling foot pronation and facilitating supination.

The results of this study corroborate those of Sarrafian (1987), who stated that the foot behaves as a twisted plate, according to the counter motion of the rearfoot and forefoot. The outcomes of the present study also reinforce the idea that the torque generated by midfoot can decelerate foot pronation and increase supination. Therefore, it seems that midfoot stiffness can play an important role in foot motion during walking, as suggested by Gomes et al. (2019), Hunt et al. (2001), Sanchis-Sales et al. (2018) and Souza et al. (2014b).

The gait speed was not different among conditions. As the amount of foot pronation/supination is influenced by gait speed (Hornestam et al., 2016), it was important to make sure that differences found were due to the orthoses' effects. Moreover, the TPS orthosis comfort was not different from that reported for the Flat orthosis. It suggests that the orthosis architecture produced in carbon fiber did not promote any additional discomfort. Although the TPS and Rigid orthoses had similar shapes, the latter orthosis was quite uncomfortable when compared to the other orthosis. The discomfort of the Rigid orthosis may have interfered in its kinematics effects (Che et al., 1994). Thus, the kinematic change observed when using the Rigid orthosis may be due to its mechanical effect of hampering pronation or due to its discomfort. Finally, the TPS orthosis characteristics did not change meaningfully throughout the study (maximum variation of 1.3° in inclination, 0.65 Nm in torque peak and 0.03 Nm/° in stiffness), which ensured its correct functioning for all participants.

The observed effects of the TPS orthosis were different from the Rigid orthosis. It should be noted that the Rigid orthosis did not intend to mimic any foot orthosis proposed in the literature, as it has a large amount of medial wedge than the ones usually used. The aim of this orthosis was solely to serve as a shape control for the TPS orthosis. Considering the rearfoot-shank in the frontal plane, the Rigid orthosis decreased eversion in comparison to the TPS orthosis. During loading response phase, the mean rearfoot-shank eversion for the Rigid orthosis was extremely limited (0.12°), indicating that this orthosis blocked eversion or the orthosis's discomfort lead the participants to avoid eversion. On the other hand, the TPS orthosis allowed rearfoot eversion to occur. Consequently, the mean rearfoot-shank eversion during loading response phase was 4.23°, which represents a desirable effect, as foot pronation is necessary to allow energy absorption (Leardini et al., 2014; McPoil and Knecht, 1985). Even though the TPS orthosis reduced rearfoot eversion when compared to the Flat orthosis, it did not block this movement, considering the values of rearfoot-shank range of motion reported in the literature (Rankine et al., 2008). Examining forefoot-rearfoot in the sagittal plane during the propulsive phase, the TPS orthosis increased forefoot-rearfoot plantar flexion peak compared to Rigid orthosis, suggesting the importance of a spring behavior to allow the hypothesized effect of raising the arch. In addition, the TPS orthosis decreased duration of rearfoot-shank eversion compared to Rigid orthosis. It seems that the TPS orthosis acted to store/return energy, anticipating foot supination. This effect indicates that the combination of the osteoligamentous plate shape with its material properties and comfort are essential to produce the desired foot behavior.

The effect size analyses of the pairwise comparison between TPS and Flat orthoses revealed that the magnitude of the effects were greater at propulsive phase ($d_z = 0.72$), followed by midstance ($d_z = 0.58$) and loading response phase ($d_z = 0.38$). This makes sense because as the greater the contact area of the anterior part of plate on the ground, the greater will be the effectiveness of the twisted plate to resist both rearfoot eversion and longitudinal arch flattening. During the beginning of loading response phase, the metatarsal heads does not contact the ground and are not stabilized and, thus, the orthosis are not in proper position to generate as much resistance moment to rearfoot eversion and longitudinal arch flattening. During midstance, the effect produced by the TPS orthosis increased, as the forefoot rested on the ground, allowing the full functioning of the spring. During the propulsive phase, the observed effect size was even greater than that of the midstance, since weight bearing is partially removed from rearfoot, allowing the orthosis to increase rearfoot inversion by means of elastic energy return. Therefore, the greater effect of the TPS orthosis was related to foot supination facilitation.

Contrary to our hypotheses, the TPS orthosis did not reduce the duration and magnitude of forefoot dorsiflexion relative to rearfoot (i.e. reduce longitudinal arch flattening) in comparison to the Flat orthosis. This finding may be explained by the heel elevation produced by the TPS orthosis, which may have increased rearfoot plantar flexion and, as a consequence, increased forefoot-rearfoot dorsiflexion. This effect of increasing forefoot-rearfoot dorsiflexion may counterbalance the hypothesized effect of increasing forefoot plantar flexion relative to rearfoot (i.e. increase longitudinal arch raising). It is possible that a stiffer TPS orthosis might be necessary to produce greater effects in forefoot-rearfoot behavior in the sagittal plane.

The present study was designed as a proof of concept about the effect of an orthosis inspired by the foot's twisted plate on foot kinematics. As a proof of concept, this study does not intend to recommend the use of the TPS orthosis. Before using this TPS orthosis in clinical practice, additional studies are required to compare this orthosis with others currently used to modify foot motion. In addition, the magnitude of the effects on kinematics using TPS orthosis was small and its clinical importance remains unclear.

The present study had some limitations. Firstly, this study did not measure the effect of TPS orthosis with different stiffness magnitudes and on springs that have hard non-linear characteristics as most human "springs". Secondly, we limited participants BMI and orthosis size, and this limits the generalization of the results, since the orthoses effects may be dependent of the subject mass and orthosis stiffness. Thirdly, the effect of the orthoses may be different if they were inserted in sneakers with different characteristics (Stacoff et al., 1991).

The orthosis inspired by the concept of a foot's twisted osteoligamentous plate was able to control foot pronation and facilitate/anticipate supination during walking. The results showed that both shape and spring like behavior are linked to these effects. The TPS orthosis proposed by this study represents a new biologically inspired approach for managing excessive foot pronation.

Acknowledgments

The authors gratefully acknowledge the financial support offered by the Brazilian government agencies CNPq, Brazil (Conselho Nacional de Desenvolvimento Científico e Tecnológico), FAPEMIG, Brazil (Fundação de Amparo à Pesquisa do Estado de Minas Gerais) and CAPES, Brazil (Coordenação de Aperfeiçoamento de Pessoal de Nível Superior).

Declaration of Competing Interest

The orthotic device tested in the current manuscript is associated with a US Patent request, in which the co-author KGH is listed as inventor. The author affirms the scientific integrity of all data presented in this manuscript and declare no other conflict of interest. The other authors affirm no conflict of interest regarding the publication of this manuscript.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.06.020.

References

Bishop, C., Arnold, J.B., May, T., 2016. Effects of taping and orthoses on foot biomechanics in adults with flat-arched feet. Med. Sci. Sports Exerc. 48, 689–696.

Blackwood, C.B., Yuen, T.J., Sangeorzan, B.J., Ledoux, W.R., 2005. The midtarsal joint locking mechanism. Foot Ankle Int. 26, 1074–1080.

Brown, G.P., Donatelli, R., Catlin, P.A., Wooden, M.J., 1995. The effect of two types of foot orthoses on rearfoot mechanics. J. Orthop. Sport. Phys. Ther. 21, 258–267.

Che, H., Nigg, B.M., de Koning, J., 1994. Relationship between plantar pressure distribution under the foot and insole comfort. Clin. Biomech. 9, 335–341.

Cheung, R.T.H., Chung, R.C.K., Ng, G.Y.F., 2011. Efficacies of different external controls for excessive foot pronation: A meta-analysis. Br. J. Sports Med. 45, 743–751.

Chuter, V.H., 2010. Relationships between foot type and dynamic rearfoot frontal plane motion. J. Foot Ankle Res. 3, 1–6.

Desmyttere, G., Hajizadeh, M., Bleau, J., Begon, M., 2018. Effect of foot orthosis design on lower limb joint kinematics and kinetics during walking in flexible pes planovalgus: a systematic review and meta-analysis. Clin. Biomech. 59, 117–129.

Eslami, M., Begon, M., Farahpour, N., Allard, P., 2007. Forefoot-rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running. Clin. Biomech. 22, 74–80.

Flanagan, S.P., 2013. Biomechanics: A Case-Based Approach. Jones & Barlett Learning, Burlington, USA.

Fleiss, J.L., 1986. Reliability of measurement. In: Fleiss, J.L. (Ed.), The Design and Analysis of Clinical Experiments. John Wiley & Sons Inc, New York, USA, pp. 32–56.

Ghoussayni, S., Stevens, C., Durham, S., Ewins, D., 2004. Assessment and validation of a simple automated method for the detection of gait events and intervals. Gait Posture 20, 266–272.

Gomes, R.B.O., Souza, T.R., Paes, B.D.C., Magalhães, F.A., Gontijo, B.A., Fonseca, S.T., Ocarino, J.M., Resende, R.A., 2019. Foot pronation during walking is associated

to the mechanical resistance of the midfoot joint complex. Gait Posture 70, 20–23.
Hornestam, J.F., Souza, T.R., Arantes, P., Ocarino, J., Silva, P.L., 2016. The effect of walking speed on foot kinematics is modified when increased pronation is induced. J. Am. Podiatr. Med. Assoc. 106, 419–426.
Houck, J.R., Tome, J.M., Nawoczenski, D.A., 2008. Subtalar neutral position as an offset for a kinematic model of the foot during walking. Gait Posture 28, 29–37.
Hunt, A.E., Smith, R.M., Torode, M., Keenan, A.M., 2001. Inter-segment foot motion and ground reaction forces over the stance phase of walking. Clin. Biomech. 16, 592–600.
Ker, R.F., Bennett, M.B., Bibby, S.R., Kester, R.C., Alexander, R.M., 1987. The spring in the arch of the human foot. Nature 325, 147–149.
Kirby, K.A., 2000. Biomechanics of the normal and abnormal foot. J. Am. Podiatr. Med. Assoc. 90, 30–34.
Leardini, A., O'Connor, J.J., Giannini, S., 2014. Biomechanics of the natural, arthritic, and replaced human ankle joint. J. Foot Ankle Res. 7, 8.
McPoil, T.G., Knecht, H.G., 1985. Biomechanics of the foot in walking: a function approach. J. Orthop. Sports Phys. Ther. 7, 69–72.
Mills, K., Blanch, P., Vicenzino, B., 2010. Identifying clinically meaningful tools for measuring comfort perception of footwear. Med. Sci. Sports Exerc. 42, 1966–1971.
Rankine, L., Long, J., Canseco, K., Harris, G.F., 2008. Multisegmental foot modeling: a review. Crit. Rev. Biomed. Eng. 36, 127–181.
Riddick, R.C., Kuo, A.D., 2016. Soft tissues store and return mechanical energy in human running. J. Biomech. 49, 436–441.
Sanchis-Sales, E., Sancho-Bru, J.L., Roda-Sales, A., Pascual-Huerta, J., 2018. Effect of static foot posture on the dynamic stiffness of foot joints during walking. Gait Posture 62, 241–246.
Sarrafian, S.K., 1987. Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading. Foot Ankle 8, 4–18.
Souza, T.R., Fonseca, H.L., Vaz, A.C.A., Antero, J.S., Marinho, C.S., Fonseca, S.T., 2014a. Between-day reliability of a cluster-based method for multisegment kinematic analysis of the foot-ankle complex. J. Am. Podiatr. Med. Assoc. 104, 601–609.
Souza, T.R., Mancini, M.C., Araújo, V.L., Carvalhais, V.O.C., Ocarino, J.M., Silva, P.L., Fonseca, S.T., 2014b. Clinical measures of hip and foot–ankle mechanics as predictors of rearfoot motion and posture. Man. Ther. 19, 379–385.
Stacoff, A., Kälin, X., Stüssi, E., 1991. The effects of shoes on the torsion and rearfoot motion in running. Med. Sci. Sports Exerc. 23, 482–490.
Stearne, S.M., McDonald, K.A., Alderson, J.A., North, I., Oxnard, C.E., Rubenson, J., 2016. The Foot's Arch and the Energetics of Human Locomotion. Sci. Rep. 6, 1–10.
Takahashi, K.Z., Gross, M.T., Van Werkhoven, H., Piazza, S.J., Sawicki, G.S., 2016. Adding stiffness to the foot modulates soleus force-velocity behaviour during human walking. Sci. Rep. 6, 1–11.
Wilken, J., Rao, S., Saltzman, C., Yack, H.J., 2011. The effect of arch height on kinematic coupling during walking. Clin. Biomech. 26, 318–323.
Willwacher, S., König, M., Potthast, W., Brüggemann, G., 2013. Does Specific Foot Wear Facilitate Energy Storage and Return at the Metatarsophalangeal Joint in Running ? 29, 583–592.
Zeni, J.A., Richards, J.G., Higginson, J.S., 2008. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait Posture 27, 710–714.