Musculoskeletal Science and Practice 42 (2019) 98–103

Contents lists available at ScienceDirect

Musculoskeletal Science and Practice

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

Original article

The clinical measure of forefoot-shank alignment partially reflects mechanical properties of the midfoot joint complex

Bruno Dayrell da Costa Paes, Renan Alves Resende, Raphael Borges Gomes, Bruna Antônia Gontijo, Fabrício Anicio Magalhães, Juliana de Melo Ocarino, Sérgio Teixeira Fonseca, Thales Rezende Souza*

Graduate Program in Rehabilitation Sciences, Universidade Federal de Minas Gerais, Brazil

A R T I C L E I N F O

Keywords:
Clinical measure
Forefoot-shank alignment
Mechanical resistance
Midfoot

A B S T R A C T

Background: The clinical measure of forefoot-shank alignment (FSA) predicts the amount of foot pronation during weight-bearing tasks. This may be mediated by a relationship between FSA and the mechanical resistance of the midfoot joint complex (*MFJC*) to forefoot inversion, which is a component of weight-bearing foot pronation.

Objective: To investigate if the clinical measure of FSA is associated with MFJC mechanical resistance to inversion.

Design: Cross-sectional observational study.

Method: Forty-six healthy individuals (27 males; 19 females) with mean age of 26.4 years (SD 5.3) participated in this study. FSA was measured with photographs. The resistance torque of the MFJC against inversion was measured with a specially designed device. Mean torque, mean torque normalized by body mass, and joint resting position were calculated as variables related to MFJC mechanical resistance. Correlation analyses were carried out to test the association between each MFJC resistance variable and the FSA (α = 0.05).

Results: /findings: There were significant moderate correlations of FSA with mean torque ($r = -0.44$, $p = 0.002$), mean normalized torque ($r = -0.42$, $p = 0.004$) and resting position (r = 0.39, $p = 0.007$). The clinical measure of FSA is associated to the mechanical resistance of the MFJC: (a) the greater the FSA, the smaller the resistance torques; (b) the greater the FSA, the more inverted the forefoot resting position.

Conclusions: These results showed that the clinical measure of FSA is moderately related to mechanical properties of the MFJC.

1. Introduction

Clinical measures of non-weight-bearing forefoot alignment in the frontal-plane (Holt et al., 1995; Mendonça et al., 2013; Monaghan et al., 2013, 2014; Souza et al., 2014) predict the amount of foot pronation during orthostatic posture (Souza et al., 2014), walking (Monaghan et al., 2013, 2014) and running (Monaghan et al., 2014). Both increased and reduced foot pronation have been associated with different orthopedic conditions (Becker et al., 2017; Gross et al., 2007; Korpelainen et al., 2001; Louw and Deary, 2014; Menz et al., 2013; Neal et al., 2014; O'leary et al., 2013; Powers et al., 2012; Sharma et al., 2010; Williams et al., 2001). These clinical measures quantify the inversion alignment of the forefoot relative to a global vertical reference. However, it is not known if the mechanical properties of the foot tissues are related to the forefoot alignment. The mechanical properties of the midfoot joint complex (MFJC) tissues are related to the magnitude of foot pronation during walking (Gomes et al., 2019) and, thus, may mediate the relationship between the clinical measure of forefoot alignment and weight-bearing foot pronation.

The forefoot-shank alignment (FSA) is one of the measures that quantify forefoot alignment (Bittencourt et al., 2012; Mendonça et al., 2013, 2018; Souza et al., 2014). The angle obtained may be influenced by the MFJC mechanical resistance against forefoot inversion (Mendonça et al., 2013; Souza et al., 2014), which is a motion-component of weight-bearing foot pronation (Neumann, 2006). Therefore, the capacity of this type of measure to predict the amount of foot pronation (Monaghan et al., 2013, 2014; Souza et al., 2014), could be, in part, due to its possible relationship with MFJC mechanical

* Corresponding author. Universidade Federal de Minas Gerais, Av. Antônio Carlos 6627, Escola de Educação Física, Fisioterapia e Terapia Ocupacional, CEP 31270-010, Belo Horizonte, MG, Brazil.
E-mail address: thalesrs@ufmg.br (T.R. Souza).

https://doi.org/10.1016/j.msksp.2019.04.016
Received 6 November 2018; Received in revised form 26 April 2019; Accepted 30 April 2019
2468-7812/ © 2019 Elsevier Ltd. All rights reserved.

properties.

This study investigated the relationship between the mechanical resistance provided by the MFJC soft tissues to forefoot inversion and the clinical measure of FSA. The study hypothesis was that larger FSA angles are related to lower MFJC resistance to forefoot inversion and to a more inverted MFJC resting position.

2. Methods

2.1. Participants

Forty-six healthy individuals (27 males and 19 females) with mean age of 26.4 years (SD 5.3) participated in this cross-sectional study. The inclusion criteria were: (i) no history of injury or any orthopedic surgery in the lower limbs; (ii) no use of foot orthoses, since it is unknown how the foot orthoses could influence the mechanical properties of the MFJC; (iii) no sports practice on the day of data collection, to avoid temporary changes in the MFJC tissues' mechanics; (iv) presenting a minimum MFJC range of motion of 20° of eversion to 50° of inversion; (v) having a minimum of 0° of dorsiflexion amplitude (i.e. a straight angle between the shank and the foot). The exclusion criteria were: (i) reporting discomfort or pain during the procedures, to avoid muscle contractions; (ii) not being able to clearly relax the foot and ankle muscles during at least one trial of the measurement of MFJC mechanical properties; (iii) absence of acceptable smoothness in all three torsimeter data curves; (iv) presenting a MFJC resistance torque greater than 10 Nm (i.e. the maximum torque measured by the device). Sample size was estimated for a correlation with a moderate size (r = 0.4 estimated based on a pilot study), a two-tailed significance level of 0.05, and a statistical power of 0.8[30]. This was computed in the G*Power software (Faul et al., 2007) and resulted in a sample size of 44 participants. The pilot study was conducted with ten participants and, besides estimating the correlations' sizes, it assessed the intratester reliability of the clinical measures and the measures representing the mechanical resistance of the MFJC to forefoot inversion. These participants were evaluated twice, one week apart, and Intraclass Correlation Coefficients (ICC) were calculated (Aquino et al., 2018). All participants signed a consent form approved by the Institution's Ethics in Research Committee (CAAE 78785717.7.0000.5149).

2.2. Procedures

2.2.1. Measurement of forefoot-shank alignment

Digital photographs of the foot were taken from a superior view, with the participant in prone (Fig. 1), using a digital camera (Nikon D-SLR D5000; Nikon Inc, Melville, NY). The procedures followed the description of Mendonça et al. (2013) (Mendonça et al., 2013). The examiner used a caliper rule to mark, on the skin of the shank posterior aspect of the participant, the central point between the tibial plateaus and between the superior boards of the malleoli. A flexible rule (80 cm) was used to connect the central marks between the tibial plateaus and malleoli and to trace a straight line on the skin of the distal portion of the shank.

A straight metallic rod was attached to the forefoot, with micropore tape, and was placed under the first and fifth metatarsal heads to represent the alignment of all the metatarsal heads. The assessed lower limb was positioned with the calcaneus surface facing upwards, to allow the marks to be seen in the center part of the camera display. Afterwards, the examiner used a goniometer to position the foot at 0° of dorsiflexion and asked the subjects to actively sustain this position. Then, the examiner took the picture of the foot-ankle alignment (Fig. 2). These procedures were performed three times. The examiner observed whether the participant extended the toes during the test, which would invalidate the assessment. All subjects were capable of avoiding toes extension. The FSA was measured as the angle between the forefoot line (rod) and a virtual line perpendicular to the shank line

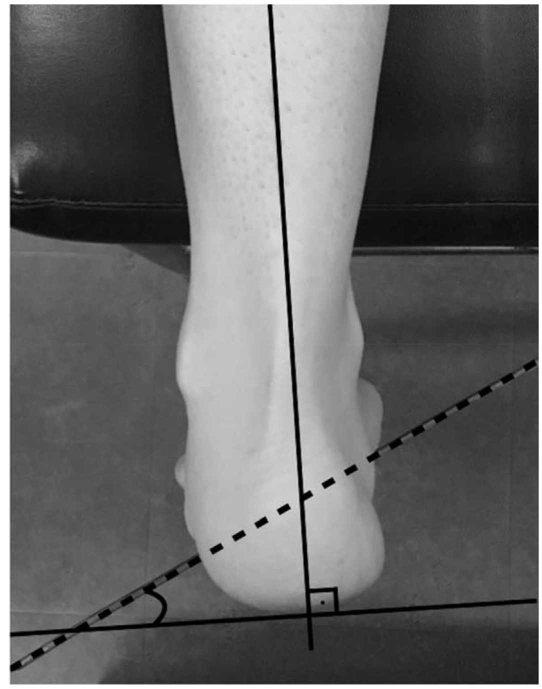

Fig. 1. Shank bisection line and forefoot-shank alignment. The alignment angle consists of the angle between a virtual line (perpendicular to the shank line) and the line of the forefoot.

(Fig. 1). The FSA angles were calculated with MATLAB® (2015b, MathWorks, USA) software.

2.2.2. Measurement of the mechanical resistance of the MFJC to forefoot inversion

The Torsimeter (patent deposited) (Fig. 3a and b) was used to measure the resistance torque and the angle of the MFJC. It has a torque meter (GSP930, Xi'an Gavin Electronic Technology Co, China) and a potentiometer (model 357, Vishal Spectrol, Israel). The participant was sitting on an adjustable chair with the posterior aspect of the lower leg and the calcaneus supported by the instrument's boot. The ankle assumed a 0° position in the sagittal plane (since the shank and foot supports have a perpendicular position relative to each other). The knee was positioned at 45° of flexion in the sagittal plane measured with the goniometer. The hip was near to the neutral position in the transverse and frontal planes. The forefoot was held with an adjustable clamp, which was attached to the torque meter and the potentiometer sensors. The potentiometer axis was aligned with the foot's second ray, which is the least mobile foot ray (Michaud, 1993) and consequently the approximate axis for forefoot movement in the frontal plane. The Torsimeter's forefoot clamp was positioned immediately proximal to the first metatarsal phalangeal joint so that the metatarsal heads were firmly attached to it (Fig. 3b). Velcro strips fixed the shank inside the boot (Fig. 3b). Clamps for the distal shank, calcaneus and metatarsal heads held them tight (Fig. 3a). Only the MFJC was free to move. After the adjustments, the participant was asked to maintain the lower leg and foot completely relaxed during all procedures and to let the examiner know whether he/she felt any discomfort or pain. The examiner moved the participant's forefoot to 20° of eversion (−) and then to 50° of inversion (+), at a slow angular velocity around 2°/s (using a real-time check method – we accepted up to 10% of variance below and above 2°/s), to favor the absence of muscle contractions (Nordez et al., 2006). Three trials were performed to accommodate soft tissues viscosity. Then, three measurement trials were performed. The torque data was displayed in real time and the trial was interrupted and discarded if any peak on the torque/angle curve was visually perceived. Analog signals

Fig. 2. Ankle positioning at 0° of dorsiflexion for the forefoot-shank alignment measurement.

from the potentiometer and the torque meter were converted to digital signals through an A/D converter (NI USB-6210, National Instruments, USA). The torque values and the angular position of the MFJC were recorded with a sampling rate of 100 Hz, using the software LabVIEW® (2012, National Instruments, USA). The torque was collected at the range of −20° to 50°, which was found to be the available range of motion for all participants. Although there was not a measure of muscle activity, caution was taken to exclude trials with any sign of muscle contraction, as explained above and in the next section.

2.3. Data processing

Torque vs angle data were filtered with a Chebyshev, 3rd-order, low-pass filter, with a 4 Hz cutoff frequency. A smoothness analysis (Gomes et al., 2019) was carried out for each torque vs angle time-series, to identify trials with large irregularities, which were more likely to have been influenced by muscle contraction. Large irregularities were defined, at each trial, as large changes in the instantaneous slopes of torque-angle curve (Gomes et al., 2019). The slopes of each torque-angle curve (Nm/°) were expressed as $\Delta\tau/\Delta\theta$, where $\Delta\tau$ is the frame-to-frame change in torque (in Nm) and $\Delta\theta$ is the frame-to-frame change in angle (in degrees). Slope changes were calculated as the moduli values of the differences between the slopes of consecutive frame-to-frame intervals, which were expressed in percentage. To classify slope changes as 'large', a quartile analysis was performed for all change values obtained from all participants and trials. Change values in the third quartile were considered large (i.e. change values greater than 18%). Thus, torque-angle curves with slope changes greater than 18% were excluded from the analyses. From 144 trials, 34 were excluded, resulting in 110 valid trials. Two subjects were excluded from the study because their three trials were excluded. Thus, from an initial sample of 48 subjects, 46 subjects remained in the final sample. Fig. 4 shows the mean torque vs. angle time-series from the final-sample participants.

2.4. Data reduction

For the clinical measure FSA, the mean score of the three measures were calculated and used for statistical analysis. In a pilot study with ten participants, the examiner demonstrated good intratester reliability, with ICC of 0.87 and a standard error of measurement (SEM) of 1.91° (Portney and Watkins, 2000). The variables extracted from the MFJC torque vs. angle time-series during forefoot inversion were: mean torque, mean torque normalized by body mass, and resting position. The mean torque (Nm) was computed as the average of each trial torque data. This variable was computed as the mean value from the test repetitions. Mean torque was then normalized by body mass (Hamill et al., 2013). The resting position was defined as the angle (°) in which the passive torque was zero and represents the angle at which the MFJC begins to resist inversion. An average resting position was also calculated from the trials. All computations were done in MATLAB® (2015b, MathWorks, USA). The intratester reliability (ICCs) for mean torque, mean torque normalized by body mass, and resting position

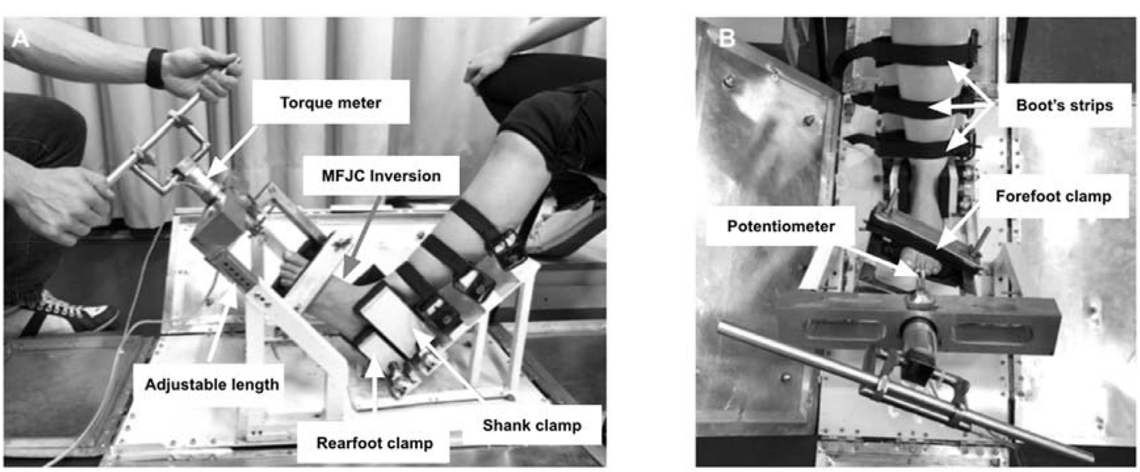

Fig. 3. Torsimeter. A) Lateral view. B) Superior view.

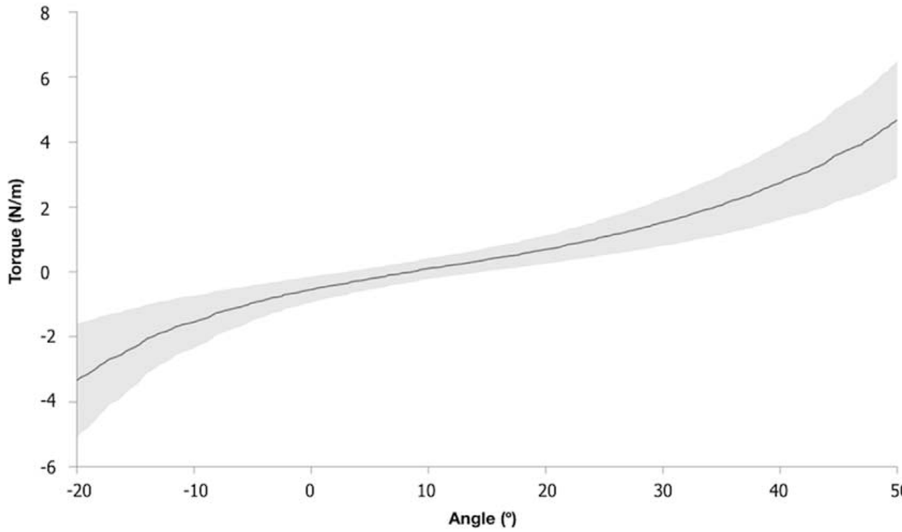

Fig. 4. Mean torque vs. angle curve from all participants. The shaded area represents one standard deviation above and under the mean torque.

were 0.85 (SEM = 0.046 Nm), 0.87 (SEM = 0.0007 Nm/kg) and 0.82 (SEM = 1.88°), respectively, which were considered good (Portney and Watkins, 2000). An average time-series was obtained from all valid torque vs. angle curves, for description.

2.5. Statistical analyses

Normality of data distribution was verified with the Shapiro-Wilk test and Pearson correlation analyses were performed between each MFJC mechanical variable and the FSA. The significance level was set at 0.05. All the analyses were carried out with SPSS 24 (SPSS Inc, Chicago, USA).

3. Results

There were significant moderate (Cohen, 1988, 1992) and negative correlations of FSA with mean torque ($r = -0.44$, $p = 0.002$), mean torque normalized by body mass ($r = -0.42$, $p = 0.004$), and a significant moderate and positive correlation of FSA with resting position (r = 0.39, $p = 0.007$) (Table 1). Table 2 presents the values of mean and standard deviation, range and ICC of the clinical measure, the mean torque, the mean torque normalized, and the resting position. The data distribution (scatterplots) for the correlations between the FSA and the MFJC mechanical variables are presented in Fig. 5.

4. Discussion

This study demonstrated that increased forefoot inversion alignment, identified with the clinical measure of FSA, was moderately associated with reduced MFJC mechanical resistance to forefoot inversion. Specifically, increased forefoot inversion alignment was associated with reduced mean torque and reduced mean torque normalized by body mass. Further, a more inverted FSA was associated with a more inverted resting position (i.e. with a resistance to inversion that begins later in the MFJC range of motion).

The clinical measure of FSA was associated to the resistance provided by the MFJC soft tissues. The contraction of the tibialis anterior (i.e. the main ankle dorsiflexor) required to maintain ankle neutral position (0° of dorsiflexion) during the FSA measurement tends to invert the forefoot (Muanjai et al., 2017). Therefore, lower MFJC resistance to forefoot inversion will lead to more forefoot inversion in the FSA measure. However, it is expected that other factors, such as bone alignment components including forefoot varus, rearfoot varus and tibial varus, influence FSA (Mendonça et al., 2013; Souza et al., 2014). This fact could help explain that a moderate correlation between MFJC mechanical resistance and FSA was found.

The present findings support the hypothesis that MFJC resistance to forefoot inversion may partly mediate the relationship between high values of weight-bearing foot pronation and high values of non-weight-bearing inversion position of the forefoot, demonstrated by previous studies (Monaghan et al., 2013, 2014; Souza et al., 2014). During weight-bearing tasks, with the whole forefoot on the ground, foot pronation, characterized by rearfoot eversion, is accompanied by motions at the MFJC (Lundgren et al., 2008; Neumann, 2006). These motions permit the metatarsal heads, as a unit, to invert relative to the rearfoot and stay horizontally supported. Thus, the MFJC resistance to forefoot inversion may also resist rearfoot eversion during weight-bearing tasks (Gomes et al., 2019; Souza et al., 2014). Additionally, the relationship between FSA and MFJC resting position suggests that FSA reflects the tendency for the open-chain position of the forefoot before it contacts the ground in support phases of weight-bearing tasks, which is associated with the magnitude of foot pronation during the stance phase of walking and running (Monaghan et al., 2013, 2014).

Traditional clinical measures of forefoot alignment (commonly considered as measures of forefoot varus) quantify the open-chain frontal-plane forefoot position relative to the calcaneus in an attempt to measure bone shapes assumed to be the origin of this alignment (Donatelli et al., 1999; Michaud, 1993; Root et al., 1977). However, previous studies have demonstrated contradictory findings regarding the ability of these measures to explain foot pronation during weight-

Table 1
Correlation between forefoot-shank alignment and the mechanical resistance of the MFJC.

		Mean torque (Nm)	Mean torque normalized (Nm/Kg)	Resting position (°)
Forefoot-shank alignment	Correlation (r)	-.44[a]	-.42[a]	.39[a]
	Significance (p)	.002	.004	.007

[a] Correlation is significant with $\alpha \leq 0.05$.

Table 2
Description and intratester reliability of the forefoot-shank alignment and the variables related to the mechanical resistance of the MFJC.

	Mean (SD)	Range	ICC[a]	SEM
Forefoot-shank alignment	17.41° (5.29)	6.70° to 31.66°	0.87	1.91°
Mean torque	0.47 Nm (0.56)	−0.54 to 1.80 Nm	0.85	0.046 Nm
Mean torque normalized by body mass	0.0074 Nm/kg (0.0085)	−0.0089 to 0.0272 Nm/kg	0.87	0.0007Nm/kg
Resting position	10.31° (4.91)	−1.85° to 18.74°	0.82	1.88°

SD: Standard deviation, ICC: intraclass correlation coefficient.
SEM: Standard error of measurement.

bearing tasks (Cornwall et al., 2004; Hamill et al., 1989; Mcpoil and Cornwall, 1996). For example, Mcpoil and Cornwall (1996) (Mcpoil and Cornwall, 1996) have found that forefoot alignment is a poor predictor of rearfoot movement during gait, while Donatelli et al. (1999) (Donatelli et al., 1999) and Silva et al. (2014) (Silva et al., 2014) reported opposite findings. On the contrary, recent studies have consistently shown that measures of forefoot alignment relative to a global vertical reference are related to foot pronation (Monaghan et al., 2013, 2014; Souza et al., 2014) and related movements, such as dynamic knee valgus (Bittencourt et al., 2012). Differently from the FSA measure, Monaghan et al. (2013, 2014) (Monaghan et al., 2013, 2014) used the treatment table as the reference for the forefoot alignment, while the subject's shank was vertically oriented relative to the table. Thus, both measures are very similar. These measures probably quantify the combination of multiple factors that may influence foot pronation during weight-bearing tasks, including bone alignment and soft tissues mechanical resistance. Thus, it is possible that increased forefoot inversion alignment reflects a combination of factors that favors increased foot pronation.

In the present study, FSA was measured by photography (Mendonça et al., 2013; Monaghan et al., 2013, 2014). In clinical settings, the use of a goniometry is more common (Souza et al., 2014). Even though there are no studies comparing the values of this measure obtained by goniometry and photography (Gross et al., 2007; Mendonça et al., 2013; Monaghan et al., 2013, 2014), they are expected to be similar considering that both use similar parameters: anatomic references, the superior view, and the subject's positioning (Mendonça et al., 2013; Souza et al., 2014). The FSA angles, in the photography method, were obtained with Matlab software. This could be easily done by using currently available, commercial smartphones apps (Stove et al., 2018).

Limitations of the study should be acknowledged. It was not guaranteed that foot-ankle intrinsic and extrinsic muscles were not active during the measurements of MFJC mechanical resistance. However, the careful procedures for the measurement of MFJC torque (e.g. the low speed) and the MFJC torque time-series showing a typical passive torque pattern (Chino and Takahashi, 2015; Muanjai et al., 2017) suggest that the foot muscles were not active or were minimally active during data collection. The low-speed movement during the measurement of the MFJC mechanical resistance was performed to avoid muscle contraction. However, this does not reflect the velocity of the MFJC during different weight bearing tasks, such as gait. In addition, it should be noted that the present results are limited to young, able-bodied and asymptomatic individuals, which applies to preventive contexts. The same investigation should be also conducted with other clinical populations.

The only moderate correlations observed indicated a limitation of the FSA measure to inform about MFJC mechanical resistance (i.e. only about 16% of explanation). Still, the significant correlations show that the FSA and the MFJC mechanical variables have a significant covariation. Considering the lack of clinical measures to assess MFJC mechanical resistance, the findings point to the use of the FSA to gain insight about this resistance, when a clinician is investigating individual

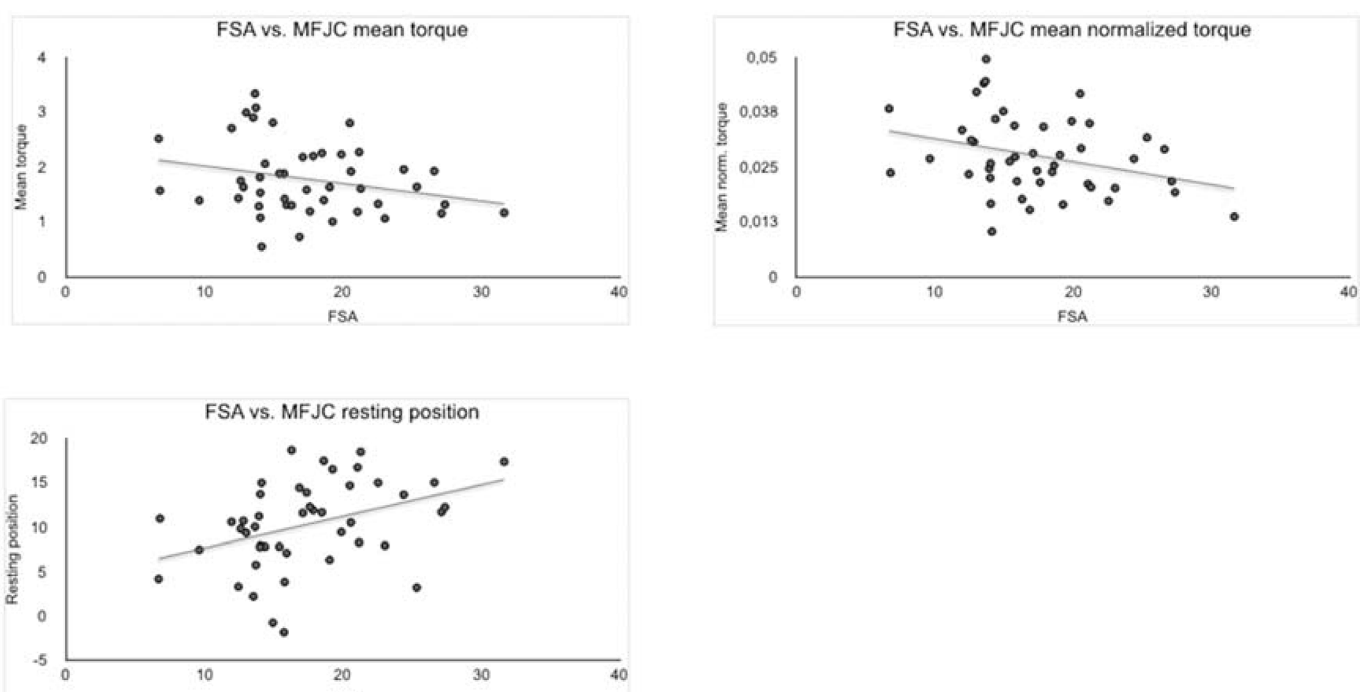

Fig. 5. Scatterplots of the relationship between the forefoot-shank alignment and the variables describing the MFJC mechanical resistance. The trend-line (least-squares fit) represents the tendency of the relationship.

factors that may be related to an observed magnitude of weight-bearing foot pronation (Gomes et al., 2019). However, improvements of this measure and the development of other measures are recommended.

5. Conclusion

The clinical measure of FSA is moderately associated to the MFJC mechanical resistance to inversion: the more inverted the alignment in the clinical measure, the smaller the torque and the more inverted the resting position. Therefore, the FSA partially reflects the mechanical properties of the MFJC.

Conflicts of interest

None.

Ethical approval

This study was approved by the Institution's Ethics in Research Committee (CAAE 78785717.7.0000.5149).

Acknowledgements

This study was financed in part by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior - Brasil (CAPES)(Finance Code 001), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), and Fundacão de Amparo a Pesquisa do Estado de Minas Gerais.

References

Aquino, M.R.C., Avelar, B.S., Silva, P.L., Ocarino, J.M., Resende, R.A., 2018. Reliability of Foot Posture Index individual and total scores for adults and older adults. Musculoskeletal Science and Practice 36 (August), 92–95.
Becker, J., James, S., Wayner, R., Osternig, L., Chou, L.S., 2017. Biomechanical factors associated with achilles tendinopathy and medial tibial stress syndrome in runners. Am. J. Sports Med. 45 (11), 2614–2621.
Bittencourt, N.F.N., Ocarino, J.M., Mendonça, L.D., Hewett, T.E., Fonseca, S.T., 2012. Foot and hip contributions to high frontal plane knee projection angle in athletes: a classification and regression tree approach. J. Orthop. Sport. Phys. Ther. 42 (12), 996–1004.
Chino, K., Takahashi, H., 2015. The association of muscle and tendon elasticity with passive joint stiffness: in vivo measurements using ultrasound shear wave elastography. Clin. Biomech. 30 (10), 230–235.
Cohen, J., 1988. Statistical Power Analysis for the Behavioral Sciences. Routledge.
Cohen, J., 1992. A power primer. Psychol. Bull. 112 (1), 155–159.
Cornwall, M.W., Mcpoil, T.G., Fishco, W.D., Hunt, L., Lane, C., O'donnel, D., 2004. The relationship between forefoot alignment and rearfoot motion during walking. Australas. J. Podiatr. Med. 38 (2), 35–40.
Donatelli, R., Wooden, M., Ekedahl, S.R., Wilkes, J.S., Cooper, J., Bush, A.J., 1999. Relationship between static and dynamic foot postures in professional baseball players. J. Orthop. Sport. Phys. Ther. 29 (6), 316–325.
Faul, F., Erdfelder, E., Lang, A.G., Buchnner, A.G., 2007. *Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behav. Res. Methods 39 (2), 75–91.
Gomes, R.B.O., Souza, T.R., Paes, B.D.C., Magalhaes, 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.
Gross, K.D., Niu, J., Zhang, Y.Q., Felson, D.T., Mclennan, C., Hannan, M.T., Holt, K.G., Hunter, D.J., 2007. Varus foot alignment and hip conditions in older adults. Arthritis Rheum. 56 (9), 2993–2998.
Hamill, J., Bates, B.T., Knutzen, K.M., Kirkpatrick, G.M., 1989. Relationship between selected static and dynamic lower extremity measures. Clin. Biomech. 4 (4), 217–225.
Hamill, J., Selbie, W.S., Kepple, T.M., 2013. Three dimensional kinematics. In: Robertson, D.G.E. (Ed.), Research Methods Biomechanics, 2ª ed. Human Kinetics Publishers.
Holt, K.G., Hamill, J., 1995. Running injuries and treatment: a dynamic approach. In: Sammarco, G.J. (Ed.), Rehabilitation of the Foot and Ankle. Mosby, St. Louis.
Korpelainen, R., Orava, S., Karpakka, J., Siira, P., Hulkko, A., 2001. Risk factors for recurrent stress fractures in athletes. Am. J. Sports Med. 29 (3), 304–310.
Louw, M., Deary, C., 2014. The biomechanical variables involved in the aetiology of iliotibial band syndrome in distance runners - a systematic review of the literature. Phys. Ther. Sport 15 (1), 64–75.
Lundgren, P., Nester, C., Liu, A., Arndt, A., Jones, R., Stacoff, A., Wolf, P., Lundberg, A., 2008. Invasive in vivo measurement of rear-, mid- and forefoot motion during walking. Gait Posture 28 (1), 93–100.
Mcpoil, T.G., Cornwall, M.W., 1996. The relationship between static lower extremity measurements and rearfoot motion during walking. J. Orthop. Sport. Phys. Ther. 24 (5), 309–314.
Mendonça, L.D.M., Bittencourt, N.F.N., Amaral, G.M., Diniz, L.S., Souza, T.R., Fonseca, S.T., 2013. A quick and reliable procedure for assessing foot alignment in athletes. J. Am. Podiatr. Med. Assoc. 103 (5), 405–410.
Mendonça, L.D.M., Ocarino, J.M., Bittencourt, N.F.N., Macedo, L.G., Fonseca, S.T., 2018. Association of hip and foot factors with patellar tendinopathy (Jumper's knee) in athletes. J. Orthop. Sport. Phys. Ther. 48 (9), 676–684.
Menz, H.B., Dufour, A.B., Riskowski, J.L., Hillstrom, H.J., Hannan, M.T., 2013. Foot posture, foot function and low back pain: the Framingham Foot Study. Rheumatology 52.
Michaud, T., 1993. Foot Orthoses: and Other Forms of Conservative Foot Care. Williams and Wilkins, Baltimore.
Monaghan, G.M., Lewis, C.L., Hsu, W.H., Saltzman, E., Hamill, J., Holt, K.G., 2013. Forefoot angle determines duration and amplitude of pronation during walking. Gait Posture 38 (1), 8–13.
Monaghan, G.M., Lewis, C.L., Hsu, W.H., Saltzman, E., Hamill, J., Holt, K.G., 2014. Forefoot angle at initial contact determines the of forefoot and rearfoot eversion during running. Clin. Biomech. 29 (8), 936–942.
Muanjai, P., Jones, D.A., Mickevicius, M., Satkunskiene, D., Snieckus, A., Rutkauskaite, R., Misckeviciene, D., Kamandulis, S., 2017. The effects of 4 weeks stretching training to the point of pain on flexibility and muscle tendon unit properties. Eur. J. Appl. Physiol. 117 (8), 713–1725.
Neumann, D.A., 2006. Ankle and foot. In: Neumann D.A. Kinesiology of the Musculoskeletal System. Mosby, pp. 480–524.
Neal, B.S., Griffiths, I.B., Dowling, G.J., Murley, G.S., Munteanu, S.E., Simth, M.M.F., 2014. Foot posture as a risk factor for lower limb overuse injury: a systematic review and meta-analysis. J. Foot Ankle Res. 7 (55).
Nordez, A., Cornu, C., Mcnair, P., 2006. Acute effects of static stretching on passive stiffness of the hamstring muscles calculated using different mathematical models. Clin. Biomech. 21, 755–760.
O'leary, C.B., Cahill, C.R., Robinson, A.W., Barnes, M.J., Hong, J., 2013. A systematic review: the effects ofpodiatrical deviations on nonspecific chronic low back pain. J. Back Musculoskelet. Rehabil. v.26, 117–123.
Portney, L.G., Watkins, M.P., 2000. Foundations of Clinical Research: Applications to Practice. Prentice-Hall, Upper Saddle River.
Powers, C.M., Bolgla, L., Callaghan, M., Collins, N., Sheehan, F., 2012. Patellofemoral pain: proximal, distal and local factors. J. Orthop. Sport. Phys. Ther. 42, A1–A20.
Root, M.L., Orien, W.P., Weed, J.H., 1977. Normal and Abnormal Function of the Foot. Clinical Biomechanics. Corp, Los Angeles.
Sharma, J., Golby, J., Greeves, J., Spears, I.R., 2010. Biomechanical and lifestyle risk factors for medial tibia stress syndrome in army recruits: a prospective study. Gait Posture 33 (3), 361–365.
Silva, R.S., Ferreira, A.L., Veronese, L.M., Serrao, F.V., 2014. Forefoot varus preditcs subtalar hyperpronation in young people. J. Am. Podiatr. Med. Assoc. 104 (6) November/December.
Souza, T.R., Mancini, M.C., Araújo, V.L., Carvalhais, V.O.C., Ocarino, J.M., Silva, P.L., Fonseca, S.T., 2014. Clinical measures of hip and foot-ankle mechanics as predictors of rearfoot motion and posture. Man. Ther. 19, 379–385.
Stove, M.P., Palsson, T.S., Hirata, R.P., 2018. Smartphone-based accelerometry is a valid tool for measuring dynamic changes in knee extension range of motion. The Knee 25 (1), 66–72 January.
Williams, D.S., Mcclay, I.S., Hamill, J., 2001. Arch structure and injury patterns in runners. Clin. Biomech. 16 (4), 341–347.