COMPUTER METHODS IN BIOMECHANICS AND BIOMEDICAL ENGINEERING
https://doi.org/10.1080/10255842.2018.1480760

Taylor & Francis
Taylor & Francis Group

ARTICLE

Foot internal stress distribution during impact in barefoot running as function of the strike pattern

Enrique Morales-Orcajo[a,b], Ricardo Becerro de Bengoa Vallejo[c], Marta Losa Iglesias[d], Javier Bayod[a] and Estevam Barbosa de Las Casas[b]

[a]Applied Mechanics and Bioengineering group (AMB) Aragón Institute of Engineering Research (I3A). University of Zaragoza, Zaragoza, Spain. Biomedical Research Networking Center in Bioengineering, Biomaterials and Nanomedicine (CIBER-BBN); [b]Group of Biomechanical Engineering UFMG - (MecBio) School of Engineering, Universidade Federal de Minas Gerais, Belo Horizonte, MG, Brazil; [c]Nursing Physiotherapy and Podiatry Faculty. Medicine Faculty, Complutense University, Madrid, Spain; [d]Faculty of Health Sciences, Rey Juan Carlos University, Madrid, Spain

ABSTRACT
The aim of the present study is to examine the impact absorption mechanism of the foot for different strike patterns (rearfoot, midfoot and forefoot) using a continuum mechanics approach. A three-dimensional finite element model of the foot was employed to estimate the stress distribution in the foot at the moment of impact during barefoot running. The effects of stress attenuating factors such as the landing angle and the surface stiffness were also analyzed. We characterized rear and forefoot plantar sole behavior in an experimental test, which allowed for refined modeling of plantar pressures for the different strike patterns. Modeling results on the internal stress distributions allow predictions of the susceptibility to injury for particular anatomical structures in the foot.

ARTICLE HISTORY
Received 12 November 2017
Accepted 21 May 2018

KEYWORDS
Foot biomechanics; Barefoot running; Running-related-injuries; Plantar sole mechanical properties; Strike pattern

1. Introduction

In recent years, it is becoming popular running barefoot or with minimalist footwear (Rixe et al. 2012), motivated by the belief in a lower risk of injury (Rothschild 2012) and by the absence of scientific evidence that heel-cushioning shoes reduce running-related injuries (Richards et al. 2009). Recent in-depth reviews regarding running-related injuries in shod and barefoot running concluded that: (1) despite advances in cushioned running shoes, runners continue to experience high injury rates; (2) human beings are designed to run, although we have run with heel-cushioning footwear only for the last four decades; (3) there is no clinical evidence that shod or barefoot runners suffer fewer injuries; (4) there are biomechanical differences between both styles of running; (5) due to this, there are certain benefits between running styles for particular injuries; (6) many unknowns about the relationship between running style and injury remain to be answered; (7) more large-scale studies are required; (8) there is no unique solution/advice for all runners (Jenkins and Cauthon 2011; Lohman et al. 2011; Altman and Davis 2012; Daoud et al. 2012; Rixe et al. 2012).

In any case, it is undeniable that more and more runners are replacing their previous heel-cushioning shoes with the new minimalist shoes (Squadrone et al. 2015). This change implies alterations in several biomechanical aspects. Starting with the fact that running barefoot tends to strike midfoot or forefoot, this modifies the stride length, which influences loading rate, plantar peak pressure, step frequency, muscular activity, leg compliance, ankle, knee and hip kinematics (De Wit et al. 2000; Divert et al. 2008; Squadrone and Gallozzi 2009; Lieberman et al. 2010; Yong et al. 2014). Despite this tendency of flatter foot placement at the landing when transitioning from shod to barefoot running, there are still barefoot runners with heel-to-toe contact pattern (Cheung and Rainbow 2014; Samaan et al. 2014). Kinematic studies showed that the strike pattern has a greater influence in the lower leg mechanics than the shod condition (Shih et al. 2013). However, approaches different from kinetic/kinematic or prospective studies have not been performed to study the mechanical effects of the strike pattern.

The kinetic/kinematic analysis describes motion and its causes considering the components as rigid solids, while the continuum mechanics approach deals with the internal stress produced in the components

CONTACT Enrique Morales-Orcajo emorales.zgz@gmail.es Ed. Betancourt, C/María de Luna s/n, Zaragoza, Spain.

due to the forces acting on them, i.e., deformable solid. Internal stresses are relevant from a biomechanical perspective since they are related to discomfort, pain and tissue damage (Jordan and Bartlett 1995; Witana et al. 2009; Fernandez et al. 2012). Therefore, a computational simulation based on the finite element method can provide information about the mechanical performance of the internal components of the foot during impact at different strike patterns. Despite the large use of finite element models in foot biomechanics, no previous simulation has analyzed foot running stresses (Morales-Orcajo et al. 2016).

The plantar sole is the component with the greatest capacity of impact absorption. Then, before analyzing the mechanics of the impact at different strike patterns, it is necessary to investigate the mechanical responses at different locations on the plantar sole. Previous mechanical tests of the plantar soft tissue have centered on the heel pad (Bennett and Ker 1990; Rome et al. 2001; Gefen et al. 2001; Miller-Young et al. 2002; Tong et al. 2003; Erdemir et al. 2006), while few studies investigated the response of the submetatarsal soft tissue (Chen et al. 2011). The comparison of results between regions in these cases is difficult due to the different test protocols employed in each study and because most of the studies were focused on structural properties (geometry dependent properties) instead of material properties (properties independent of the geometry). Only one study investigated the material properties of the plantar soft tissue in six different locations, but they removed the skin of the specimens and tested loads up to 20% of the body weight (BW) (Ledoux and Blevins 2007). Hence, before the running strike analysis, an experimental characterization of the mechanical properties of the different plantar sole regions was conducted for the present study.

The aim of this study was to conduct a biomechanical analysis of foot impact during barefoot running for different strike patterns to evaluate potential risks of injury for new adopters of barefoot and minimalist running. A finite element simulation was performed in a foot model at rearfoot strike (RFS), midfoot strike (MFS) and forefoot strike (FFS) positions to compare the different stress distributions. Furthermore, the effect of the landing angle and the surface stiffness in the reduction of the stresses for RFS was evaluated.

2. Material and Methods

2.1. Experimental plantar sole testing

In order to analyze the mechanical properties of the human plantar sole, the soft tissue sole from five frozen-stored feet male elder donors was tested after thawing. Using a scalpel, the skin and fat pad was removed together, beginning from the heel to the forefoot along the plantar aspect of the fascia and the fat pad of the toes was released of the flexor tendons until the distal aspect of the toes. The specimens were tested in compression in a universal material testing machine (Instron Ltd., U.K., model 5548) shortly after dissection. Each plantar sole (n = 5) was tested in 15 different locations, as shown in Figure 1, so that fat tissues kept their natural confinement. A 40mm-diameter rounded stainless steel platen was used to test the heel pad and a 10 mm-diameter cylindrical indenter was employed for the other points (Figure 2). The experiments had the approval of the Bioethical Research Committee of the Hospital Clinico San Carlos at Complutense University (n° 12/210-E).

The testing machine's scale was set to zero when the upper plate was touching the base plate. The undeformed thickness of the specimens was measured with the vertical position of the upper platen when the initial stress reach ∼1.25KPa (1.6N for Ø40 and 0.1N for Ø10) (Ledoux and Blevins 2007). Compression tests were performed at 0.1mm/s of vertical displacement (quasi-static) until load reached 1KN at room temperature (∼25 °C). Soft tissues soles were kept hydrated during testing. For each test, the testing machine recorded a load-deformation curve. These curves were normalized to stress-strain curves and sorted into four groups: rearfoot, midfoot, forefoot and toes (Figure 1). The average curve of each of the four groups was used to compare mechanical behavior among different plantar sole regions.

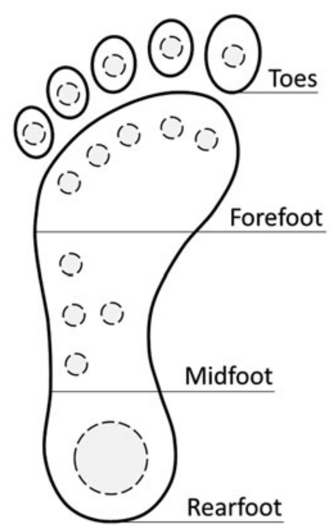

Figure 1. Size and location of the indentations performed in the plantar sole sorted by regions.

2.2. Experimental results

The four plantar sole regions exhibited a strong non-linear stress-strain response up to a limit of 100KPa, after which the stiffening ratio softened. Three different mechanical responses were observed, as shown in Figure 3. Heel pad showed the stiffest response, while the mid-arch region showed the softest. The indentations under metatarsal heads and toes exhibited very similar responses. Plantar sole thickness decreased from rearfoot to forefoot. The average thickness measured for heel pad samples was 16.8 ± 1.2 mm. In the midfoot, the average thickness was 10.5 ± 2.8 mm, and in the forefoot, the average thickness was 6.7 ± 2.1 mm under the metatarsal heads and 6.7 ± 0.8 mm for the toes.

2.3. Plantar sole characterization

In view of the experimental results and considering that the weight is borne mainly in the areas under the calcaneus and the metatarsal heads, plantar sole responses were parametrized for rearfoot and forefoot separately. The former was based on heel pad tests and the latter was based on the test for submetatarsal heads and the toes. Three different hyperelastic constitutive models had been used previously for plantar soft tissue: a Polynomial model (Lemmon et al. 1997), the Yeoh model (Spears et al. 2005) and the Ogden model (Erdemir et al. 2006; Chokhandre et al. 2012; Chen et al. 2014; Telfer et al. 2015). Rearfoot and forefoot stress-strain curves obtained from the experiments were introduced in the software ABAQUS (ABAQUS Inc., Pawtucket, RI, USA) to calculate the parameters of the three models that best fit the experimental curves. A Poisson's ratio of 0.475 was defined to set the compressibility parameter (Chokhandre et al. 2012; Chen et al. 2014). A computational simulation of the compression tests was subsequently performed to evaluate the structural response of the three constitutive models. Ogden's model showed the closest response to the experimental values. Therefore, the first order Ogden constitutive model was chosen to represent the non-linear response of the plantar sole tissue obtained from the experiments. This model describes a hyperelastic behavior of rubber-like materials accounting for compressibility. Its strain energy density function U in terms of generalized strain is:

Figure 2. Left: Heel pad indentation. Right: Forefoot indentations on metatarsal heads marks.

$$U = \frac{\mu}{\alpha^2}\left(\lambda_1^{\alpha} + \lambda_2^{\alpha} + \lambda_3^{\alpha} - 3\right) + \frac{1}{D}(J-1)^2$$

where μ is the initial shear modulus, α is the strain hardening exponent, and D is the compressibility parameter. The adjusted parameters are given in Table 1.

2.4. Finite element model

A three-dimensional finite element model of the musculoskeletal human foot was employed in this study (Morales-Orcajo et al. 2017). The model included the bones, cartilages, muscles, tendons and the plantar fascia that forms the foot-ankle-complex, all surrounded by a bulk soft tissue that represented fat and skin. The finite element mesh contained 805,792 tetrahedral elements with element size ranging from

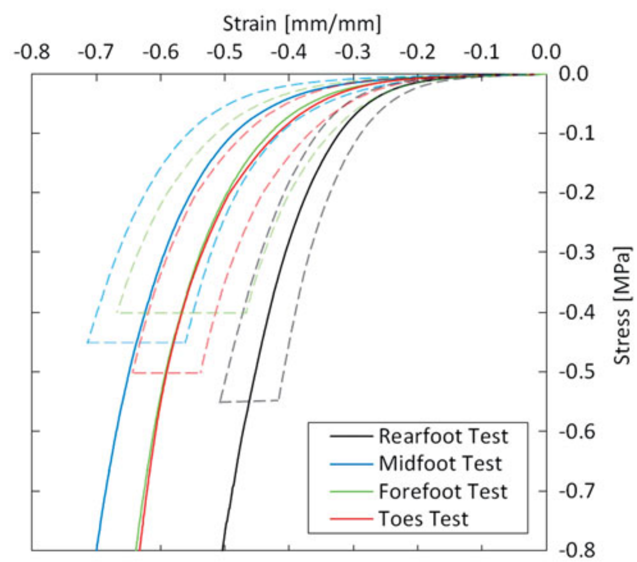

Figure 3. Average stress-strain curves of the four plantar sole regions. Mean (solid lines) and standard deviation (dashed lines). Standard deviations were cut to improve visualization.

Table 1. Material parameters of the tissues simulated with hyperelastic properties. The parameters correspond to the first order Ogden constitutive model.

Tissue	μ (MPa)	α	D (mm^2 N^{-1})
Tendon	33.1622	24.899	0.0001207
Muscle	0.05737	28.820	1.1283
Forefoot sole	0.01199	9.919	8.4822
Rearfoot sole	0.01995	15.011	5.0963

1 to 5 mm as a function of geometric requirements. The model had been previously validated for plantar pressure and structural displacement (Morales-Orcajo et al. 2017).

Two isotropic linear material models were used to simulate bone tissue properties. Cortical bone properties (17GPa for Young's modulus and 0.3 for the Poisson ratio) were applied to the external layer of elements while for the internal elements, properties corresponding to trabecular bone were considered (0.7GPa for Young's modulus and 0.3 for the Poisson ratio (Morales-Orcajo et al. 2015)). The elements between bones were modeled with linear properties of articular cartilage (0.01GPa of Young's modulus and 0.4 of Poisson ratio (Morales-Orcajo et al. 2015)). The tissue properties of the plantar fascia were defined based on the experiment performed by Wright and Rennels (1964) with values of 0.35 GPa and 0.3 for Young's modulus and Poisson ratio respectively. The non-linear compressive response of the intrinsic muscles was simulated with the optimized parameters calculated by Petre et al. (2013) (Table 1). Tendons were modeled with a hyperelastic material model fitted in the previous study (Morales-Orcajo et al. 2017) (Table 1). Finally, the tissue properties of the plantar sole for rear and forefoot regions were characterized as described in the previous section.

2.5. Quasi-static strike analyses

A solid block composed of hexahedral elements was created under the foot to simulate different running strikes patterns over different types of ground. Three barefoot running strikes were configured based on the running strike pattern observations of Lieberman et al. (2010). Barefoot RFS running was simulated with an impact force of 1.89BW and an impact angle of $\theta = 16.4°$, which corresponds to the impact angle of habitually shod adults running barefoot (Lieberman et al. 2010). Barefoot FFS running was set to an impact force of 0.58BW and $\theta = -1.13°$ as impact angle. This configuration corresponds to barefoot runners that never were been shod (Lieberman et al. 2010). Barefoot MFS running was considered with the average impact force of RFS and FFS (1.24BW) and a landing parallel to the ground ($\theta = 0°$). Additionally, a moment in the ankle was defined in order to include running dynamic forces. This moment was estimated according to ankle moment measurements during running (Novacheck 1998) and introduced in the model by applying an axial force at the end of the Achilles tendon. The proximal extreme of the tibialis anterior was constrained to have zero displacement, to simulate the muscle tuning that occurs prior to impact (Nigg 2001). A description of the boundary conditions is shown in Figure 4. A quasi-static analysis was performed for all cases with 700 N of BW.

Two additional scenarios were analyzed for runners that do not transition to MFS/FFS when running barefoot: RFS at lower angles of impact and RFS over softer surfaces. A moderate impact angle was set at $\theta = 8.2°$, keeping the rest of the parameters equal to the previously defined RFS settings. Running on different surfaces was simulated by adjusting the elastic properties of the ground block. An elastic modulus of 30GPa was employed in all cases to simulate the hard surfaces of modern infrastructures such as roads or sidewalks (Demir 2005). Softer elastic moduli were defined to emulate natural environments: 1GPa for compact soil, 0.2GPa for dense sand and 0.05GPa for loose sand (Bowles 1996; Zhang et al. 2014).

3. Computational Results

The impact force in RFS was transmitted vertically through hindfoot bones, from heel to tibia. Figure 5 shows a sagittal section of the hindfoot where this vertical distribution of the compressive stresses is observed. In the heel pad, the stresses under the calcaneal tuberosity were higher than at the ground surface (i.e. internal stress were higher than at the external layers). Despite the cushioning properties of the heel pad, high stresses were predicted for the cortical layer of the calcaneus. This effect was due to the fact that the cortical layer has a much greater stiffness than the plantar sole. Although the cortical bone supported most of the load, the trabecular bone was also stressed. At the dorsal part of the calcaneus, high compressive

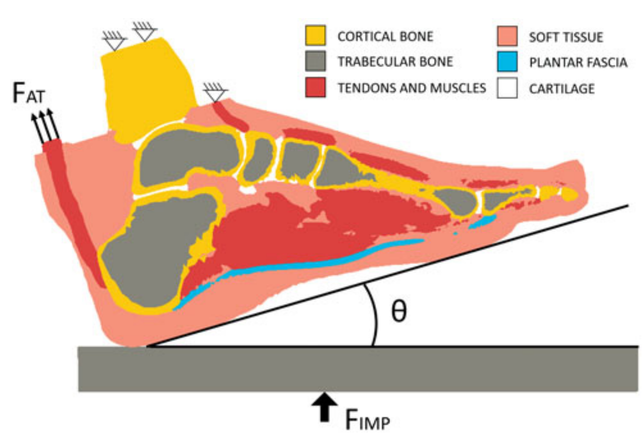

Figure 4. Boundary conditions of the model in the cross-sectional view of the second radius. Nodes at the proximal end of the Tibia and tibialis anterior were clamped. F_{AT}: force applied at Achilles tendon to simulate ankle joint moment; F_{IMP}: impact force applied in each case; θ: strike angle.

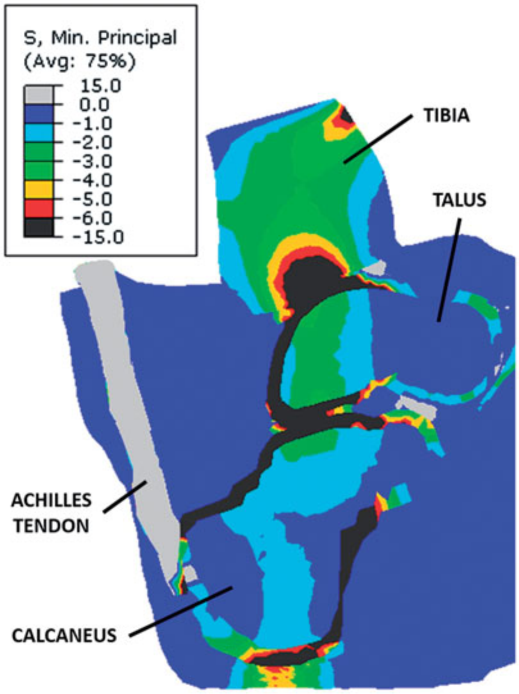

Figure 5. Compressive stress distribution of the hindfoot during RFS in barefoot running.

stresses appeared due to the bending forces. Because of the rigidity of the system in RFS, high pressures take place at the talocrural and subtalar joints. The heel pad was deformed up to 64.7% in the thinnest part just under calcaneal tuberosity. A maximum plantar pressure of 1.4MPa was predicted at the base of the heel. The peak plantar pressure was calculated by averaging the contact pressure of the elements located under the calcaneus tuberosity as the measured local pressures in barefoot running (De Wit et al. 2000). Regarding the tendon work, the tibialis anterior presented high traction stresses in opposition to the ankle joint movement.

The impact force in FFS was distributed among the five rays. High stresses appear in the metatarsals due to the bending forces derived from the running advance. This condition also produced a stretch of the plantar fascia. Plantar sole under metatarsal heads presented peak plantar pressures of 0.48 MPa and strains up to 52.9%, with a greater contact surface than in RFS. MFS analysis presented very similar stress distribution to FFS. In that case, the contact surface was twice that of FFS, which yielded maximal plantar pressure values similar to the FFS. The deformation of the plantar sole was 55.6% under the calcaneus and 54.7% under the metatarsal heads. The remaining components of the foot were under stress levels analogous to FFS.

The additional scenarios considered for RFS condition showed an 18% reduction of the maximal plantar pressure when halving the impact angle. However, the effect of elastic properties of the ground in the reduction or the peak plantar pressure was not significant. Running over compact soil did not reveal relevant differences in peak plantar pressure and stress distribution compared to landing on hard surfaces. Reduction of contact pressure similar to halving the landing angle was predicted for soil elastic modulus lower to 0.2 GPa, i.e., sand.

4. Discussion

Running barefoot is catching on in recent years, leading many runners from change heel-cushioning shoes to minimalist shoes. This transition requires adaptation of the running style in order to avoid possible injuries. The main difference between running shod and barefoot or minimalist is the midsole cushioning layer of modern running shoes. This layer provides an extra aid for absorbing impact forces, which have modified the running technique to landing with high angles. Then, when this cushioning layer is removed, runners should return to the original running technique. If this transition is not well addressed, it could result in impact related injuries (Samaan et al. 2014; Altman and Davis 2015). In order to analyze impact absorption mechanisms of the foot, a number of simulations were performed for different strike patterns while barefoot.

Since each strike pattern makes initial ground contact with a different part of the plantar sole, detailed information of the mechanical properties of the plantar sole in different regions was important. Fifteen locations of the plantar sole were indented to obtain the stress-strain curve. The subcalcaneal tissue showed a stiffer response than the submetatarsal tissue, while the forefoot was stiffer than the soft tissue in the mid-arch. These differences seem to be due to the distinct structure and composition of the plantar sole in each location. Chen et al. (2002) investigated differences in strength at the plantar skin interface and found similar results. They reported that the strength in the heel region was significantly higher than in all other regions and forefoot higher than in mid-arch region. In the same line, Ledoux and Blevins (2007) determined in a stress relaxation study of the plantar soft tissue that the subcalcaneal tissue is different from other plantar soft tissue areas. Although tests were not performed in living subjects, it has been proven that plantar sole specimens stored frozen yield similar mechanical properties (Bennett and Ker 1990).

The plantar sole strains and pressures predicted in the simulation were in the range of the values reported in actual measurements of barefoot runners. De Clercq et al. (1994) measured a deformation of 60.5 ± 5.5% of the heel pad *in vivo* on a barefoot running step comparable to the 64.7% predicted in the present study. De Wit et al. (2000) measured an average maximal pressure of 0.97 ± 0.35MPa for seven volunteers running barefoot at 4.5 m/s with an average angle landing of 6.6 ± 5.6°. The peak pressure predicted in the present study for a RFS angle of 8.2° was similar (1.1MPa), and in a standard deviation (1.4 MPa) for striking at 16.4°.

From the perspective of impact mechanics, MFS and FFS presented analogous response in the internal and external stress distribution. However, significant differences were found compared with RFS. In a RFS, the foot presents a stiffer response where only the heel pad absorbed the energy of the impact, producing high compressive stresses in the hindfoot bones and cartilaginous joints. However, when the foot lands flat (MFS) or plantarflexed (FFS), the system gained compliance through flexure of the metatarsals aided by the elastic response of the fascia, which improves the absorption of the impact. Understanding these two mechanisms of impact absorption would help to predict which foot anatomical structure is more prone to injury.

Based on present results and considering that a touchdown is a repetitive event during running, different risks of impact related injuries can be associated with each strike pattern. The internal stress predicted for each foot component in the RFS, MFS and FFS simulations suggest that RFS runners could be more prone to suffer calcaneal stress fracture, cartilage damage, tibialis anterior tendinitis and heel ulceration, whereas MFS/FFS runners might have a greater risk of metatarsal stress fracture, plantar fasciitis, and submetatarsal ulceration.

In the present study, stress in the metatarsals was predicted higher for MFS and FFS compared to RFS, what can be presumed for runners that transitioned from shod to barefoot running since barefoot runners tend to land flatter i.e., less RFS (Lieberman et al. 2010). A recent clinical report alerted to two cases of metatarsals stress fracture in experienced runners who transitioned to minimalist footwear (Giuliani et al. 2011). It was proposed that the transition without specific training was a factor leading to injury. The plantar fasciitis risk predicted for MFS and FFS is based on two different principles. In FFS, the plantar fascia is stretched due to the bending of the foot; the foot arch compliance is greater, which produces a stretch of the fascia. However, in MFS the compressive stress originates at the base of the calcaneus at the plantar fascia insertion. This stress also occurs in RFS at very low impact angles. Plantar fasciitis has been correlated with load rate (Pohl et al. 2009), which is associated to RFS but also was correlated with low arch index (Pohl et al. 2009), which is related to excessive strain.

Another important factor when transitioned from shod to barefoot running is the intrinsic foot musculature. An increase in the size of these muscles, when transitioned to barefoot or minimalist running, has been hypothesized (Mullen et al. 2014), but only measured for training periods over 12 weeks (Miller et al. 2014; Campitelli et al. 2015; Johnson et al. 2015; Chen et al. 2016). The three largest plantar intrinsic foot muscles, abductor hallucis, flexor digitorum and quadratus plantae, have been demonstrated to have the capacity to control foot posture and longitudinal arch stiffness (Kelly et al. 2014), but no study has been found to analyze the role of the internal foot musculature as function of the strike pattern. Only one study has investigated the differences in the muscle activity between RFS and FFS runners (Yong et al. 2014). They measured electromyographic patterns of ten lower limb muscles from natural RFS and FFS runners founding the Tibialis Anterior muscle more active for RFS runners in the late swing phase, i.e. right before touchdown, whereas the Gastrocnemius more active for FFS runners.

Two attenuating factors were evaluated for RFS barefoot runners: the landing angle and the surface stiffness. The results suggested that the angle of impact plays a more important role in reducing plantar pressure than the surface stiffness. A previous correlation between the angle of impact and the local pressure under the heel was found in a kinematic study coupled with plantar pressure measurements (De Wit et al. 2000). It was assumed that barefoot runners strike in flatter positions to avoid high plantar pressure during impact. Our simulation confirmed this reduction of peak plantar pressures from severe (16.4°) to moderate (8.2°) impact angles. The slight influence of ground stiffness in stress levels could explain the absence of injury reduction when running on soft surfaces compared with harder surfaces (Nigg 2001).

The barefoot simulations and discussion of strike effects can be extended to the minimalist running since the minimalist shoes do not provide any additional cushioning support. Therefore, the mechanics of impact for barefoot and minimalist running can be assumed to be equal. The preference to run minimalist instead of barefoot is due to the protective layer that provides the sole to avoid minor skin injuries such as cuts or abrasions.

There are some limitations in the simulation that have to be considered. The model employed did not include the lower extremity which is responsible for part of the primary impact attenuation in running strike. To compensate for this, different boundary conditions were defined as a function of the strike pattern. Furthermore, the simulations were reduced to the quasi-static analysis of the impact moment. That is, dynamic factors such as the loading rate or the complete analysis of the stance phase were not analyzed. Lastly, the compressive response of healthy runners' plantar soles could provide better impact absorption capacity than the elder donor specimens tested.

In summary, most of the biomechanical studies of barefoot running were focused on kinematic and kinetic measurements. The present study was intended to analyze the internal stresses of the foot components at the moment of impact as a function of the strike pattern. Two different mechanisms of impact absorption were described and associated with injury risks. It was found that the landing angle has more relevance in reducing impact stresses than the stiffness of the surface. In addition, the model presented in this study is a step forward in the field of foot computational modeling involving a complete three-dimensional foot model with the real geometry of tendons and muscles, introducing non-linear properties for all soft tissues except plantar fascia and differentiating between rearfoot from forefoot plantar soft tissue responses.

Acknowledgments

The authors gratefully acknowledge the support of the project DPI2013-44987-R funded by the Spanish Ministry of Economy and competitiveness and the Brazilian funding agency CAPES.

Disclosure statement

No potential conflict of interest was reported by the authors.

ORCID

Enrique Morales-Orcajo http://orcid.org/0000-0001-9209-1373
Ricardo Becerro de Bengoa Vallejo http://orcid.org/0000-0003-1568-7602
Marta Losa Iglesias http://orcid.org/0000-0001-7588-2069

References

Altman AR, Davis IS. 2012. Barefoot running: Biomechanics and implications for running injuries. Curr Sports Med Rep. 11:244–250.

Altman AR, Davis IS. 2015. Prospective comparison of running injuries between shod and barefoot runners. Br J Sports Med. 1–6. http://www.ncbi.nlm.nih.gov/pubmed/26130697

Bennett MB, Ker RF. 1990. The mechanical properties of the human subcalcaneal fat pad in compression. J Anat. 171:131–138.

Bowles JE. 1996. Foundation analysis and design fifth edition. [place unknown]: McGraw-Hill.

Campitelli NA, Bernhard KN, Kidon A, Heard K, Spencer SA, Kawalec JS. 2015. Effect of vibram fivefingers minimalist shoes on the abductor hallucis muscle. J Am Podiatr Med Assoc. 106:344–351.

Chen TLW, Sze LKY, Davis IS, Cheung RTH. 2016. Effects of training in minimalist shoes on the intrinsic and extrinsic foot muscle volume. Clin Biomech. 36:8–13.

Chen W-M, Lee S-J, Lee PVS. 2014. The in vivo plantar soft tissue mechanical property under the metatarsal head: implications of tissues′ joint-angle dependent response in foot finite element modeling. J Mech Behav Biomed Mater 40:264–74.

Chen W-M, Phyau-Wui Shim V, Park S-B, Lee T. 2011. An instrumented tissue tester for measuring soft tissue property under the metatarsal heads in relation to metatarsophalangeal joint angle. J Biomech. 44:1801–4.

Chen Y, Davis B, Valdevit A. 2002. Regional differences in strength at the plantar skin-fat interface. In: Proc IEEE 28th Annu Northeast Bioeng Conf. [place unknown]: Ieee; p. 69–70.

Cheung RTH, Rainbow MJ. 2014. Landing pattern and vertical loading rates during first attempt of barefoot running in habitual shod runners. Hum Mov Sci. 34:120–127.

Chokhandre S, Halloran JP, van den Bogert AJ, Erdemir A. 2012. A three-dimensional inverse finite element analysis of the heel pad. J Biomech Eng 134:31002.

De Clercq D, Aerts P, Kunnen M. 1994. The mechanical characteristics of the human heel pad during foot strike in running: An in vivo cineradiographic study. J Biomech. 27:1213–1222.

Daoud AI, Geissler GJ, Wang F, Saretsky J, Daoud Y a., Lieberman DE. 2012. Foot strike and injury rates in endurance runners: A retrospective study. Med Sci Sports Exerc. 44:1325–1334.

Demir F. 2005. A new way of prediction elastic modulus of normal and high strength concrete-fuzzy logic. Cem Concr Res. 35:1531–1538.

Divert C, Mornieux G, Freychat P, Baly L, Mayer F, Belli A. 2008. Barefoot-shod running differences: Shoe or mass effect? Int J Sports Med. 29:512–518.

Erdemir A, Viveiros ML, Ulbrecht JS, Cavanagh PR. 2006. An inverse finite-element model of heel-pad indentation. J Biomech. 39:1279–86.

Fernandez JW, Ul Haque MZ, Hunter PJ, Mithraratne K. 2012. Mechanics of the foot Part 1: a continuum framework for evaluating soft tissue stiffening in the pathologic foot. Int j numer method biomed eng. 28:1056–70.

Gefen a, Megido-Ravid M, Itzchak Y. 2001. In vivo biomechanical behavior of the human heel pad during the stance phase of gait. J Biomech. 34:1661–5.

Giuliani J, Masini B, Alitz C, Owens BD. 2011. Barefoot-simulating footwear associated with metatarsal stress injury in 2 runners. Orthopedics. 34:e320–3.

Jenkins DW, Cauthon DJ. 2011. Barefoot running claims and controversies: a review of the literature. J Am Podiatr Med Assoc. 101:231–246.

Johnson AW, Myrer JW, Mitchell UH, Hunter I, Ridge ST. 2015. The Effects of a Transition to Minimalist Shoe Running on Intrinsic Foot Muscle Size. Int J Sports Med. 37:154–158.

Jordan C, Bartlett R. 1995. Pressure distribution and perceived comfort in casual footwear. Gait Posture. 3:215–220.

Kelly LA, Kuitunen S, Racinais S, Cresswell AG. 2012. Recruitment of the plantar intrinsic foot muscles with increasing postural demand. Clin Biomech. 27:46–51.

Kelly LA, Cresswell AG, Racinais S, Whiteley R, Lichtwark G. 2014. Intrinsic foot muscles have the capacity to control deformation of the longitudinal arch. J R Soc Interface. 11:20131188–20131188.

Ledoux WR, Blevins JJ. 2007. The compressive material properties of the plantar soft tissue. J Biomech. 40:2975–2981.

Lemmon D, Shiang TY, Hashmi A, Ulbrecht JS, Cavanagh PR. 1997. The effect of insoles in therapeutic footwear—A finite element approach. J Biomech. 30:615–620.

Lieberman DE, Venkadesan M, Werbel W a, Daoud AI, D'Andrea S, Davis IS, Mang'eni RO, Pitsiladis Y. 2010. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature. [cited 2015 Feb 2]; 463:531–535.

Lohman EB, Balan Sackiriyas KS, Swen RW. 2011. A comparison of the spatiotemporal parameters, kinematics, and biomechanics between shod, unshod, and minimally supported running as compared to walking. Phys Ther Sport. 12:151–63.

Miller-Young JE, Duncan N a, Baroud G. 2002. Material properties of the human calcaneal fat pad in compression: experiment and theory. J Biomech. 35:1523–31.

Miller EE, Whitcome KK, Lieberman DE, Norton HL, Dyer RE. 2014. The effect of minimal shoes on arch structure and intrinsic foot muscle strength. J Sport Heal Sci. 3:74–85.

Morales-Orcajo E, Bayod J, Barbosa de Las Casas E. 2016. Computational Foot Modeling: Scope and Applications. Arch Comput Methods Eng. 23:389–416.

Morales-Orcajo E, Bayod J, Becerro-de-Bengoa-Vallejo R, Losa-Iglesias M, Doblare M. 2015. Influence of first proximal phalanx geometry on hallux valgus deformity: a finite element analysis. Med Biol Eng Comput. 53:645–653.

Morales-Orcajo E, Souza TR, Bayod J, Barbosa de Las Casas E. 2017. Non-linear finite element model to assess the effect of tendon forces on the foot-ankle complex. Med Eng Phys. 49:71–78.

Mullen S, Cotton J, Bechtold M, Bruce Toby E. 2014. Barefoot running: The effects of an 8-week barefoot training program. Orthop J Sport Med. 2:1–5.

Nigg BM. 2001. The role of impact forces and foot pronation: a new paradigm. Clin J Sport Med. 11:2–9.

Novacheck TF. 1998. The biomechanics of running: Review Paper. Gait Posture. 7:77–95.

Petre M, Erdemir A, Panoskaltsis VP, Spirka T A, Cavanagh PR. 2013. Optimization of nonlinear hyperelastic coefficients for foot tissues using a magnetic resonance imaging deformation experiment. J Biomech Eng. 135:61001–12.

Pohl MB, Hamill J, Davis IS. 2009. Biomechanical and Anatomic Factors Associated with a History of Plantar Fasciitis in Female Runners. Clin J Sport Med. 19.

Richards CE, Magin PJ, Callister R. 2009. Is your prescription of distance running shoes evidence-based? Br J Sports Med. 43:159–162.

Rixe JA., Gallo RA., Silvis ML. 2012. The barefoot debate: Can minimalist shoes reduce running-related injuries? Curr Sports Med Rep. 11:160–165.

Rome K, Webb P, Unsworth A, Haslock I. 2001. Heel pad stiffness in runners with plantar heel pain. Clin Biomech. 16:901–905.

Rothschild CE. 2012. Primitive Running: A Survey Analysis of Runners' Interest, Participation, and Implementation. J Strength Cond Res. 26.

Samaan CD, Rainbow MJ, Davis IS. 2014. Reduction in ground reaction force variables with instructed barefoot running. J Sport Heal Sci. 3:143–151.

Shih Y, Lin KL, Shiang TY. 2013. Is the foot striking pattern more important than barefoot or shod conditions in running? Gait Posture. 38:490–494.

Spears I, Miller-Young J, Waters M, Rome K. 2005. The effect of loading conditions on stress in the barefooted heel pad. Med Sci Sports Exerc. 37:1030–1036.

Squadrone R, Gallozzi C. 2009. Biomechanical and physiological comparison of barefoot and two shod conditions in experienced barefoot runners. J Sports Med Phys Fitness. 49:6–13.

Squadrone R, Rodano R, Hamill J, Preatoni E. 2015. Acute effect of different minimalist shoes on foot strike pattern and kinematics in rearfoot strikers during running. J Sports Sci. 33:1–9.

Telfer S, Woodburn J, Cavanagh PR. 2015. Footwear embedded ultrasonography to determine plantar soft tissue properties for finite element simulations. Footwear Sci. 7:9424280.

Tong J, Lim C., Goh O. 2003. Technique to study the biomechanical properties of the human calcaneal heel pad. Foot. 13:83–91.

De Wit B, De Clercq D, Aerts P. 2000. Biomechanical analysis of the stance phase during barefoot and shod running. J Biomech. 33:269–278.

Witana CP, Goonetilleke RS, Xiong S, Au EYL. 2009. Effects of surface characteristics on the plantar shape of feet and subjects' perceived sensations. Appl Ergon. 40:267–79.

Wright DG, Rennels DC. 1964. A Study of the Elastic Properties of Plantar Fascia. J Bone Joint Surg Am. 46:482–92.

Yong JR, Silder A, Delp SL. 2014. Differences in muscle activity between natural forefoot and rearfoot strikers during running. J Biomech. 47:3593–3597.

Zhang R, Haibao X, Zhang S, Zeng G, Li J. 2014. Finite Element Analysis in the Characteristics of Ostrich Foot Toenail Traveling on Sand. Appl Mech Mater. 461: 213–219.