

# FLEX CEUs



## Running and the Effect of Footwear Insoles



# Influence of custom-made and prefabricated insoles before and after an intense run

## Abstract

Each time the foot contacts the ground during running there is a rapid deceleration that results in a shock wave that is transmitted from the foot to the head. The fatigue of the musculoskeletal system during running may decrease the ability of the body to absorb those shock waves and increase the risk of injury. Insoles are commonly prescribed to prevent injuries, and both custom-made and prefabricated insoles have been observed to reduce shock accelerations during running. However, no study to date has included a direct comparison of their behaviour measured over the same group of athletes, and therefore great controversy still exists regarding their effectiveness in reducing impact loading during running. The aim of the study was to analyse the acute differences in stride and shock parameters while running on a treadmill with custom-made and prefabricated insoles. Stride parameters (stride length, stride rate) and shock acceleration parameters (head and tibial peak acceleration, shock magnitude, acceleration rate, and shock attenuation) were measured using two triaxial accelerometers in 38 runners at 3.33 m/s before and after a 15-min intense run while using the sock liner of the shoe (control condition), prefabricated insoles and custom-made insoles. No differences in shock accelerations were found between the custom-made and the control insoles. The prefabricated insoles increased the head acceleration rate (post-fatigue,  $p = 0.029$ ) compared to the control condition. The custom-made reduced tibial (pre-fatigue,  $p = 0.041$ ) and head acceleration rates (pre-fatigue and post-fatigue,  $p = 0.01$  and  $p = 0.046$ ) compared to the prefabricated insoles. Neither the stride nor the acceleration parameters were modified as a result of the intense run. In the present study, the acute use of insoles (custom-made, prefabricated) did not reduce shock accelerations compared to the control insoles. Therefore, their effectiveness at protecting against injuries associated with elevated accelerations is not supported and remains unclear.

**Competing interests:** The authors have declared that no competing interests exist.

## Introduction

Running is a type of physical activity that involves the athlete striking the ground about 600 times per kilometer [1,2]. Each foot strike during running there is a rapid deceleration of the lower-limb that results in a shock wave that is transmitted from the foot to the head [3]. On its way upwards to the head, this shock is partly absorbed by the ground, the running shoes and the musculoskeletal system in a process known as shock attenuation [4,5]. However, even though the musculoskeletal system is prepared to deal with each one of these contacts, their repetitive and cumulative effect on the human body could overload and fatigue the musculoskeletal system, especially of the lower leg, and lead to increased risk of overuse injuries such as patellofemoral pain syndrome, tibial stress fractures, plantar fasciitis, metatarsalgia and Achilles tendinitis [6–9]. In this sense, the analysis of the shock attenuation, the loading rate and the magnitude of the shock wave (also called impact or shock acceleration) during running is drawing the attention of the research community as a consequence of their relationship with tibial stress fractures [10,11], performance [4,5], and lower-limb comfort [12].

Repeated exposure to shock accelerations, as experienced by long distance runners, is believed to increase the incidence of injury as a result of the reduced ability of the musculoskeletal system to absorb these shock waves [13]. The ability of the musculoskeletal system to attenuate these accelerations decreases with fatigue, and therefore the articular cartilage and ligaments become more vulnerable to excessive loading stress loading [14]. In this sense, previous studies have observed that shock acceleration increases with speed and fatigue [2,15,16] and suggest that muscle fatigue also plays a role in overloading the musculoskeletal system leading to overuse injury [10].

Considering that shock accelerations are inherent to running, different strategies including modifying foot strike pattern [17], footwear [18,19], compressive garments [2], gait retraining [20] or insoles [21] are being investigated aiming to reduce such accelerations and therefore decrease the risk and frequency of injury in runners. Insoles are in-shoe devices widely prescribed by podiatrists to reduce or eliminate pathological stresses to the foot or other portions of the kinetic chain [22]. However, there is a great controversy between the effectiveness of over the shelf (prefabricated) insoles chosen by taking solely into account the individual's foot size and custom-made insoles. On the one hand, custom-made insoles are devices built by a podiatrist from a three-dimensional representation of the individual's foot and their use has been associated with pain relief [23,24], improved comfort [12], reduced plantar pressure [25], impact magnitude, and loading rate [26–28]. On the other hand, prefabricated insoles are mass-produced devices at a fraction of the cost of custom-made insoles and therefore it is not surprising that their use is expanding. Prefabricated cushioning insoles have also been associated with reduced plantar pressure [29], shock accelerations [27], impact forces, and loading rates [30]. However, there is a paucity of studies analysing the efficiency at attenuating the running-related accelerations of these types of insoles (custom-made vs prefabricated insoles) and compared to a control situation (sock liner of the running shoe) in the same population. Although individualised prescription is recognised to be a gold standard and it would be reasonable to expect that a custom-made insole adapted to the individual's foot would better fulfil the runner's expectations and provide more protection than a prefabricated insole [31], scientific evidence to demonstrate its benefits is needed. As a result, the aim of the present study was therefore to determine the effects of different insoles (custom-made, prefabricated, control) on stride and shock acceleration parameters before and after an intense run. It was hypothesised that the use of custom-made insoles would reduce shock accelerations compared to the control and prefabricated insoles. It was also hypothesised that runners would exhibit greater shock acceleration after the intense run independent of the insole condition.

## Methods

### Participants

A sample size of 34 participants was estimated using G\*Power 3 software for a desired power of 80% from the results of published work which studied similar dependent variables [3,27]. As a result, thirty-eight recreational runners recruited from local running clubs participated in the study: 20 males and 18 females ( $29.8 \pm 5.3$  years;  $170.3 \pm 11.4$  cm;  $65.4 \pm 10.1$  kg, weekly running distance:  $36.5 \pm 7.2$  km/week, best time in 10k race:  $53.6 \pm 9.4$  min). Inclusion criteria were: I) no injuries in the last year, II) no previous lower-limb surgery in the last 3 years, III) no previous use of insoles, and IV) a training routine of at least 20 km / week. All runners provided written informed consent before participation. The study procedures complied with the Declaration of Helsinki and were approved by Fernando Alejo Verdú Pascual, acting secretary of the University ethics committee (Comité Ético de Investigación en Humanos de la Comisión de Ética en Investigación Experimental de la Universidad de Valencia, approval number H1411628681304).

### Insole conditions and customisation

Participants carried out the study under three different conditions: the original sock liner of their running shoes (control), the prefabricated insoles (<http://www.herbitas.com/plantilla-tecnoped-especial-running-p-4-50-2549/>) (Tecnoped Run, Herbitas, Valencia, Spain) and the custom-made insoles (<http://sidas.eurowintuecommerce.com/articulo/SPCTR-L-opctrunnings4243.html>) (OPCT Run, Sidas S.L., Barcelona, Spain) (Fig 1). The custom-made insoles were initially the same in terms of material and properties, and their shape was afterwards customized based on the feet of the participants. For the customisation of the custom-made insoles, as described in detail previously [32], participants stood on a Printlab2 platform (Podiatech, Sidas Technologies, Voiron, France), which consisted of a pair of silicon vacuum bags that allowed an experienced podiatrist to create a plaster mould based on the plantar print of the participants taking into account their foot morphology. Afterwards, through a thermo-welding process, the custom-made insoles were warmed-up and adapted with a vacuum system (Mobilab2, Podiatech, Sidas Technologies, Voiron, France) to the exact shape of the heel, midfoot and forefoot of each participant using their individual feet plaster moulds. As a result, the insoles fitted perfectly to the plantar surface of the feet of the participants: feet with slightly lower medial arch resulted in insoles with lower height support in this area, whereas participants with slightly higher medial arch resulted in insoles with a higher support.

### Protocol

Participants performed three running tests on different days and the total duration of the study was 2 weeks (Fig 2). All running tests were carried out on a treadmill (Excite Run 700, TechnoGymSpa, Gambettola, Italy). As described elsewhere [32], in the first laboratory session, participants performed an incremental test to determine their lactate threshold speed right below 4-mM blood lactate concentration [33]. This test involved a 5-min warm-up at 2.78 m/s followed by 0.56 m/s speed increments every 3 min. Blood samples were taken from the ear lobe at the end of each stage [33] and blood lactate concentration was determined using a Lactate Pro Analyzer (Arkay Factory Inc., Shiga, Japan). Blood lactate concentration was used as the physiological parameter for determining their individual lactate threshold speed as it is considered a useful tool to effectively predict exercise performance [34]. Then, the speed of the last stage before reaching 4 mM of blood lactate concentration was written down and, later on, in the laboratory sessions 2 and 3, was used as the fatiguing speed for the



Custom-made Insoles

- Top layer: Podiamic 160 polyethylene + ethyl-vinyl acetate (EVA), 2.5 mm thick, hardness 30° Shore A.
- Sole reinforcement (white): Polyester resin Transflux® 1.0 mm thick.
- Forefoot insert: Synthetic Viscotene® 2.5 mm thick, hardness 30° Shore A.
- Rearfoot insert: Podiaflex® resin 0.9 mm thick.
- Rearfoot reinforcement: Polyester resin Transflux® 1.0 mm thick.



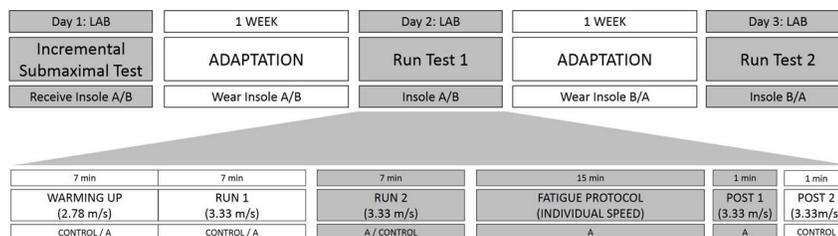
Prefabricated Insoles

- Forefoot reinforcement of polyurethane foam, hardness 15-25° Shore A.
- Rearfoot reinforcement of polyurethane foam, hardness 15-25° Shore A.
- Extra support under medial arch (10 cm long and 3.5 mm high) of Techcarbon.

**Fig 1. Characteristics of the insoles.**

doi:10.1371/journal.pone.0173179.g001

15-min intense run. Following the incremental test of the first laboratory session, a pair of insoles (custom-made, prefabricated) was randomly given to each participant using a research randomizer program [35]. During the adaptation week, participants were asked to run 3 times using the assigned insoles and to lead their daily routine during this week (using the insoles with their sport footwear when going for a walk, in their leisure time, etc.) for adaptation purposes and return to the lab for session 2 (Fig 2: Run Test 1) after this adaptation week.



**Fig 2. Study Design and Protocol.**

doi:10.1371/journal.pone.0173179.g002

Since the use of insoles was a new situation for the participants, they were asked to wear their own running footwear during the adaptation week and throughout the tests in order to introduce no further changes in their running customary condition, as recommended by previous studies [36,37]. After the first familiarisation week, participants came to the lab to perform the laboratory session 2 (laboratory sessions 2 and 3 were identical with the only exception of the insole being used and measured, custom-made or prefabricated). In these laboratory sessions 2 and 3, participants performed a 7-min warm-up at 2.78 m/s with the sock liners of the shoe (control) or the study insoles of that session (custom-made or prefabricated) at random. Following the warm-up, participants ran for 7 min at 3.33 m/s and shock accelerations and stride parameters were measured within the last minute of the run. After this running bout, the insoles inside the footwear were replaced by the second condition (control or study insoles, depending on the initial order) and the 7-min run at 3.33 m/s was repeated in order to measure shock acceleration and stride parameters again. Afterwards, participants ran for 15 min (intense run) at their individual lactate threshold speed ( $4.04 \pm 0.36$  m/s). All participants were able to finish the intense run and the rating of perceived exertion between 6 and 20 [38] was also reported during the last minute of the run. Immediately after the intense run, acceleration and stride parameters were measured again during two 1-min runs at 3.33 m/s (post-fatigue control and post-fatigue insole conditions). The time between measurements was not longer than 1 minute (time needed to change the insoles inside the running shoes).

At the end of the laboratory session 2, participants received the second pair of study insoles (custom-made or prefabricated, depending on the initial randomisation) and repeated this running protocol with the control and the second pair of study insoles (laboratory session 3) after another adaptation week.

## Data collection

Accelerations were measured during 10 seconds using two lightweight tri-axial accelerometers (Sportmetrics, Spain; mass: 2.5 g; dimensions: 40 mm × 22 mm × 12 mm; sampling frequency 500 Hz). As explained in detail elsewhere [2], the accelerometers were attached to the skin as tight as possible to the participants' tolerance with double-sided adhesive tape and secured via elastic belts around the proximal anteromedial aspect of the tibia and around the forehead. The vertical axis of the accelerometer was aligned to be parallel to the long axis of the shank (Fig 3). Acceleration data were filtered (8-order low-pass digital Chebyshev type II filter, stop-band edge frequency 120 Hz, stop-band ripple 40 dB) [39] and analysed using Matlab (The Math Works Inc., Natick, MA, USA). From the acceleration signal, stride frequency was calculated as the time between consecutive leg impacts, whereas stride length was obtained by dividing running speed by stride rate [5]. On the other hand, the following acceleration parameters were also calculated [2]: head and tibia peak acceleration (maximal amplitude), acceleration magnitude (difference between the positive and the negative peak), acceleration rate (slope from ground contact to peak acceleration), and shock attenuation (reduction in peak acceleration from the tibia to the head as a percentage of the head acceleration).

## Statistical analysis

A commercial statistical package (SPSS 18.0, SPSS Inc., Chicago, IL, USA) was used for statistical analyses. After checking the normality of the variables (Kolmogorov–Smirnov), a descriptive analysis of the data was performed. The sphericity assumption was verified by the Mauchly test. Then, a two-way repeated-measures ANOVA with insole (control, prefabricated, custom-made) and fatigue (pre- and post- intense run) as intra-subject factors and



**Fig 3. Accelerometer placement on the tibia.**

doi:10.1371/journal.pone.0173179.g003

acceleration and stride parameters as dependent variables was performed. Bonferroni post-hoc was carried out to provide details as to the whereabouts of significant differences. Significance was set at  $\alpha = 0.05$ . Data are presented as mean  $\pm$  95% confidence intervals (95% CI).

**Table 1. Mean (95% confidence intervals) of the stride parameters for the different insole conditions and fatigue state.**

	PRE			POST		
	Control	Prefabricated	Custom-made	Control	Prefabricated	Custom-made
Stride Rate (stride/s)	1.41 (1.39–1.44)	1.42 (1.39–1.44)	1.41 (1.37–1.44)	1.42 (1.39–1.44)	1.42 (1.39–1.44)	1.37 (1.28–1.49)
Stride Length (m/stride)	2.36 (2.32–2.41)	2.36 (2.31–2.40)	2.36 (2.31–2.41)	2.36 (2.31–2.41)	2.36 (2.31–2.41)	2.37 (2.32–2.42)

PRE: pre-fatigue; POST: post-fatigue. No significant difference was found between the pre-fatigue and the post-fatigue values.

doi:10.1371/journal.pone.0173179.t001

## Results

### Effect of the insole condition

The different insoles did not influence stride rate and stride length ( $p > 0.05$ ) (Table 1). However, the insole conditions did affect the shock accelerations during running (Table 2). In the pre-fatigue state, the use of custom-made insoles reduced the head acceleration rate ( $p = 0.041$ , mean difference: 6.3, 95%CI mean difference: 0.21–12.48) and the tibial acceleration rate ( $p = 0.014$ , mean difference: 85.38, 95%CI mean difference: 14.56–156.20) compared to the prefabricated insoles. Moreover, in the post-fatigue state, the prefabricated insoles increased the head acceleration rate compared to the custom-made ( $p = 0.046$ , mean difference: 6.84, 95%CI mean difference: 0.11–13.59) and the control insoles ( $p = 0.029$ , mean difference: 6.97, 95%CI mean difference: 0.56–13.38). No difference was observed between the custom-made and the control insoles for any of the parameters analysed ( $p > 0.05$ ).

### Effect of the intense run

Participants considered that the intense protocol was ‘Hard’ as they reported a rating of perceived exertion of 14.34 (13.40–15.42) within the last minute of the intense run. Stride rate and stride length were not influenced by the intense run ( $p > 0.05$ ) (Table 1). Similarly, the intense run did not modify any of the shock acceleration parameters measured in the study ( $p > 0.05$ ) (Table 2).

**Table 2. Mean (95% confidence intervals) of the acceleration parameters for the different insole conditions and fatigue state.**

	PRE			POST		
	Control	Prefabricated	Custom-made	Control	Prefabricated	Custom-made
Max Tibia (G)	7.89 (7.00–8.78)	8.13 (7.15–9.11)	7.69 (6.93–8.44)	7.75 (6.73–8.77)	8.59 (7.55–9.63)	7.96 (6.91–9.00)
Max Head (G)	2.37 (2.20–2.54)	2.38 (2.15–2.60)	2.31 (2.13–2.49)	2.25 (2.01–2.48)	2.34 (2.06–2.63)	2.27 (2.08–2.47)
Magnitude Tibia (G)	8.54 (7.63–9.46)	8.63 (7.56–9.69)	8.61 (7.79–9.44)	8.50 (7.48–9.52)	9.31 (8.19–10.42)	9.05 (7.96–10.13)
Magnitude Head (G)	2.43 (2.26–2.60)	2.41 (2.19–2.63)	2.41 (2.23–2.60)	2.31 (2.09–2.53)	2.38 (2.11–2.65)	2.36 (2.17–2.56)
Tibia Rate (G/s)	272.28 (200.67–343.90)	319.99 (236.96–403.02)	<b>234.61<sup>*b</sup></b> (173.53–295.70)	257.03 (186.06–328.01)	340.06 (237.50–442.61)	287.50 (187.58–387.41)
Head Rate (G/s)	55.05 (48.96–61.14)	58.33 (50.40–66.26)	<b>51.98<sup>*b</sup></b> (44.93–59.04)	51.34 (43.86–58.82)	<b>58.31<sup>*a</sup></b> (48.73–67.90)	<b>51.47<sup>*b</sup></b> (44.06–58.87)
Attenuation (%)	66.43 (62.52–70.34)	67.37 (62.94–71.80)	65.78 (60.33–71.23)	66.82 (61.71–71.92)	70.55 (66.89–74.20)	64.85 (55.54–74.16)

PRE: pre-fatigue; POST: post-fatigue.

<sup>\*a</sup>  $P < .05$ . significant difference compared to control insoles for the matching fatigue condition.

<sup>\*b</sup>  $P < .05$ . significant difference compared to prefabricated insoles for the matching fatigue condition. No significant difference was found between the pre-fatigue and the post-fatigue values.

doi:10.1371/journal.pone.0173179.t002

The effect of the insole intervention on the stride and shock accelerations parameters was not modified by the fatigue state, as no significant interaction ( $p > 0.05$ ) was observed between the two factors (insole, fatigue).

## Discussion

This study analysed the effects of prefabricated and custom-made insoles on stride and shock acceleration parameters before and after an intense run. Our main finding suggests that even though the use of custom-made insoles reduced the acceleration rate at the tibia and head compared to prefabricated insoles, no major differences were observed between the study insoles (custom-made, prefabricated) and the control insoles.

Prolonged and elevated magnitudes of shock accelerations have been associated with increased risk of injuries [11]. The use of different strategies including gait retraining, compressive garments or cushioned shoes or insoles have aimed to reduce these shock accelerations during running [2,20,27,30,40]. In the present study, it was hypothesised that custom-made insoles would reduce the shock acceleration experienced by the runner compared to the prefabricated and the control insoles. However, this hypothesis was only partly supported as the use of custom-made insoles only led to a lower acceleration rate compared to prefabricated insoles, whereas no differences with the control condition were observed. Moreover, no alterations of the tibial and head peak accelerations were observed when running with the study insoles (custom-made, prefabricated) compared to the control condition, and therefore these findings question the efficacy of insoles when aiming to reduce shock accelerations during running.

The use of insoles has been suggested as a strategy to reduce the shock accelerations associated with running, thereby decreasing the risk of overuse injuries [27,41]. However, while most of the previous studies analysed the effect of insoles compared to a control situation (running with insoles versus running without insoles) [27,30,42], to the authors' knowledge this is the first study to analyse the effect of custom-made insoles before and after an intense run on impact accelerations during running compared to prefabricated and control insoles.

In the present study neither the acceleration peaks nor the acceleration magnitudes on the tibia and head were altered when running with insoles compared to the control condition. Although this result is in accordance with Laughton et al. [42], who did not find differences in tibial peak accelerations when running with and without customised insoles; it is also in contrast with two previous studies, who observed reduced tibial peak accelerations when running with cushioned insoles [21] and semi-rigid prefabricated insoles inside military boots [30]. One of the reasons that may explain the differences among studies is the cushioning system of the footwear. In this sense, running shoes have inherently greater shock attenuation properties than street shoes or military boots, and consequently the overall effect of the shoe-insole complex may vary depending on the footwear [41]. Another reason that will likely explain the controversy among studies is the different materials and the thickness of the layers used to build the insoles of each study. In this sense, in contrast with the polyethylene + EVA (custom-made) and polyurethane foam with Techcarbon (prefabricated) of the insoles used in this study, previous studies have used insoles based on a number of materials such as polyurethane foam + Poron foam [21], Trocellen foam with polypropylene [30], or suborthelene covered with a neoprene pad [42]. As a result, the behaviour of the different materials against vibrations and accelerations may explain the differences between studies.

Recent studies are emphasising the role of loading rate rather than peak acceleration values when analysing the effects of the resulting shock wave following exercise on the musculoskeletal system [43,44]. Repetitive, rapidly applied loads are more associated with joint degeneration

than slowly applied loads of equal or even greater magnitudes [45] and a recent study has found a positive relationship between greater acceleration rate and stress fractures [11]. Moreover, loading rate may describe better than the acceleration magnitude the capacity of the cushion structure (footwear, insole) to reduce the rate at which the shock acceleration is transmitted to the lower extremity and may be a better indicator of cushioning performance [46]. In the present study, it was observed that the prefabricated insoles increased by 27% and 11% the tibial and head acceleration rates compared to the custom-made insoles and by 12% the head acceleration rates compared to the control condition, which contrasts with a recent study who found a reduction of the tibial acceleration rate during running with insoles compared to running without insoles [42]. The differences between studies may be explained by the deformation of the materials of the insoles, as it has been previously suggested that the materials of shoe-insole complex determine the spring stiffness of the footwear-insole-foot system and ultimately influence their behaviour against accelerations during running [47–50]. Therefore, and taking into account that the acceleration rates may represent the cushioning performance of the structure and influence the risk of overuse running injuries, the use of prefabricated insoles as a protective mechanism against accelerations during running is not supported. However, as custom-made insoles decreased both the tibial and the head acceleration rates compared to the prefabricated insoles, if a runner would need to use insoles for a given biomechanical reason (comfort, motion control, plantar redistribution), the use of custom-made insoles would behave better at attenuating shock accelerations than the prefabricated insoles and could be more effective as a protective strategy to reduce the risk of overuse running-related injuries or as a conservative treatment for the rehabilitation of runners after an overuse running injury, similar to those studies observing lower shock accelerations resulting from gait retraining [20,51]. However, this hypothesis remains just a speculation and future studies should investigate it.

On the other hand, no difference between the custom-made and the control insoles was observed for any of the shock acceleration parameters, which indicates that even though the use of custom-made insoles has been observed to effectively relieve pain [23,24], improve comfort [12], and redistribute plantar pressure [25], their role as a shock-absorbing strategy during running is not supported either. Only two studies have observed a reduction of shock acceleration during running with insoles compared to running without insoles [27,41]. However, the insoles used in those studies were described as cushioned insoles (3–6 mm thick with foam cover) [27] and shock-absorbing insoles (1–6 mm thick with foam support) [41]. Whereas the insoles in the present study were made of harder and stiffer materials and may have stabilise better the movement of the rearfoot. Taking into account that foot pronation is considered a shock-absorbing mechanism [52,53], it could be speculated that the control provided by the use of insoles could reduce the pronation of the foot, thereby reducing the efficiency of this shock-absorption mechanism and lead to greater shock accelerations. However, foot pronation was not measured in this study and this speculation needs to be further investigated.

Of special relevance is the recent publication by Nigg et al.[54], who stated that there is still no evidence to confirm the relationship between certain factors that traditionally were believed to increase injury risk such as pronation or shock accelerations and the probability of suffering a running-related injury. These authors indicate that studies on this field to support this association are insufficient and those who observed a relationship between shock accelerations and injury risk had a small sample size. Therefore, there is still controversy nowadays regarding the role that accelerations play during running and their effect on the human body overtime. As a result, future studies analysing the effects or long-term exposure to shock accelerations on the human body are encouraged to throw some light into this interesting matter.

The majority of the running-related studies are conducted in a non-exerted state. Although difficult, analysing the effects of the fatigue is important because it is a regular state experienced by all runners and it is when the athlete is fatigued that most overuse running-related injuries are thought to occur [2,55]. In the present study it was hypothesised that the fatigue state provoked by the intense run would increase shock acceleration. However, our results showed no changes in peak acceleration and acceleration rate with the development of the fatigue state. Previous studies have found an increase [2,15,16,56] as well as a reduction [57] of shock accelerations with fatigue. These authors suggested that a change in the attenuation properties of the body as a result of muscle fatigue could be due to the loss of the shock-absorbing capacity of muscles or to alterations in the lower extremity kinematics to compensate for the change in muscle ability [15,58]. In this sense, a decrease in stride rate leading to a greater shock acceleration was reported after a fatigue run [16,56]. These authors suggested that the alteration of the 'optimal' stride rate could have influenced shock transmission. However, the runners in our study, in agreement with Mercer et al. [58], did not make any adjustments to stride rate in response to fatigue. This result may indicate that runners in the present study were able to maintain their optimal stride rate and it could explain why the shock accelerations were not modified after the intense run. Discrepancies in the shock acceleration behaviour after the intense run can be attributed to the differences in the fatigue protocols used between studies. In the current study, in order to have a greater ecological validity, participants ran for 36 minutes (21 minutes resulting from the pre-fatigue running conditions plus 15 minutes of the intense run) at a training pace, which is a fatigue state more commonly reached within the recreational running population, rather than an incremental running protocol to exhaustion. On the other hand, other studies measured shock acceleration on a runway after a 20-min and a 40-min run [33], on a treadmill after a 30-min run [16,56], or throughout an increasing protocol until exhaustion [15,58]. Thus, the actual level of fatigue attained by the participants and the type of exercise chosen to reach the fatigue state (short protocols at high intensity versus longer protocols at lower intensity) may account for the inconsistent results observed in the literature.

Running on a treadmill could be considered a limitation of the study. Even though a treadmill was used in order to better control some variables (running speed, hardness and slope of the running surface), running on a treadmill could lead to different running biomechanics compared to overground running [3]. Moreover, the running pattern of the athletes (rearfoot, midfoot, forefoot) and the cushioning system of the athlete's footwear was not controlled (standard shoes were not provided) in order not to alter further their running customary conditions, but these factors may influence shock accelerations and future studies should look at these parameters while controlling running pattern and footwear. The two models of insoles (custom-made, prefabricated) were chosen based on their popularity among runners and podiatrists. While these results are interesting because they come from analysing two very popular types of insoles, caution is advised when interpreting these results as the differences in materials and stiffness of the insoles and running shoes were not taken into account, which may have influenced the results. As a result, future studies should control the materials and properties of the insoles and running shoes. Finally, participants in our study reported an average RPE value of 14 (Hard) after the intense run, which indicates that the intense run may have not been fatiguing enough to provoke some of the biomechanical adaptations observed in previous studies. Therefore, in future studies it would be of interest to investigate the effects of custom-made and prefabricated insoles on shock acceleration during overground running or after more extenuating running tests in order to provide a better insight into the shock attenuation mechanisms of these types of insoles and their potential role as an injury-prevention strategy.

## **Conclusion**

This study demonstrated that the acute use of insoles (both custom-made and prefabricated) did not reduce shock accelerations compared to the control condition. However, it was observed that custom-made insoles reduced tibial and head acceleration rate compared to prefabricated insoles. Although the effectiveness of insoles at reducing shock accelerations during running remains unclear, the custom-made insoles led to lower shock acceleration rates than the prefabricated insoles and therefore showed a better shock attenuation behaviour.



# Effects of a leaf spring structured midsole on joint mechanics and lower limb muscle forces in running

## Abstract

To enhance running performance in heel-toe running, a leaf spring structured midsole shoe (LEAF) has recently been introduced. The purpose of this study was to investigate the effect of a LEAF compared to a standard foam midsole shoe (FOAM) on joint mechanics and lower limb muscle forces in overground running. Nine male long-distance heel strike runners ran on an indoor track at  $3.0 \pm 0.2$  m/s with LEAF and FOAM shoes. Running kinematics and kinetics were recorded during the stance phase. Absorbed and generated energy (negative and positive work) of the hip, knee and ankle joint as well as muscle forces of selected lower limb muscles were determined using a musculoskeletal model. A significant reduction in energy absorption at the hip joint as well as energy generation at the ankle joint was found for LEAF compared to FOAM. The mean lower limb muscle forces of the m. soleus, m. gastrocnemius lateralis and m. gastrocnemius medialis were significantly reduced for LEAF compared to FOAM. Furthermore, m. biceps femoris showed a trend of reduction in running with LEAF. The remaining lower limb muscles analyzed (m. gluteus maximus, m. rectus femoris, m. vastus medialis, m. vastus lateralis, m. tibialis anterior) did not reveal significant differences between the shoe conditions. The findings of this study indicate that LEAF positively influenced the energy balance in running by reducing lower limb muscle forces compared to FOAM. In this way, LEAF could contribute to an overall increased running performance in heel-toe running.

## Introduction

From a biomechanical perspective, the mechanical energy generated by the muscles of the lower limb joints enables the runner to fulfil the movement task and determines running performance. In the literature, three major strategies have been proposed to improve the mechanical energy cost in running to enhance running performance: (1) storage and return of energy (2) optimization of the muscle functions by enabling muscles to work at an optimal force-velocity and force-length relationship, and (3) minimization of energy loss [1, 2].

Running shoe designs have had limited success in applying the concept of energy return [1, 3] and the concept of functional optimization of the musculoskeletal system lacks scientific

**Competing interests:** Please note that the study was supported by adidas FUTURE Team, adidas AG, Herzogenaurach, Germany. adidas had no involvement in the collection, analysis, or interpretation of any data. The results of the current study do not constitute endorsement of the product by the authors or the journal. There are no patents, products in development, or marketed products to declare. This does not alter the authors' adherence to all PLOS ONE policies on sharing data and materials. Therefore, none of the authors of this article has any conflict of interest.

support [4]. Therefore, Nigg and Segesser [2] suggested that running shoe designs should focus on strategies to minimize energy loss. Consequently, material [5, 6] and structural changes of running shoe midsoles have been considered [7, 8].

One of these shoe designs is a leaf spring structured midsole shoe (LEAF, Fig 1A). In contrast to a standard foam midsole shoe (FOAM, Fig 1B) the midsole of LEAF consists of a series of non-linked leaf springs. The LEAF shoe uses the midsole deformation induced by the vertical ground reaction force for shifting the shoe anteriorly during the first part of stance phase in heel-toe running [7]. When using LEAF compared to FOAM in treadmill running, the anterior foot shift resulted in increased stride length and improved running economy [7].

Based on these findings, one could presume a positive effect of LEAF compared to FOAM on the mechanical energy cost of the human locomotor system. However, it is important to determine the sources and magnitudes of energy absorption and generation to achieve a deeper understanding of the potential mechanisms on how this specific footwear affects the overall energy cost in running [9]. Independent of the footwear joint power analysis during stance phase has indicated that in each joint energy is either absorbed (time-integrated negative power) or generated (time-integrated positive power) in distinct phases [9]. It has been shown that LEAF leads to an anterior foot shift during the first part of stance [7], therefore, it can be hypothesized that LEAF compared to FOAM affects especially the first part of stance showing reduced energy absorption of the lower limb joints.

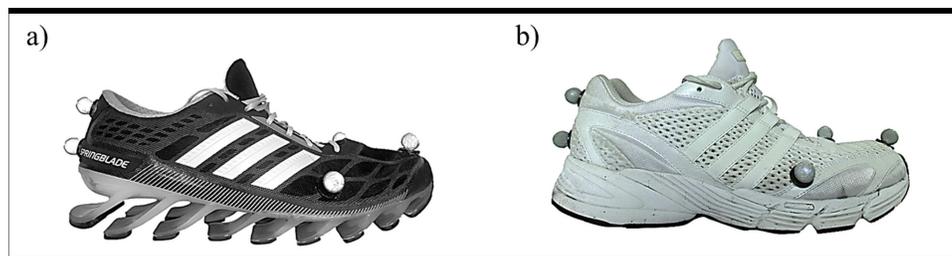
While the absorption of energy is considered to be eccentric muscle activity, the generation of energy is related to concentric muscle activity [9]. To determine the forces produced by the lower limb muscles during running, inverse dynamic musculoskeletal models can be used [10–13]. These models are constrained to estimate the raw muscle forces from the previously calculated joint moments [14]. In this way musculoskeletal models can help to identify those single muscles differing in running with LEAF compared to FOAM [9]. In case that LEAF affects the energy contribution of hip, knee and ankle joint it seems plausible that also lower limb muscle forces are affected by the midsole design. Because muscles consume metabolic energy [15], these differences in muscle forces could be relevant for explaining differences in running economy between LEAF and FOAM [7]. Within lower limb muscles the m. biceps femoris was identified to have the greatest impact on running economy [16].

Based on these considerations it was hypothesized that running with LEAF compared to FOAM shoes in overground running (1) increases the anterior foot shift, (2) alters the energy contributions of hip, knee and ankle joint, and (3) reduces the lower limb muscle forces of selected hip, knee and ankle joint muscles.

## Materials and methods

### Participants

Nine male, non-professional, long-distance runners (mean  $\pm$  SD: age  $32.9 \pm 6.1$  y, height  $1.78 \pm 0.04$  m, mass  $75.7 \pm 5.6$  kg, leg length  $0.94 \pm 0.03$  m) volunteered to participate in the study. All participants had previously completed the treadmill study of Wunsch et al. [7], and were familiarized in running with LEAF and FOAM footwear. All participants were heel strikers with a foot-ground angle at touch-down of at least 10 degrees, which was checked beforehand as an inclusion criterion [7]. Prior to the measurements, the participants were informed about the potential risks and discomforts and completed a written informed consent document. The study was conducted in accordance with the Declaration of Helsinki and approved by the ethics committee of the University (Approval: 13BM-12).



**Fig 1. Midsole designs.** (a) Leaf spring structured midsole shoe (LEAF). (b) Standard foam midsole shoe (FOAM). The shoes are shown with reflective markers attached.

doi:10.1371/journal.pone.0172287.g001

## General overview

After a 10-min individual warm-up, the participants ran on a 40-m indoor track with a force plate imbedded at 30-m of the runway, using both LEAF (size US 9) and FOAM (size US 9) in randomized order. Both shoes featured similar geometrical characteristics and had a mass of 327 g (LEAF) and 338 g (FOAM). To ensure consistency with previous work on treadmill running [7], a constant speed of 3 m/s was chosen. The running speed was constrained by an acoustic pacemaker (signal every 5 m) and was controlled by photocells positioned 2.5 m before and 2.5 m after the center of the force platform. The participants completed four trials with each shoe, ensuring a full foot contact on the force platform and maintaining a speed range of  $3.0 \pm 0.2$  m/s. A period of 5 min between the shoe conditions was provided for changing the shoes.

## Instruments and calculations

Reflective markers (diameter of 15 mm) were attached to the participants according to the Cleveland Clinic Marker set (Motion Analysis Corp, Santa Rosa, USA). Kinematic and kinetic data were collected simultaneously by an eight camera three-dimensional motion analysis system (200 Hz, Vicon MX 1.3, Oxford Metrics Ltd, UK) and a force plate (1000 Hz, AMTI, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA). Kinematic and kinetic data were processed using Vicon Nexus 1.7.1 software (Vicon, Oxford Metrics Ltd, UK). Further calculations were completed using MATLAB (R2013a) and an inverse dynamic musculoskeletal modelling software (AnyBody 6.0, AnyBody Technology A/S, Aalborg, Denmark).

Raw data were filtered using a sixth order zero-lag Butterworth low pass filter with a cut-off frequency of 12 Hz for the kinematic data and 50 Hz for the kinetic data [17]. The time events of 'heel-strike' (HS) and 'toe off' (TO) used to define the stance phase were determined by a 10-N threshold applied to the vertical ground reaction force [17]. 'Heel off' (HO) was defined as the event when the vertical velocity of the heel marker changed from negative to positive [7].

The anterior shift of the foot was calculated as the displacement of the heel marker in the anterior-posterior direction from HS to HO [7]. Joint moments, joint angles and muscle forces were calculated using the musculoskeletal model (AMMR 1.6.2, MoCapModel) available in the AnyBody Modeling System. This model includes the Twente Lower Extremity Model (TLEM), which is based on the morphological data set for the lower extremities by Klein Horsmann et al. [18]. The data set was used to model mass, moments of inertia, and muscle sites/geometry for all segments. The model consisted of 11 body segments: head, trunk, pelvis, right and left femur, patella, tibia, talus, and foot. Each leg contained 55 muscles and mechanical effects

were modelled by 159 simple muscle slips [18]. The model was scaled to match each participant's anthropometry [19]. Inverse dynamics were performed and a third order polynomial muscle recruitment criterion was used [10, 11].

Joint power ( $P_j$ ) for the joint  $j$  (hip, knee and ankle joint) was calculated using the equation:

$$P_j = M_j \cdot \omega_j \quad (1)$$

with  $M_j$  as internal joint moment and  $\omega_j$  as joint angular velocity retrieved from the model.

The amount of generated and absorbed energy at the hip, knee and ankle joint was calculated using trapezoidal numerical integration of the power-time curve. For each trial, the absolute values of all power and energy data were normalized to body mass.

Based on the model, lower limb muscle forces during stance phase were determined for: m. gluteus maximus (GM), m. biceps femoris (BF), m. rectus femoris (RF), m. vastus medialis (VM), m. vastus lateralis (VL), m. gastrocnemius medialis (GM), m. gastrocnemius lateralis (GL), m. soleus (SO) and m. tibialis anterior (TA). For each trial, all muscle forces were normalized to body weight (BW).

## Model validity and influence of shoe marker placement

The TLEM model was validated during walking and showed a good agreement between the estimated muscle forces and measured EMG data [20]. Furthermore, the TLEM model was previously used to analyze sprint running [12]. No studies were found using this model for analyzing midsole designs in running shoes. The calculated time courses of hip, knee and ankle power in this study, however, corresponded well with those from previous studies [21, 22] and the analyzed lower limb muscle forces showed that almost all muscles acted similarly to patterns reported the literature [11, 23].

Except the shoe marker (Fig 1) no marker were replaced between the shoe conditions. While both shoes featured similar geometrical characteristics and specific care was taken to place the markers at the identical position, a perfect match cannot be guaranteed. Therefore, a sensitivity analysis on marker placement was conducted. For one typical trial on one participant, the shoe markers were mathematically repositioned within a range of 7 mm, which represents alterations clearly above the assumed error of marker displacement by the investigator. To determine the effect of these repositions on joint energy and muscles forces the modified trials were compared with the original trial. The differences calculated for each joint were: 1) energy absorption: < 3% at the hip, < 1% at the knee and < 1% at the ankle joint; 2) energy generation: < 3% at the hip, < 1% at the knee and < 3% at the ankle joint; 3) mean muscle forces: < 1% for GM, < 3% for BF, < 2% for RF, < 1% for VM, < 1% for VL, < 2% for GM, < 2% for GL, < 1% for SO and < 4% for TA.

## Statistics

The course of the joint power data and muscle forces of all four trials for each participant and shoe condition were time normalized over the entire stance phase. Ensemble mean curves were calculated for each participant and shoe condition. Additionally, group means along the stance phase were calculated and presented as mean  $\pm$  standard error curve.

For analysis, mean ( $\pm$  standard error) values were reported for: anterior foot shift, peak power absorption/generation and energy absorption/generation at the hip, knee and ankle joint, average muscle force during stance for each analyzed muscle. For each variable normal distribution was confirmed by the Shapiro Wilk test. A paired sample t-test was applied for analyzing a shoe difference for the variable anterior foot shift. With respect to the hypotheses of this study, for the remaining variables functional groups were built, more specifically 'hip

joint power and energy' (G1), 'knee joint power and energy' (G2), 'ankle joint power and energy' (G3), 'hip joint muscles' including GM, BF, RF (G4), 'knee joint muscles' including RF, VM, VL, GM, GL (G5) and 'ankle joint muscles' including GM, GL, SO, TA (G6). For each of these six groups (G1-G6), separate MANOVAs were calculated. In case of significance, a univariate ANOVA was performed. The level of significance for the univariate tests was Bonferroni-adjusted according to the number of variables in each group. Cohen's  $d_z$  was used to describe the effect size and practical relevance of differences. Effect sizes for each comparison were described as: small for  $d_z$  between 0.20 and 0.49, medium for  $d_z$  between 0.50 and 0.79 and large for  $d_z > 0.80$  [24]. Level of significance was set at  $\alpha < 0.05$ . The Statistical Package for the Social Sciences (Version 24.0; SPSS Inc., Chicago, IL, USA) was used.

## Results

The anterior shift of the foot was increased by  $8 \pm 1$  mm for LEAF compared to FOAM ( $p < 0.001$ ,  $d_z = 2.36$ ) (Table 1 and S1 File).

All of the calculated MANOVAs, except for G2, revealed significant multivariate main effects for the depended variable shoe (G1: Wilks'  $\lambda = 0.033$ ,  $F = 36.89$ ,  $p = 0.001$ , G2: Wilks'  $\lambda = 0.419$ ,  $F = 1.74$ ,  $p = 0.278$ , G3: Wilks'  $\lambda = 0.031$ ,  $F = 38.94$ ,  $p = 0.001$ , G4: Wilks'  $\lambda = 0.224$ ,  $F = 6.94$ ,  $p = 0.022$ , G5: Wilks'  $\lambda = 0.028$ ,  $F = 17.61$ ,  $p = 0.019$ , G6: Wilks'  $\lambda = 0.025$ ,  $F = 48.23$ ,  $p < 0.001$ ). Detailed information is presented in Table 1.

The time profiles of the joint power for the hip, knee and ankle joint are presented in Fig 2. The differences between LEAF and FOAM occurred at the hip joint primarily during the braking phase and at the ankle joint predominantly during the push-off phase. At the hip, a reduction of 32% in the peak power absorption ( $p < 0.001$ ,  $d_z = 2.50$ ) and 11% in the energy absorption ( $p = 0.010$ ,  $d_z = 1.11$ ) was found for LEAF compared with FOAM, whereas at the ankle joint, a reduction of 17% in peak power generation ( $p < 0.001$ ,  $d_z = 2.08$ ) and 13% in the energy generation ( $p < 0.001$ ,  $d_z = 4.29$ ) occurred (Table 1 and S1 File).

The trajectories of the muscle forces during stance phase of BF, GL, GM and SO are presented in Fig 3 indicating that both average and peak muscle forces were affected by the midsole design. The average muscle forces (Table 1) revealed a reduction for LEAF compared to FOAM in GL (15%,  $p < 0.001$ ,  $d_z = 2.16$ ), GM (9%,  $p = 0.009$ ,  $d_z = 1.15$ ), SO (8%,  $p = 0.001$ ,  $d_z = 1.69$ ) and a trend towards reduction was found for BF (12%,  $p = 0.031$ ,  $d_z = 0.87$ ).

## Discussion

This study investigated the effects of LEAF compared to FOAM shoes on anterior foot shift, joint energy and lower limb muscle forces in overground running at a constant speed of 3 m/s. The main findings were that the participants responded to the structured midsole design showing (1) an increased anterior foot shift (hypothesis accepted), (2) a reduced energy absorption at the hip and energy generation at the ankle joint (hypothesis partly rejected) and (3) reduced lower limb muscle forces, particularly for three muscles around the ankle joint (GL, GM, SO) and a trend toward a reduction for BF (hypothesis partly rejected).

The increase in anterior foot shift in running with LEAF compared to FOAM was similar to treadmill running ( $8 \pm 1$  mm [7]). This shows that, on the one hand, the participants responded to LEAF in overground running similarly as in treadmill running. On the other hand, this indicates that the mechanical behavior of the midsole deformation during ground contact in treadmill and in overground running are similar.

The effect of this response on the energy demand in running was determined within a first step using a joint level approach. This approach was used to identify the sources and magnitudes of mechanical joint power and the contribution of energy absorption and generation to

**Table 1.**

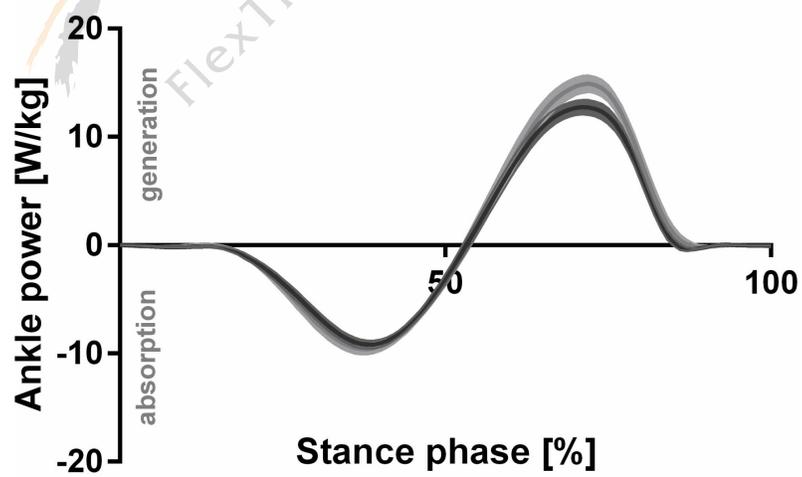
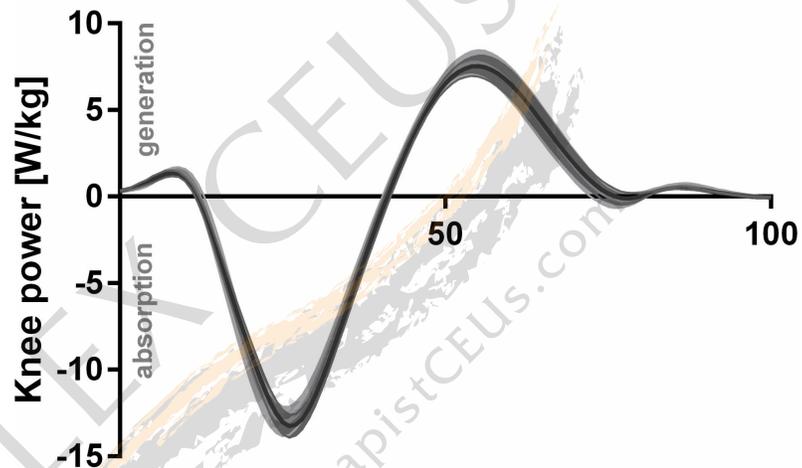
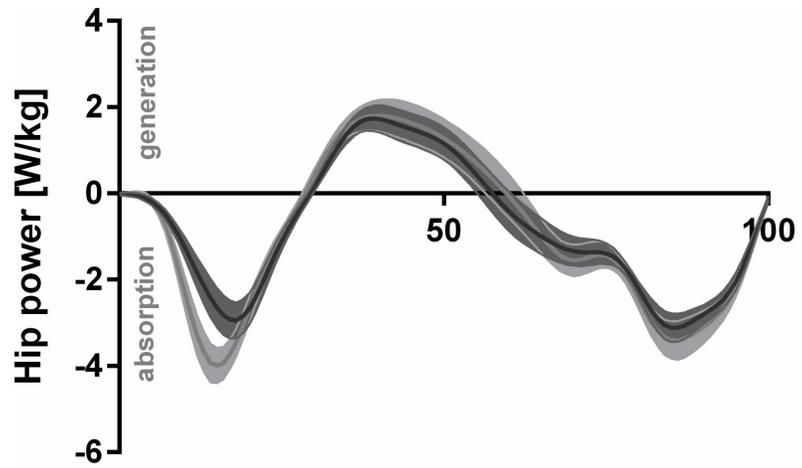
	LEAF	FOAM	diff (LEAF-FOAM)	P	d <sub>z</sub>
Anterior foot shift [mm]	20 ± 1	12 ± 1	8 ± 1	<0.001	2.36
<i>Hip joint (G1)</i>					
<i>Adjusted level of significance: α &lt; 0.0125</i>					
Peak power absorption [W/kg]	3.25 ± 0.45	4.28 ± 0.45	-1.03 ± 0.14	<0.001*	2.50
Peak power generation [W/kg]	2.01 ± 0.29	2.06 ± 0.42	-0.05 ± 0.23	0.853	0.06
Energy absorption [J/kg]	0.33 ± 0.02	0.37 ± 0.02	-0.04 ± 0.01	0.010*	1.11
Energy generation [J/kg]	0.10 ± 0.02	0.11 ± 0.03	-0.01 ± 0.01	0.500	0.24
<i>Knee joint (G2; MANOVA not sig.)</i>					
Peak power absorption [W/kg]	13.68 ± 0.65	13.36 ± 0.81	0.32 ± 0.22		
Peak power generation [W/kg]	7.87 ± 0.61	8.13 ± 0.70	-0.26 ± 0.16		
Energy absorption [J/kg]	0.62 ± 0.03	0.61 ± 0.04	0.01 ± 0.01		
Energy generation [J/kg]	0.48 ± 0.04	0.50 ± 0.05	-0.02 ± 0.01		
<i>Ankle joint (G3)</i>					
<i>Adjusted level of significance: α &lt; 0.0125</i>					
Peak power absorption [W/kg]	9.40 ± 0.38	9.76 ± 0.51	-0.36 ± 0.38	0.364	0.32
Peak power generation [W/kg]	13.08 ± 0.66	15.27 ± 0.81	-2.19 ± 0.35	<0.001*	2.08
Energy absorption [J/kg]	0.56 ± 0.02	0.57 ± 0.03	-0.01 ± 0.02	0.581	0.19
Energy generation [J/kg]	0.71 ± 0.03	0.81 ± 0.03	-0.10 ± 0.01	<0.001*	4.29
<i>Mean muscle force during stance</i>					
<i>Hip joint muscles (G4)</i>					
<i>Adjusted level of significance: α &lt; 0.0125</i>					
M. gluteus maximus [× BW]	0.18 ± 0.01	0.19 ± 0.01	-0.01 ± 0.00	0.248	0.41
M. biceps femoris [× BW]	0.25 ± 0.02	0.28 ± 0.02	-0.03 ± 0.01	0.031 <sup>t</sup>	0.87
M. rectus femoris [× BW]	0.43 ± 0.03	0.43 ± 0.04	0.00 ± 0.01	0.953	0.02
<i>Knee joint muscles (G5)</i>					
<i>Adjusted level of significance: α &lt; 0.010</i>					
M. rectus femoris [× BW]	0.43 ± 0.03	0.43 ± 0.04	0.00 ± 0.01	0.953	0.02
M. vastus medialis [× BW]	0.59 ± 0.04	0.61 ± 0.05	-0.02 ± 0.01	0.090	0.64
M. vastus lateralis [× BW]	1.28 ± 0.09	1.32 ± 0.10	-0.04 ± 0.02	0.085	0.66
M. gastrocnemius medialis [× BW]	0.61 ± 0.04	0.67 ± 0.04	-0.06 ± 0.02	0.009*	1.15
M. gastrocnemius lateralis [× BW]	0.24 ± 0.01	0.28 ± 0.01	-0.04 ± 0.01	<0.001*	2.16
<i>Ankle joint muscles (G6)</i>					
<i>Adjusted level of significance: α &lt; 0.0125</i>					
M. gastrocnemius medialis [× BW]	0.61 ± 0.04	0.67 ± 0.04	-0.06 ± 0.02	0.009*	1.15
M. gastrocnemius lateralis [× BW]	0.24 ± 0.01	0.28 ± 0.01	-0.04 ± 0.01	<0.001*	2.16
M. soleus [× BW]	2.79 ± 0.11	3.01 ± 0.13	-0.22 ± 0.04	0.001*	1.69
M. tibialis anterior [× BW]	0.11 ± 0.01	0.11 ± 0.01	0.00 ± 0.00	0.242	0.42

\* significant difference

<sup>t</sup> trend

doi:10.1371/journal.pone.0172287.t001

the total energy needed for performing the running task [9, 21]. Differences between LEAF and FOAM were found at the hip and the ankle joint. While running with LEAF, the reduction of hip joint energy was primarily found in the first half of the braking phase, the reduction of the ankle joint energy occurred in the second half of stance indicating that also the push-off phase was affected by the midsole design (Fig 2). As hypothesized, the reduced energy loss during the braking phase seemed to be derived from the rearfoot leaf springs leading to an anterior shift by 8 mm and a reduction of the horizontal braking force on the centre of mass [7, 25]. During the push-off phase, the short forefoot leaf springs contact ground and no additional anterior foot shift can be observed [7]. Thus, the reduced energy loss during the push-off phase seems to be also related to the energy saving during the braking phase. It can be



— LEAF  
— FOAM

**Fig 2. Joint power.** Joint power trajectories during stance (mean  $\pm$  standard error) for the hip, knee and ankle joint in running with LEAF and FOAM shoes.

doi:10.1371/journal.pone.0172287.g002

concluded from the reduced horizontal braking of the centre of mass that less energy is needed during the push-off phase to accelerate the centre of mass for maintaining the constant running speed. Therefore, the midsole design of LEAF appeared to successfully exploit the concept of minimizing energy loss during running [1, 2].

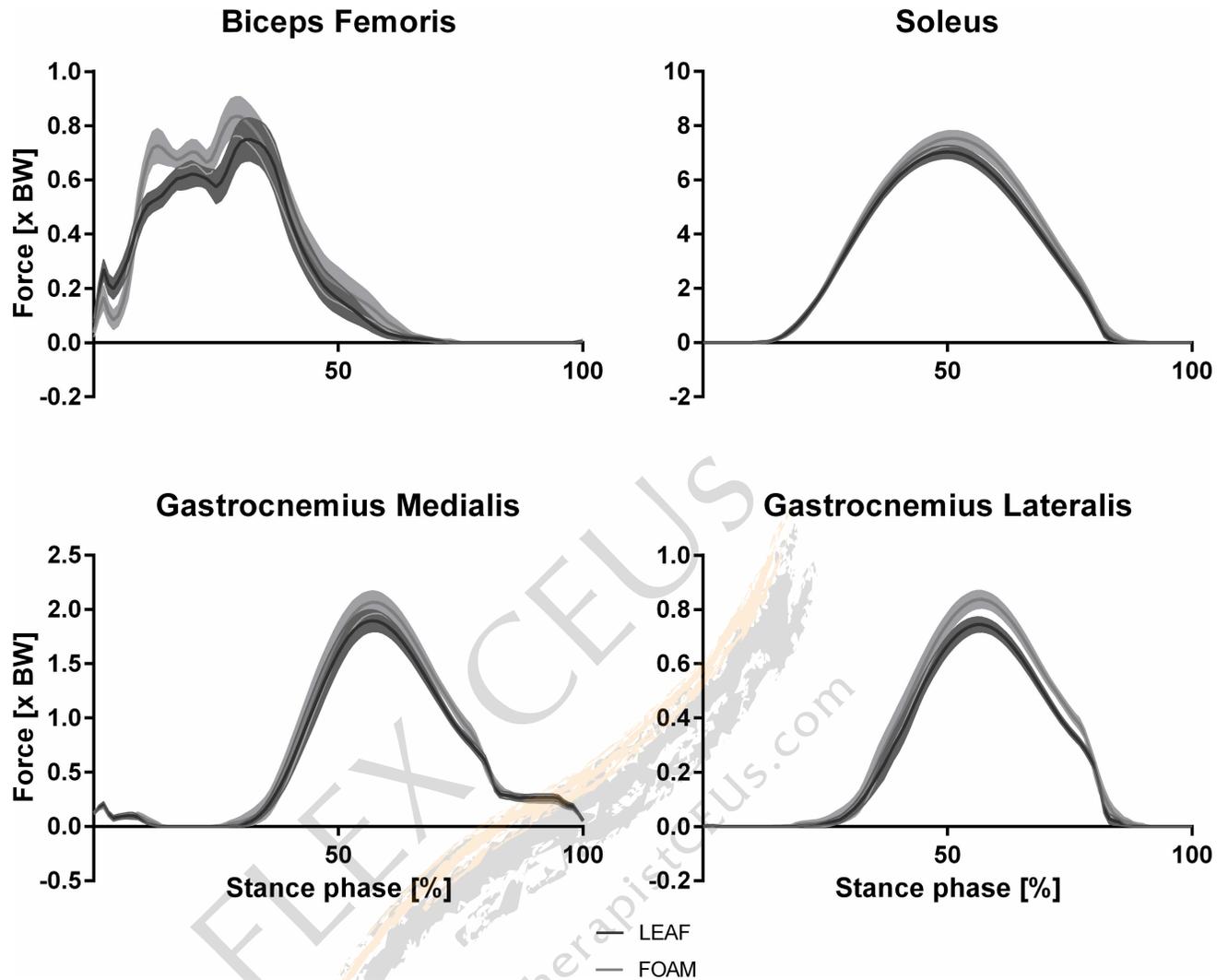
The differences in lower limb muscle forces between LEAF and FOAM indicate which muscles were adjusted by the locomotor system in response to the midsole design for generating the movement output. Significant reductions of muscle forces comparing LEAF and FOAM occurred in GL (15%), GM (9%) and SO (8%), which accounted for the reduced energy generation at the ankle joint. Furthermore, a trend towards a reduction of lower limb muscle forces between LEAF and FOAM was found in BF (12%), explaining the reduced energy absorption at the hip joint. Previous studies showed that participants adapted the activity of lower limb muscles in response to the midsole stiffness and to varying midsole wedges [26, 27]. Based on the current study, it can be concluded that also structural changes of the midsole have the potential to affect the lower limb muscle mechanics.

In general, muscles consume metabolic energy to generate joint energy [15]. Thus, the observed reductions and trends in muscle forces can affect the total metabolic energy expenditure in running with LEAF compared to FOAM. According to Kyrolainen et al. [16] the BF provides the greatest impact on economy and is one of the largest extensor muscles at the hip, which consumes a considerable amount of metabolic energy when active due to its high muscle mass [22, 28]. Therefore, it is favorable to conserve metabolic energy by reducing the hip extensor muscle activity [22], which was demonstrated in this study by the trend towards reductions of muscle forces.

The muscles contributing primarily to the horizontal braking and propulsion of the center of mass during stance have a substantial impact on running economy [16]. During the braking phase, the quadriceps muscle group is the largest contributor to deceleration of the horizontal motion of the center of mass [21, 29] and during the propulsion phase the triceps surae muscle group is the greatest contributor to forward acceleration of the centre of mass [23, 29]. In running with LEAF compared to FOAM shoes, RF, VL and VM showed no differences in muscle forces during the braking phase. The muscle force of the triceps surae group (GL, GM and SO; Fig 3), however, was reduced by 11% when running with LEAF compared with FOAM. Consequently, this group produced lower muscle forces for propulsion in running at constant speed.

For the same participants with the same test shoes (LEAF and FOAM) at the same running speed (3 m/s), a reduction of oxygen consumption of 2% with LEAF shoe was found in treadmill running [7]. Hence, the positive effect of LEAF on oxygen consumption in this study can be at least partly explained by the observed reductions of the lower limb muscle forces combined with the reductions of energy absorption and generation at the hip and ankle joint. This is supported by the consideration that for marathon running, a total physiological energy consumption during one foot contact is estimated to be 6.61 J/kg [1]. The sum of measured reductions for LEAF compared to FOAM shoes in absorbed energy at the hip (0.04 J/kg) and generated energy at the ankle (0.11 J/kg) was 0.15 J/kg. This represents 2% of the total energy required and presents a similar reduction of oxygen consumption [7]. It has to be noted, however, that further relevant energy forms, e.g. thermodynamic, chemical, etc. [1] were not taken into account.

Finally, the gained results could provide an important background concerning injury prevention. Muscles have been shown to be the major contributors to the joint contact forces [30–



**Fig 3. Muscle forces.** Muscle force trajectories during stance (mean  $\pm$  standard error) for m. biceps femoris (BF), m. gastrocnemius medialis (GM), m. gastrocnemius lateralis (GL) and m. soleus (SO).

doi:10.1371/journal.pone.0172287.g003

[32]. Thus, reduced muscle forces could lead to reduced joint load. Sinclair [33] has shown that the peak Achilles tendon force was significantly reduced using LEAF compared to conventional footwear. These results are in line with the reduced triceps surae muscle forces found in the current study.

One limitation of this study was the small sample size. It should be noted, however, that the presented results are well in line with previous studies analyzing the same test shoes [7, 34]. Thus, it can be concluded that for runners responding to LEAF the results found in this study are plausible. The second limitation was the slightly reduced shoe mass of 11 g for LEAF compared to FOAM. This explains approximately 0.11% of the differences in oxygen consumption [35] and may also contribute to a small extent to the observed changes in joint mechanics and muscle forces. Furthermore, this and the previous studies comparing LEAF with FOAM shoes only investigated the midsole effects in non-fatigued running. From a practical perspective it would be interesting to examine the joint power joint energy and leg muscle forces under fatigued conditions [36].

## Conclusion

Knowledge of the energy generation and absorption of the lower limb joints and the biomechanical function of the lower limb muscles is important for improving the understanding of performance in running with different running shoes. Hereby musculoskeletal models could provide valuable information for the sport shoe research. This study showed that structural changes of the midsole have the potential to affect joint energy and lower limb muscle forces in running. Furthermore, these findings are consistent with previous studies indicating that running with LEAF compared to FOAM enhances running economy [7]. Thus, the structured midsole shoe seems to be suitable for heel strike runners and may enhance running performance by reducing lower extremity joint energy due to force reductions in lower extremity muscles.



# Increased Vertical Impact Forces and Altered Running Mechanics with Softer Midsole Shoes

## Abstract

To date it has been thought that shoe midsole hardness does not affect vertical impact peak forces during running. This conclusion is based partially on results from experimental data using homogeneous samples of participants that found no difference in vertical impact peaks when running in shoes with different midsole properties. However, it is currently unknown how apparent joint stiffness is affected by shoe midsole hardness. An increase in apparent joint stiffness could result in a harder landing, which should result in increased vertical impact peaks during running. The purpose of this study was to quantify the effect of shoe midsole hardness on apparent ankle and knee joint stiffness and the associated vertical ground reaction force for age and sex subgroups during heel-toe running. 93 runners (male and female) aged 16-75 years ran at  $3.33 \pm 0.15$  m/s on a 30 m-long runway with soft, medium and hard midsole shoes. The vertical impact peak increased as the shoe midsole hardness decreased (mean(SE); soft: 1.70BW(0.03), medium: 1.64BW(0.03), hard: 1.54BW(0.03)). Similar results were found for the apparent ankle joint stiffness where apparent stiffness increased as the shoe midsole hardness decreased (soft:  $2.08\text{BWm}^{\circ} \times 100$  (0.05), medium:  $1.92\text{BWm}^{\circ} \times 100$  (0.05), hard:  $1.85\text{BWm}^{\circ} \times 100$  (0.05)). Apparent knee joint stiffness increased for soft ( $1.06\text{BWm}^{\circ} \times 100$  (0.04)) midsole compared to the medium ( $0.95\text{BWm}^{\circ} \times 100$  (0.04)) and hard ( $0.96\text{BWm}^{\circ} \times 100$  (0.04)) midsoles for female participants. The results from this study confirm that shoe midsole hardness can have an effect on vertical impact force peaks and that this may be connected to the hardness of the landing. The results from this study may provide useful information regarding the development of cushioning guidelines for running shoes.

## Introduction

Impact forces during heel-toe running have been discussed in the scientific literature for many years. Some have argued that increased impact forces are associated with the development of specific running injuries [1-4]. As a result, the development of shoe cushioning

guidelines evolved as repetitive loading became a concern for injury risk during running. Several strategies have been proposed to reduce impact loading. One of the most popular approaches was to change the hardness of the shoe midsole [5–8]. However, this strategy was associated with the surprising result that in many studies, shoe midsole hardness had little to no effect on impact force peaks during landing [5–7]. This result is surprising because it was assumed that one could easily reduce the impact force peaks with soft shoe midsoles. However, the results of many initial studies have shown that this is not the case [5–7]. It may be that the results have been influenced by typically homogeneous test subject groups, which were mostly young sporty male university students. For this well-trained population, shoe midsole hardness may not be enough to influence running style. It may be that the result would be different for less trained or for elderly subjects [5–7]. However, the influence of shoe midsole hardness on impact loading during over ground running across different age and sex subgroups has yet to be examined.

Another possible reason for the lack of change in impact forces found with midsole hardness may be that subjects change something else during the initial ground contact in a softer shoe. There are different candidates for this change. One is to change the landing velocity when changing the hardness of the midsole. Another is to change the stride length [9–11]. However, it has been shown that there is no support for these two possible interpretations [12,13]. Another possibility is that subjects adjust their landing mechanics when running in different midsole shoes, changing their effective mass during landing by making their ankle and/or knee joint more stiff when the shoe sole becomes softer [14–16]. A stiffer limb would result in a “harder” landing (higher impact forces), counteracting any influence of the shoe cushioning on impact absorption. This has been supported by previous studies that have demonstrated increased leg or joint stiffness when running or hopping on a softer surface or a shoe with a thicker midsole [13, 16–19]. One method to quantify the hardness of a landing is to determine the relationship between the landing kinematics and kinetics, or the apparent stiffness of the joint [15, 20, 21]. This will provide an indication of how much joint displacement occurs for a given external force (moment) at the joint. The influence of shoe midsole cushioning on landing mechanics and the hardness of the landing may differ across age and gender subgroups. Landing mechanics can be influenced by a variety of factors, including lower extremity kinematics, the active forces of the muscles and the external forces applied to the body. Previous studies have demonstrated that age, sex, and footwear influence these factors [22–25]. Therefore, it may be speculated that the influence of shoe midsole hardness on landing mechanics may change across gender and sex subgroups.

While landing kinematics and kinetics have been investigated independently when running in shoes with different midsole properties, the relationship between the two has yet to be quantified. Understanding how shoe midsole hardness influences ankle and knee joint apparent stiffness and the resulting vertical impact peaks during heel-toe running and whether or not this influence differs across age and sex subgroups will provide useful information for the understanding of the landing during heel-toe running. This understanding would provide a small but significant addition to the understanding of the control strategies of the human system. The human system adapts as the boundary condition (e.g. footwear) changes. Investigating the different adaptation processes across a wide age range could shed some light onto the control strategies of the human system. Understanding these landing mechanisms due to shoe properties may allow for the construction of footwear that ultimately optimizes the running pattern or reduces the incidence of injuries.

Therefore, the purpose of this study was to characterize the effect of shoe midsole hardness on the apparent joint stiffness at the ankle and the knee and the resulting vertical ground reaction force during heel-toe running for different age and sex subgroups.

Based on previous results for differences in kinematics and muscular strength across the tested groups [22–25], it was hypothesized that:

- (H1). Apparent ankle joint stiffness increases as shoe midsole hardness decreases for all age and sex groups.
- (H2). Apparent knee joint stiffness will not be different between the shoe conditions.
- (H3). Vertical impact force peaks will not be different between shoe conditions.

## Methods

### Subjects

Ninety-three recreational runners (47 male, 46 female) who ran at least 30 minutes per week participated in this study (Table 1). All participants provided written informed consent in accordance with the University of Calgary’s policy on research using human subjects and approval for this research project was obtained from the University of Calgary’s Conjoint Health Research Ethics Board. Written informed consent was provided by the legal guardian of the minors that participated in this study. All subjects were free from injury or pain at the time of testing. Four age groups ranging across the lifespan development stages were defined as follows: Group 1 (G1, adolescence) 16–20 years; Group 2 (G2, early adulthood) 21–35 years; Group 3 (G3, middle age) 36–60 years; and Group 4 (G4, older age) 61–75 years.

### Experimental setup

Three different shoe conditions provided by Decathlon (now Oxyane Group, France) that differed only in their midsole hardness were investigated: Asker C-40 (Soft), Asker C-52 (Medium) and Asker C-65 (Hard). The shoes were identical with respect to all other footwear properties besides the midsole hardness. Kinematic data were collected using 12 retro-reflective markers mounted on the pelvis and right lower extremity (Fig 1) to measure three-dimensional movements of each segment using an eight-camera, 240 Hz motion capture system (Motion Analysis, CA). Kinetic data was collected simultaneously using a Kistler force plate at a sampling frequency of 2400 Hz (Kistler Instruments AG, Winterthur, Switzerland) embedded within the laboratory floor. A static trial was taken with markers placed over the right greater trochanter, medial and lateral knee joint axis, and medial and lateral malleoli in order to define the joint centers. Position

**Table 1. Subject characteristics.**

	Age Group	Number of Subjects	Age		Height		Mass	
			Average	SEM	Average	SEM	Average	SEM
Male	16–20	13	17.9	0.5	178.0	1.6	69.6	2.5
	21–35	13	25.5	1.1	179.7	2.0	74.0	2.0
	36–60	11	48.5	1.8	175.3	1.3	77.5	2.0
	61–75	10	66.9	1.5	174.5	1.8	74.6	3.0
Female	16–20	12	18.1	0.4	162.9	2.0	55.4	1.7
	21–35	12	26.2	0.9	166.8	2.4	62.5	2.1
	36–60	11	49.6	1.5	165.2	1.4	64.5	2.6
	61–75	11	65.5	1.5	162.0	1.6	55.6	1.5

Subject characteristics for each sex and age subgroup used in this study.



**Fig 1. Marker Set Up.** Frontal view of retro-reflective marker placement on the shoe, shank, thigh and pelvis.

doi:10.1371/journal.pone.0125196.g001

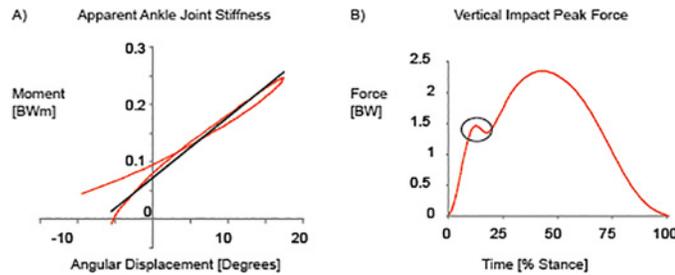
data was collected for a static neutral trial for each of the shoe-subject conditions in order to define the segment coordinate system. Joint center markers were removed for the running trials.

Subjects performed five successful heel-toe running trials ( $3.33 \pm 0.15$  m/s) for each of the three different shoe conditions on a 30 m running lane in the lab. The order in which the shoes were tested was randomly selected for each subject. Subjects were allotted at least three practice trials for familiarization prior to data collection for each of the shoe conditions.

## Data analysis

Markers were identified and their three-dimensional coordinates were tracked using EVaRT Real Time software (Version 5.0.4, Motion Analysis, CA, USA). Data were filtered using a low pass fourth order Butterworth filter with a cutoff frequency of 12 Hz for the kinematic data and 100 Hz for the kinetic data. Joint angular displacements and resultant moments were calculated using the kinematic and ground reaction force data (Kintrak, HPL, University of Calgary). Specifically, ankle dorsiflexion/plantar flexion and knee flexion/extension angular displacement and resultant moments were calculated. All variables were clipped to the stance phase of the step with the right foot on the force plate with heel contact and toe-off determined using a 15 N threshold in the vertical ground reaction force.

The average apparent joint stiffness in the sagittal plane during the loading phase of stance for the ankle and knee joint were defined as the ratio of the change in joint moment ( $\Delta M$ ) to the joint angular displacement ( $\Delta\theta$ ) [18]. For the ankle, this corresponded to the change in ankle dorsiflexion/plantar flexion moment to the change in dorsiflexion/plantar flexion angular displacement. For the knee joint, this corresponded to the change in knee flexion/extension



**Fig 2. Methods.** A) Illustration of apparent ankle joint stiffness (stiffness = slope), and B) vertical force during running highlighting the impact peak during the first part of stance.

doi:10.1371/journal.pone.0125196.g002

moment to the change in flexion/extension angular displacement. A least squares linear regression equation was used for the resultant joint-moment versus joint-angle curves for the loading portion of the stance phase (10%–50% of stance) and the slope of this line was identified as the apparent joint stiffness [20] (Fig 2A). The vertical impact force peak was calculated as the local maximum of the vertical ground reaction force during the first 50 ms following heel strike [26] (Fig 2B). A footwear condition was not included if an impact peak was not present in at least three of the five running trials.

A mixed model was used to perform a repeated measures ANCOVA (IBM SPSS Statistics 20.0, IL, USA) using a within subject factor of footwear (3 levels: soft, medium, and hard) and between subjects factors of age (4 levels: 16–20 years ( $n = 25$ ), 21–35 years ( $n = 25$ ), 36–60 years ( $n = 22$ ), 61–75 years ( $n = 21$ )) and sex (2 levels: male ( $n = 47$ ), female ( $n = 46$ )) for each of the three variables measured in this study in order to determine main effects and interaction effects. Height and weight were included as covariates. A pairwise comparison with a Bonferroni correction was then used if any significant main effects were found for footwear, age, or sex. All reported  $p$ -values are Bonferroni-corrected  $p$ -values and, thus, statistical significant was set at the  $\alpha = 0.05$  level.

## Results

Descriptive statistics for the variables tested in this study can be seen in Table 2. There was a significant main effect for footwear for the vertical impact peak ( $F = 54.877$ ,  $p < 0.001$ ). Post-hoc comparisons revealed that all footwear conditions were significantly different from each other (all  $p < 0.001$ ). The magnitude of the vertical impact peak increased as the shoe midsole hardness decreased with the soft midsole shoe having the largest vertical impact peak (mean (SE): 1.70BW (0.03)) followed by the medium midsole shoe (mean (SE): 1.64BW (0.03)) and finally the hard midsole shoe (mean (SE): 1.54BW (0.03)) (Fig 3).

There was a significant main effect for footwear for the apparent ankle joint stiffness ( $F = 55.409$ ,  $p < 0.001$ ). Post-hoc comparisons revealed that all footwear conditions were significantly different from each other (all  $p \leq 0.001$ ). Apparent ankle joint stiffness increased as shoe midsole hardness decreased. The average apparent stiffness during the loading phase of stance was highest for the soft midsole shoe (mean (SE): 2.08BWm/degrees  $\times$  100 (0.05)) followed by the medium midsole shoe (mean (SE): 1.92BWm/degrees  $\times$  100 (0.05)) and finally the hard midsole shoe (mean (SE): 1.85BWm/degrees  $\times$  100 (0.05)) ( $p < 0.001$ ) (Fig 4). There were no significant main effects for age or sex of the runner.

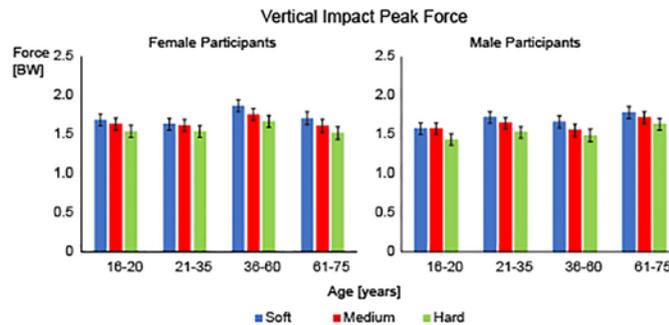
Apparent knee joint stiffness showed a sex-dependent effect due to the shoe condition. There was a significant shoe sex interaction ( $F = 6.336$ ,  $p = 0.002$ ). For female subjects, the apparent knee joint stiffness increased in the soft midsole shoe (mean (SE): 1.06BWm/degrees  $\times$

**Table 2. Statistics.** Descriptive statistics (mean and standard deviation (SE)) for the variables tested.

Outcome Measure	Shoe Condition	Mean (SE)	Shoe Effect (F,[p])	Sex Effect (F,[p])	Age Effect (F,[p])	Weight Effect (F,[p])	Height Effect (F,[p])	Shoe Sex Interaction (F,[p])	Shoe Age Interaction (F,[p])	Sex Age Interaction (F,[p])	Shoe Sex Age Interaction (F,[p])
Apparent Ankle Joint Stiffness (BWm <sup>o</sup> x 100)	Soft	2.08 (0.05)	55.409 [p<0.001]	0.174 [0.678]	0.545 [0.653]	0.957 [0.331]	4.067 [0.047]	2.155 [0.119]	1.060 [0.389]	1.454 [0.233]	1.142 [0.340]
	Medium	1.92 (0.05)									
	Hard	1.85 (0.05)									
Apparent Knee Joint Stiffness (BWm <sup>o</sup> x 100)	Soft	0.96 (0.02)	26.254 [p<0.001]	6.260 [0.014]	1.574 [0.202]	0.891 [0.348]	13.593 [p<0.001]	6.336 [0.002]	1.138 [0.342]	0.970 [0.411]	0.338 [0.916]
	Medium	0.89 (0.02)									
	Hard	0.91 (0.02)									
Vertical Impact Peak (BW)	Soft	1.70 (0.03)	54.877 [p<0.001]	0.329 [0.568]	0.690 [0.561]	0.215 [0.644]	1.656 [0.202]	0.103 [0.903]	0.743 [0.616]	1.867 [0.142]	0.613 [0.720]
	Medium	1.64 (0.03)									
	Hard	1.54 (0.03)									

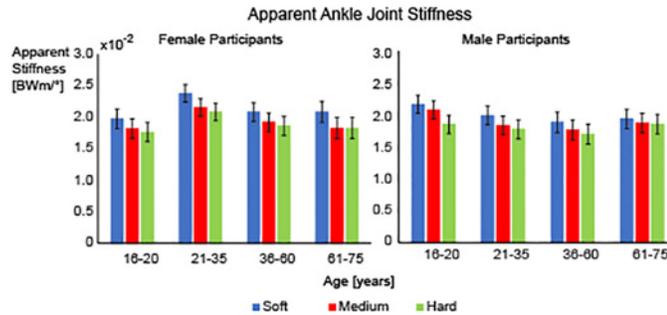
doi:10.1371/journal.pone.0125196.t002

100 (0.04)) compared to the medium midsole (mean (SE): 0.95BWm/degrees x 100 (0.04)) and the hard midsole shoe (mean (SE): 0.96BWm/degrees x 100 (0.04)) (both p<0.001). For male participants, there was a significant difference between the soft midsole shoe (mean (SE): 0.87BWm/degrees x 100 (0.04)) and the medium midsole shoe (mean (SE): 0.82BWm/degrees x 100 (0.04)) (p = 0.006) but there were no differences with respect to the hard midsole shoe (mean (SE): 0.85BWm/degrees x 100 (0.04)) (Fig 5).



**Fig 3. Vertical Impact Peak.** Average vertical impact peak force (mean ± SEM) for the soft (blue), medium (red) and hard (green) midsole shoes for the female participants (left) and male participants (right). Significant differences were found between soft, medium and hard midsole shoes. Covariates were evaluated at the following values: weight = 66.7kg, Height = 170.9cm.

doi:10.1371/journal.pone.0125196.g003



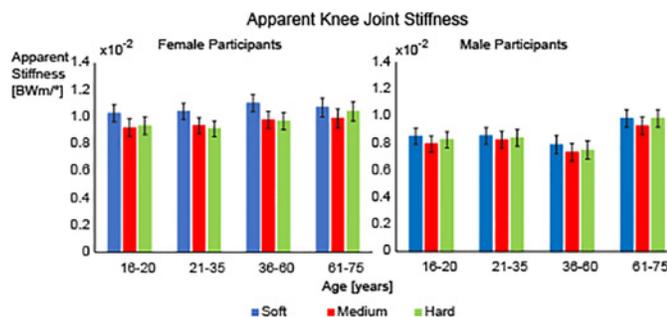
**Fig 4. Apparent Ankle Stiffness.** Average apparent ankle joint stiffness (mean  $\pm$  SEM) for the soft (blue), medium (red) and hard (green) midsole shoes for the female participants (left) and male participants (right). Significant differences were found between soft, medium and hard midsole shoes. Covariates were evaluated at the following values: weight = 66.7kg, Height = 170.9cm.

doi:10.1371/journal.pone.0125196.g004

## Discussion

The current study has two major results that are of specific interest with respect to impact forces. First, the study shows an effect of midsole hardness on vertical impact force peaks. Second, the study shows a potential connection between midsole hardness and apparent ankle/knee joint stiffness.

The influence of shoe midsole hardness on vertical impact peaks during running has been a question of high interest in running research [5–8]. Most studies report that there is no correlation between midsole hardness and vertical impact force peaks [6,7]. There is one study that indicates that the vertical impact force peaks increase for very soft and very hard midsole hardness [8]. However, these specific results have not been discussed in the literature. The results from the current study indicate that wearing soft midsole shoes can result in increased vertical impact force peaks. There are different possible interpretations of this result. First, it may be that the higher vertical impact force peaks are a result of “bottoming out” of the midsole. Second, it may be that the higher impact force peaks are a result of the increased apparent stiffness at the ankle and knee joints. Individuals in the present study adjusted their apparent joint stiffness when running in shoes with different midsole properties. Specifically, the majority of the tested individuals increased the apparent stiffness in their joints when running in softer shoes. The resulting increase in the vertical impact force peaks during landing could imply an increase of the loading on the tissues when running in softer midsole shoes.



**Fig 5. Apparent Knee Stiffness.** Average apparent knee joint stiffness (mean  $\pm$  SEM) for the soft (blue), medium (red) and hard (green) midsole shoes for the female participants (left) and male participants (right). Significant differences were found between the soft midsole and the medium and hard midsoles for the female participants and between the soft midsole and the medium midsole for the male participants. Covariates were evaluated at the following values: weight = 66.7kg, Height = 170.9cm.

doi:10.1371/journal.pone.0125196.g005

Apparent joint stiffness was sensitive to the shoe midsole hardness, specifically at the ankle joint. The apparent ankle joint stiffness increased as the shoe midsole hardness decreased (Fig 4) and the effects were systematic across sex and age groups. Similar findings have been found when comparing barefoot and shod running where ankle stiffness was higher in the softer cushioned footwear condition compared to barefoot running [19]. Overall leg stiffness has also been shown to increase during softer cushioned footwear conditions compared to barefoot running [16]. It has been speculated previously that the apparent stiffness of the lower extremity increases as the demand of the activity increases, as seen through increased apparent joint stiffness with increasing running velocity [20]. Thus, one could speculate that greater apparent joint stiffness may be required in order to control the joint movements in a softer midsole. Increased apparent joint stiffness could increase the decelerated mass at impact, which may explain the increase of the vertical impact peak for the soft midsole shoe condition.

The influence of shoe midsole hardness on apparent knee joint stiffness depended on the sex of the runner. Specifically, apparent knee joint stiffness was increased in the soft midsole condition compared to the medium and the hard midsole condition for the female participants. For male participants, there was a significant difference between the soft and the medium midsole shoe but not with the hard midsole shoe. For the soft and medium midsole shoes, the female participants also had more apparent knee joint stiffness than the male participants. For the hard midsole shoe, there was no difference between the males and females with respect to apparent knee joint stiffness. Thus, female participants appear to be more sensitive at the knee joint to the midsole hardness conditions used in this study. This may be related to other biomechanical and neuromuscular differences that have been found between males and females including muscle strength, proprioception, and joint laxity [27–30]. Females have been shown to have reduced joint position sense, lower hamstring to quad strength ratios, increased muscular co-contraction prior to landing and increased joint laxity [27–30]. It has been speculated that the altered muscle activity and increased co-contraction prior to landing in females may be compensatory mechanisms to improve joint stability due to their proprioceptive deficits and increased joint laxity [31–33]. It may be that the softer midsole shoe conditions used in this study were in a range that also required greater apparent knee joint stiffness as a mechanism for the female participants to increase joint stability and control the movements at the joint.

The results from the apparent ankle and knee joint stiffness also indicate that the ankle joint is more sensitive than the knee joint to changes in shoe midsole properties. Similar results have been found previously when examining the influence of shoe midsole hardness on lower extremity kinematics and kinetics. For example, Hardin and colleagues examined kinematic adaptations at the ankle, knee, and hip joint due to shoe midsole hardness and found that footwear only influenced movements at the ankle joint [13]. Similar results were also found when using a vector-based approach [34]. Von Tschärner and colleagues used an iterative support vector machine in order to identify kinematic differences due to footwear with different midsole properties. They found that differences due to footwear were located at the ankle joint and that there were no differences at the knee or hip joint due to footwear with different midsole properties. These kinematic results support the current findings that footwear appears to influence the distal ankle joint more strongly than the proximal knee joint.

Certain limitations exist for this study. The models used for apparent joint stiffness are rather simplistic and provide only an approximation of stiffness. Apparent joint stiffness has been termed as “quasi-stiffness” as it is not a true mechanical representation of stiffness [35]. However, this measure provides an indication of how much joint displacement occurs for a given external force (moment) at the joint, which in turn provides information about the hardness of the landing during over ground running. In addition, while the overall sample for shoe effects was large, the individual age and subgroup samples were still relatively small.

## Concluding remarks

Shoe midsole properties used in this study had a significant influence on the local measurement of apparent joint stiffness. Shoe midsole hardness affected the distal ankle joint more than the proximal knee joint, where differences depended on sex of the runner. The increase in apparent joint stiffness may have resulted in an increase in the decelerated mass at landing, effectively increasing the impact peak magnitude in the softer midsole conditions. This study provides experimental evidence that shoe midsole hardness can in fact affect vertical force impact peaks during running. Even more importantly, the results from this study showed that softer midsole shoes can actually increase external vertical force impact peaks. This contradicts the popular belief that softer midsole shoes should reduce impact peaks during running. The results from this study may provide useful information regarding the development of cushioning guidelines for running shoes. The shoe midsole properties used in this study encompass the midsole stiffness properties of shoes currently on the market. However, in order to develop more specific guidelines for midsole cushioning, a larger study would need to be conducted in which a variety of midsole stiffness levels are tested in order to determine if there is an ideal level of shoe midsole hardness.

The results from this study may also have important implications for the loads imposed on certain tissues of the leg and foot while running in different shoe midsoles. The studies concentrating on the effects of impact forces on injuries can be grouped into studies that claim that impact forces are associated with the development of specific running injuries [1–4], studies that claim that there is no connection between impact forces and injury development [36, 37] and into a few studies that claim that subjects with high impact forces or loading rates are less likely to be injured in heel-toe running [38–40]. All of these impact-related injury studies have the shortcoming that they typically use small sample sizes of subjects and, therefore, that the results are not conclusive. Future studies would need to investigate whether the changes in local measures of apparent joint stiffness due to shoe midsole hardness are related to either changes in performance or changes in the risk of injuries. This would allow for shoe manufacturers to develop cushioning guidelines based on injury prevention requirements.

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