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Guest Editors: *Anton Arndt, Wolfgang Potthast and Helen Woo*

Articles

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Experimental and computational analysis of orthotic medial longitudinal arch support height

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Keywords: foot; footwear; orthoses; arch support; plantar pressure; orthopaedic shoes

Introduction

Lower extremity injuries (LEI) in the foot and ankle region are the most common injuries experienced by physically active groups (including sports and military personnel). Classical treatment for LEI involves the fabrication of custom moulded or fitted foot orthoses. The arch height is a critical parameter for an orthosis at mid-stance. Typically, the standard procedure for fabricating a custom-made foot orthoses starts with using moulds or surface data captured in a neutral position of the foot based on Root's subtalar neutral theory (Root, 1973, 1997). More recent research (Laughton, Davis, & Williams, 2002), however, reported a significant difference in forefoot and rearfoot geometric measurements between different casting techniques (plaster cast, foam cast and laser scan), and arch height varied up to 2 cm among fabrication techniques.

A low arch height would lead to an orthosis with no functional orthotic benefits, and a high arch height would injure the subject instead of providing injury prevention or rehabilitation benefits. In all, an objective means of obtaining or forming orthotic shape has yet to be developed.

Purpose of the study

The purpose of the study is that we want to objectively determine the optimum range of orthotic arch height which controls navicular displacement and therefore the plantar pressure redistribution.

Methods

In this research a 3D finite element foot model is developed, including details of bones, soft tissue, ligament and cartilages. A series of virtual orthotic shape is obtained by capturing the plantar foot surface profile when the foot is subjected to a portion of body weight under balance standing loading. The obtained virtual orthotic shapes are then integrated with the foot model and treated as rigid. Vertical navicular displacement during the loading serves as an indicator for determining the optimum orthotic arch height.

An adjustable arch height orthoses is developed. A fiberglass/epoxy orthosis is fabricated through the vacuum bagging method. A longitudinal cut-out is made just lateral to the first ray of the foot. The adjustment of flexible orthosis height is achieved by adjusting the length of the arch span through a lead screw mechanism under the hallux. An experimental study is performed on a range of individuals by varying the orthotic arch height. Displacement of the navicular is measured during balance standing with the orthotic changes. Plantar pressure redistribution is measured during walking by Novel plantar pressure measurement system for a range of walking speeds.

Results

The computational model gives the navicular displacement during balance standing versus orthotic height relation as shown in Figure 1.

Experimentally, the mean pressure in forefoot, midfoot and rearfoot is evaluated and the mean mean pressure over the walking cycles is shown in Figure 2.

Discussion and conclusion

According to both the computation model and the experimental study, the navicular displacement and plantar pressure are sensitive to orthotic arch height. The navicular displaces vertically downward when the orthotic arch

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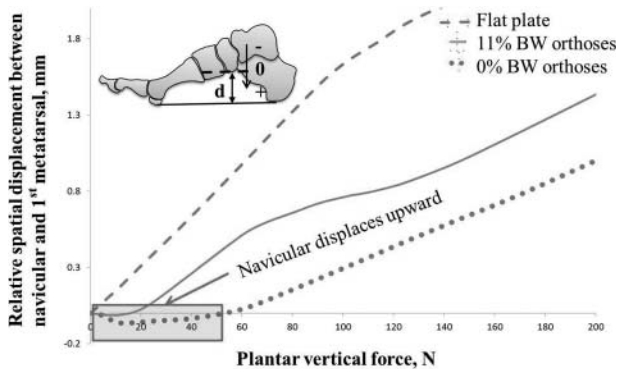

Figure 1. Navicular displacement during balance standing with varying orthotic height.

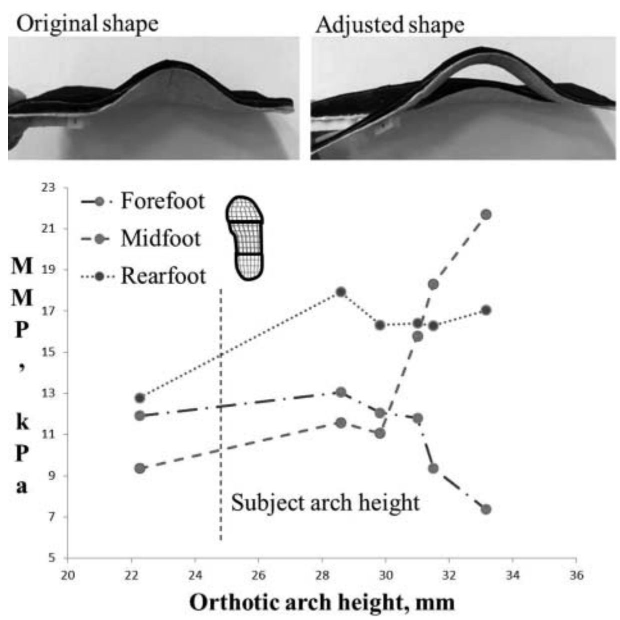

Figure 2. Mean mean pressure over all frames in forefoot, midfoot and rearfoot with varying orthotic arch height.

height is low compared with the subject's arch height. Alternatively, when the orthotic arch height is high, the navicular displaces vertically upward, causing discomfort. Significant pressure redistribution is observed as shown in Figure 2 from forefoot to midfoot as the arch height is increased. A 1.5 times increase in mean pressure in the midfoot is obtained for a 4 mm change in arch height, for example.

The findings and results of this study guide foot specialists to design orthoses with effective arch heights for skeletal foot control and therefore better treatment of the LEI. The developed adjustable medial longitudinal arch orthosis serves for orthosis arch height selection in two ways: (1) the subjects can select orthosis arch height based on subjective comfort ratings, and (2) physicians may select a control range for the navicular displacement, plantar pressure or other biomechanical parameters.

Disclosure statement

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Influence of shoe colour on perceived and actual jumping performance

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Keywords: basketball; footwear; jump; perception; ground reaction forces

Introduction

The effects of colour have been examined in multidisciplinary areas. Red has a subtle but important influence on psychological and basic motor functions, increasing one's motivation and altering one's perception of performance (Feltman & Elliot, 2011). In sports, previous research has shown that wearing red enhanced one's perception of relative dominance and threat, while viewing an opponent in red heightened the perception of the opponent's dominance and threat (Greenless, Leyland, Thelwell, & Filby, 2008; Greenless, Eynon, & Thelwell, 2013; Krenn, 2014).

In the sports context it appears that wearing red gains an advantage over comparable opponents, as observed in football, boxing, taekwondo, Greco-Roman wrestling and free-style wrestling (Attrill et al., 2008; Feltman &, Elliot, 2011; Greenless et al., 2008; Hill et al., 2005; Piatti, Savagem, & Torgler, 2012). However, most studies had primarily focused on colour of apparel, and no scientific enquiry on colour effect of shoes has been done. Furthermore, previous studies examined only behavioural data, and no evaluation of biomechanical or performance data has been conducted.

Purpose of the study

This study aimed to examine if shoe colour would influence perceptual and biomechanical variables in vertical countermovement jump.

Methods

Thirty-six male university basketball players, with no leg injuries or colour blindness, were recruited to perform perception and jumping tasks, when wearing each of three identical shoes with different shoe upper colours (red, blue and black). Prior to data acquisition, participants indicated their preference among red, blue and black shoes.

Participants performed three vertical countermovement jumps with maximum-effort on a force platform. Maximum jump height was defined as vertical difference of the knee joint marker at the highest point relative to standing (Figure 1(A)). Flight time, propulsion peak and impulse were determined using ground reaction force (GRF) data. Immediately after each jump, participants were asked to use a laser pointer to indicate their perceived body height and maximum reach height on a wall from a fixed distance of 10 m, using cinematographic analysis (Figure 1(B)). Shoe sequence was randomized, and an 8-min rest was given between shoe colours.

Best trials of each variable were determined for further statistical analysis (alpha = 0.05). One-way repeated measures ANOVA was performed to determine if shoe colour had an effect on perception and jumping performance. The participants were grouped by colour-preference to determine if individual preference influenced colour effects.

Results

One-way ANOVA (Table 1) showed no shoe effect on all perceptual and biomechanical variables with all participants combined ($P > 0.05$). When split into colour-preference groups, a significant shoe colour effect on perceived body height ($P < 0.05$) was found in the red-preference group; a significant shoe colour effect on flight time ($P < 0.05$) was found in the black-preference group.

Pairwise comparisons showed that red-preference participants perceived themselves taller when wearing red shoes (1.73 m) compared to black shoes (1.70 m, $P < 0.05$). Black-preference participants displayed longer flight time when wearing red shoes (648 ms) compared to blue shoes (642 ms, $P < 0.05$).

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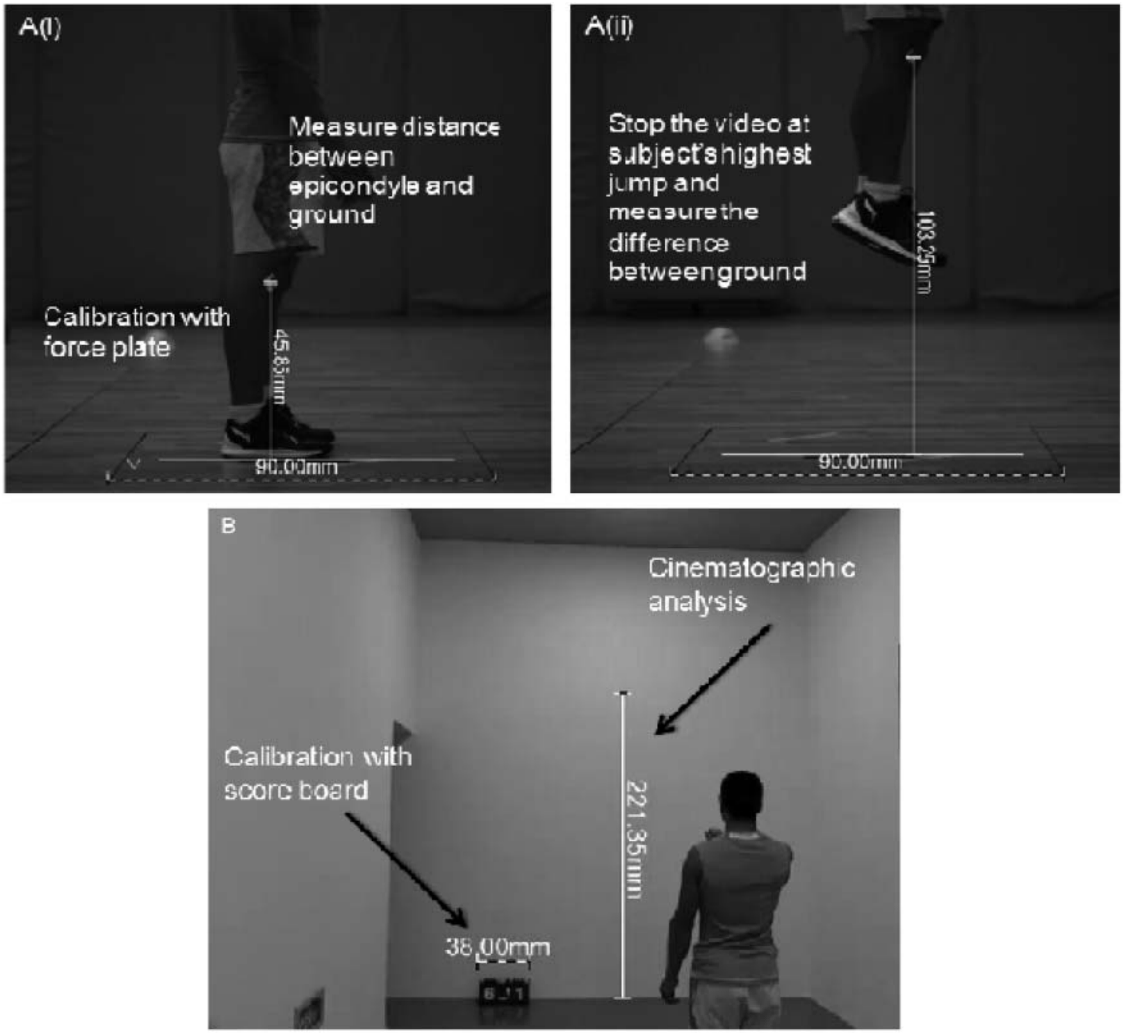

Figure 1. Jump preparation (Ai) and flight (Aii); perceived jump height estimation (B).

Discussion and conclusion

Wearing red shoes may have advantages for some basketball players, as suggested by higher perceived body height and longer flight time after take-off. However, individual colour-preference seems to be influencing the colour-associated psychomotor effects in jumping. These findings may contribute to footwear design and manufacturing, and sports coaching. Further research on the relationship

Table 1. Main shoe effects on variables.

	Main shoe-color effects (P value)			
Variable	All ($n = 36$)	Red ($n = 10$)	Blue ($n = 17$)	Black ($n = 9$)
Perceived body height	0.14	*0.02	0.31	0.99
Perceived max jump height	0.60	0.59	0.75	0.36
Propulsion impulse	0.39	0.77	0.06	0.12
Max propulsion GRF	0.82	0.66	0.91	0.46
Flight time	0.53	0.75	0.32	#0.02
Measured jump height	0.10	0.35	0.17	0.24

*Difference between red and black shoes.
#Difference between red and blue shoes.

among colour, individual preferences and performance in different motor tasks is needed to improve applications of colour to functional footwear.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Running shoe quality perception of runners can be predicted from biomechanical variables

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Keywords: running shoe; perception of shoe quality; biomechanical testing; field testing; regression analysis

Introduction

From 1991 to 2015, 10 running shoe tests were carried out at the biomechanics laboratory of the University Duisburg-Essen for a German government supported consumer product agency (Stiftung Warentest). Using material, biomechanical, and field tests, a comprehensive evaluation of footwear properties was accomplished.

Purpose of the study

The purpose of this study was to investigate which biomechanical footwear properties are related to shoe quality judgement by runners.

Methods

From the 10 shoe surveys, the last 4 with an identical testing protocol were selected for our analysis. Experienced runners (20–25) ran in each shoe model a distance of 10 km on a specified course with varying terrain. Subjects filled out a questionnaire with 15-point perception score ratings immediately after finishing their run. They evaluated overall performance (liking) of the shoe, perceived shock absorption and pronation control features, fit and comfort, traction properties, and foot climate. The anchored 15-point scale ranged from very, very good (value = 1) to very, very bad (value = 15). After having used each shoe in the field test, all subjects participated in a biomechanical study, running with all shoes across a force platform (3.3 m/s). An electrogoniometer served to measure pronation and pronation velocity and a miniature accelerometer (Entran EGAX-25) was attached to the medial aspect of the tibia. Miniature piezoceramic pressure transducers were taped to the foot sole to measure heel, midfoot, and forefoot pressures. Two adjacent 'Kistler' force plates (120 cm × 40 cm) were used as target for foot landing of the runners. Running speed was controlled by a photo cell arrangement. Different shoe types (motion control, cushioning) of following companies were used in our tests (Adidas, Asics, Brooks, New Balance, Lunge, Mizuno, Nike, Puma, Karhu, Reebok, Saucony, discounter shoe products). Sixty-five different shoe models were tested in our studies between 2005 and 2015.

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Results and discussion

Correlation analyses revealed a high match of the overall liking of shoes with all other field test questionnaire items. Based on our study from 2009 overall liking correlated highly with fit and comfort ($r = +0.90$), shock attenuation ($r = +0.95$), and pronation control ($r = +0.96$). It appears that runners judge all other shoe features positive if they like the shoe.

Reductions in peak tibial accelerations and maximum vertical Ground Reaction Force rates showed the highest correlation values ($p < 0.01$) with better shoe liking (lower values). The pronation variables show very little and the peak plantar pressures indicate a medium relationship between better shoe likings.

Further analyses showed that the results are almost identical for both, cushioning as well as motion control shoes, when analysing them as two separate groups. Performing a z-transformation for our results in the four studies allowed an analysis across all 65 shoe models (Table 1). Comparing the overall liking scores with the biomechanical results following determination values (r^2) were found: peak tibial acceleration (0.47), max GRF rate (0.36), max pronation (0.01), max pronation velocity (0.01), peak heel pressure (0.22), peak forefoot pressure (0.21). Performing stepwise regression analyses, peak tibial acceleration (prime factor) was complemented by peak forefoot pressure as a second-step variable (Table 2).

Because heel pressures have a stronger relationship with the peak tibial acceleration, forefoot pressure information seems to add valuable information in predicting, how much footwear is liked by runners.

Table 1. Correlation coefficients of biomechanical variables versus shoe liking.

Overall liking versus	2005	2007	2009	2015
Peak tibial acceleration (g)	+0.63	+0.70	+0.61	+0.75
Max GRF rate (bw/s)	+0.64	+0.74	+0.61	+0.60
Max pronation (d)	−0.16	−0.38	−0.12	+0.28
Max pronation velocity (d/s)	+0.42	−0.25	+0.32	−0.22
Peak heel pressure (kPa)	+0.45	+0.47	+0.49	+0.49
Peak forefoot pressure (kPa)	+0.32	+0.50	+0.58	+0.62

Table 2. Stepwise regression determination coefficients with two factors (peak tibial acceleration, peak forefoot pressure).

Overall liking	2005	2007	2009	2015
R-squared (all $p < 0.01$)	0.62	0.54	0.55	0.71

Conclusion

The company name has a big influence on consumer judgement of product quality (Hennig & Schulz, 2011). Therefore, it is surprising to find the high probability of approximately 60% for predicting overall quality rating of footwear by runners. Independent of footwear type (cushioning, motion control), runners prefer shoes with good shock absorption properties and low plantar pressures. This was consistently found in four studies between 2005 and 2015.

Disclosure statement

No potential conflict of interest was reported by the author.

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Minimalist running: evolution of spatiotemporal parameters and plantar pressure following a training of specific running technique in novice subjects

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(*Received 19 January 2017; accepted 27 March 2017*)

Keywords: running; kinematics; centre of pressure; forefoot strike; injuries; mid-foot strike

Introduction

Running is a part of the most spread sports activities in our societies. It does not require much equipment and is not submitted to schedule constraints, for example. However, the risk of injuries is not negligible. Minimalist running is very fashionable and the object of numerous recent researches, because of its potential to reduce the risks of running-related injuries. It is also said to be more natural (Altman & Davis, 2012; Crowell & Davis, 2011, Giandolini, Horvais, Farges, Samozino, & Morin, 2013; Honeine, Schieppati, Gagey, & Do, 2014; Lieberman et al., 2010; Nunns, House, Fallowfield, Allsopp, & Dixon, 2013; Squadrone & Gallozzi, 2009).

The approach chosen within the framework of this study (learning of the minimalist running technique by novices wearing conventional running shoes) is in line with recent training approaches (Giandolini et al., 2013; Warne et al., 2014) and could allow contributing to promote the inclusion of running technique exercises in jogging training programs.

Purpose of the study

The purpose of this study was to investigate if it was possible to teach the technique of minimalist running with standard shoes, to novices in running. The second purpose was to verify if it reduces the risks of pain to the knee, shin, metatarsus and Achilles.

Methods

During 12 weeks, a group M (25 men and 17 women) followed a session of training of running technique as well as a session of 30 minutes of running every week, while a group C (7 men and 21 women) had two running sessions a week without instruction. They both were novices in the field of the running and ran with their own standard running shoes during training and experimental tests.

At the beginning (T1) and at the end (T2) of the study, participants ran three times on a 6.1 m long instrumented walkway (GAITRite Gold, CIR Systems Inc.; or Zeno, ProtoKinetics LLC) (Egerton, Thingstad, & Helbostad, 2014; Humphrey, McDougal, Cook, Vallabhajosula, & Freund, 2016) with imbedded pressure sensors sampling at 100 Hz. Spatio-temporal and plantar pressure parameters were sampled and analysed with the PKmas software (ProtoKinetics LLC). An algorithm allows calculating spatial and temporal parameters, the centre of pressure (COP) as well as relative pressure for each footfall and step.

A questionnaire recording the type and location of possible pains and injuries occurring every month, the study was completed by the subjects of each group at the end of the study.

The evolution between pre-test and post-test was assessed by non-parametric tests of Wilcoxon for the within-group comparisons or Mann–Whitney for between-group comparisons.

Results

For group M, an increase of the cadence and a decrease of the stride width were observed, as well as a lesser footprint length used (57% used less than 67% of the total length of their foot) after the intervention. COP excursion decreased significantly in antero-posterior (AP) direction and increased in medio-lateral (ML) direction in group M after the intervention (Figure 1). A decrease in average COP velocity was also observed in this group.

The incidence of pain and injuries incidence was globally limited, especially when considering month 3, where less calf, tibia and knee pain were reported (Figure 2).

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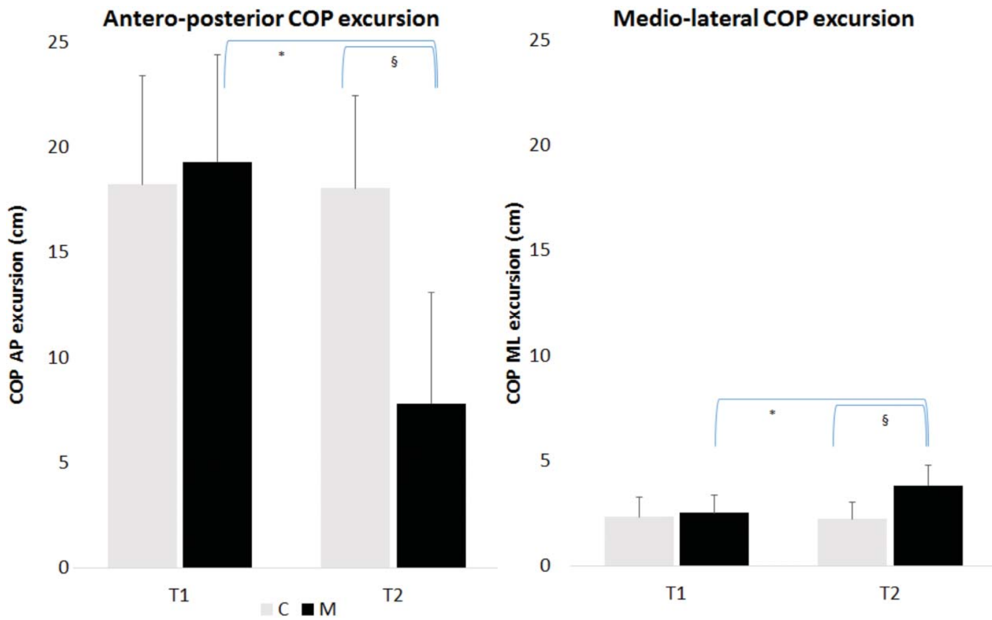

Figure 1. COP excursion in both groups and at both test times. Left: antero-posterior excursion; right: medio-lateral excursion.

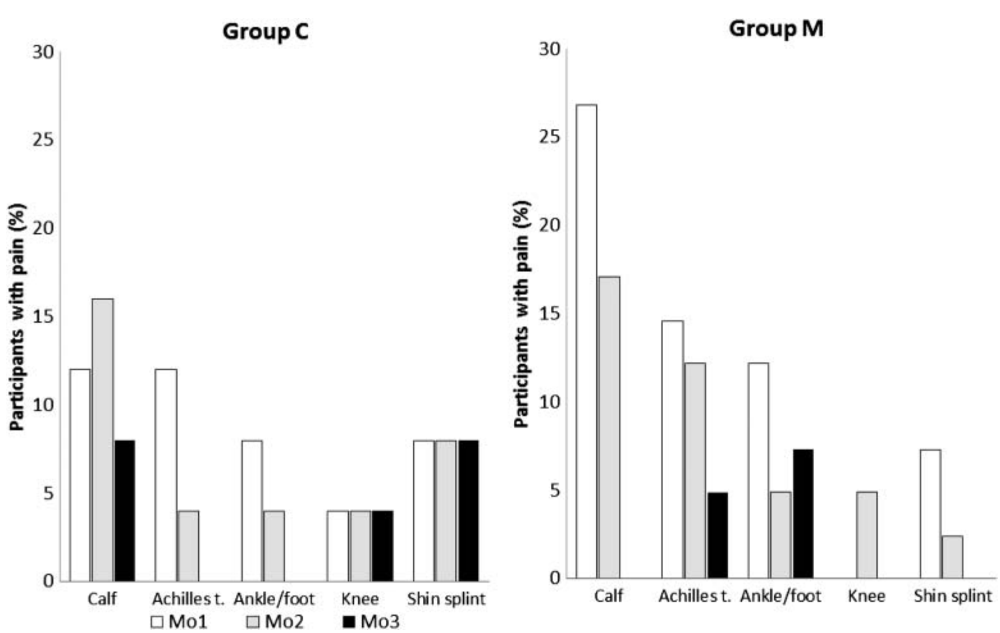

Figure 2. Incidence of injuries and pain during each month of the study.

Discussion and conclusion

Minimalist running is, among others, characterized by a forefoot strike pattern. We registered a decrease in length foot used and a lesser COP AP excursion, indicating a transition to a forefoot strike pattern in group M at the end of our training program. Moreover, the increase of ML COP excursion could indicate an increase of the used forefoot area to compensate the AP decrease.

The decrease of COP velocity suggests a slower ankle dorsiflexion and consequently a lesser solicitation of Achilles' tendon.

We conclude that it is possible to teach novice recreational runners to apply the minimalist technique with standard shoes. This may contribute to reduce the risk of injury at the knee, shin and metatarsus. It is suggested that exercises such as those used in our study could be included in jogging programs for beginners.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The role of the free moment in the perception of rotational friction

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Keywords: free moment; rotational friction; dry friction; statistical parametric mapping; perception

Introduction

In sports, the friction of the shoe–floor interface governs both safety and performance. Recently, the relation between the perception of performance (Starbuck et al., 2016) and safety (Morio, Bourrelly, Sissler, & Gueguen, 2017) with biomechanical measurements of friction has been demonstrated. Therein properties of the shoe grip have been mainly studied through linear friction. However, sport implies a lot of rotational movements or pivot steps, especially in dance, gym courses and indoor team sports. Although previous studies have investigated rotational traction of cleated footwear (i.e. Wannop & Stefanyshyn, 2016), there is to our knowledge, only one study about rotational friction for indoor sport shoes (Nigg, Stefanyshyn, Rozitis, & Mundermann, 2009).

Purpose of the study

The aim of this study was to highlight the relation between the perception of rotational friction and the free moment time series (1D). Then, this relation was used to identify the key variables (0D) which are associated with the rotational friction.

Methods

Fifteen females participated in the study (27 ± 4.5 years; 168 ± 4.6 cm; 61 ± 8.1 kg). They were regularly involved in gym, aerobic or indoor team sports where rotation movements were needed in normal use. Three different shoes (Rubber, TPE and TPU outsoles) with four different floors (rubber, parquet, linoleum and steel) were tested to

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get 12 different shoe–floor combinations. Participants were asked to move laterally from one side to the other of the forceplate by performing a 180° rotation with the right foot over the forceplate. As it was covered with different floors, the forceplate measurements were corrected for each different floor thickness. The free moment was measured for five repetitions for each shoe–floor combination.

The perception of rotational friction was studied with subjective ratings calculated as the mean of all individual perception scores for each shoe–floor combination. The scores of −1, 0 and +1 corresponded to 'not enough grip to turn', 'just enough grip to turn' and 'too much grip to turn', respectively.

The differences in free moment time series were assessed with a statistical parametric mapping (1D SPM) two-way repeated-measures ANOVA (3 shoes × 4 floors). Bonferroni corrections were applied on post-hoc comparisons. Then, a 1D SPM correlation was performed between the free moment and the mean perception scores (Pataky, Vanrenterghem, & Robinson, 2015). Finally, to compare these results with conventional 0D statistics, Pearson correlations were used with local maximum values of significant clusters of the 1D SPM correlation.

Results

The 1D SPM ANOVA revealed significant ($p < 0.001$) main effects of the shoes (26%–88% of the stance duration) and the floors (41%–100%) as well as an interaction (23%–87%). Significant differences were found on 56 out of the 66 post-hoc comparisons.

The 1D SPM correlation (Figure 1) revealed a significant relation between the free moment and the rotation perception at 0%–3%, 11%–18% and 28%–100% of the stance duration ($p = 0.04$, $p < 0.001$ and $p < 0.001$, respectively).

The first and the third significant clusters of the 1D SPM correlation with a positive free moment were kept

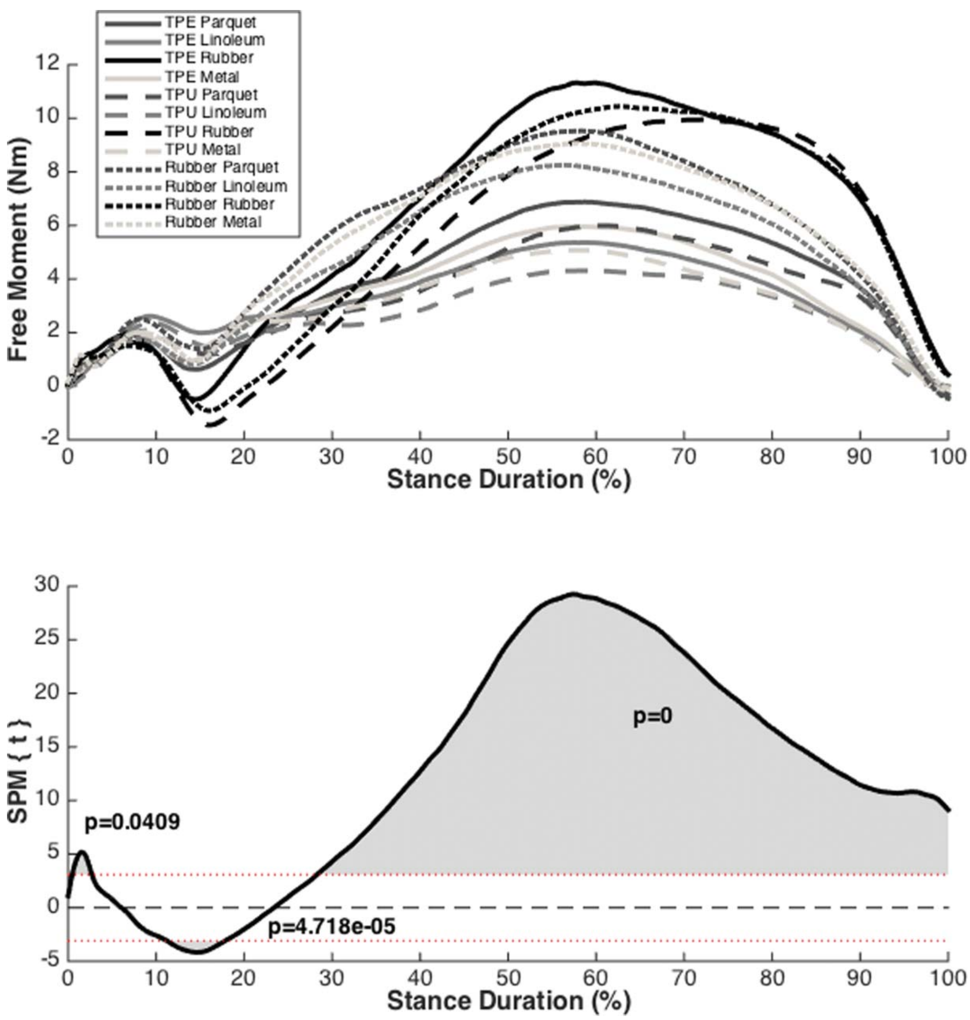

Figure 1. The top chart presents the free moment time series for the different shoe–floor combinations. The bottom chart presents the SPM{t} statistics of the correlation between the free moment time series and the rotation perception scores.

for 0D analyses. Thus, the maximum of free moment in the early 0%–3% and the late 28%–100% part of the stance duration was significantly correlated with the rotation perception scores ($r = 0.76$, $p = 0.004$ and $r = 0.98$, $p < 0.001$, respectively).

Discussion and conclusion

A significant relation was found between the free moment time series and the perception scores of rotational friction. Indeed, the smaller the free moment, the easier the rotation. This relation was corroborated by the simple correlation on early (0%–3%) and late (28%–100%) part of the stance duration. These simpler variables could be considered as static and dynamic rotational friction coefficients, respectively.

Similarly to rotational traction where the shoe studs penetrate the grass (Wannop & Stefanyshyn, 2016), the rotational friction of indoor shoe outsoles should be taken into account in the study of non-contact injuries.

To conclude, the free moment is an interesting variable to investigate the rotational friction of indoor sports shoes. Future research should focus on its effect on transverse and frontal lower limb kinetics, and might try to define safe values or thresholds like it is the case for linear friction.

Disclosure statement

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Effects of shoe energy return and bending stiffness on running economy and kinetics

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Keywords: footwear; running economy; ground reaction forces; bending stiffness; energy return

Introduction

Sport shoes are expected to contribute to performance. A common indicator of running performance is running economy (RE; Fletcher, Esau, & MacIntosh, 2009), and a number of studies have investigated different shoe features to improve RE. Hence, RE could be increased by around 1% (mean over participants) with soft and resilient compared to traditional running shoes (Worobets, Tomaras, & Stefanyshyn, 2014), or with an intermediate bending stiffness (BS) compared to low or high BS shoes (Roy & Stefanyshyn, 2006). These shoe features could act in increasing the energy return (EnR) or restricting the energy loss at joint levels by the optimal use of the musculoskeletal system (Roy & Stefanyshyn, 2006). Potentially due to a lack of available methodological solutions, ground reaction force (GRF) was never linked to performance changes following midsole modifications. Statistical parametric mapping (SPM; Pataky, Vanrenterghem, & Robinson, 2015), as an emergent approach, could help to highlight the effects of shoe features on GRF, and thus help the understanding of mechanisms behind these physiological alterations.

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Purpose of the study

The purpose of this study was to investigate the effects of midsole EnR and BS on RE and GRF during running.

Methods

Nineteen healthy male recreational runners took part in three separate sessions with 48 hours rest between each one. First, participants ran with their own shoes on a 250-m outdoor track during an incremental test until the first ventilatory threshold (VT1) was reached. Two submaximal tests were then performed, where four shoe conditions were tested in a random order. Shoes had identical geometry, mass and midsole stiffness, and only differed in terms of EnR (a low EnR and a 29% higher EnR) and BS (a low BS and a 135% higher BS). For both submaximal tests, participants ran continuously at 90% of the VT1 during 8 min with 5 min rest between each shoe condition. Expired gas analysis by indirect calorimetry (K5, Cosmed) was used to record breath-by-breath data between 6 min and 7 min 30 s while participants ran on the outdoor track. RE (expressed in caloric unit cost) was computed during this time interval. GRF data were recorded in laboratory during running in a 50-m hallway with the force plate (9287CA, Kistler) located at 35 m from the start. Five valid trials from the 6th min were acquired at 2000 Hz for GRF data. Gas exchanges data were not filtered. A critically damped low-pass filter was performed on GRF data with a 40-Hz cut-off frequency.

Two-way repeated measures ANOVAs ($p = 0.05$) were performed to evaluate the effects of EnR and BS on RE and GRF through classical (0D) and SPM (1D) procedures, respectively (Pataky et al., 2015).

Results

RE of the whole group was not affected by the EnR or BS ($p > 0.05$). A main effect of the EnR factor was found on anteroposterior (23%–25% of stance phase) and vertical (15%–18% of stance phase) GRF (Figure 1). Force traces on both GRF components were lower with the low EnR.

A main effect of the BS factor was found on anteroposterior (71%–88% of stance phase) and vertical (52%–93% of stance phase) GRF. Force traces on both GRF components were lower with the high BS.

Discussion and conclusion

Regarding running performance, RE was not altered despite high variations in EnR and BS of midsoles. Nonetheless, effects of midsole mechanical properties were observed on GRF with EnR altering the braking phase and BS altering the propulsion phase. The low EnR (thus a high absorbed energy) decreased GRF values after the

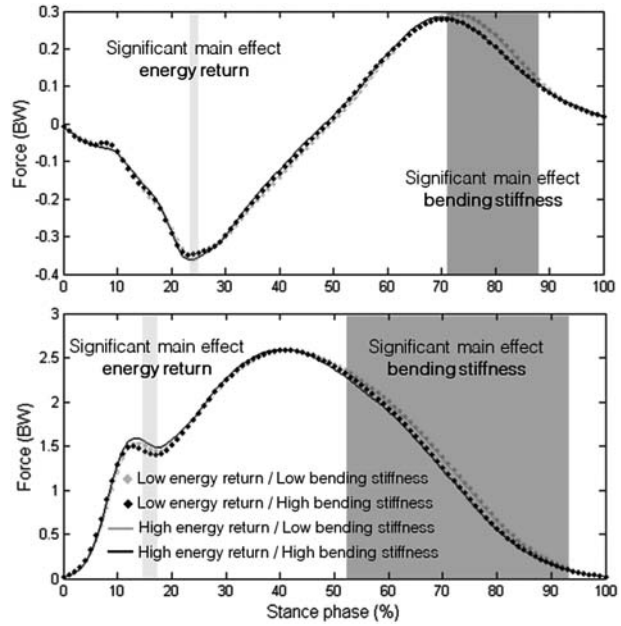

Figure 1. Significant main effects of EnR (light grey areas) and BS (dark grey areas) on anteroposterior (top) and vertical (down) GRF from SPM analysis. Standard deviations and SPM curves were not presented for more clarity.

vertical first peak and at the anteroposterior braking peak. Changes in midsole mechanical properties could induce biomechanical adaptations, which could decrease repeated loadings of the anatomical structures. The high BS decreased GRF values after the vertical and anteroposterior propulsion peaks. The high BS could be beneficial because less propulsion force is needed to keep an identical running speed. Acting during key phases of stance, suitable EnR and BS may decrease loadings but did not seem to save energy cost during running.

Further studies should analyze kinematic and electromyogram patterns to help in the understanding of GRF alterations observed in the present study. Further studies could also look at alterations caused by such footwear during prolonged running.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Long-term effects of gradual shoe drop reduction on young tennis players' kinematics

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Keywords: tennis; biomechanics; shoe; children; adaptation

Introduction

Worldwide, tennis is practiced by millions of children, several thousands being intensive players. The participation in high-performance training programmes and competition places them at risk to experience overuse injuries (Pluim, Loeffen, Clarsen, Bahr, & Verhagen, 2015). Indeed, this sport is characterized by repeated accelerations–decelerations, and changes of direction which generate large biomechanical loads on internal structures (Abrams, Renstrom, & Safran, 2012).

Shoe drop (i.e. difference between heel and forefoot height) was reported to influence the young tennis players' kinematics (Herbaut et al., 2016). Zero-drop shoes induced an immediate shift from a rearfoot- (RFS) to a forefoot strike (FFS) for about 30% of young tennis players to perform an open stance forehand. However, the long-term influence of wearing low-drop shoes was still unknown.

Purpose of the study

The aim of this study was to examine the long-term effects of a gradual shoe drop reduction on the biomechanics of young tennis players performing an open stance forehand.

Methods

Thirty young tennis players (age = 10.7 ± 0.7 years, height = 1.44 ± 0.04 m, body mass = 34.8 ± 5.2 kg) participated in this study. They came thrice to the laboratory. First, an inclusion visit confirmed that they met the inclusion criteria: minimum of 3 h/week tennis practice, aged between 8 and 12 years. All the participants systematically adopted a RFS to perform an open stance forehand after a 2.5-m run (Herbaut et al., 2015).

They were equally distributed in two groups: control (CON) and experimental (EXP) (Figure 1). CON received twice 12-mm drop shoes (D12) while EXP received 8-mm drop shoes (D8), then 4-mm drop shoes (D4) (Figure 1). In POST testing session, 6 out of the 12 EXP switched towards an FFS (EXP-FFS) while the other 6 kept a RFS (EXP-RFS).

A motion capture system (500 Hz, VICON) and two force plates (1000 Hz, Kistler) allowed the recording of both kinematic and kinetic data. The main parameters (expressed as mean ± standard deviation of 10 successful trials) were the foot vs. ground angle and ankle flexion at foot strike (FS). Two independent *t*-tests were performed between each group for each testing session (INT: intermediate testing session and POST: post testing session). Significance threshold was set at 0.025 (0.05/2, Bonferroni correction).

Results

The main results are depicted in Figure 2.

Discussion and conclusion

Even though no difference in foot vs. ground angle and ankle flexion at FS was observed in INT with the use of D8, significant changes appeared in POST with the use of D4.

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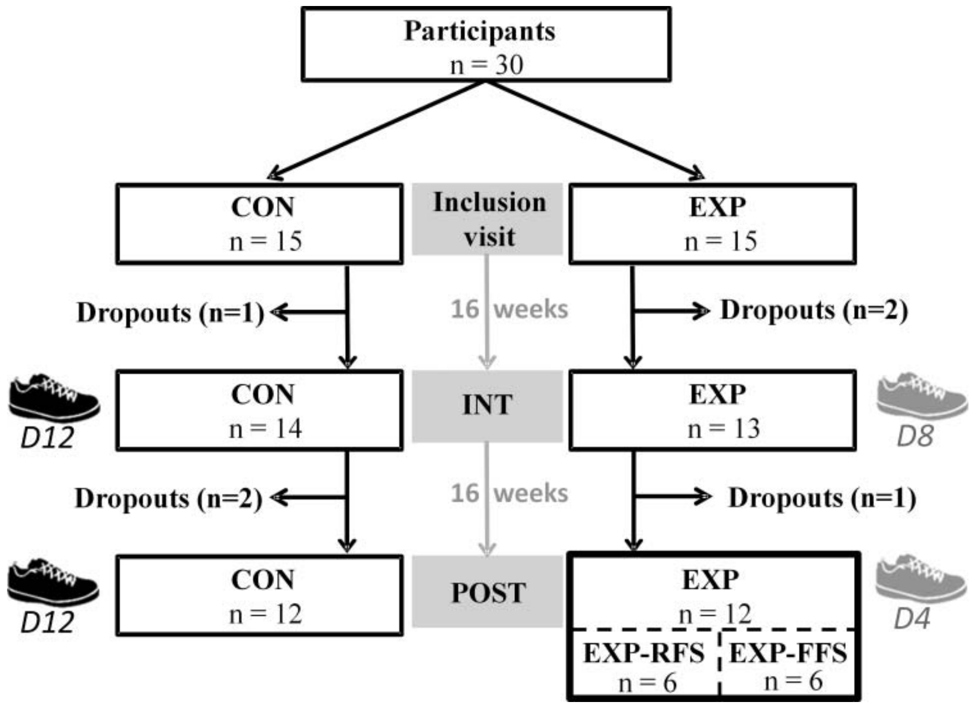

Figure 1. Flow diagram of the participants (CON: control group; EXP: experimental group; RFS: rearfoot strikers; FFS: forefoot strikers; INT: intermediate testing session; POST: post-testing session; D12: 12-mm drop; D8: 8-mm drop; and D4: 4-mm drop shoes).

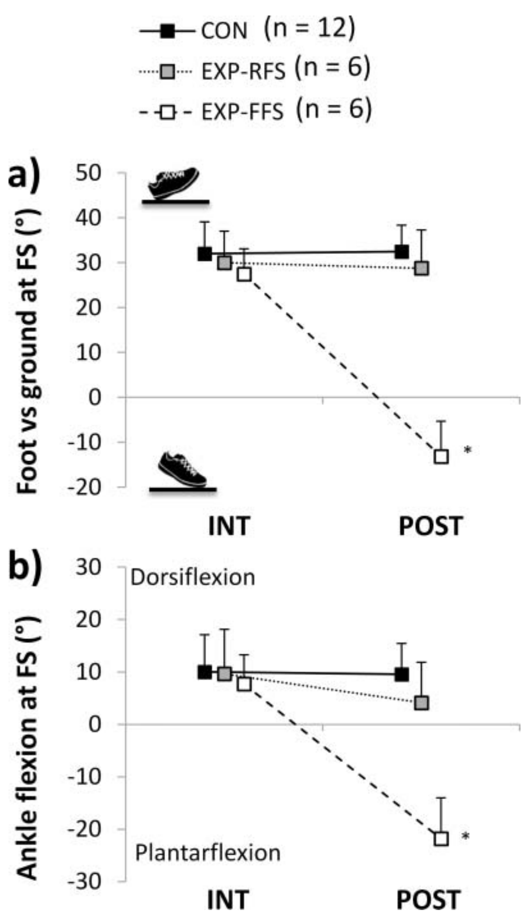

Figure 2. (a) Foot vs. ground angle and (b) ankle flexion (expressed in degrees) at foot strike in INT and POST testing session for experimental and control groups. $^{*}p < 0.025$.

Six out of the 12 EXP transitioned from RFS towards FFS pattern with the ankle plantar-flexed to perform the open stance forehand with D4 (EXP-FFS; Figure 2). Such a change in the landing strategy suggests the existence of a drop/cushioning threshold under which young tennis players feel discomfort or pain when impacting the ground with the heel. A forefoot landing strategy may be beneficial to reduce heel compressive forces at ground contact and to avoid heel pain. However, it requires an eccentric work of the plantarflexor muscles to control heel lowering after ground contact, which might be deleterious.

The other half of EXP kept a RFS despite the lower shoe drop in D4. Although the amount of cushioning was low, it was likely sufficient for the participants to bear the biomechanical load resulting from a rearfoot landing. Further investigation is needed to better understand why half of children switch towards an FFS landing strategy and why others do not.

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Disclosure statement

No potential conflict of interest was reported by the authors.

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Correlation between foot pressure and comfort in recreational and advanced tennis players

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Keywords: shoe; comfort; tennis; pressure distribution; ergonomics

Introduction

Footwear comfort is essential to practise sport with satisfaction (Au & Goonetilleke, 2012). Llana, Brizuela, Dura, and Garcia (2002) reported that one-third of the tennis players judged their shoes uncomfortable or just acceptable. They identified the medial aspect of the first metatarsal (MT1) and the lateral aspect of the fifth metatarsal (MT5) as the most influential areas on general discomfort. Although footwear comfort is difficult to assess due to personal preferences, some researchers observed significant correlations between comfort and pressure applied on the foot (Cheng & Hong, 2010).

A previous study showed that advanced tennis players preferred a tighter fit than recreational tennis players (Herbaut, Foissac, Jurca & Guéguen, 2016). However, the pressure on tennis players' feet has never been measured yet.

Purpose of the study

The aim was to examine the effect of tennis player's skill (recreational or advanced) on the perception of shoe comfort.

Methods

Twenty male tennis players (age: 29.4 ± 7.8 years; height: 1.78 ± 0.04 m; body mass: 75.7 ± 7.2 kg) were distributed into 2 groups: 10 advanced players (AP) and 10 recreational players (RP).

Three tennis shoes (Eu43) with the same upper construction but made with three lasts of different forepart width and girth were assessed by the participants.

A sock equipped with textile sensors (Texisense®, France), validated in a previous study (Herbaut, Simoneau-Buessinger, Barbier, Cannard, & Guéguen, 2016), was used to measure the pressure applied on MT1 and MT5 with the participants standing in static position.

Pearson correlations between pressure and comfort were computed for AP and RP.

Results

Correlations between comfort and pressure on MT1 were strong for AP ($r = -0.705$) and very weak for RP ($r = -0.289$) (Figure 1).

Correlations between comfort and pressure applied on MT5 were moderate for AP ($r = -0.505$) and very weak for RP ($r = -0.284$) (Figure 2).

Discussion and conclusion

The first outcome was that regardless of the foot location or players' skill, correlations between pressure and comfort were all significantly negative. It means that, even though a minimum amount of pressure is necessary to ensure a proper fit and security in use, tennis shoe manufacturers should avoid using too narrow lasts and too rigid upper compounds if they aim to design a comfortable shoe.

The second outcome was that the mean pressure applied on MT5 (56.2 kPa) was larger than the mean

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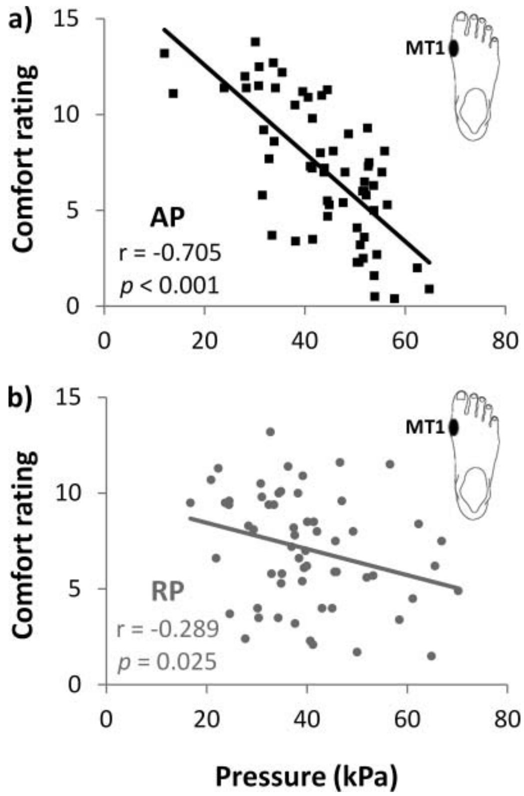

Figure 1. Correlations between pressures on the first metatarsal head (MT1) and comfort for (a) advanced players (AP) and (b) recreational players (RP).

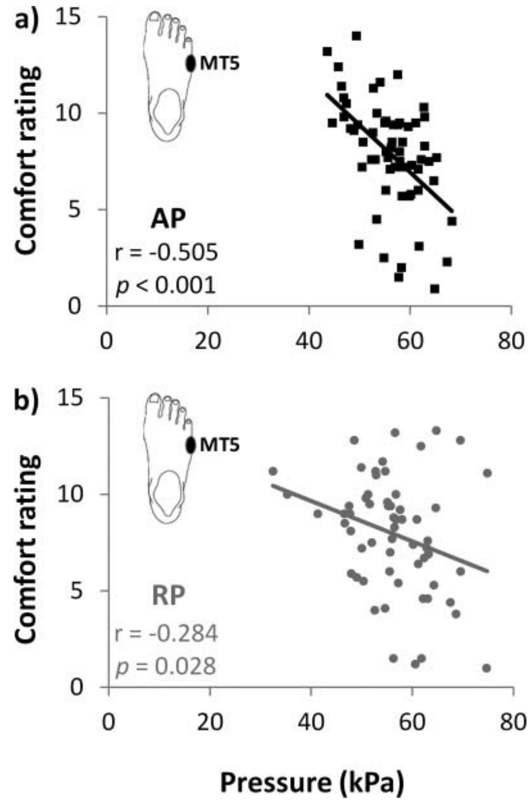

Figure 2. Correlations between pressures on the fifth metatarsal head (MT5) and comfort for (a) advanced players (AP) and (b) recreational players (RP).

pressure on MT1 (41.7 kPa). It was consistent with Cheng and Hong's study (2010) and may be due to the more protruding shape of MT5 compared with MT1.

The third outcome was that the correlation coefficients between pressure and comfort on both MT1 and MT5 were higher for AP than for RP. It was consistent with a previous study aiming to determine the optimal inner-shoe volume both for AP and RP, where RP had a larger variability in their rating (Herbaut et al., 2016). This suggests that AP have a better knowledge of their needs in terms of footwear fitting and comfort. Therefore, AP were more consensual in their rating than RP.

A future study is needed to examine the differences between AP and RP in terms of pressure and comfort perception in dynamic situation to ensure a sufficient support of the foot in usage.

Acknowledgement

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Disclosure statement

No potential conflict of interest was reported by the authors.

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Patient-specific foot orthotics improves postural control of rheumatoid arthritis patients: a pilot study

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Keywords: rheumatoid arthritis; postural sway; insoles; orthoses; elderly

Introduction

Rheumatoid arthritis (RA) is an autoimmune disease often causing pain and foot problems. Daily pain, stiffness, fatigue and physical disabilities are among the common features of RA. Over 85% of RA patients experience painful feet and ankles during the course of the disease. In RA, synovitis, effusion and eventually erosive arthritis are thought to cause clinically recognizable valgus heel or pes planovalgus deformity. Foot pain, especially for the older population, is associated with decreased ability to perform activities of daily living (Bowling & Grundy 1997), problems with balance and gait and increased risk of falls (Gorter, Kuyvenhove & de Melker, 2000; Menz & Lord 2001). With the intention to stabilize and align the foot, foot orthotics are often prescribed to RA patients.

Purpose of the study

The purpose of this pilot study was to investigate foot orthotics effect on the postural control of RA patients compared with a control insole.

Methods

Five RA patients (4 females and 1 male) were recruited for this study (age 59.8 ± 6.95 years, Height 172 ± 6.95 cm, body mass 69.2 ± 15.68 kg).

A subject-specific foot orthosis (FO) was made for each patient by an experienced orthotist using a weight-bearing corrected footprint technique in a cast form. The foot print was 3D surface scanned and manufactured in a milling cutter in a hard material.

Each subject completed six trials single-legged postural control test: three with the FO and three with a control (C) condition (Nike Pegasus standard insole, USA). Subjects were instructed to stand as still as possible for 20 seconds. All trials were performed with a standardized neutral shoe type (Nike Pegasus, USA). Centre of pressure (COP) data were collected on a force plates installed in the floor (1000 Hz, Gain 2000) (AMTI, USA).

Results

Average medio-lateral COP range was reduced with the FO: left leg C: 20.7 ± 3.2 cm, FO: 18.6 ± 2.8 cm; right leg C: 23.3 ± 3.4 cm, FO: 19.2 ± 2.2 cm. The differences in percentage were 22.2% ± 7.3% for the right leg and 11.4% ± 10.1% for the left leg.

Average antero-posterior COP range was reduced with the FO: left leg C: 24.2 ± 4.5 cm, FO 21.8 ± 3.9 cm; right leg C: 24.5 ± 2.6 cm, FO: 22.3 ± 2.9cm. The differences in percentage were 10.4% ± 8.2% for the right leg and 11.1% ± 6.3% for the left leg. Figures 1 and 2 show a line plot for each subject for medio-lateral and antero-posterior for the right leg, respectively.

Discussion and conclusion

The results from this pilot test suggest reductions in medio-lateral and antero-posterior COP range for both left (all but one participant) and right legs (all participants) when using FO. The reduced COP range indicates that subjects were more stable while standing on one leg while wearing the FO.

Because of the FO support, it is likely that the difference found in COP range is caused by reduced muscle activity in the lower extremity, because the muscles do not need to stabilize as much to keep a stable posture.

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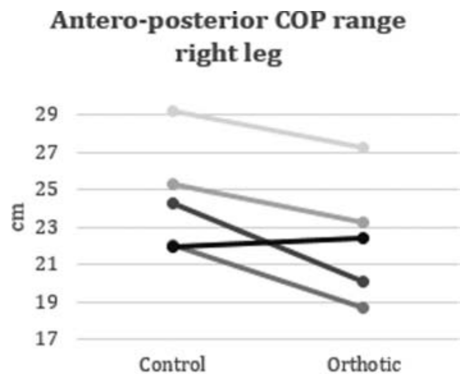

Figure 1. Antero-posterior COP range for during left right test.

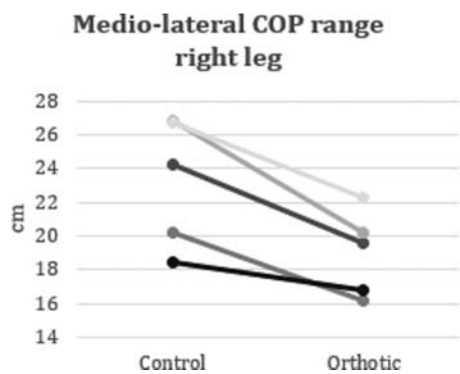

Figure 2. Medio-lateral COP range for during right leg test.

The difference found in this study is likely to be permanent with the use of FO. The motivation for prescribing foot orthotics for the RA patient group is often because of an intention to reduce pain. However, potential improvement of balance and stability could also be a reason for prescribing foot orthotics. Further investigation into the effect of foot orthotics on postural control could help define new clinical guidelines, thereby potentially reducing the risk of falling for elderly RA patients.

Disclosure statement

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Women perception of shoe cushioning as a function of mechanical properties of footwear

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Keywords: user perception; sensory characterization; mechanical characterization; footwear cushioning; women

Introduction

Many key features of footwear are assessed through field or lab tests. On one hand, field tests well reflect real use but are time-consuming and can bring less reliability. On the other hand, mechanical lab tests bring more repeatability in less time, but are sometimes less realistic. User perception obtained from field tests is a key aspect in shoe conception. Being able to predict those feelings using a quick lab test is crucial. Goonetilleke (1999) reported high correlations ($r = -0.99$) between

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mechanical impact test variables and user perception of shoe cushioning. However, sensory variables are often collected with limited methods (e.g. Dinato et al., 2014; Goonetilleke, 1999), thus making conclusions limited. Moreover, the fields of cosmetics or foods usually use expert method to characterize complex sensory aspects of products. This method consists in training participants for the sensory characterization of products in order to ensure data reliability.

Purpose of the study

The purpose of this study was to determine the relation between women sensory characterization of cushioning and mechanical properties of running shoes.

Methods

Heel part of 32 various running footwear models were mechanically characterized using an impact test (ASTM F1614-99, 2006). Based on principal components analysis and hierarchical clustering of the mechanical data, 10 footwear models were retained for next steps. Shoe covers were used to avoid any unwanted effect of design on sensory data (Figure 1).

Ten recreational women runners (27.4 ± 6.6 years, 63.5 ± 6 kg, 167.6 ± 5.2 cm) were trained to become experts of shoes sensory characterization. First, simple semantics and gestures were highlighted to make descriptors definition explicit and understood by all participants. Then, participants were trained to become discriminant, repeatable and in agreement for the rating of descriptors. Finally, each participant ran 15 min at 2.7 $m{\cdot}s^{-1}$ around an indoor sport playground with the 10 shoe models. Participants wore models twice in order to test for repeatability. After running with a shoe model, participants rated the descriptors' intensity (from 1 to 9) on linear scales. The order of shoe models was controlled to avoid any effect of rank, sequence and fatigue. Each footwear model was available in four copies to limit alterations of shoe properties due to repetitive use. Participants rating performances were verified with a three-ways 'shoe × participant × session' analysis of variance. The highest correlation between the sensory descriptor and mechanical variables was searched. Critical threshold was set to 0.05.

Results

Shoes mechanical properties were distributed over a broad range (Figure 2).

'Heel softness' was an explicit descriptor understood by all to describe shoe cushioning. Panel performances showed a significant effect of shoe ($p = 9 \times 10^{-11}$), no significant effect of shoe × session interaction ($p = 0.119$) and no significant effect of shoe × participant interaction ($p = 0.067$) on the rating of heel softness. A significant correlation between the rating of heel softness and the shoe soles maximal displacement in impact test was found ($r = 0.86$, adjusted $r^2 = 0.75$, $p = 6.9 \times 10^{-4}$).

Discussion and conclusion

Data processing enabled to well identify 10 footwear whose mechanical properties were spread out over a broad range (Figure 2). Semantics, gestures and training made

Figure 1. Shoes with shoe covers.

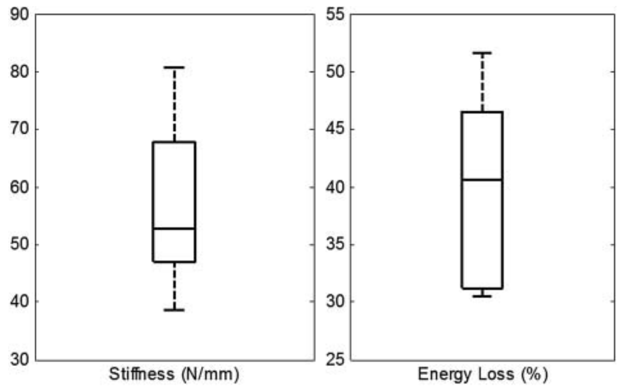

Figure 2. Boxplots of stiffness and energy loss of the ten shoe models.

the participants discriminant (significant effect of shoe), repeatable (no significant effect of shoe × session interaction) and in agreement (no significant effect of shoe × participant interaction). These results ensured data reliability for further analysis. The highlighted correlation enabled predicting 75% of the perceived heel softness variance from mechanical data. Even with a lower correlation coefficient, this result seemed more reliable than those reported by Goonetilleke (1999) since they are not exclusively based on extreme mechanical properties. This was due to participants performances in the sensory evaluation of descriptors (i.e. footwear features), especially their capacity to discriminate between shoe models. More broadly, by ensuring data reliability, expert panel is a powerful method for the quantitative assessment of footwear features perception. Since this study focused on European females, future studies should look at user perception in other populations.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Foot shape, perceived comfort, and plantar pressure characteristics during long-distance running

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Keywords: foot morpohology; footwear; marathon; plantar pressure; running

Introduction

Long-distance running is an increasingly popular past time globally. As reported, finishers of full marathons increased 32% and half marathons increased 307% over the past decade in the United States alone. In China, participation hit the highest record of 2.8 million in long-distance events during 2016. While running such a long distance either in 10 km (mini marathon), half marathon, or full marathon, shoe fit and comfort are the main considerations of most participants. The foot is the primary link of the human motor system with its external environment (Mei, Fernandez, Hume, & Gu, 2016), and participants tend to select running shoes with specific functions like cushioning, stability, or motion control (Mundermann, Nigg, Stefanyshyn, & Humble, 2002).

However, foot shape differs under certain situations, like obesity, pregnancy, and long-time physical activities (Cowley & Marsden, 2013). Foot shape changes may lead to ill-fitting shoes. Ill-fitted footwear has been shown to be related with dermatologic problems, which leads to

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abnormal compensatory movements, thus increasing risks of musculoskeletal injuries.

Purpose of the study

This study aimed to analyse foot shape changes during long-distance running, and recorded the subjective perceived shoe comfort and measured plantar pressure distribution characteristics.

Methods

A total of 26 male recreational runners were recruited to join in the running test, with age of 28.6 ± 3.4 years, height of 173.6 ± 5.4 cm, weight of 72.7 ± 4.8 kg, and weekly running distance of 45 ± 15 km. They all had the same shoe size of 8 (US) or 41 (Europe). During the running test, they all wore the same shoes made from a fixed shoe (same as their shoe size). Subjective perceived shoe comfort was measured with a 150 mm visual analogue scale (VAS) by Munermann et al. (2002).

Before the test, foot shape data under normal static standing conditions were collected with a 3D foot scanner (YETITM, VORNM, Canada), with accuracy of 0.1 mm, and scanning volume of 400 × 800 × 1000 mm^3.

Participants were instructed to run on the outdoor playground at their self-selected speed. Foot morphology data were collected immediately after finishing 10 and 20 km separately. Plantar pressure after 10 and 20 km (within two minutes) were also recorded with a Novel Pedar insole measurement system.

The repeated measures ANOVA from SPSS 19.0 was taken to analyse the significance of foot shape data, perceived comfort and plantar pressure among pretest, 10 and 20 km. The significance level was set at 0.05.

Results

Foot shape showed alterations compared with normal static conditions, particularly ball width, ball girth, arch height, and foot volume with significance. Ball width reduced to 99.7 ± 1.6 mm after 10 km, comparing with normal static conditions of 100.5 ± 1.6 mm, with $p = 0.008$. Also, ball girth significantly ($p = 0.029$) reduced to 242.6 ± 2.1 mm, comparing with 244.7 ± 2.5 mm in normal conditions. Arch height dropped significantly to 11.7 ± 3.3 mm ($p = 0.039$) versus normal conditions (13.1 ± 3.2 mm). Foot volume reduced significantly from 908,960 ± 28,674.6 mm^3 in normal conditions to 893,533.7 ± 22,752.3 mm^3 (after 10 km, $p = 0.033$), and 884,549.2 ± 31,817.8 mm^3 (after 20 km, $p = 0.008$).

Measured perceived shoe comfort changed accordingly, especially in overall comfort, forefoot width, and forefoot sole comfort. Consistent with recorded plantar pressure, M1 (first metatarsal), MM (medial midfoot) and H (heel) presented significantly increased peak pressure (Figure 1).

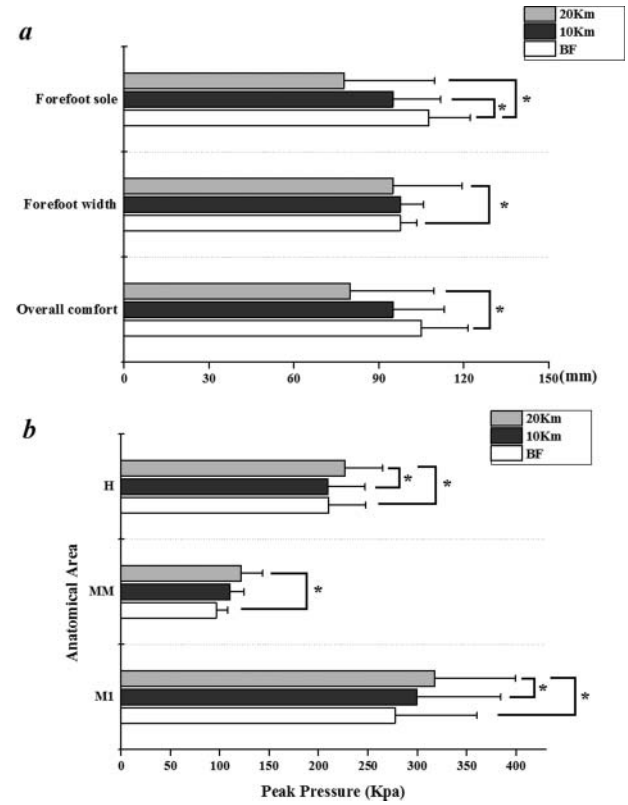

Figure 1. The illustration of foot shape (a) and plantar pressure (b) data with significance.

Discussion and conclusion

The objective of this study was to measure the foot shape changes, perceived shoe comfort, and plantar pressure distribution characters after 10 and 20 km running. Previous studies have indicated that shoe fit and comfort are closely linked with foot shape and plantar pressure (Cowley & Marsden, 2013; Hoerzer, Trudeau, Edwards, & Nigg, 2016; Kouchi, Kimura, & Mochimaru, 2009; Mundermann et al., 2002). In this study, foot shape changes and subjective perceived comfort from VAS showed consistency with plantar pressure. The reduced ball width and girth, and VAS score could be explained by increased peak pressure to M1. This would explain observed blisters and callus to the medial and beneath forefoot. Also, the shrinking of foot volume would lead to shoe support instability, combined with poor motion control from muscle fatigue after long-distance running. This potentially increases the chance of overuse injuries in the lower extremity. Knowledge of foot shape and perceived comfort changes after long-distance running would bring about some implications for running shoes design from the perspective of biomechanics.

One limitation that should be considered in this study is the lack of lower extremity joint kinematics, a future objective of this work.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The influence of motion control, neutral and cushioned running shoes on foot kinematics

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Keywords: running; running shoe; biomechanics; foot; kinematics

Introduction

Footwear biomechanics research typically focuses on the assessment of frontal plane rearfoot (RF) motion when determining the influence of footwear on foot motion (Cheung & Ng, 2007; Lilley, Stiles, & Dixon, 2013). The key limitation of this approach is that it only describes one aspect of foot motion. Application of a multi-segmental foot model to the assessment of the shod foot could more accurately describe how footwear influences foot motion. There is currently limited exploration of the influence of running shoes on intersegmental foot kinematics due to the challenges of modelling the foot within the shoe.

Purpose of the study

To determine the influence of running shoes on foot kinematics.

Methods

Twenty-eight active males (26 ± 7 years, 1.77 ± 0.05 m, 79 ± 9 kg) ran at a self-selected pace (2.9 ± 0.6 m/s) on a treadmill, in standardised ASICS motion control, neutral and cushioned running shoes. Kinematic data were collected using a VICON motion analysis system, operating at 200 Hz. Ankle joint kinematics were calculated using a two segment model of the foot and shank. Intersegmental foot motion was calculated using the 3DFoot model (Leardini et al., 2007). Incisions were made within the shoe to enable direct tracking of the shod foot (Langley, Cramp, Moriyasu, Nishiwaki, & Morrison, 2015). Angles at initial contact (IC), peak angle and time to peak angle were extracted for analysis. One-way repeated measures ANOVA and Friedman's ANOVA were used to explore differences between conditions.

Results

There were significant increases in RF dorsiflexion upon IC ($p = .01$, $W = .16$) and peak dorsiflexion ($p = .02$, $W = .14$) when running in the neutral shoe compared to the motion control and cushioned shoes. Peak RF eversion was significantly ($p = .04$, $W = .12$) increased when running in the motion control shoe compared to the

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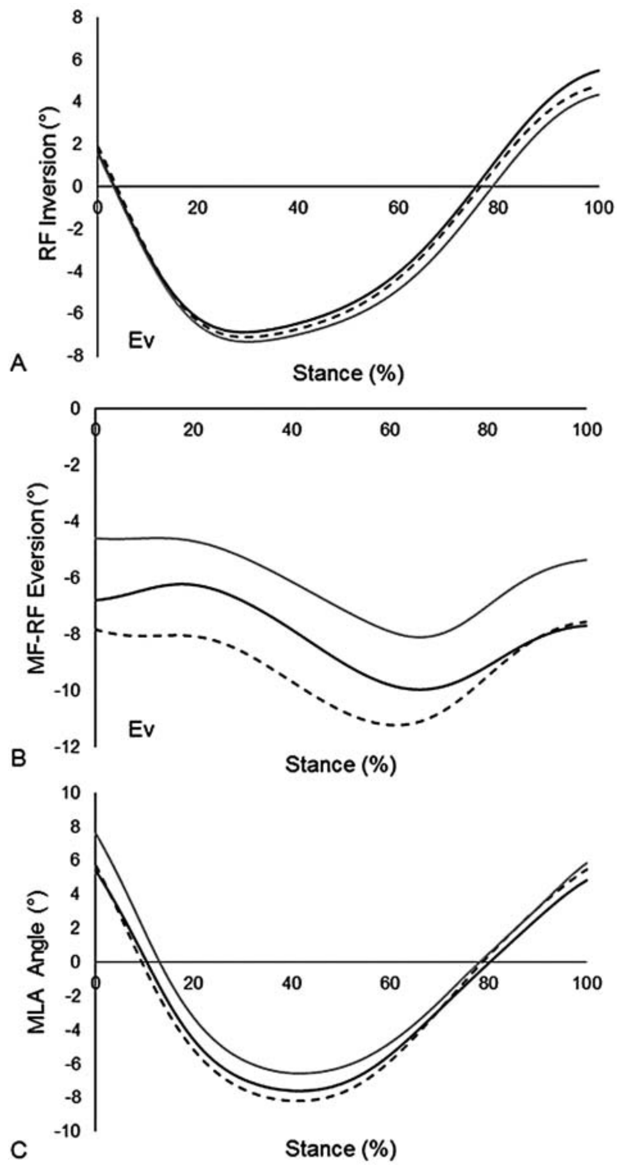

Figure 1. (A) Frontal plane rearfoot, (B) frontal plane midfoot to rearfoot and (C) medial longitudinal arch kinematics over the stance phase of running in motion control (solid grey line), neutral (solid black line) and cushioned (dashed black line) running shoes. Data averaged across all participants ($n = 28$).

cushioned shoe. There was a significant increase in RF adduction upon IC ($p = .03$, $\eta^2 = .12$) and reduction in peak RF abduction ($p = .01$, $\eta^2 = .15$) when running in the neutral shoe compared to the motion control shoe. There were significant reductions in MF-RF eversion upon IC ($p = .01$, $W = .16$) and MLA deformation ($p = .04$, $W = .11$) when running in the motion control shoe compared to the neutral and cushioned shoes. No other significant differences were reported.

Discussion and conclusion

Motion control running shoes were shown to reduce aspects of intrinsic foot motion such as medial longitudinal arch deformation and MF-RF eversion (Figure 1(B) and 1(C)). These findings were expected based on the design features of the shoe in comparison to the neutral and cushioned shoes. However, the motion control shoe did not reduce peak RF eversion (Figure 1(A)). This finding is in contrast to the previous literature (Cheung & Ng, 2007; Lilley et al., 2013) and the design aims of the shoes. This disparity between the current study and the literature is likely due to this work assessing the motion of the foot within the shoe. The findings of this study suggest that motion control shoes were better at controlling intrinsic foot rather than RF motion. However, it should be noted that while statistically significant, both the magnitude of change ($\leq 3.32°$) and the effect sizes ($\leq .16$) were small. The novel information developed within this study provides a more comprehensive understanding of how footwear influences foot motion which may aid in shoe development and potentially a reduction in running injury rates.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Spatial distribution of impact intensity under the shoe in different foot strike patterns

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Keywords: foot strike; impact forces; athletic footwear; plantar pressure; running

Introduction

It is generally known that the foot strike pattern during running relates to initial foot and ankle kinematics and impact intensity (Breine et al., 2016). Most researchers observed higher impact intensities in rear-foot strike patterns (RFS) when compared with mid-foot (MFS) or fore-foot strike patterns (FFS).

In a previous study (Breine, Malcolm, Frederick, & De Clercq, 2014) we have distinguished Typical and Atypical RFS, based on the centre of pressure (COP) trajectory. An Atypical RFS is identified by an initial COP position at the lateral side of the rear-foot immediately followed by a fast anterior displacement along the lateral shoe margin towards the mid-foot zone, resulting in an early first metatarsal contact. The relevance of distinguishing Atypical and Typical RFS is that these Atypical RFS showed the highest impact intensity, measured with the vertical instantaneous loading rate of the ground reaction force (GRF). Impact intensity is an important variable as an high impact intensity has been related to an increased stress fracture injury susceptibility (van der Worp, Vrielink, & Bredeweg, 2016).

Information about the magnitude and spatial distribution of the impact intensity under the shoe sole in different foot strike patterns could be used in the design of foot strike-specific footwear. The shoe zones with the highest impact intensity could be targeted for adding passive cushioning for impact intensity reduction.

Purpose of the study

The purpose of this study is to compare different foot strike patterns in the spatial distribution of the impact intensity over four different zones under the shoe. Impact intensity will be measured by determining the peak magnitude of the high frequency (>10 Hz) component of the vertical GRF per foot zone, which has been shown to be a more reliable parameter than peak vertical force and loading rate to assess passive cushioning properties and impact intensity (Shorten & Mientjes, 2011).

Methods

Forty-nine runners (37♂ and 12♀) ran at 3.2 m·s^{-1} wearing the same running shoe (Li Ning Magne) that was optimized for plantar pressure measurements by substituting a flat outsole. Three left foot contacts were recorded. GRF and plantar pressures were recorded by a 2 m pressure plate (500 Hz, Footscan ®, rs scan) mounted on top of a 2 m force plate (500 Hz, AMTI). Plantar pressures were dynamically calibrated with the measured GRF data from the force plate. Foot strikes were categorized as Typical RFS, Atypical RFS or MFS (Breine et al., 2014). Only one FFS trial was recorded which was not retained for further analysis.

A four-zone footmask (Figure 1), determined in the Footscan® software and manually corrected by expert, allowed to determine the local vertical GRF per foot zone. A fast Fourier transform analysis transformed the local GRF signals into the frequency domain. With an inverse fast Fourier transform analysis, the data with a frequency content above 10 Hz was recomposed into the time domain as the high frequency (~impact) component of the vertical GRF. The peak magnitude of this high frequency component was determined per foot zone as local impact intensity measure.

ANOVAs with *post hoc* analysis (Bonferroni, $p < 0.05$) were conducted to assess between foot strike pattern differences in the local peak magnitude of the high frequency GRF component, per foot zone.

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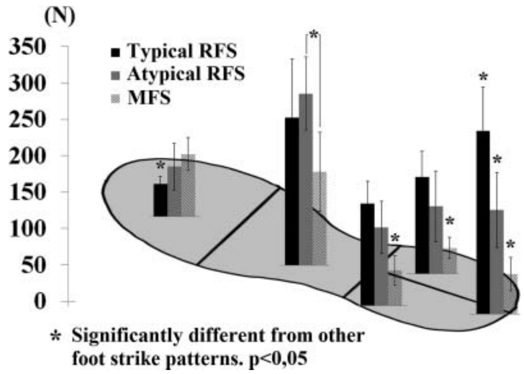

Figure 1. Peak magnitude of the high frequency component of the vertical GRF per foot zone for the different foot strike patterns. The most right bars on the figure indicate the values for the merged lateral and medial rear-foot zone.

Results

Figure 1 shows the peak magnitude of the high frequency component of the vertical GRF, per foot zone (fore-foot, mid-foot, lateral and medial rear-foot and merged rear-foot zones) for the different foot strike patterns. The highest peak values were found under the rear- and mid-foot zone for the Typical RFS, under the mid-foot zone for the Atypical RF and under the mid- and fore-foot zone for the MFS.

Discussion and conclusion

The magnitude and the spatial distribution of the impact intensity over the different foot zones are related to foot strike pattern. Different foot strike patterns could benefit from different amounts of cushioning in different zones of the shoe to reduce impact intensity. This concept could be incorporated into individualized footwear.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Footwear influences soft-tissue vibrations in rearfoot strike runners

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Keywords: running shoe; acceleration; electromyography; muscle tuning; materials

Introduction

Muscle activity is tuned in response to ground reaction force to dampen soft-tissue vibrations (Wakeling, Von Tscharner, Nigg, & Stergiou, 2001) during whole-body vibrations (Wakeling, Nigg, & Rozitis, 2002) and running (Boyer & Nigg, 2004). A model study demonstrated this mechanism may be affected by fatigue and shoe hardness (Nikooyan & Zadpoor, 2012). Footwear may then

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influence soft-tissue vibrations, and thus the part of muscle activity affecting the muscle tuning mechanism.

Purpose of the study

To investigate the effect of midsole material (ethylene vinyl acetate – EVA – versus expanded thermoplastic polyurethane – eTPU) on (1) soft-tissue vibrations and (2) muscle activity.

Methods

Eight rearfoot strike runners performed two 2-min treadmill runs at 3.33 m s^{-1} in two different footwear conditions: EVA (Salomon XScream: mass = 300 g, drop = 9 mm; EVA midsole: hardness = 60C), and eTPU (Adidas Ultra Boost: mass = 302 g, drop = 8 mm; eTPU midsole: hardness = 40C).

Triaxial accelerometers (2560 Hz, 4524B, Brüel & Kjaer) were set onto *tibialis anterior* (TA), *gastrocnemius lateralis* (GL), *rectus femoris* (RF) and *biceps femoris* (BF). EMG activity was recorded at 2000 Hz for TA, GL, RF and BF with four SX-230 EMG sensors connected to a wireless DataLOG MWX8 biometrics system. Recordings were performed during the last 30 s of each trial.

Accelerations were converted in power spectral density (PSD) curves using a fast Fourier transform. *K*-means clustering was first used to discriminate high-PSD frequencies from low-PSD frequencies through the frequency range from 10 to 100 Hz for each acceleration component of each muscle. Frequencies below 10 Hz were excluded since they are related to the active motion. Soft-tissue vibrations were then quantified for each acceleration component of each muscle by calculating PSD within the respective high-PSD frequency range previously identified.

Muscle activity was quantified by calculating the root mean square (RMS) of EMG signals within the 100-ms pre-contact period, a 100-ms post-contact period and a global period gathering the two previously mentioned periods.

Paired *t*-tests were computed between EVA and eTPU for soft-tissue vibrations, and RMS ($p = 0.05$). The shoe effect was also assessed by calculated the Cohen's d coefficient.

Results

No difference in muscle activity was found. High-PSD frequency ranges identified were reported Table 1. There were no major discrepancies regarding the high-PSD frequencies identified in EVA and eTPU.

Vertical TA vibrations were greater for eTPU by 26.7 ± 20% ($p = 0.003$, $d = 0.74$, Figure 1). Anteroposterior RF was greater for eTPU by 42.6 ± 35.5% ($p = 0.02$, $d =$ 0.93). While there was no significant difference, medium and large effects were noted for the vertical and mediolateral RF vibrations ($p = 0.11$ and 0.09, $d = 0.52$ and 1.32, respectively). Similarly, a medium effect was observed for anteroposterior GL vibrations despite a non-significant difference ($p = 0.07$, $d = 0.62$). No significant effect or difference was found for BF.

Table 1. High-PSD frequency ranges (in Hz) identified from the cluster analysis for the three acceleration components of the four muscles.

	Vertical	Anteroposterior	Mediolateral
TA	[10:30]	[10:60]	[10:52.5]
GL	[10:60]	[10:57.5]	[10:40]
RF	[12.5:37.5]	[10:62.5]	[10:47.5]
BF	[10:37.5]	[10:52.5]	[10:37.5]

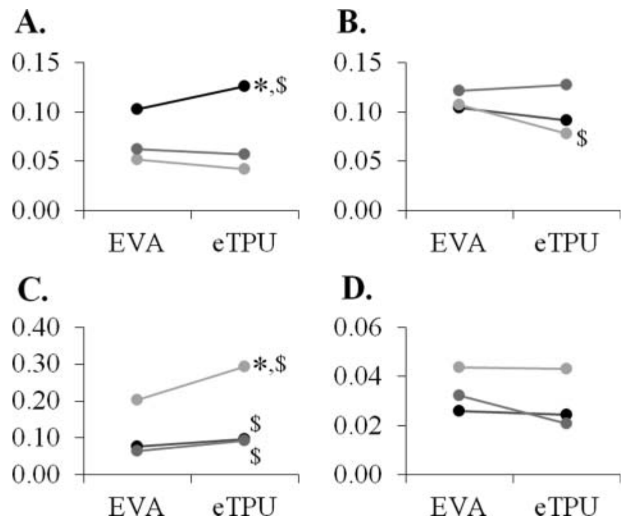

Figure 1. PSD (in g^2/Hz) for vertical (black dots), anteroposterior (light grey dots) and mediolateral (dark grey dots) acceleration components for TA (A), GL (B), RF (C) and BF (D) within their respective high-PSD frequency ranges. Significant differences and effects were noted as * and $, respectively.

Discussion and conclusion

This study demonstrates that footwear has a substantial effect on soft-tissue vibrations. It can be hypothesized that midsole properties (e.g. material, hardness) affect soft-tissue vibrations by changing the mechanical properties of the 'shoe-runner' system and/or by changing the input signal (Boyer & Nigg, 2004). In this study, an elastic midsole increased the power of soft-tissue vibrations.

The absence of change in muscle activity highlights that (1) footwear did not change the overall running kinematics emphasizing that the change in vibrations could be mainly attributed to footwear, and (2) there may be a certain threshold of increase in soft-tissue vibrations inducing a protective response at the muscle level.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Preliminary evaluation of prototype footwear and insoles to improve balance and prevent falls in older people

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Keywords: aging; falls; orthoses; functional footwear; gait analysis

Introduction

Footwear has the potential to influence balance in either a detrimental or beneficial manner, and is therefore an important consideration in relation to falls prevention. Based on previous studies, it has been suggested that the ideal safe shoe for older people at risk of falling should have a low, broad heel, a thin, firm midsole, a high collar and a textured, slip-resistant outersole (Menant, Steele, Menz, Munro, & Lord, 2008). However, although this recommendation is a valid summary of the available literature, few commercially available footwear styles incorporate all these features, particularly with regard to female footwear. Furthermore, in order for such a recommendation to be widely adopted, such footwear needs to be acceptable to older people from the perspective of comfort, ease of use and aesthetics.

Purpose of the study

The purpose of this study was to (1) evaluate balance ability and gait patterns in older women while wearing prototype footwear and insoles designed to improve balance, and (2) investigate older women's perceptions of the footwear.

Methods

Older women ($n = 30$) aged 65–83 years (mean 74.4, SD 5.6) performed a series of laboratory tests of balance ability (postural sway on a foam rubber mat, limits of stability and tandem walking, measured with the Neurocom® Balance Master (Natus Medical Inc., Pleasanton, CA, USA) and gait patterns (walking speed, cadence and step length measured with the GAITRite® walkway [CIR Systems, Inc., Franklin, NJ, USA]). Participants were tested in: (1) their own footwear, (2) flexible footwear and (3) prototype footwear and insoles designed to improve dynamic balance. Perceptions of the footwear were documented using a structured questionnaire (van Netten, Hijmans, Jannink, Geertzen, & Postema, 2009). The flexible footwear (Dunlop Volley™, Pacific Brands, Australia) had a rubber sole of uniform 18 mm thickness, a hardness of Shore A 35 and lace fixation. The prototype footwear was based on an existing model (Dr Comfort® Vigor, Mequon, WI, USA) and had a Shore A 55 rubber sole of 25 mm thickness under the heel and 18 mm under the forefoot, laces plus Velcro® fastening, a high collar to support the ankle, and a firm heel counter. The outersole was modified to optimise slip resistance by grinding a 10 degree bevel into the heel (Menz, Lord, & McIntosh, 2001), placing

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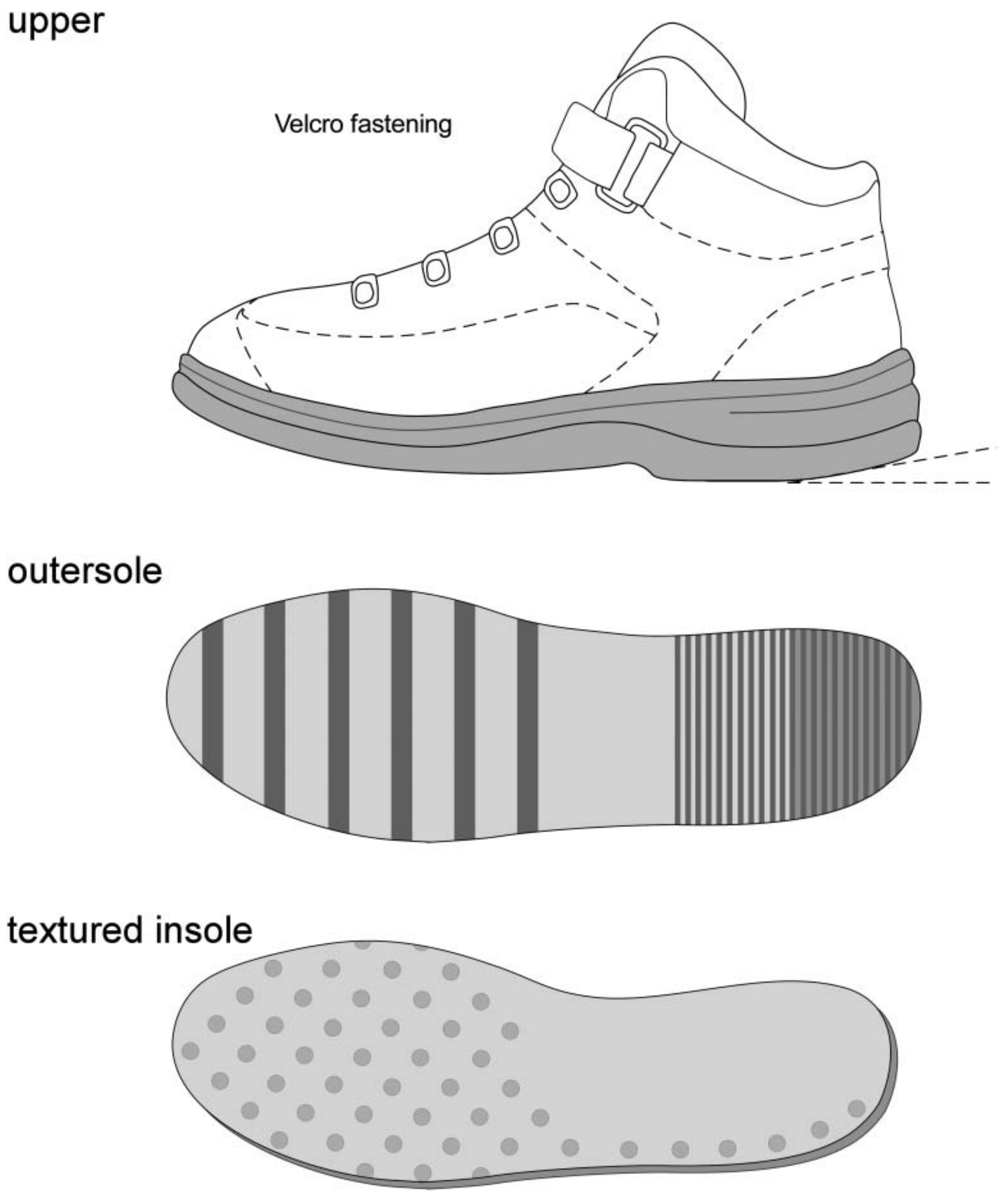

Figure 1. Prototype footwear and insoles.

grooves perpendicular to the sole across the heel (Liu, Lee, Lin, Li, & Chen, 2013) and placing perpendicular grooves across the rest of the sole (Li & Chen, 2004, 2006). The footwear also incorporated a textured insole, constructed from 4 mm thick ethyl vinyl acetate (Shore A 25) with dome-shaped projections (3 mm high and 8 mm diameter, Shore A 85) placed across the forefoot in a 15 mm diamond pattern and along the lateral border, extending to the heel (see Figure 1).

Results

There was no difference in postural sway, limits of stability or gait patterns between the footwear conditions. However, when performing the tandem walking test, there was a significant reduction in step width ($F = 9.3$, $P = 0.001$, $\eta^2 = 0.40$, large effect size) and end sway ($F = 5.6$, $P = 0.009$; $\eta^2 = 0.29$, large effect size) when wearing the prototype footwear compared to both the flexible footwear and participants' own footwear. Participants perceived their own footwear to be more attractive, comfortable, well-fitted and easier to put on and off compared to the prototype footwear. Eighteen participants (60%) reported that they would consider wearing the prototype footwear to reduce their risk of falling.

Discussion and conclusion

The prototype footwear and insoles used in this study improve balance when performing a tandem walk test, as evidenced by a narrower step width and decreased sway at completion of the task. However, further research is required to evaluate the footwear under more challenging conditions and to modify the design to make the footwear acceptable to older women from the perspective of aesthetics and comfort. Finally, to determine whether wearing such footwear can contribute to the prevention of falls, a randomised trial using prospectively documented incident falls as the primary outcome measure would need to be conducted.

Disclosure statement

The prototype footwear tested in this study was manufactured by a footwear company (Dr Comfort®, Mequon, WI, USA) with a view to making the shoes commercially available if they are effective, and the company believes there is a sufficient market for them. No commercial arrangements or royalty agreements have been made, as this is an early proof-of-principle study. However, there is a possibility that the researchers, in their capacity as staff members of La Trobe University, may obtain some commercial benefit if the footwear reaches the market in the future. The authors have not received any reimbursements and do not hold any stocks or shares in the company, and are not currently applying for any patents related to the content of the manuscript. There are no non-financial competing interests.

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Identifying representative test parameters to assess skin laceration injury risk for individual studs

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Keywords: rugby; biomechanics; boots; cleats; impact testing; sports injuries

Introduction

Skin injuries account for ~6% of all injuries in rugby union. Skin lacerations resulting from stud–skin interactions in rugby union are frequently caused by stamping in the ruck (Oudshoorn, Driscoll, Dunn, & James, 2016). Stud design is regulated by World Rugby's Regulation 12, but no supporting evidence currently exists for the selected test parameters used in these standards. Ideally, mechanical tests that assess injury risk should replicate conditions observed during play (Ura & Carré, 2016). Relevant mechanical test parameters, such as foot inbound velocity, stud impact energy, inclination angle and effective mass, can be derived through biomechanical analysis of rugby stamping. However, due to human movement

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Table 1. Stud impact kinetics and kinematics during rugby stamps (mean ± standard deviation).

Participant cluster	Stud energy (J)	Stud mass (kg)	Inbound velocity (m/s)	Inbound velocity angle (°)	Stud angle (°)
1	6.0 ± 1.9	1.4 ± 0.4	2.9 ± 0.6	33.6 ± 12.2	−4.3 ± 5.1
2	6.1 ± 3.0	1.5 ± 0.7	2.8 ± 0.4	25.2 ± 13.6	17.8 ± 4.7
Cluster A	6.0	1.5	2.9	29.4	6.7
3	8.2 ± 1.5	0.7 ± 0.2	4.9 ± 0.6	59.9 ± 6.5	27.4 ± 5.5
4	9.1 ± 1.9	0.8 ± 0.2	4.8 ± 0.5	36.4 ± 4.3	8.3 ± 4.7
Cluster B	8.7	0.8	4.8	48.2	17.9
5	11.0 ± 4.2	1.8 ± 0.5	3.5 ± 0.6	51.4 ± 12.3	−4.2 ± 3.9
6	11.0 ± 4.3	1.6 ± 0.6	3.9 ± 1.0	37.5 ± 9.2	25.5 ± 6.5
Cluster C	11.0	1.7	3.7	44.5	10.6
7	12.0 ± 3.3	0.9 ± 0.4	5.3 ± 0.9	46.6 ± 7.6	1.5 ± 8.0
8	12.0 ± 2.4	0.9 ± 0.2	5.4 ± 0.6	61.7 ± 5.1	3.4 ± 8.5
Cluster D	12.0	0.9	5.4	54.2	2.4

variability, the measured kinetics and kinematics of stamping impacts can have a large range and replicating all possible parameters within a mechanical test device is unfeasible. Identifying different stamp techniques by clustering provides an economical solution.

Purpose of the study

The purpose of this study was to identify representative impact values from rugby stamps for use in future mechanical tests.

Methods

Eight participants (mean ± standard deviation: age: 27.1 ± 4.4 years; stature: 174.1 ± 5.1 cm; mass: 76.2 ± 8.2 kg) were recruited; all procedures were approved by the Ethics committee of Sheffield Hallam University. During a rucking scenario, participants were asked to perform 10 stamps on an anthropomorphic test device (Hybrid III 50th percentile male), used as a surrogate player. Two high-speed cameras (Phantom Miro Lab 320) recorded the three-dimensional position of three shoe markers, used to determine shoe kinematics. Stud inclination angle was calculated using a modified approach to that of Driscoll, Kelly, Kirk, Koerger, and Haake (2015). Two pressure sensors (Tekscan, F-scan, 3000E 'Sport') recorded stamp pressure, from which force was derived. Effective mass (m_e, each stud) was calculated using the following equation, adapted from Neto, Silva, de Miranda Marzullo, Bolander, and Bir (2012):

$$m_e = \frac{\int_{t1}^{t2} F dt}{\Delta v} \tag{1}$$

with Fdt being stud force over time, t_1 is the time at the first impact, t_2 is the time at which the foot velocity is ~0, and Δv the velocity difference between t_0 and t_1. The mean and standard deviation of stud energy, inbound velocity magnitude, inbound velocity angle, stud angle and stud mass of each participant were calculated. Inter-participant parameters were clustered using impact energy (respective means) and test parameters for each cluster were calculated.

Results

Four impact clusters were identified (Table 1): 6 J (cluster A), 9 J (cluster B), 11 J (cluster C) and 12 J (cluster D). Clusters C and D have similar stud energies; however, impact energy of cluster C was associated with a lower inbound velocity (3.7 m/s) and higher effective stud mass (1.7 kg). Cluster D exhibited high inbound velocity (5.4 m/s) combined with low stud effective mass (0.9 kg).

Discussion and conclusion

Large variations in impact parameters, such as stud mass and inbound velocity, were observed during rugby stamping impacts. Clustering participants based on stud energy showed four generic movement solutions were used during stamping, ranging from 6 to 12 J. The identified clusters provide a combination of test parameters that can be used in a mechanical test to assess laceration injury risk of studs. Using clusters of impact parameters provides an economical means to determine the laceration injury risk of a stud, whilst maintaining fidelity to the conditions observed during play.

Disclosure statement

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Striking the ground with a neutral ankle angle results in higher impacts in distance running

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Keywords: foot strike pattern; impact forces; sagittal plane kinematics; atypical rearfoot strike; ankle angle; plantar pressure

Introduction

Running style is most often defined by foot strike pattern: rearfoot (RFS), midfoot (MFS) or forefoot strike (FFS). Different foot strikes have been related to differences in impact intensity. Impact intensity, measured as the instantaneous loading rate of the vertical GRF (VILR), has retrospectively been associated with an occurrence of lower limb stress fractures (van der Worp, Vrielink, & Bredeweg, 2016). In a previous study (Breine, Frederick, & De Clercq, 2014), we have discerned in distance runners Typical and Atypical RFS based on the centre of pressure (COP) trajectory. An Atypical RFS is identified by an initial COP position at the lateral side of the rearfoot immediately followed by a fast anterior displacement along the lateral shoe margin towards the midfoot zone resulting in an early first metatarsal contact. The relevance of distinguishing Typical and Atypical RFS lies in their associated VILR: Atypical RFS showed 40% higher VILR. Moreover, after distinguishing Typical and Atypical RFS, no difference in VILR was found between Typical RFS and MFS.

It is of specific interest if differences in known kinematic impact-reducing strategies between foot strikes can explain the higher VILR in Atypical RFS. Such strategies are an initial ankle plantar flexion and deformation of the cushioned rearfoot part of the shoe and the fat pad under the heel in RFS or an initial ankle dorsiflexion in MFS and FFS (Daoud et al., 2012).

Purpose of the study

The purposes of this study are (1) to assess kinematic differences between Atypical RFS, Typical RFS and MFS and (2) to correlate the observed kinematic differences with VILR.

Methods

Fifty-two runners (39♂, 13♀) ran over a 25-m track at 3.2 $m{\cdot}s^{-1}$ wearing the same running shoe (Li Ning Magne). Three left foot contacts were recorded. GRF and plantar pressures were recorded with a 2-m pressure plate (500 Hz, Footscan®, rs scan) mounted on a 2-m force plate (1000 Hz, AMTI). 3D-kinematics were recorded at 200 Hz with a 14-camera passive marker motion capture system (Qualisys AB, Gothenburg, Sweden). A 3-segment left leg kinematic model (foot, shank, thigh) was constructed in Visual 3D (C-motion, Germantown, MD, USA). Foot strikes were categorized as Typical RFS, Atypical RFS or MFS. Only one FFS trial was recorded which was not retained for further analysis. ANOVAs with *post hoc* analysis (Bonferroni, $p < 0.05$) were conducted to assess differences between foot strike patterns.

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Pearson correlation coefficients (r) and multiple linear regression models between VILR and selected kinematic variables were calculated. The correlation and multiple regression models were constructed for the Atypical and Typical RFS subgroup and for the Atypical RFS and MFS subgroup. These analyses were not conducted for the entire subject group since Typical RFS and MFS use opposing ankle strategies to reduce impact intensity which would impede linear modelling.

Results

For a description of the kinematic differences between the foot strike patterns, we refer to Breine et al. (2016).

In the subgroup of Typical and Atypical RFS, the regression model with the highest adj. R^2 of 0.526 revealed foot angle at touchdown (FA) and contact time (CT) as significant predictors of VILR:

$$\mathrm{VILR} = (-2.15 \times \mathrm{FA}) - (757.023 \times \mathrm{CT}) + 350.39.$$

This model was almost equally strong when using ankle plantar-dorsiflexion angle at touchdown instead of foot angle. In the subgroup of Atypical RFS and MFS, the model with the highest adj. R^2 of 0.319 used ankle angle at touchdown (AA) as significant predictor of VILR:

$$\mathrm{VILR} = (3.60 \times \mathrm{AA}) + 148.706.$$

Figure 1 shows the relationship between ankle angle at touchdown and VILR. Also, the significant correlation (r) between ankle angle and VILR is shown separately for both subgroups. Note that the direction of the correlation is reversed, which indicates the nonlinear relationship between ankle angle at touchdown and VILR.

Discussion and conclusion

Typical RFS, Atypical RFS and MFS are notably different running styles that differ both in kinematics and VILR. The different foot strike patterns use different impact-reducing ‘strategies’. The high VILR of Atypical RFS can be explained by shorter contact times and the nearly ‘flat’ foot and neutral ankle plantar-dorsiflexion position at touchdown. These mostly distal kinematics limit the use of known impact-reducing kinematics such as initial ankle plantar flexion in Typical RFS or initial ankle dorsiflexion in MFS.

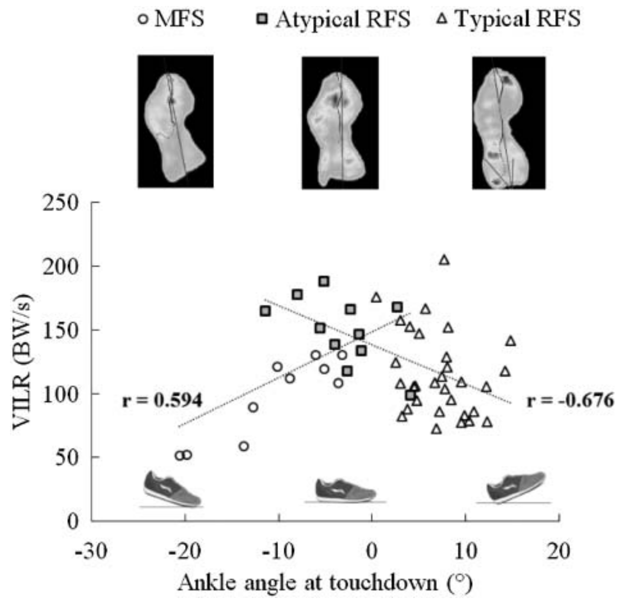

Figure 1. VILR and ankle angle at touchdown.

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Evaluating running footwear

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Keywords: injury risk; loading; pronation; preferred motion path; deviation; Joint coordination

Introduction

The explosion in the popularity of running and physical activity began in the 1970s. Along with this interest in physical activity came the development of sports shoes with an emphasis on the biomechanics of footwear. Since this interest in physical activity began, the evaluation of running footwear focused on two guiding principles: (1) decreasing the risk of running related injuries; and (2) improvement of performance. Most footwear researchers have focused on the former while fewer have concentrated on the latter.

The purpose of this lecture is to present an overview of the parameters used to evaluate footwear in an attempt to reduce the risk of injury leaving out the focus on performance improvement. I will present what parameters most footwear scientists investigated in the early years of footwear research and why we sought to link these parameters to injury. Lastly, I will try to look into the future and suggest some possible parameters that we may use.

Footwear evaluation in the early years

In the area of reducing the probability of running footwear as a risk factor for running-related injuries, two foci were studied extensively: (1) reducing the load on the body (i.e. cushioning the landing at the foot/ground contact); and (2) medio-lateral stability (i.e. controlling rearfoot calcaneal eversion or pronation).

The load on the body and the attenuation of this load via footwear primarily used ground reaction force (GRF) data. It was thought that the first peak and the loading rate of the vertical GRF component were critical factors in reducing the risk of impact-type injuries (Frederick & Hagy, 1986). However, we ultimately learned that high impact peaks or high loading rates did not relate to injury and softer materials did not necessarily result in better cushioning (Nigg, 2001).

There has also been a substantial amount of research on the injury risk associated with rearfoot stability with particular focus on the measurement of rearfoot eversion. The theory of rearfoot stability as a risk factor involved the coupling and timing of calcaneal eversion, tibial internal rotation and knee flexion during support. It was thought that too much eversion (i.e. 'excessive' pronation) and extraneous rearfoot movements increased the risk of injury. Neither of these assumptions has been shown to be true (Nigg, 2001).

Current evaluation of footwear

In many aspects, the current state of footwear evaluation has not changed the focus of footwear research. Many researchers are still using the cushioning or stability paradigm as a basis of injury risk reduction. A large number of these studies have contributed to interesting and beneficial footwear technologies but the ultimate goal has still not been achieved. What has changed is that the instruments and methodologies have allowed significant questions to be asked but still based on the paradigm previously used. Today, researchers use three-dimensional kinematics and kinetics, EMG, MRI, optimization, forward dynamics, etc., to analyse footwear. At the 2015 FBG Symposium, Dr. Darren Stefanyshyn, in his keynote lecture, suggested that quite possibly we have been looking at the 'wrong measures' in the evaluation of footwear. In the future, new measures or new paradigms must be evaluated if the goal of reducing the risk of injury is to be achieved.

Future possibilities in footwear evaluation

In his 2010 book (Nigg, 2010), *The Biomechanics of Sport Shoes*, Professor Benno Nigg suggested that we re-think our notions of cushioning and stability as risk factors for running-related injuries. Nigg's alternative paradigm was

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what he termed the 'preferred movement path'. He suggested that there is a subject- and task-specific locomotion pattern that is determined by many factors and that this preference for a particular motion path may explain why shoes have little effect on lower extremity kinematics. This new paradigm has given researchers 'food for thought' and has led to the new assessment tools such as how footwear may affect the determination of a runner's deviation from the runner's preferred path.

A runner's 'preferred movement path' may be interpreted in many different ways, one of which relates to the movement of the whole body system rather than a single joint movement. Joints and segments of the lower extremity interact to produce a smooth and efficient movement path. Thus, for an individual, there will be a movement path unique to them and, in Nigg's words, would result in significant muscle adaptations if there are deviations from the ideal path. Such an analysis of the coordination of the body when running gives rise to analysis techniques based on Dynamical Systems Theory (Hamill, van Emmerik, Heiderscheit, & Li, 1999). It may be this systems approach could lead to a more fruitful evaluation of running footwear that what was used in the past.

As an example, Taunton et al. (2002) identified the knee as the most prevalent site for running injuries. However, focusing on a single parameter such as calcaneal eversion may be too narrow a view and, as we have seen, produce mixed results. It may be that the hip or ankle is affected by footwear, and these joints cause deviations at the knee, with the knee being the site of the injury. By analysing how the knee interacts with other joints using a holistic or coordination analysis, we may then present a reason, for example, for the number of so many knee injuries in running. Then we may answer the question: can footwear reduce the risk of these injuries? The holistic analyses may also help reveal functional groups of runners to identify which runners are more or less susceptible to injury.

Conclusions

While running footwear can be a risk factor for injury, it is not the 'main' or most significant factor. The research conducted in the past has produced excellent footwear but we cannot state definitively that footwear has reduced the risk of running injuries.

Disclosure statement

No potential conflict of interest was reported by the author.

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Is consumer behaviour towards footwear predisposing for lower extremity injuries in runners and walkers? A prospective study

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Keywords: consumer behaviour; footwear; sports injuries; running; walking

Introduction

When looking for a new pair of running or walking shoes, the customer is overwhelmed by the possible choices. Advertisements with commitments of better stability, lower impact forces next to increasing speed, distance and performance, less fatigue, and reduction of injuries are

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legion. Although those promises, still many running- and walking-related injuries occur. Not all can be related to the footwear since the aetiology of those injuries is multifactorial (Hreljac, 2004). However, little is known about the relationship between the consumer behaviour towards footwear and the development of those injuries.

Purpose of the study

The purpose of this investigation was to investigate if consumer behaviour towards footwear is a risk factor for lower extremity injuries.

Methods

The consumer behaviour towards running and walking footwear was investigated in 300 runners and 280 walkers by means of a baseline questionnaire which included 28 questions concerning the basic decisions and influencing factors. The basic decisions imply those choices that a runner/walker makes before he gets to the store, including the place of acquisition, whether or not undergoing a gait analysis, the price, a second-hand buy, the replacement after a specific distance or time, the reason of acquisition, the influence of advice of others, and impulsiveness. The influencing factors include colour, model, material, closure mechanism, presence of specific properties, price, quality, price–quality ratio, sales and discounts, brand, fashion, advertisement, comfort, necessity, sport specificity, right fitting, technology, and store service. Information on injuries sustained during a 24-week period after the baseline questionnaire was obtained using a 2-weekly questionnaire. A running or walking injury was defined as a self-reported injury on muscles, joints, tendons, and/or bones of the lower extremities that the participant attributed to running or walking. The problem had to be severe enough to cause a reduction in the distance, speed, duration or frequency of running or walking, or treatment of the injury was carried out. Binary logistic regression analysis was used to identify risk factors for lower extremity injuries in the consumer behaviour.

Results

Data of 104 walkers and 104 runners who responded to all injury questionnaires and who did not change footwear during the follow-up period was used for further analysis. Forty-nine (24%) subjects suffered a self-reported lower extremity injury. Thirty-five injuries occurred in runners and 14 among the walkers. Logistic regression analysis showed that a gait analysis before buying footwear, not caring for the model or the closure mechanism of the shoe and feeling very much concerned about price–quality ratio were risk factors for lower extremity injuries. Buying shoes specific for the requested sport activity and buying the correct size decreased the risk.

Discussion and conclusion

The results of this study showed that buying footwear after a gait analysis increased the risk for a lower extremity injury in runners and walkers. Runners might think that after a gait analysis, they are protected against injuries but the contrary seems to be true. A possible explanation might be that they presume to have the perfect shoes with optimal protection against injuries after such an analysis. Consequently, they become unconsciously imprudent and take more risks. Another explanation might be that those subjects who have an injury history are more likely to undergo a gait analysis, hoping not to develop any further injuries by procuring individually adapted shoes. But in fact, they have a higher risk of developing another injury because injury history is the most predisposing factor for a new injury.

Next to that, results of this study also showed that not caring for the model and the closure mechanism of the shoe were risk factors for a lower extremity injury, while buying sport-specific shoes and correct size were rather protective against developing injuries. Therefore, runners and walkers should pay attention to the model, the sport specificity, the closure mechanism, and the correct size when buying footwear.

Since consumer behaviour towards footwear is a risk factor for lower extremity injuries, physical therapists and other medically trained co-workers might assist runners/walkers in buying correct footwear to prevent these injuries.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Improvement of fit of security shoes – evaluation of dynamic foot structure

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Introduction

The fit of footwear has been generally accepted as a key factor for foot health. One indicator for the quality of fit is an optimal match of the static shape of the foot and the shape of the last (Mauch, Grau, Krauss, Maiwald, & Horstmann, 2009). High correlation between the perceived fit of a shoe and the match of a last with the static foot shape was reported (Witana, Feng, & Goonetilleke, 2004).

Recent studies with children and adolescent feet have shown the need of dynamic foot shape information in order to describe foot structure more precisely (Barisch-Fritz, Schmeltzpfenning, Plank, & Grau, 2014) and the necessity of adjusting lasts with dynamic foot structure information for a better fitting of shoes (Barisch-Fritz, Plank, & Grau, 2016).

Research about the improvement of fitting of shoes was mainly done in sports and children but never at the workplace. Nevertheless, workers wear their shoes 8–10 hours a day and it is assumed that bad-fitting shoes are a causal factor in work-related injuries and overuse syndromes that will lead non-productive time at the workplace.

Purpose of the study

The purpose of this study was to investigate the dynamic foot structure at the workplace in order to optimize the lasts as well as shoe parts (e.g. upper) to improve the fitting of security shoes.

Methods

One thousand and twenty-four male and female workers with various work areas were measured at different companies in Germany. The scanner system, DynaScan4D, was used within this study to capture images of the foot during standing and natural walking. The measurement system and data processing as well as the description of foot measures and evaluation routines were in accordance with the description of Barisch-Fritz et al. (2014). Workers had to walk with a standardized speed of 4.5 km/h ± 5%. Static measures were conducted in a half weight bearing position.

The final sample sizes comprised 912 subjects (592 men and 320 women) after drop-outs. All foot measures were tested for normality by the Shapiro–Wilk test. Foot length, width, height and girth measures were normalized to the respective foot length (FL). To test the differences between loaded static and dynamic foot measures, a one-way ANOVA with paired Student's *t*-test was calculated. The level of significance was adjusted by Bonferroni correction. Finally, a multiple linear regression analysis was conducted for the dynamic maximum value (MaxDyn) and the difference to the corresponding static value (MaxDyn-HWB) of each foot measure. The variables age, BMI, gender and the respective static value for each foot measure were added to the model by the stepwise forward method to identify their influences on the outcome measures MaxDyn and MaxDyn-HWB.

Results

All length, width, height and angular foot measures increased from static to dynamic loading, whereas all circumference measures decreased. The overall decrease of the circumference measures was not systematic for all measures, as parts of the foot decreased very much and others only little or not at all. Age, BMI and gender had no influence on the outcome.

Nevertheless, there is some variation of each foot measure with respect to the differences between static and dynamic consideration.

Discussion and conclusion

The increase of length, width, and angular foot measures from static to dynamic loading might be explained by the additional loading of the foot during walking. The increase of the height measures as well as the decrease of

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all circumference measures might be explained by the muscular contraction during dynamics. The non-systematic decrease of the circumference measures might be due to the non-symmetric shape of the foot in the midfoot region.

Further investigation should be spent into the variability of the adaptation process from static to dynamic of each foot measure to find out if specific feet (e.g. flat feet, high-arched feet) behave differently in this process. Additionally, adaptations to the last and upper of a shoe, based on these findings – need to be tested to find out if the fit was practically improved.

Disclosure statement

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Elten Security Shoes.

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The influence of an off-the-shelf lateral wedge orthotic on knee loading during running

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Keywords: knee injury; insoles; joint loading; running; kinematics

Introduction

Knee injuries are prevalent in sports and associated with degenerative changes to the joint (Silverwood et al., 2015). High external knee adduction moments (EKAMs), knee adduction angular impulses (KAAIs) and knee flexion moments (KFMs) have been associated with increased cartilage deterioration (Chehab, Favre, Erhart-Hledik, & Andriacchi, 2014).

Lateral wedge insoles (LWIs) have demonstrated reductions in biomechanical loading (EKAM, KAAI) associated with osteoarthritis progression during walking in individuals with osteoarthritis (OA; Jones, Chapman, Forsythe, Parkes, & Felson, 2014). Younger individuals who sustain a knee injury during sport are likely to return to physical activity following treatment (Kim, Nagao, Kamata, Maeda, & Nozawa, 2013). With increased risk of developing knee OA, identifying preventative measures to delay the progression of OA during dynamic tasks such as running is required.

Previously, customised LWI have demonstrated reduced knee loading when compared to medial wedge insoles during running (Lewinson, Fukuchi, Worobets, & Stefanyshyn, 2013) but no difference compared with neutral insoles. Yet, participants reported discomfort with increased wedge thickness. An off-the-shelf LWI with medial arch support has shown improved comfort, most likely to ankle joint changes, whilst maintaining similar reductions in knee loading to LWI (Jones et al., 2014). This device offers the advantage of being available to all without requiring access to specialist podiatric or orthotic skill-sets. However, no data exists in more dynamic activities.

Purpose of the study

The aim of this study was to examine the effect of arch-supported LWI on knee loading during running.

Methods

Nine healthy individuals (age 25.1 ± 2.2 years, mass 68.2 ± 11.6 kg, height 1.7 ± 0.1 m), five males and four

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females, who ran at least 15 km per week for at least three months prior to data collection volunteered for this study. Two footwear conditions, a standard trainer (Decathlon Kalenji Running Shoes) and the standard trainer plus the arch-supported LWI (SalfordInsole™, UK), were assessed. Familiarisation to the conditions was given. For each condition, participants completed five successful 25-m running trials at 3.5 ± 0.2 m/s on a running track.

The CAST technique (Cappozzo, Catani, Croce, & Leardini, 1995) was employed to collect lower limb kinematic (10 Qualisys ProReflex; 240 Hz, Qualisys AB, Sweden) and kinetic (three force plates; 3600 Hz, AMTI, USA) data. A window was made in the heel counter to accommodate an additional wand marker on the lateral calcaneus defining calcaneus motion independent of the condition. Foot strike patterns for individuals were classified using the strike index and kinematic approach (Altman & Davis, 2012). Comparisons between conditions were assessed using dependent *t*-tests.

Results

Frontal and sagittal lower limb motion and moments were similar between conditions (Table 1). Center of pressure (COP) excursion demonstrated similar results between the two conditions. Foot strike patterns differed between the participants: five participants rearfoot, three midfoot and one forefoot.

Discussion and conclusion

The current study assessed the use of an arch-supported LWI on knee loading during running. The study showed that running with this device demonstrated no changes in lower body biomechanics.

Unlike the current study, Lewinson et al. (2013) reported lower knee loading with increased LWI thickness when compared with a neutral condition, although not significant. The variation in lower limb motion between individuals in the current study is likely due to the varied foot strike patterns reported.

Our findings suggest that arch-supported LWI do not reduce frontal plane knee loading in healthy individuals with varied foot strike patterns. Further evidence is needed to identify interventions on those predisposed to degenerative changes following a knee injury and effect of such interventions on pain during dynamic tasks.

Table 1. Kinematic and kinetic variables between the two conditions.

	Trainer	Insole
EKAM (N m/kg)	0.52 ± 0.23	0.54 ± 0.22
KAAI (N m/kg·s)	0.06 ± 0.03	0.06 ± 0.03
KFM (N m/kg)	2.65 ± 0.54	2.86 ± 0.60
Peak knee flexion (deg)	41.68 ± 5.13	43.05 ± 4.82
Ankle moment (N m/kg)	0.79 ± 0.18	0.79 ± 0.22
Maximum eversion (deg)	−7.68 ± 4.01	−9.21 ± 4.56
COP excursion (mm)	54.51 ± 17.52	55.07 ± 16.79

Disclosure statement

No potential conflict of interest was reported by the authors.

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Do minimalist shoes improve balance and foot strength in older adults?

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Keywords: minimal footwear; elderly; falls; foot; gait; hallux

Introduction

30% of people over 65 years old and 50% of people over 80 years old have at least one fall per year (NICE, 2013). One avenue explored to attempt to reduce this fall risk is footwear.

Minimalist footwear is a way to simulate walking barefooted, offering no constriction to the foot allowing it to function naturally whilst providing a protective surface. Aside from allowing the foot to spread it is suggested, the thin flat sole and absence of restricting motion control structures increases the sensitivity of the sensory mechanisms and better activates the foot and lower leg muscles (Mckeon, Hertel, Bramble, & Davis, 2015).

Previous research has indicated that both enhanced sensory feedback sensitivity (Machado, da Silva, da Rocha, & Carpes, 2017) and increased foot strength (Spink et al., 2011) improve balance in older adults.

Purpose of the study

To investigate if wearing minimalist shoes in daily life for a four-month period leads to improvements in balance and foot strength.

Methods

Twenty-six physically active adults (13 in intervention group, 4 males, 71 ± 4 years, 25.13 ± 4.36 BMI; 13 in control group, 3 males, 71.8 ± 3.7 years, 23.09 ± 3.43 BMI) volunteered.

All participants were assessed on static balance (postural sway; 3 × 30 seconds, parallel stance, eyes closed), dynamic balance (functional reach test (FRT); 3 × both sides), gait (3 × 10 m walk test, 3D motion capture), foot mobility magnitude and hallux strength (3 × hallux plantar flexion MVIC on each foot) at baseline and after four months of wearing a minimalist shoe (women's: Vivobarefoot™ Joy; men's: Vivobarefoot™ RA) in daily life.

Footwear use (time spent wearing the minimalist shoes/day) and habitual activity (step count/day) were recorded for one week intervals within each month.

Qualitative data was collected prior to the start of the follow-up session to evaluate their perceptions of the shoes and any changes felt in balance or confidence over the four-month period.

Statistical comparisons were made using paired sample *t*-tests ($p < .05$).

Results

The following results are from the 13 participants in the intervention group. Participants showed a good level of activity with an average of 8360 ± 4830 steps per day and spent 414 ± 151 minutes per day wearing the minimalist shoes.

There was an increase in FRT performance from baseline (30.45 ± 6.36 cm) to follow-up (33.98 ± 4.00 cm); $t(12) = 2.43$, $p = .032$ (Table 1). This corresponded with increased strength in the left hallux from baseline (37.54 ± 17.12 N) to follow-up (48.74 ± 24.00 N); $t(12) = 3.34$, $p = .006$ (Figure 1).

There were no significant differences in postural sway, foot mobility, gait timing variability, gait speed, step length, step width or stance phase duration.

Twelve out of the 13 participants reported improved confidence in their balance with 8 of those feeling their balance had improved since wearing the minimalist shoes and the remainder reporting no change.

Reasons they gave for the improvement were greater awareness of the surface as well as improved weight/pressure distribution across the foot due to the flat and wide shoe.

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Table 1. Functional reach test performance (cm) at baseline and follow-up (means ± s.d.).

	Right	Left	Average
Baseline	30.8 ± 6.7	30.0 ± 6.5*	30.4 ± 6.4*
Follow-up	34.1 ± 4.1	33.8 ± 4.5*	34.0 ± 4.0*

*Sig. diff. from baseline to follow-up ($p < .05$).

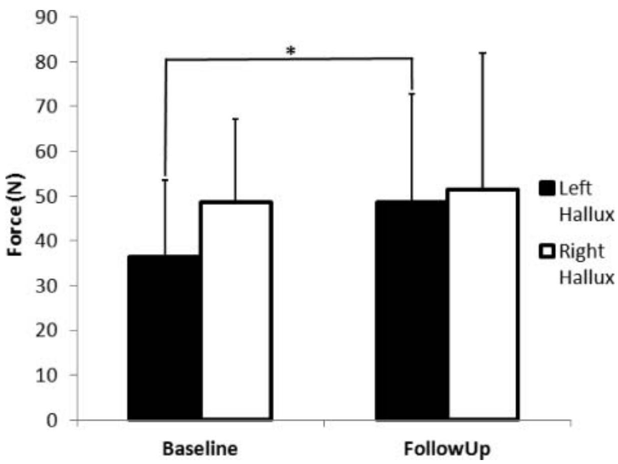

Figure 1. Hallux plantarflexor maximum strength at baseline and follow-up (means with error bars depicting s.d.) * = sig. diff. from baseline to follow-up ($p < .05$).

Discussion and conclusion

This study demonstrated that daily use of minimalist footwear could have a beneficial effect on the balance of older adults. Increased hallux strength and reach ability further suggest that walking in minimalist footwear activates the foot muscles to a greater extent, thus leading to improvements in foot strength and consequently balance.

Whilst we observed no improvements in postural sway or the gait measures associated with poor balance control, the qualitative data suggest that outside of a laboratory environment improvements have been witnessed in both confidence and balance itself.

The participants in this study were habitually very active with good baseline performance thus greater improvements may be seen in the less active amongst this population.

The intervention group will be compared to the control group upon completion of data collection.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Influence of the composition of artificial turf on rotational traction and athlete biomechanics

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Keywords: footwear traction; biomechanics; artificial turf; football; joint loading

Introduction

Advances in artificial turf technology enable these surfaces to behave similarly to natural grass; debate remains as to whether artificial turf is as safe to play on as natural grass. Previous research has shown that low rotational traction is associated with reduced lower extremity injury

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Table 1. Composition of different artificial turf surfaces for mechanical testing.

Surface #	Fibre length (mm)	Fibre density (oz)	Infill material	Compaction
1	12	40	Rubber	New field
2	19	40	Rubber	New field
3	25	40	Rubber	New field
4	19	40	Rubber	New field
5	19	50	Rubber	New field
6	19	50	TPE	New field
7	19	50	Cork	New field
8	19	40	TPE	New field
9	19	40	Cork	New field
10	19	40	Rubber	Compacted
11	19	40	Rubber	Heavily compacted
12	19	50	Rubber	Compacted
13	19	50	Rubber	Heavily compacted

(Wannop, Luo & Stefanyshyn, 2013). Therefore, to reduce athlete injury risk, sport surfaces should have low rotation traction. Artificial surfaces have a potential advantage as the individual components can be altered to change the surface properties and rotational traction. However, the degree to which each aspect of an artificial turf system influences footwear traction and corresponding athlete biomechanics is unknown.

Purpose of the study

To determine the influence that different components of artificial turf have on rotational footwear traction and lower extremity joint loading.

Methods

The project consisted of a mechanical testing phase and a biomechanical testing phase. For mechanical testing, the rotational traction of 13 different artificial turf surfaces were measured using a robotic footwear testing machine. Four components of the surface were systematically altered: fibre density (40 or 50 oz), fibre length (12 or 19 or 25 mm), infill material (rubber or thermoplastic elastomer (TPE) or cork), and compaction level (new field, compacted, heavily compacted) (Table 1).

The robotic testing machine consisted of a six degree of freedom P2000 servo-driven parallel link hexapod (Mikrolar Inc., Hampton, VA, USA). The turf to be tested was rigidly attached to the movable platform and a right prosthetic foot (fitted with an adidas 16.4 FXG US 10 cleat) was rigidly attached in series to a triaxial load cell. Rotational traction was tested with a normal load of 650N while the platform was rotated 20^{o} at a speed of 75^{o}/second. Rotational traction between the shoe and surface was defined as the peak moment about the vertical axis. A total of eight trials were performed at different locations on each surface. Following mechanical testing, three surfaces were selected for biomechanical testing: the conventional infill surface (control = surface 4), the surface that resulted in the highest rotational traction (high = surface 13) and the surface that resulted in the lowest rotational traction (low = surface 8).

Data were collected on eight male athletes to determine the peak moments and angular impulse at the knee and ankle when performing on the three different surfaces. Athletes performed a maximal effort running v-cut while kinematic and kinetic data were collected using an eight-camera motion capture system (240 Hz) and a force platform (2400 Hz). The box containing the surfaces (same surfaces used for mechanical testing) was securely attached to the force platform, while FieldTurf was laid along the path of motion creating a lab runway. For inverse dynamics analysis, the forefoot, rearfoot and shank segments were defined for the right leg of each athlete.

For mechanical testing, a two-way Univariate ANOVA with the level of significance set at $\alpha = 0.05$ was used. The dependent variable was rotational traction, while the independent variables were the fibre density, fibre length, infill material and compaction. For biomechanical testing, data were compared using a paired t-test at a significance level $\alpha = 0.10$. All comparisons were made relative to the Control condition.

Results

When investigating the overall influence of each compositional element of artificial turf, fibre density ($F = 20.873$, $p < 0.001$), infill material ($F = 3.208$, $p = 0.026$) and surface compaction level ($F = 11.813$, $p < 0.001$) all significantly altered rotational traction. The results of biomechanical testing indicate that the alterations in sport surface composition and the associated changes in rotational traction were sufficient to alter joint loading at the knee and ankle

Table 2. Ankle and knee joint moments and angular impulse. Bold indicates $p < 0.10$, * indicates $p < 0.15$.

		Peak moment (Nm)			Angular impulse (Nms)		
		Low	Control	High	Low	Control	High
Ankle	Transverse plane	**67**	75	82	12	13	14
	Frontal plane	111	120	129	14	15	16
Knee	Transverse plane	49	51	54	6.8	7.3	7.4
	Frontal plane	84	84	99*	18*	20	20

joint (Table 2), in addition to altering the ankle inversion angle during the cutting movement (Low = 16°, p = 0.138, control = 18° and high = 20°, p = 0.042).

Discussion and conclusion

The results of this study indicate that systematic alterations in artificial turf components can alter rotational traction and these changes in rotational traction are substantial enough to result in alterations in the joint loading of the ankle and knee. Using this data, artificial turf companies can alter the composition of sport surfaces to reduce the joint loading of athletes performing on these surfaces. Reduction of the joint loads should result in a reduction in knee and ankle injuries providing a distinct advantage for artificial turf when compared to natural grass.

Disclosure statement

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Subjective and biomechanical assessment of 'ride' during running

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Keywords: footwear; running; gait transition; plantar pressure; velocity

Introduction

A fundamental aspect, when choosing a running shoe, is the way that the shoe feels during a run. This topic is discussed regularly on running blogs and by footwear manufacturers and it is generally accepted that the 'smoother the feeling', the better the shoe. A term that is commonly used to describe the smooth feeling of the shoe during a run is 'ride', however, to our knowledge, this term has never formally been defined. We propose the following definition: 'ride is the feeling of the transition from heel to forefoot during the stance phase of gait' and suggest that an appropriate method to evaluate 'ride' is to quantify the peak anterior posterior (AP) velocity of the centre of pressure (CoP) between heel contact and toe-off.

AP velocity of the CoP has previously been used to characterize different phases of gait (Han, Paik, & Im, 1999), to distinguish between shoes and has even been used to define foot function (De Cock, Vanrenterghem, Willems, Witvrouw, & De Clercq, 2008), however, it has never been used to communicate the sensation of a shoe.

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Furthermore, a separate study collected subjective data on the smooth transition during gait, and related their findings to the kinematic rotation of the shoe (Sterzing, Schweiger, Ding, Cheung, & Brauner, 2013). Kinematic rotation only provides an indication of the movement of the shoe, whereas a more representative measure would be to examine the interaction of the foot within the shoe according to the pressure applied at the plantar aspect of the foot.

Purpose of the study

To investigate the association between the subjective assessment of 'ride' and the peak AP velocity of the CoP for two shoes with different midsole hardness.

Methods

Ten healthy, active males (30 ± 6 years) took part in this proof of concept study. They evaluated the ride of two running shoes (which differed only in the stiffness of the midsole: one stiff and the other, soft) while running over 160 m at approximately 3.5 m/s. Prior to running, participants were explained the definition of 'ride' and instructed to concentrate on the sensation of the foot transitioning through the stance phase of gait as they ran. Upon completing the running trials for each shoe, they evaluated the subjective 'ride' using a 15 cm visual analog scale (VAS).

PEDAR pressure insoles (sampled at 200 Hz; Novel GmbH, Germany) were placed in both shoes and plantar pressure was sampled from the right foot only. CoP displacement traces were calculated for each step, which were then time-normalized to 100 data points from heel contact to toe-off and amplitude normalized to the insole dimensions. Forty steps were selected and averaged for each shoe condition and participant.

Peak AP velocity was calculated as the peak slope of the normalized AP CoP displacement as the foot transitioned from heel to forefoot during the first 30% of stance. A paired t-test was used to determine if there was a difference in the peak AP velocity between the higher- and lower-rated 'ride' shoes ($\alpha = 0.05$).

Results

The shoe that had a higher VAS for 'ride' had significantly lower peak AP velocity (2.21% ± 0.09% insole length/% stance time) than the shoes with a lower rating of ride (2.48% ± 0.08% l/% s; $t_{1,9} = -3.93$, $p = 0.003$). This was true for 9/10 of the participants (Figure 1).

There was not a particular midsole stiffness that was consistently rated as a better 'ride'. In fact, half of the subjects (5/10) found the stiff shoe to have a smoother 'ride' (based on VAS) while the other half favoured the softer shoe.

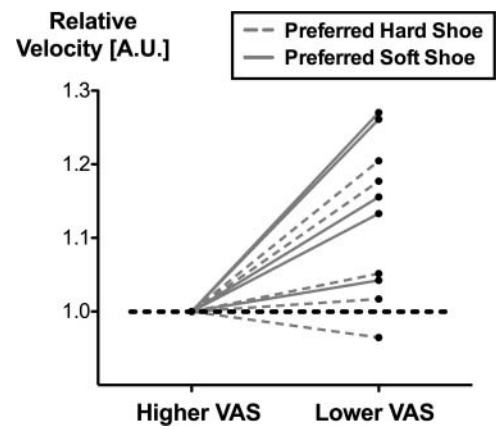

Figure 1. Relative AP velocity of the CoP (A.U.) normalized to the slope of the shoe with the higher rated 'ride' VAS. Dashed lines represent subjects who preferred the hard midsole shoe and the solid lines are subjects who preferred the soft midsole shoe.

Discussion and conclusion

For 9 out of 10 runners, a higher 'ride' rating coincided with a lower AP velocity of the CoP, verifying the association between the subjective and the proposed objective variables. Interestingly, it was not always the shoe with the softer midsole that had a smoother 'ride'. This implies that a shoe's ride is not simply driven by the construction of the shoe but-rather by the pairing of the shoe's design to an individual's anthropometrics or preferred movement path (Nigg, Baltich, Hoerzer, & Enders, 2015).

The outcome of this research could offer a novel approach for footwear companies to help identify appropriate footwear for an individual.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Adidas.

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Check for updates

Acute adaptability to barefoot running among professional AFL players

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Keywords: sports biomechanics; musculoskeletal loading; forefoot strike; technique; sports injuries; impact forces

Introduction

Match analysis of a typical Australian Football League (AFL) player shows frequent changes in movement direction, intermittent high intensity running, plus a high volume of low intensity jogging (Coutts, Quinn, Hocking, Castagna, & Rampinini, 2010). Many lower limb injuries in this sport can be traced to running technique and the impact forces between the foot and the ground (Zadpoor & Nikooyan, 2011). Loading rate (LR) is a variable commonly used to characterize the nature of the impact period. LR control is thought to be more effective when adopting a forefoot strike (FFS) technique compared to a rearfoot strike (RFS) technique (Lieberman, 2012). For shod athletes who run with an RFS technique, the LR is predominantly controlled by footwear. Habitual FFS runners have a reduced likelihood of injuries compared to runners that wear conventional running shoes and RFS (Altman & Davis, 2015). However, this theory is untested when elite athletes from football codes transition from hard to soft turf surfaces and from conventional running shoes to football boots.

Purpose of the study

We tested the hypothesis that players would demonstrate adaptability when transitioning from conventional shod to barefoot running, where: (a) foot strike patterns associated with high LR will be avoided; and (b) landing strategies when barefoot will be just as effective as conventional shod.

Methods

Thirty AFL players ran shod and barefoot across a 30-m runway striking two embedded force platforms (10 trials per condition). Full body kinematics was recorded using Nexus (Oxford Metrics Ltd, UK). Kinetics were also recorded. Data were exported in Visual 3D (C-Motion, USA) for analysis and calculation of LR and foot strike. LR was defined as the rate of change to GRFz between two threshold events relating to 20% and 100% of participant's body weight. This section of the force–time curve was selected because it avoids non-linearity (Samaan, Rainbow, & Davis, 2014). Foot strike was defined by identifying the ankle joint moment in the sagittal plane at initial foot contact with the force platform; an FFS pattern was identified when there was an internal plantar-flexor moment. Players were further categorized on their ability to reduce LR while running barefoot.

Results

When considering all trials as individual cases ($n = 600$), 70% of trials in the shod condition were RFS ($n = 211$), and 71% of the trials in the barefoot condition were FFS ($n = 214$). A higher LR in the FFS trials compared to the RFS for the shod condition was observed (effect size = 1.35), but FFS was lower than RFS in the barefoot condition (effect size = 0.43). Data were then stratified according to change in foot strike technique. Based on this criterion, each limb was classified into three sub-groups: persistent forefoot strike (pFFS); persistent rearfoot strike (pRFS); or an adopted forefoot strike (aFFS) if a case changed from an RFS when shod to an FFS when barefoot (Figure 1). These subgroups were further stratified according to LR response, where an ineffective response occurred if there was a significantly greater LR in the barefoot condition (i.e. when LRBF > LRSH; $p < .05$).

Discussion and conclusion

While runners can control the external GRF by utilizing both inherent neuro-mechanics and external attributes of footwear properties, the majority of AFL players from our sample mostly rely upon footwear. The shod FFS group had a higher LR compared to the shod RFS group, indicating that footwear appears to reward runners who select

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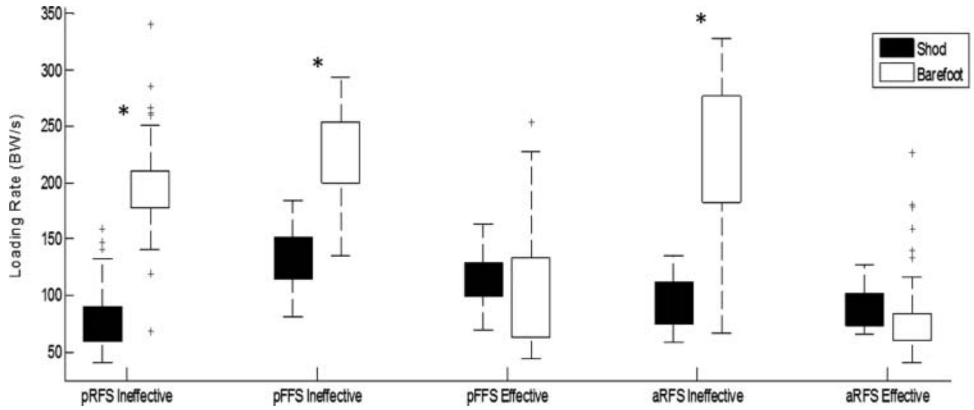

Figure 1. Box plots for LR. Groups 1–5 are based on: (i) LR response, (ii) foot strike type, and (iii) footwear. *Significant difference between shod and barefoot conditions, $p < .05$.

RFS over FFS. Other studies show that when shod, large dorsi-flexion foot strike angles correspond to small LRs. However, this reward might hinder adaptability to control the external ground reaction forces when foot-ground substrate conditions change.

We conclude there is no benefit to the athlete who adopts an FFS technique if they do not have the ability to adapt (as defined by decreased LRs) to that foot strike technique. The findings of this study have implications for understanding individual responses and running-related injuries for football code athletes when faced with stiff substrates, and potentially plays a role when advising running technique and footwear selection.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The relationship between footwear comfort and variability of running kinematics

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Keywords: running shoe; comfort; kinematics; biomechanics; running; Athletic footwear

Introduction

Footwear comfort plays an important role in enhancing running performance and in reducing movement-related injuries (Nigg, Baltich, Hoerzer, & Enders, 2015). However, the influence of footwear comfort on running kinematics is unknown leaving the benefits of comfortable footwear unexplained.

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In ergonomics, perceived comfort during sitting tasks has been linked to the degree of postural variability (Søndergaard, Olesen, Søndergaard, de Zee, & Madeleine, 2010). We hypothesized that a similar association is present between perceived footwear comfort and the variability of running kinematics.

Purpose of the study

The purpose of this study was to investigate the relationship between perceived comfort of running footwear and the variability of running kinematics.

Methods

Thirty-six recreational athletes (18 females and 18 males (mean ± SD) age: 25.4 ± 3.5 years) completed two separate testing sessions, 2–3 days apart. During each session, participants assessed the perceived comfort of five common running shoes with varying construction features. The comfort assessments included a comfort rating on a 10-cm visual analogue scale (VAS) as well as a comfort ranking of the five shoes. The most and least comfortable shoes were determined for each participant based on the highest and the lowest average comfort ranks across day 1 and day 2.

On day 2, each participant ran a total of 180 m in each shoe at a controlled speed of 3.5 m/s. Running kinematics were obtained using one inertial measurement unit (IMU, 3D accelerometer + 3D gyroscope, 250 Hz) securely attached to the laces on the dorsum of the right foot. For each shoe, 45 gait cycles were identified between two consecutive heel strikes and time normalized to 100% of the gait cycle duration. This analysis focused on the swing phase, defined as 50%–95% of the gait cycle, to avoid artificially high variability arising from signal noise around heel strike and toe-off.

For each participant, shoe, and IMU signal, kinematic variability was calculated across the swing phases of 45 gait cycles using a relative variability (RV) measure according to the following equation (Enders, Maurer, Baltich, & Nigg, 2013):

$$RV = \frac{RMS_{\text{Standard deviation}}}{RMS_{\text{Mean}}} \quad (1)$$

Comfort and relative variability were compared between the most and least comfortable shoe conditions for each subject using paired t-tests (normally distributed data) or Wilcoxon signed-rank tests (normality violated) ($\alpha = 0.05$).

Results

On average, the VAS comfort ratings within an individual for the most comfortable shoe were significantly higher by 3.6 cm ($p < 0.001$) compared to the least comfortable shoe condition, demonstrating relevant changes in perceived comfort (Mills, Blanch, & Vicenzino, 2010).

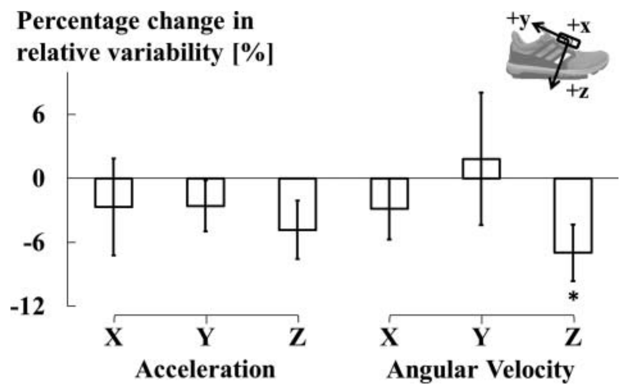

Figure 1. Mean (±SE) percentage change in relative variability of swing phase kinematics for six IMU channels in the least comfortable compared to most comfortable shoe ($n = 36$).

On average, five out of six IMU channels showed a reduction in relative variability in the least comfortable compared to the most comfortable shoe condition during the defined swing phase (Figure 1). For the angular velocity about the sensor z-axis, the variability in the least comfortable shoe was significantly reduced by 7% ($p = 0.005$). Similarly, the relative variability of the accelerations in sensor x- and z-directions were reduced by 2.7% ($p = 0.065$) and 4.8% ($p = 0.098$), respectively, showing a trend for reduced variability in the least comfortable shoe.

Visual inspection of the IMU signals indicated that signal waveforms exhibited a more repetitive pattern in the least comfortable compared to the most comfortable shoe, particularly during the late swing phase (75%–95% of gait cycle, Figure 2).

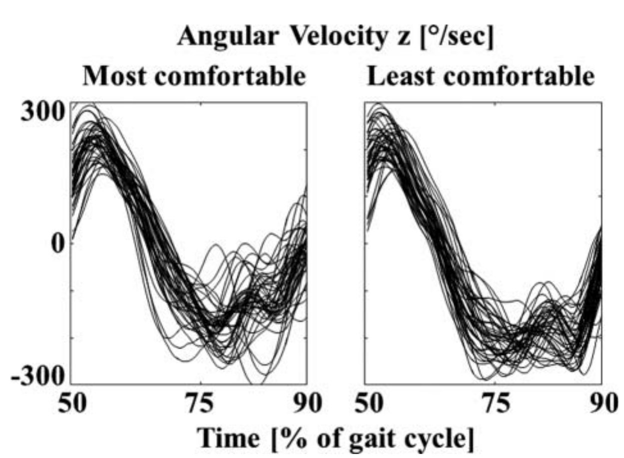

Figure 2. Angular velocity z for 45 overlaid gait cycles in the most and least comfortable shoe for one representative runner.

Discussion and conclusion

The findings of this study indicate that lower perceived footwear comfort is associated with a reduction in the variability of IMU-based running kinematics, particularly during the late swing phase of running. We speculate that a less comfortable shoe offers a lower number of solutions for a runner to execute the running movement comfortably, leading to a more repetitive kinematic pattern. This may be particularly true for the preparation of heel strike during the late swing phase.

These results warrant further research into kinematic variability as a variable that may partially explain the beneficial effects of high perceived footwear comfort with respect to running performance and injury risk.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Variability in foot contact patterns in independent walking in infants

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Keywords: pressure distribution; walking; methodology; ageing; biomechanics

Introduction

Independent walking in infants can be defined as the ability to walk 5–10 steps without support. Plantar pressures in independent walking in infants exist in the literature (Bertsch, Unger, Winkelmann, & Rosenbaum, 2004; Müller, Carlsohn, Müller, Baur, & Mayer, 2012), but data has been captured in environments which impact on the infant's typical gait patterns; for example, walking in straight lines for set numbers of steps being dictated and encouraged by researchers. This is likely to lower the gait variability, for example, by reducing the variability in progression angles, standardising the direction they are facing and increasing the number of steps taken in one walking bout (phase or trial). Thus producing data with low external validity in terms of the loading on the plantar surface of the infant foot in real-world environments. Additionally, mean data for groups of infants is presented in the existing literature, with no measure of the within-infant variability. This is despite consensus that reducing variability is key in development of experienced walking (Bertsch et al., 2004; Hallemans, D'Aout, De Cleroq, & Aerts, 2003).

Purpose of the study

The purpose of this work was to quantify plantar pressures and contact areas (and variability thereof) in natural independent walking in infants.

Methods

Data were collected on four infants walking independently in an infant lab (8–12 weeks of independent walking; 3 female). An Emed-xl platform (100 Hz; Novel, Munich) with a resolution of 4 sensors per cm^2 was positioned in the centre of a 2.0 × 2.3 m crèche scene. Two

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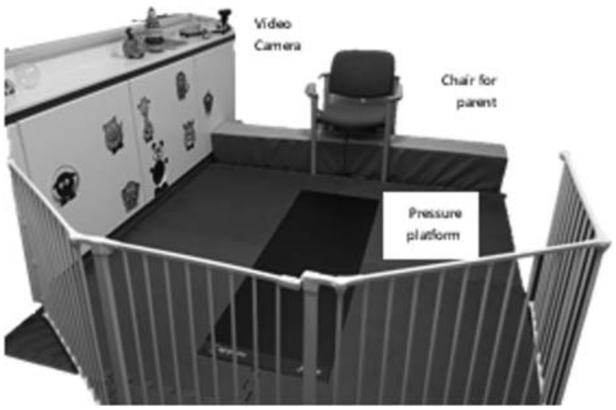

Figure 1. Laboratory environment for pressure data collection.

video cameras (50 Hz) in corners of the area were synchronised with the pressure platform. The environment included foam flooring, imagery, toys and cubes akin to a crèche (Figure 1).

Infants were accompanied by a parent in the cordoned (with baby gates) area for approximately 5 minutes while data were collected continuously. Infants were encouraged to move by the researcher and parent from standardised positions away from the platform. Otherwise, the infant was left to play and move around as they wished. Steps were manually extracted and analysed using Novel Multimask software. Steps analysed were dynamic steps where the infant was ambulating and foot contact was not on the platform border (confirmed using video), irrespective of velocity or direction of progression. Total foot contact area and peak pressures were evaluated; medians, inter-quartile ranges (IQR) and ranges are presented for comparison. No statistical comparison was undertaken at this stage.

Results

Nineteen to thirty-five steps met the criteria for inclusion from each infant (example 6 steps from one infant; Figure 2).

Mean total foot contact area was 45.1 cm^2 (range 42.6–46.4 cm^2) and IQRs for each infant were 9.9%–13.1% of the magnitude of the median total contact area (42.5–46.4 cm^2). The difference between the minimum and maximum contact patterns across steps for each infant was 22.8%–38.7% of the median total contact area for each infant – an average difference of 14.8 cm^2.

Median peak pressures across the total foot and all steps for each infant ranged from 105 to 165 kPa with IQRs of 35–92.5 kPa. Total foot peak pressure values ranged from 70 to 330 kPa across all steps and all infants.

Figure 2. Representative 6 peak plantar pressure steps of the total 24 analysed from infant 2.

Discussion and conclusion

Capturing ‘typical’ foot loading patterns during independent walking in infants denotes a highly variable contact pressure on the plantar surface of the foot. Mean pressure and contact area for the infants were similar to those calculated using the mean of 3–5 steps in a two-step protocol (Müller et al., 2012) and the mean of 5 steps collected in the middle of straight line walking (Bertsch et al., 2004). However, our work demonstrates that there is a large range of loading on the plantar foot within individuals around these mean values.

The infant who produced the maximum peak pressure value of 330 kPa moved faster in the lab and demonstrated the lowest contact area IQR. Supporting that variability of measures is relevant to quantify walking development maturity (Hallemans et al., 2003).

This research is limited by no observational analysis of infants in a familiar environment, e.g. at home. Additionally, capturing variability in a prescribed methodology would have aided a direct comparison and a true reflection of the threat to external validity the current methodologies incorporate. Our sample size is an evident limitation, which will be increased in continuing work.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Transfer of under-foot load and mechanisms of control in dart sports

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Keywords: darts; throwing; rearfoot; stability; forefoot

Introduction

In darts, as a sport, an athlete has an aim to score on targeted areas of the dart board, from a certain distance. During a competition, each player performs 150 throws in a real situation (Walsh, Tyndyk, Barton, O'Flynn, & O'Mathuna, 2011). On a professional level of competing, dart throwing movements last a very short amount of time <150 ms (Cordo, Carlton, Bevan, Carlton, & Kerr, 1994). Since the throwing is performed in standing position, it is believed that an athlete needs to have ideal balance in every possible aspect, and that the posture is 50% of the success. There are three basic stance postures: sideways, basic and frontal (Shilin & Kanevskaya, 2003). The basic stance is the one being used by over 90% of athletes. The basic weight of the body, in a basic stance, should be on the leaning foot, which should not be lifted from the floor during the throwing, and should not be changed from throw to throw. Whereas, the back foot is used for balancing and carries a needed amount of body weight in order to keep the ideal balance. With proper stance, the hand, forearm, upper arm, thigh and foot should be lined up. Feet stand at a shoulder width. During throwing, basic load is transferred on the frontal part of the foot (Shilin & Kanevskaya, 2003).

Purpose of the study

The study was performed with an objective to determine a mechanism of contact with the ground within dart athletes, during the throwing action.

Methods

The participant sample consisted of 27 healthy right-handed male dart players from the Serbian Second league dart competition 92.1 ± 14.9 kg, Sport shoes size EU 44 ± 1.6, height 182.5 ± 7.5 cm, BMI 27.8 ± 4.05, 32.1 ± 8.1 age (in years). The competitors used their metal darts, within the standard of this competing level (21.5 ± 2.16 g). For the dart throwing task, the height of the target centre was 1.73 m (+0.14 m) and participants stood 2.37 m away from the board; again, these are the standard measures according to international dart rules from the World Darts Federation. The throw was registered in real time in the room where the competition (classic dart 501) was taking place. In the study, we used a podometric platform RSscan, size 1.0 m × 0.5 m × 0.14 m, with a maximum operating frequency of 500 Hz. The platform was placed on the throwing line. The target was raised by 1.4 cm, which was the thickness of the platform. The first task the athletes were asked to do was to throw three darts in zone 20 (T1), in a 5 sec time period, where the second task was to use three darts to aim for zones 20–15–16 (T2), also within 5 sec. The athletes used their usual stance, as they would during a competition. The results of each measurement are registered in the software package RSscan. Balance was afterwards exported for further statistical analysis.

Results

During T1 throw, the sum of speed indicators CoP was registered for the participants 337.2 ± 169.1 mm/s, where

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Table 1. Parameters of pressure distribution by zones.

	Task 3 × 20					Task 20–15–16				
	Prepare		Realization		*t*-test	Prepare		Realization		*t*-test
	Mean	SD	Mean	SD	*p*	Mean	SD	Mean	SD	*p*
Left forefoot (%)	12.5	7.7	7.7	8.3	0.030	13.6	8.7	7.4	9.1	0.014
Right forefoot (%)	52.4	16.5	71.7	20.1	0.000	50.4	13.9	70.0	19.4	0.000
Left rearfoot (%)	0.07	0.4	0.0	0.0	0.322	0.1	0.6	0.0	0.0	0.322
Right rearfoot (%)	34.5	14.8	20.8	17.4	0.003	35.2	14.1	22.4	17.9	0.005

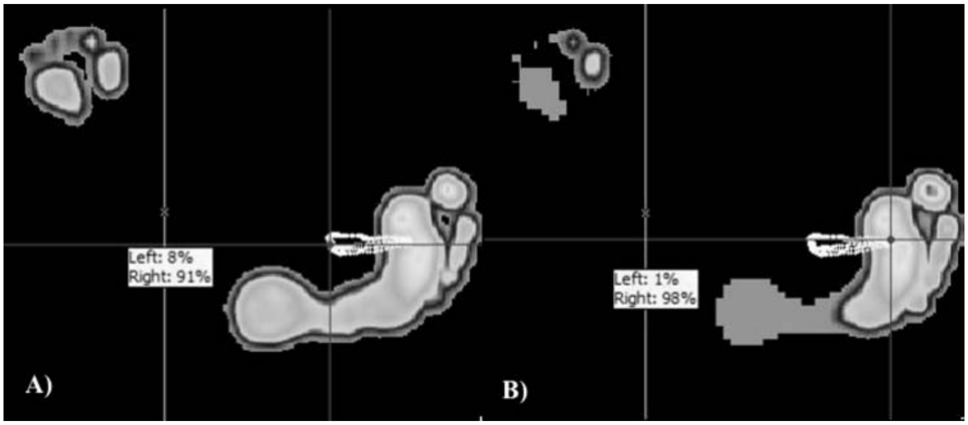

Figure 1. Preparation (A) and realization (B) phases while throwing.

for task T2, values 330.3 ± 157.5 mm/s, and they are showing no statistical significance. The parameters of mediolateral shifting (T1: 61.8 ± 43.3 mm; T2: 58.7 ± 32.5 mm) and anterio-posterior (T1: 30.5 ± 13.9 mm; T2: 30.3 ± 15.9 mm) as well as the length parameter CoP (T1: 299.4 ± 147.5 mm; T2: 92.4 ± 149.3 mm) were also not showing any statistical significance.

The registered parameters used during performing of the throws have shown that there are two phases: preparatory and realization phase. By all these means, it was found that the under-foot pressure parameters, within these (Table 1). It should be noted that no statistical significance in under-foot pressure parameters was found between tasks T1 and T2. In both cases, it was found that the greatest load was during the realization phase. The load is taken over by right foot (more than 90%), while the contact of the left foot is considered minor, since left rearfoot has practically no contact with the base/platform (Figure 1).

Discussion and conclusion

Comparing the results obtained in different tests, it has been shown that depending on the target needed to be hit on the dart board, the effect of mechanism is not reflected on registered parameter changes on platform. During the action of aiming and throwing, in realization phase, right-handed athletes transfer the whole load of their body on the front area of the right foot.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Influence of a training session on redistribution of underfoot pressure in basketball players

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Keywords: increase in volume; training session; right handed; basketball; foot pressure

Introduction

Certain habits acquired during sports, as well as the newly created conditions in which the athletes can be found on a flat, hard surface (Marchetti, Hartigan & Duarte, 2012), influence the sensorimotor system and the universal balance control, which generally can reflect in the abilities concerning balance (Bressel et al., 2007). Physical activity, which takes place in direct contact of both feet with surface, with the active involvement of the entire body and loads of lower extremities, while locomotion leads to an increase in volume and the support surface, which has already been confirmed in some studies (Cloughley & Mawdsley, 1995; McWhorter et al., 2006; Nagel et al., 2008).

Purpose of the study

The objective of the study was to confirm the influence of the effect of one training session on parameters of underfoot pressure and postural stability of female basketball players.

Methods

The participant sample consisted of 13 young healthy female basketball players from the Serbian First league basketball competition 69.5±10.1 kg, Sport shoes size EU 41.6±1.6, hight 176±8.78 cm, BMI 22.38±2.49, 16± 2.8 age (in years). Participants were eligible to take part in this study if they had no history of pathological change or structural damage to their lower limbs and had no known history of systemic musculoskeletal pathology. They had daily practice for at least 90 minutes and have participated in league competitions with at least one match per week. Measurements were performed before and immediately after afternoon training session, at usual training session time, in the middle of the competition micro cycle. In this study, we used a podometric platform RSscan, size 0.40 m x 0.5 m, with a maximum operating frequency of 300 Hz. The results of each measurement are registered in the software package RSscan. Balance results were later on exported for further statistical analysis. The testing, with open eyes, was conducted prior to and directly after the training session, where the respondents had to be maximally focused on a point that is at the level of their eyes, at a distance of 2 m. The duration of measurement was 100 s. When running, the test statistical evaluation of training data was done by *t*-test.

Results

Table 1 shows the differences found in the obtained results before and after training sessions. Four parameters have been indicated with statistical significance: left forefoot (%), right rearfoot (%), velocity of CoP, left foot velocity CoP, and top velocity CoP. Significantly higher load of the front of the left foot and significantly less load of the rear of the right foot – after training, it could be considered a direct consequence of the impact of training operators. The technical-tactical activities of both phases of the basketball game (defence and offense) are derived from the basic stance, which also includes the principle that the front part of the foot is more loaded. In this sense, a parallel can be withdrawn with the activity of marathon runners, in which the load and surge pulses increase and move, due to the impact of the marathon race in the front part of the foot (Nagel et al., 2008).

Discussion and conclusion

The increase of the volume of a foot, caused by training, reflects on an increase in support area, which directly

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Table 1. Parameters of pressure distribution and drift rate CoP.

	Before		After			
	Mean	Std. Dev.	Mean	Std. Dev.	*t*-Test	*p*
Left forefoot (%)	15.5	8.8	21.6	10.0	−2.97*	0.012
Right forefoot (%)	20.8	8.9	26.0	10.4	−2.07	0.061
Left rearfoot (%)	28.8	11.9	26.9	12.3	0.83	0.420
Right rearfoot (%)	34.9	11.4	25.7	8.5	2.82*	0.016
Dx (mm)	11.8	5.5	13.3	6.1	−0.63	0.540
Dy (mm)	21.8	16.0	27.0	14.5	−1.04	0.317
CoP(mm)	478.0	126.5	464.7	133.0	0.55	0.590
Velocity of CoP (mm\s)	4.26	1.23	3.90	1.12	2.39*	0.034
Left foot velocity CoP (mm\s)	4.32	1.37	3.68	1.16	2.36*	0.036
Right foot velocity CoP (mm\s)	3.70	1.49	3.61	1.41	0.37	0.717
Top velocity CoP (mm\s)	9.18	5.17	6.35	3.82	2.94*	0.012
Bottom velocity CoP (mm\s)	5.04	1.59	5.76	2.19	-1.81	0.096

influences a decrease of spatiotemporal parameters in those tests that were done after the training session. It has been found that the frontal area of the foot is by 11% more under load. Increased load of frontal area of the foot is a consequence of training on the mechanism of posture management, while redistribution of load of the right foot heel – before training session, on frontal area of the right foot – after training session, is a consequence of the influence of specific basketball activities applied during training sessions. This indicates that the dominant role of posture management process, as well as the program specific to basketball players, can be traced through the dominant role of the left foot in right-handed basketball players. Results lead us to think that, when choosing footwear for activities, it is necessary to take into account the influence activity has on distribution of underfoot pressure and its distribution in heel-toe relation. Undisputedly, an increase in volume is found, which should not be ignored when designing and choosing footwear.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Check for updates

Does a less torsionally stiff cycling shoe reduce knee moments during cycling?

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Keywords: cycling; knee injury; musculoskeletal loading; athletic performance; efficiency

Introduction

Knee pain is one of the most common overuse injuries across all levels of cycling (Clarsen, Krosshaug, & Bahr, 2010). Research has shown that the shoe-pedal interface is a critical factor that contributes to knee pain and that a pedal free to internally/externally rotate (i.e., float system) and/or abduct/adduct can reduce knee moments while cycling (Francis, 1986; Ruby & Hull, 1993). Changing characteristics of the cycling shoe, for example less torsional stiffness, may function similarly to a pedal that allows additional internal/external or abduction/adduction rotation and potentially reduce knee moments. How torsional stiffness of the shoe influences knee moments, however, is currently unknown.

Purpose of the study

This study determined whether a less torsionally stiff shoe can change a person's knee moments and performance while cycling.

Methods

Eight trained male cyclists (29 ± 8 years, 75.5 ± 7.5 kg) performed two tests on a cycle ergometer on two separate days (Velotron). On the first visit, power at lactate threshold (P_{LT}) was determined from an incremental exercise test to volitional exhaustion. P_{LT} was considered the power output at which blood lactate concentration rose >1 mM from the previous stage.

On the second visit, kinematic and force data were collected using a motion capture system (Motion Analysis Corporation) and 3D instrumented force pedals (Sensix), while participants cycled at 90% of P_{LT} in either a relatively torsionally flexible or stiff shoe. Gross efficiency was measured as a ratio of the work accomplished relative to the metabolic energy expenditure. Knee internal/external (I/E), adduction/abduction (Ad/Ab), and flexion/extension (F/E) moments were computed for the right leg. Moment profiles were computed for the right leg and averaged across 10 pedals revolutions. Differences between shoe conditions were determined by examining the peak moments and the summed moment difference (i.e., sum of the absolute difference in moment values at each point of the pedal stroke).

Results

There were no significant differences in the peak moments across the different shoe conditions ($F_{(2.3,15.9)} = 0.60$, $p = 0.70$). Individual analysis revealed that ~50% of subjects showed positive effects of wearing the more torsionally flexible shoe (i.e., responders). Specifically, the summed moment difference indicated that 4 of the 8 participants showed reduced Ad/Ab moments in the more torsionally flexible shoe (Figure 1) and similar results emerged for other moments (i.e., 4 of 8 F/E, 5 of 8 I/E).

It is apparent that only some individuals responded (i.e., benefitted) to the shoe technology. We defined responders as individuals that showed reductions in two of the three moment types. Table 1 displays the gross efficiency of the responders vs. non-responders for the two shoe conditions.

It is evident that 75% of the responders to the more torsionally flexible shoe condition showed improvements in gross efficiency that is likely attributed to a reduction in the knee moments produced while cycling.

Discussion and conclusion

There were no group differences in the knee moments when cycling in different footwear, however, individuals responded differently to the two shoe conditions. It is suspected that a more torsionally flexible shoe may be beneficial only for a specific group of individuals. For these individuals, this flexible shoe may reduce the risk of

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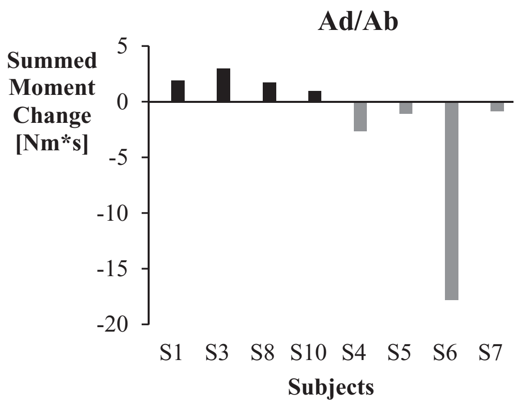

Figure 1. Summed difference for the Ad/Ab moment. The grey and black bars indicate participants that had lower and higher values for the more torsionally flexible shoe, respectively.

injury because of the lower knee moments (Wheeler, Gregor, & Broker, 1995) and may improve their performance because of a reduced energy cost associated with cycling in this footwear.

The results of this study suggest that certain shoe characteristics may influence individuals differently because these participants may belong to different 'functional groups' (Nigg, 2010). Thus, functional groups may explain why certain individuals respond positively to a less torsionally stiff shoe and others do not. It is proposed that these functional groups should be identified.

Table 1. Differences in gross efficiency for the responders (R) and non-responders (NR). A positive value indicates an increase in efficiency with the torsionally flexible shoe.

Subject type	Gross efficiency difference
NR	0.60
NR	−0.60
NR	−0.01
NR	−0.10
R	0.50
R	0.90
R	2.00
R	0.01

Disclosure statement

No potential conflict of interest was reported by the authors.

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Push-off apparent time delay based on COP trajectory of gait

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Keywords: gait; COP trajectory; apparent time delay; forefoot (foot); medial foot axis

Introduction

Currently, a number of pressure-based measures can be obtained from pedobarographic images, notably PP, PTI, area of contact, and COP displacement (Chiu, Wu, Chang, & Wu, 2013). However, data reduction that involves regions of interest (ROI), such as regional peak pressure (RPP), requires accurate region boundary definition which is typically based on regionalization schemes (Pataky et al., 2014) with COP trajectories (COPT) determined from the captured pressure (Khoury et al. 2015). In Khoury et al. (2015), the authors demonstrated that placement of orthotics controlled the COP trajectory and this feature could have implications for implant and orthotic design and practice. To fully benefit from this feature, it is

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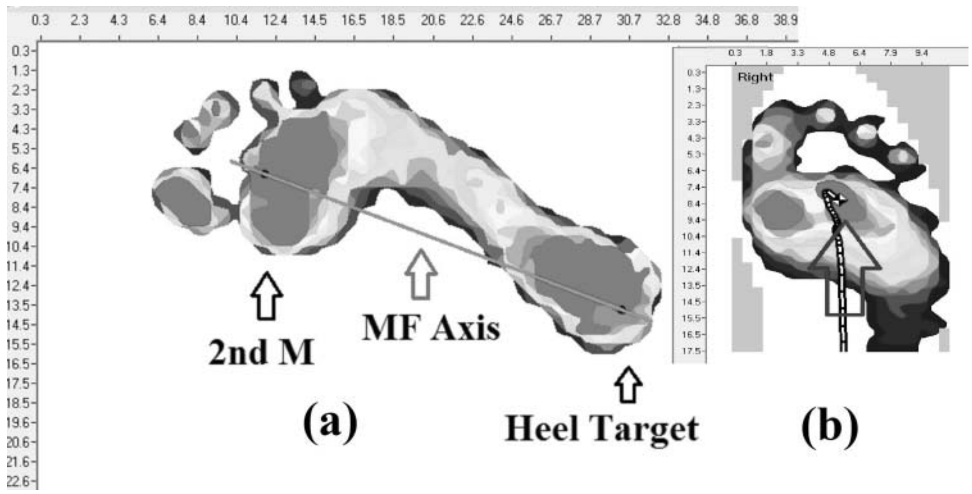

Figure 1. (a) Target imprint and the medial foot axis; (b) COP trajectory anomaly (arrow).

crucial to develop additional accurate trajectory-based and time-based measures, particularly during push-off at the forefoot where older adults experienced stiff toes. This proposed new measure could have clinical implications, particularly in fall prevention.

Purpose of the study

The purpose of this study was to investigate a method for determining an apparent delay measure (ATD) of the forefoot movement during push-off.

Methods

The gait time (GT) between the second MH contact and the foot leaving the ground was determined from the number of recorded image frames. The coordinates of the second MH and the terminus of the COP trace were used to calculate the apparent push-off distance (APoD). Gait speed at push-off was calculated by sampling a portion of the COP trajectory after it passed through the second MH separation to calculate distance travelled (SD), foot travel time (ST) and gait speed (GS). Apparent time (AT) was calculated using APoD and GS with apparent time delay (ATD) calculated by subtracting the AT value from the GT value. An individual's ATD was considered 'gait speed' normalized using the total time from heel-strike to push-off and was presented as a percentage value.

Three healthy subjects (age: 43, 68, 51 years; weight (kg): 62, 59, 66; height (cm): 165, 172, 167) were recruited. The second subject suffered from stiff (arthritic) large toe. After locating the second MH on the dorsal side of the foot, its position on the plantar surface was determined from a set of custom-fabricated devices. The pternion was marked and its conjugate position was extended to the edge of the plantar surface and then transferred to the medial foot axis 2 cm from the edge on the pedobarographic image. Image processing revealed the imprints in circular forms (Figure 1). Targets were affixed onto the plantar surface at these locations. Plantar pressure data were collected at the 20th step of a 25-step gait trial. Three sets of gait trials were collected per session/week for four weeks.

Results

Sample data are presented below (Table 1).

Discussion and conclusion

Often, 3D video and 3D analyses reveal gait aberrations at push-off where push-off occurs partially, as commonly found in older adult gait. Reported push-off impairments were related to fore-foot and toe stiffness, knee and ankle stiffness, postural stiffness, and shoe bending stiffness. The capacity to determine accurate ATD was made possible by precision targeting of the second MH and an accurate imprint of these targets on the pedobarographic image. The distance travelled by the plantar surface during push-off is related to the length of individual foot. This variation was normalized by applying the distance between the second MH and heel target. Gait speed of individual subjects affected the ATD value; however, this value was normalized by expressing it as the ratio of the ATD to time taken to complete the measured gait phase. This new, normalized, ATD measure could be used to study the effect of shoes-stiffness, guide placement of

Table 1. The mean ATD and normalized ATD measure (mean ± SD; max, min).

ID	ATD (s)	Time per phase	ATD measure (%)
1	0.01 ± 0.01	0.62	1.5 ± 1.2; 4, 0
2	0.02 ± 0.02	0.58	7.2 ± 2.8; 12, 3
3	0.01 ± 0.01	0.60	1.9 ± 1.5; 5, 0

orthotics in the forefoot area and assist in the design and determination of proper stiffness of orthotic materials.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The effect of foot type on the foot morphology and foot pressure of obese children

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Keywords: 3D scanning; foot morphology; children; obesity; flat foot; plantar pressure

Introduction

Previous study showed that obese children displayed flatter feet compared to their normal-weighted counterparts (Riddiford-Harland, Steele, & Storlien, 2000). Flatfoot presents a very high incidence rate in obese children. Therefore, obese children with flatfoot suffer not only from the excessive body mass, but also deformity of the foot structure.

Flatfoot, as a typical consequence of foot deformity in obese children, should be further investigated in the future study. However, to date, the variations in foot morphology and dynamic plantar pressure distribution of obese children with flatfoot have not been discussed in previous study. This study was designed to investigate the variations in foot morphology and plantar pressure of obese children with flatfoot compared to obese children with normal arch foot.

Purpose of the study

The purpose of this study was to investigate the effect of foot type (lower arch and normal arch) on the foot morphology and dynamic plantar pressure distribution of obese children. Are obese children with flatfoot at a high risk for foot deformity and foot injuries?

Methods

All the participants were selected from a foot morphology database of totally 551 children aged 7–16 years, including 280 boys and 271 girls. The arch structure was calculated simultaneously by Foot Angle and Chippaux–Smirak Index.

Totally, 30 obese children with flatfoot (OFF group) aged 7–14 years (9 girls and 21 boys) without other foot diseases and health problems were selected from the database. Another 30 obese children with normal arch foot (ONAF group), matched to their counterparts for age, gender, height and body mass index (BMI), were also collected from the same database. Obesity was defined by the BMI reference norm established by Group of China Obesity Task Force.

A Infoot® scanning system was used to scan and measure the external foot morphology. Totally, 10 observed values are supposed to be valid in this study. A footscan® plate system was used to test the plantar pressure distribution during walking barefoot.

The author chose right foot to perform the statistical analyses. Statistical analysis was conducted by SPSS17.0. The level of $\alpha = 0.05$ was perceived as significant.

Results

When the foot measures were normalized to the length of the right foot of each subject, significantly lower instep height and longer instep circumference were observed in the ONAF group.

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Table 1. Normalized foot measures (mean ± SD) for obese children with flatfoot and obese children with normal arch foot.

Items	ONAF group	OFF group	*p*-Value
IC	102.2 ± 3.0	105.7 ± 4.0	<0.001
IH	26.8 ± 1.7	25.4 ± 1.8	<0.01
NH	32.8 ± 4.2	31.1 ± 3.5	0.08

IC, instep circumference; IH, instep height; NH, navicular height.

The OFF group displayed significantly higher pressure rate and greater contact area beneath the midfoot region, while the ONAF group presented significantly higher pressure–time integral and force–time integral beneath the fifth metatarsal region.

Discussion and conclusion

It is speculated that as a crucial foot anthropometry, longer instep circumference in OFF group was found as a consequence of the increase of midfoot width. Therefore, small variations in foot morphology must be incorporated to the last design to meet the comfort and functionality requirements of specialized shoe for obese children with flatfoot.

The results of the study indicated a greater force–time integral beneath the midfoot region of obese children with flatfoot, which implied a potential higher risk for midfoot injuries than obese children with normal arch foot. We can speculate that collapse of arch structure deteriorates the plantar load condition of obese children with flatfoot during walking. Obese children with flatfoot could be at an increased risk for midfoot injuries such as stress fractures. The significantly decreased pressure–time integral and force–time integral for obese children with flatfoot beneath the fifth metatarsal. These results were in line with the previous literature which indicated that individuals with a flat foot could be at a lower risk for lateral column metatarsal stress fractures (Chuckpaiwong, Nunley, Mall, & Queen, 2008).

Table 2. Descriptive statistics of pressure–time integral and force–time integral for obese children with flatfoot and obese children with normal arch foot.

Items	ONAF group	OFF group	*p*-value
Pressure–time integral (N/cm^2 s)			
M5	1.95 ± 0.90	1.51 ± 0.60	0.032
MF	1.22 ± 0.57	1.38 ± 0.48	0.225
Force–time integral (N· s)			
M5	19.71 ± 10.44	15.10 ± 6.97	0.050
MF	47.48 ± 29.85	64.0 ± 28.99	0.034

M5, fifth metatarsal; MF, midfoot.

Disclosure statement

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Correlation of a non-weight bearing foot position to the neutral calcaneal stance position in an adult population

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Keywords: foot; foot measures; mid-foot; biomechanics; arch (foot)

Introduction

To correctly assess abnormalities within the foot, reliable and accurate measurements are necessary for clinicians to prescribe foot orthoses that effectively manage malalignment (Elveru et al., 2003). However, it has been highlighted that the clinical measurements used to

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quantitatively classify foot type and distinguish abnormalities have inadequate reliability and validity (Menz, 1998).

In 1971, Root and colleagues revolutionized foot biomechanics by assessing the level of deformity against an 'ideal' foot position (Root et al., 1971). In clinical practice, the 'ideal' position is weight bearing and is known as the neutral calcaneal stance position (NCSP) (Root et al., 1971, 1977). Clinicians compare the NCSP to a resting calcaneal stance position (RCSP) allowing compensations to be determined (Root et al., 1971, 1977). NCSP requires subjective positioning of the subtalar joint, by placing it in a theoretical neutral. Subsequently, authors have identified concerns relating to the Root paradigm's reliability and validity (McPoil & Hunt, 1995; Menz, 1998). As a consequence, researchers have proposed improved or alternative techniques for clinical, research or both purposes. Improvements in terms of reliability and validity with the proposal of the Navicular Drop Test and navicular drift test by Brody (1982) and Menz (1998), respectively, led to McPoil et al. (2008, 2009) introducing the foot mobility magnitude. This involved the assessment of dorsal arch height (DAH) and midfoot width (MFW) with the foot in RCSP and a non-weight bearing (NWB) position (McPoil et al., 2008, 2009). In comparison to the Root Paradigm, reliability and validity had significantly improved (McPoil et al., 2008, 2009). By using the NWB position in comparison to a relaxed weight bearing position, the issues confronting the Root paradigm may potentially provide an alternative 'ideal' position. Enabling the NWB position to replace the NCSP and enhance the efficacy and accuracy of lower limb interventions.

Purpose of the study

The purpose of this study was to determine whether a significant correlation between the NCSP and NWB foot position exists for the measurements of DAH and MFW.

Methods

Eighty healthy university students participated in this study. Apparatus used was constructed as described by McPoil et al. (2009). The NCSP was obtained by an experienced clinician (R. Scharfbillig), and confirmed by a third experienced clinician. A second examiner (H. Walker) conducted NWB measures by using a height adjustable plinth, to allow the foot to just be in contact with the surface of the apparatus. Measures of total foot length (TFL) and 50% TFL were recorded and marked on the left foot only. Using a digital caliper, measures of DAH (vertical arch height) and MFW (arch width) were obtained at the 50% TFL mark in NCSP and NWB positions. Reliability was assessed for the NWB position on nine participants by three examiners of varying clinical experience (0–30 years). Statistical analysis involved ICC (one-way random) for reliability and a Pearson product-moment correlation coefficient for correlational analysis (alpha ≤ 0.05).

Results

High intra-rater reliability was found for DAH and MFW with results of 0.90–0.99 (ICC) and 0.96–0.99 (ICC), respectively. Inter-rater reliability also had high reliability, with results of 0.90 (ICC) and 0.96 (ICC) for DAH and MFW, respectively. Analysis indicated a highly significant correlation between NCSP and NWB for the measures of DAH and MFW, with coefficient results of 0.820(r) and 0.86(r), respectively (significance = 0.000).

Discussion and conclusion

Results indicated that NCSP and NWB positions were significantly correlated when the measures of DAH and MFW were conducted. The high reliability indicates that these operator-dependent measures are reliable irrespective of examiner clinical experience. Validation of DAH has been investigated, with criterion validation via radiographs, however MFW validity is yet to be established (McPoil et al., 2008).

Clinical significance of these findings indicates that the NWB could potentially replace the NCSP as an 'ideal' position in comparison to a resting. By replacing an unreliable and invalidated measure and clinicians would be able to provide more accurate and consistent treatment strategies and improve the individuals overall outcome.

Future investigations should investigate the criterion validity of MFW and normal values for different populations and ages.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Available and proposed foot measures in terms of reliability, validity and applicability on an adult population: a systematic review

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Keywords: arch (foot); biomechanics; foot; foot measures; mid-foot

Introduction

For the prescription of foot orthoses that effectively manage malalignment, reliable and accurate measurements are necessary (Elveru et al., 1988). Foot biomechanics currently utilizes a method introduced by Root et al. (1971) of assessing the level of deformity in comparison to an 'ideal' position. This position is known as the Neutral Calcaneal Stance Position (NCSP), requiring the subtalar joint (STJ) being in its neutral position. The NCSP is then compared to a Resting Calcaneal Stance Position (RCSP) allowing abnormalities to be determined (Root et al., 1971). Due to the STJ neutral being a subjective, theoretical position, the authors have identified concerns with the reliability and validity of this approach (McPoil & Hunt, 1995; Menz, 1998). Subsequently, alternative and modified techniques have been proposed to improve this approach due to the implications posed on clinical and research practice (Elveru et al., 1988; Evans et al., 2003; Smith-Oricchio & Harris, 1990). Alternative techniques have included the Foot Posture Index (FPI), proposed by Redmond, Crosbie, and Ouvrier (2006), to classify foot type in a resting position. The Navicular Drop Test, proposed by Brody (1982), assesses arch height difference from 'ideal' to resting positions, to determine sagittal plane motion. A similar assessment, navicular drift, assesses the transverse motion (Menz, 1998).

McPoil et al. (2008, 2009) proposed the foot mobility magnitude (FMM). By assessing dorsal arch height (DAH) and midfoot width (MFW) in a non-weightbearing (NWB) and RCSP position, the vertical and medial–lateral mobility of the midfoot is quantified (McPoil et al. 2008; McPoil et al. 2009). With the numerous techniques of foot assessment accessible to clinicians and researchers within the literature, there is a potential to confront the Root paradigm and provides a reliable and validated alternative.

Purpose of the study

The purpose of this study was to critically appraise the available literature pertaining to the reliability and validity of foot measures that quantify or describe foot pronation/mobility, to determine if they are clinically and/or research appropriate tools.

Methods

A systematic search was conducted on six relevant databases. Reviewers (HW and RS) included quantitative full-text studies investigating the reliability and validity of static, clinically relevant foot assessments that were conducted on an adult population. Radiological and pressure investigations were excluded. A critical appraisal tool was implemented to assess articles for risk of bias (Brink & Louw, 2012).

Results

Fifty-five articles underwent data extraction, identifying 44 different foot assessments. Population characteristics included an age range of 14.1–85.1 years, suggesting a mature foot type. Six main categories of measurements were identified. These included those under the Root paradigm, Brody paradigm, Redmond paradigm, FMM/McPoil paradigm, footprint-based measures and 'Other' for measures that did not fit within these categories.

Discussion and conclusion

The six main paradigms reported in the literature all have a range of reliability and validity. While nearly all measures investigated reliability, less than half had validity investigations. This represents a considerable imbalance

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within research investigations, as while a measure's reliability is an important feature, validity of that measure is equally important to understand and determine if it is truly assessing what it proclaims to. The commonly used measurements under the Root and Brody paradigm contain questionable reliability and validity, undermining their current clinical use. The FMM/McPoil measurements are both reliable and valid and use NWB and weightbearing positions as comparisons, in a similar manner to which the NCSP is the 'ideal' position in the Root/Brody paradigm. The measurements under the FPI-6 and footprint-based are reliable and validated at classifying foot type as supinated, neutral or pronated, but assess the foot at one moment in time and do not compare it to an 'ideal' position. This makes it difficult to clinically classify as applicable and it does not aid or contribute to the treatment process of orthotic devices or other management therapies. It, therefore, may only be useful for understanding and classifying the individual's foot type for both clinical and research purposes.

Results indicate the need to determine if the method utilized by McPoil et al. (2008) (NWB compared to RCSP), by showing improvements in relation to reliability and validity, correlates to the 'ideal' position, NCSP and determine if it has the potential to be implemented within clinical practice. Furthermore, it is noted that more rigorous and succinct methodological procedures should be conducted to investigate the reliability and validity of foot posture measurements to improve overall podiatric care to patients.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Competitive study of stud characteristics on the penetrability

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Keywords: stud shape; stud area; impactor; football surfaces; football boots; penetrability

Introduction

Football boots as the only interface between surface and athlete play a decisive role in performance and injuries. The studs on football boots penetrate the playing surface and generate traction relative to the surface. Using football boots with an inappropriate stud length or area affects penetrability and consequently traction, stability and comfort (Dé & James, 2014; Hennig & Sterzing, 2010). The importance of stud characteristics is evident when we consider that excessive or poor traction are strongly associated with knee injuries (Alentorn-Geli et al., 2009). On the other hand, focusing on the issue of boot–surface interaction without considering playing surface is insufficient. Penetration behaviour is also influenced by the type of surface. Some studies have investigated appropriate studs for particular surfaces, but the lack of detailed

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understanding of the mechanism behind stud–surface interaction is still poorly understood.

Purpose of the study

The purpose of the study is to compare effect of different stud shape and the area changes (face area and base area) on penetration characteristics on two different surface conditions.

Methods

Stud: a single traditional football boot stud (Puma, King top), which is located at the medial forefoot was considered. Data from the face area and base area were calculated with pixel segmentation using a self-made program. The dimensions are as follows: face area = 77.24 mm^2 and base area = 195.76 mm^2. The face area and base area were magnified three and six times (Figures 1 and 2), and classified into the following three groups (although the length was only enlarged three times).

Group 1: Face area × 3 – Base area × 3
Group 2: Face area × 3 – Base area × 6
Group 3: Face area × 6 – Base area × 3

The three groups of studs were designed using Catia software in two different shapes: triangle and round. Eight studs were produced on a CNC machine.

Surface: A tray 25 cm by 20 cm was filled with sand. Particle size distributions of the surface material were defined according to DIN 18123. The main part of the surface was 52% medium size sand with particle sizes between 0.2 and 0.63 mm. The density was defined according to DIN 18127. Accordingly, it was densified with 15.29% water content to achieve a density of 1.72 g/m^3. A flat metal plate was used to compact the surface in the tray. Another experimental surface, Astroplay™ outdoor synthetic surface consisting of rubber particles, was also used in the experiment. The custom-made studs were mounted to an impactor hammer device and dropped from 20 cm onto the two different surfaces: synthetic turf and sand surfaces. The acceleration profile acquired data (6000 Hz) for three steps: (1) the starting point, (2) interaction with the surface and finally (3) back to the starting point. The maximum acceleration rate during interaction was measured for five trials.

Results

The maximum acceleration rate of each trial in group 1 and group 4 (section two) was compared with the *T*-test. The results revealed that there was significant difference between groups 1 and 3 in round shape ($P = 0.011$) and triangular shape ($P = 0.015$) only on the sand surface.

The effect of different perimeter length (shape) was evaluated for each group. In both surfaces, round and triangular stud shape were analysed and compared with the *T*-test. There was no significant difference when changing the perimeter length from triangle to round for each group on both surfaces.

Discussion and conclusion

In the first step, the effects of increasing base and face area were compared between groups (Figure 2). The findings illustrate that changing the face area can significantly affect penetrability as compared to the base area. Our study confirms the previous finding by Kirk (2008) who determined that there is a significant relationship between penetrability and cross-sectional area.

In the second step, the effect of stud shape was investigated using perimeter length for each group. The acceleration rates of round and triangular shapes do not reveal a significant difference. However, Déa and James (2005) showed a significant relationship between penetration and stud shapes by dropping the studs from a height of 60 cm. So, future research should be focused on different height ranges of dropping studs with respect to different shapes.

Disclosure statement

No potential conflict of interest was reported by the authors.

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A shoe with random midsole deformations increases ankle joint variability and muscle activations during the forward lunge

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Keywords: instability; footwear; kinematics; electromyography; lunge; dynamical systems

Introduction

Unstable shoes (USs) are designed to cause instability, which is suggested to elicit training benefits. However, studies of US are limited to standing, walking and running, and not all report increased muscle activations (Sacco et al. 2012; Stöggl, Haudum, Birklbauer, Murrer, & Müller, 2010) and balance enhancements (Ramstrand, Thuesen, Nielsen, & Rusaw, 2010). Perhaps US may provide a greater training stimulus during more challenging dynamic movements, involving longer balancing phases. The forward lunge is a closed kinetic chain exercise, frequently utilized in exercise regimes that is applicable to daily life and sports. An US with random midsole deformations increases shank muscle activations and joint variability during walking and running, which according to dynamics systems theory would unlock the degrees of freedom and increase adaptability to perturbations (Apps et al., 2017). It was hypothesized that this shoe with irregular midsole (IM) would also increase instability during lunges.

Purpose of the study

To compare the instability of the developed IM shoe and a control shoe (CS) during forward lunges.

Methods

Seventeen young, female gym class participants completed two sets of 10 right leg forward lunges in CS and IM in randomized order. Lunge frequency was controlled by a drumbeat (20 bpm), while step length was self-selected. The IM was created by placing freely moving ball bearings and cube shapes inside three-highly flexible rubber bags. The CS midsole was weight and medio-lateral shape matched to IM, both attached to the same shoe upper (Apps et al., 2016).

Three-dimensional motion capture (300 Hz, Vicon Peak, UK) and surface electromyography (3 kHz, Tele-Myo DTS, Noraxon Inc., USA) of five right lower limb muscles (tibialis anterior, peroneus longus, gastrocnemius medialis, vastus medialis and biceps femoris) were synchronized and recorded. Lunge characteristics of step length, step width and lunge contact time were derived. The variability (coefficient of variation (CV)) of sagittal and frontal ankle and sagittal knee joint ranges of motion were calculated in two phases during stance: initial ground contact until maximum sagittal ankle/knee angle (ROM1) and from this instant until toe-off (ROM2). EMG signals were normalized to the average CS peak values and mean values were obtained over the 100 ms before initial contact (pre-activation), as well as over ROM1 and ROM2 to compliment kinematics. Participant means were collapsed across both lunge sets and compared for the two shoe conditions by paired *t*-tests ($p < .05$).

Results

Stance time increased in IM ($p = .006$), but there was no difference in step length ($p = .537$) or step width ($p = .738$). Maximum sagittal angle increased at the ankle in IM, but were no different at the knee. Frontal ankle variability increased in ROM1 ($p = .001$) and ROM2 ($p = .0025$) and sagittal ankle variability in ROM1 ($p = .001$), but not ROM2 ($p = .100$) during IM lunges (Table 1). No knee joint variability differences occurred.

Peroneus longus (21%, $p = .006$) and gastrocnemius medialis (31%, $p = .040$) activations increased during ROM2 in IM, although there was large inter-subject variability (Figure 1). Gastrocnemius pre-activation also increased in IM (29%, $p = .038$). No differences were observed in the other muscles.

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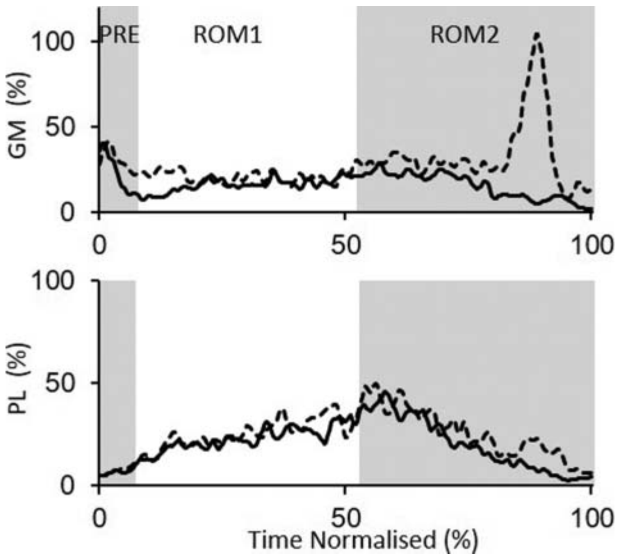

Figure 1. Mean normalized gastrocnemius medialis (GM) and peroneus longus (PL) activation during IM (dotted) and CS (solid) of a participant with increased activations in IM. Time is normalized between pre-activation, ROM1 and ROM2.

Discussion and conclusion

As hypothesized, IM increased instability compared to regular gym footwear during forward lunges. The increased frontal plane ankle variability caused greater peroneus longus activations to help stabilize the joint, which has implications for prevention and rehabilitation of ankle sprain injuries, as well as, activities of daily life and sports. The large gastrocnemius medialis activations in ROM2 in some participants relates to a plantarflexed ankle position during push-off. Future work should investigate if long-term training effects on ankle muscle strength and balance are greater in IM compared to regular gym footwear.

Table 1. Mean (SD) of kinematic variability (CV) of joint ranges of motion across participants. Significant differences in bold.

	Phase	CS	IM
Sagittal ankle	ROM1	14.0 (4.8)	17.0 (6.5)
	ROM2	15.3 (6.0)	19.7 (10.4)
Frontal ankle	ROM1	21.4 (8.3)	31.0 (9.5)
	ROM2	29.8 (14.6)	37.4 (11.9)
Sagittal knee	ROM1	4.8 (1.9)	5.5 (2.2)
	ROM2	5.9 (1.7)	6.0 (1.3)

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effects of habitual running shoe type on foot soft tissues' morphology

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Keywords: running shoe; ultrasound; plantar fascia; Achilles tendon; heel pad

Introduction

As running shoes are associated with changes in kinematics, kinetics and muscle activation during movement, habitual use of a specific type of shoe may also cause structural adaptations of the foot tissues. Soft tissues' morphological changes have been documented in response to altered loading patterns (Kjaer et al., 2006). Abnormal morphological changes in these tissues can be related to the risk of overuse injuries. However, there is little study to determine the effects of habitual use of

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different running shoes on the morphology of foot soft tissues.

Purpose of the study

The aim of the current study was to determine the association between running shoe type and foot soft tissues' morphology and arch deformation in healthy recreational runners.

Methods

Twenty-eight healthy recreational runners were recruited, whose habitual running shoes were categorized as conventional modern shoes, minimalistic shoes and conventional shoes with custom-made insoles.

An ultrasound system was used to capture images of foot soft tissues using a 10-MHz linear wideband array transducer. Ultrasound images of the following soft tissue foot structures were captured: plantar fascia (PF) at its proximal, middle and distal part; the Achilles tendon (AT) at its calcaneal insertion; and heel pad.

The arch height index (AHI) was calculated as the arch height at 50% the total foot length divided by truncated foot length in single-leg standing. The relative arch deformation (RAD) was obtained by dividing the difference of the unloaded arch height (sitting position) and arch height in single-leg standing by the unloaded arch height.

The differences of arch index, foot soft tissues' thickness between three groups were analyzed using one-way ANOVA.

Results

Recreational runners using minimalistic shoes demonstrated stiffer longitudinal arch than runners using conventional shoes did (Table 1).

The minimalistic shoe group also showed thinner plantar fascia (PF) at the proximal site and thicker Achilles tendon (AT) than other shoe groups (Table 2). Insole group had thinner heel pad than conventional shoe group.

Table 1. General information and arch height index for each shoe group.

	Conventional	Minimalistic	Insole
Number	11	7	10
BMI	22.7 ± 2.3	22.2 ± 2.3	21.7 ± 1.9
Age	24.6 ± 6.0	28.3 ± 8.4	25.1 ± 5.2
AHI	0.39 ± 0.03	0.41 ± 0.02	0.40 ± 0.04
RAD	9.6 ± 2.8	6.3 ± 3.3*	8.6 ± 2.7

Significant difference: *conventional shoe vs. minimalistic shoe.

Table 2. Mean (±SD) value of the thickness of plantar fascia, Achilles tendon and heel pad between shoe conditions (mm).

Soft tissue	Conventional	Minimalistic	Insole
PF–proximal	3.2 ± 0.4	2.9 ± 0.1*§	3.0 ± 0.2
PF–middle	1.7 ± 0.2	1.7 ± 0.2	1.7 ± 0.2
PF-distal	0.9 ± 0.1	0.9 ± 0.1	0.8 ± 0.1
AT	4.2 ± 0.5	4.6 ± 0.4*	4.2 ± 0.2**
Heel pad	11.9 ± 1.1	11.4 ± 1.4	10.3 ± 1.3***

Significant difference: *conventional shoe vs. minimalistic shoe; **minimalistic shoe vs. insole; ***conventional shoe vs. insole.

Discussion and conclusion

Our results showed that even though the arch index during standing was similar between the three shoe groups, the longitudinal arch stiffness in runners using minimalistic shoes was larger compared with runners wearing conventional shoes. The stiff sole of conventional shoes like conventional shoes will likely replace part of the work that should be done by the foot structures reducing the need for a stiff arch.

Since the minimalistic shoe runners run with their rearfoot not touching the ground (Lieberman et al., 2010), the calcaneal attachment site of the plantar fascia experiences little compressive force, which may result in a thinner PF. These runners also demonstrated thicker Achilles tendon compared with other runners. Probably, the non-rearfoot strike pattern adopted by minimalistic shoe runners increases the forces delivered by the calf muscles, and thereby, the Achilles tendon is thicker. The thinner heel pad observed in the custom-made insole group in this current study might be explained by the capacity of the heel cup to replace part of the function of the heel pad.

In conclusion, this study suggests that habitually running in a specific running shoe type is associated with the morphology of foot soft tissues. Therefore, this cross-sectional study suggests that runners should be aware of these morphological differences in foot soft tissue, especially when changing their running shoe type.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effects of shoe collar height on sagittal ankle mechanics during weight-bearing dorsiflexion movement and lay-up jump

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This study aims to investigate the effects of wearing high-top and low-top basketball shoes on the ankle joint kinematics, kinetics and performance in the sagittal plane during weight-bearing dorsiflexion (WB-DF) movement and lay-up jump. Twelve subjects performed WB-DF movement and lay-up jumps (LU) in two shoe conditions. Wearing HS can significantly reduce ankle joint excursion in WB-DF. No significant differences were found in jumping height and kinematics between the two shoes. In LU, peak plantarflexion torque and power were significantly lower in HS. The high-top shoes adopted in this study did not restrict the ankle dorsiflexion performance during actual jumping. But it is suggested that the greater power in LS during LU may have a greater potential in the actual game with different conditions, such as the distance of the shot and opposition. Thus, high shoe collar height would be applied to practical with the caution of affecting the partial kinetic characteristics of the ankle joint in the sagittal plane.

Keywords: shoe collar height; ankle joint; kinetics; kinematics; lay-up jump

Introduction

Jumping manoeuvres are basic forms in basketball training and competitions. These jumping/landing movements combined with high-intensity confrontation are major risk factors of ankle injury among basketball players (Leanderson, Nemeth & Eriksson, 1993). To reduce the risk of ankle sprains, high-top basketball shoes have been applied and showed potential in preventing ankle sprains in basketball. However, there is an allegation that high-top shoes may limit ankle dorsiflexion range of motion in certain scenarios (vertical jump and leap), and subsequently negatively affect athletic performance (Brizuela, Llana, Ferrandis, & García-Belenguer, 1997). Meanwhile, some studies have reported that high-top shoes and ankle braces restricted ankle dorsiflexion kinematics during landing and cutting manoeuvres (Lam, Park, Lee, & Cheung, 2015).

Purpose of the study

This study aims to investigate the effects of wearing high-top and low-top basketball shoes on: (1) ankle angle excursion during weight-bearing dorsiflexion movement; (2) ankle joint kinematics (contact angle and range of motion), kinetics (torque, power and stiffness) and sports performance (jump height) in the sagittal plane during lay-up jumps (LU).

Methods

Twelve male collegiate basketball players (age: 23.7 ± 0.6 years, height: 180.0 ± 4.6 cm, body mass: 73.6 ± 6.9 kg) with a minimum of four years of experience in basketball event were recruited for this study. High-top (HS) and low-top basketball shoes (LS) used in this study were obtained from the same footwear manufacturer. Both shoes had an identical design except for a 7 cm difference in shoe collar height. Weight-bearing dorsiflexion (WB-DF) test: each participant was required to stand in natural stance position with feet apart in parallel. After receiving the instruction 'Start' from the experimenter, the participant squatted with ankle dorsiflexed gradually until he could no longer subjectively squat or a heel off (lifting from the floor) could be observed. For the lay-up (LU), a common point-scoring manoeuvre in basketball, all participants stepped on the force platform with the second contralateral step after the first forward step, and then took-off to release the ball into the goal. 3D kinematics (Vicon, 120 Hz) and ground reaction force (Kistler, 1200 Hz) were measured simultaneously. The variables of interest included: the ankle angle at touchdown (θ_0), the maximum and minimum ankle angles ($\theta_{\min}$, $\theta_{\max}$), the ankle range of motion ($\theta_{ROM} = \theta_{\max} - \theta_{\min}$), the ankle angle excursion during the downward phase ($\Delta\theta = \theta_0 - \theta_{\min}$), the peak torque ($M_{\max}$), peak power ($P_{\max}$), and joint stiffness ($k$) of ankle joint and jump height (h), paired t-tests were used to determine the differences between

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high-top shoe and low-top shoe conditions (17.0, SPSS Inc., Chicago, IL, USA). The significance level was set at $\alpha = 0.05$.

Results

Sagittal plane ankle kinematics

In the WB-DF test, the $\theta_{\min}$ was significantly greater in HS compared to LS with a difference of 7.9° while the θ_{ROM} was significantly lower in HS compared to LS a difference of 5.2. However, during LU, no significant differences in the ankle angle at touchdown (θ_0), the minimum ($\theta_{\min}$) and maximum ($\theta_{\max}$) ankle angle, the ankle range of motion (θ_{ROM}) and ankle angle excursion ($\Delta\theta$) were observed between the two shoes. In addition, no significant differences were observed in the jump height between the two shoes.

Sagittal plane ankle kinetics

In LU, the peak torque ($M_{\max}$) and the peak power ($P_{\max}$) were significantly greater in LS compared to HS (Figure 1)

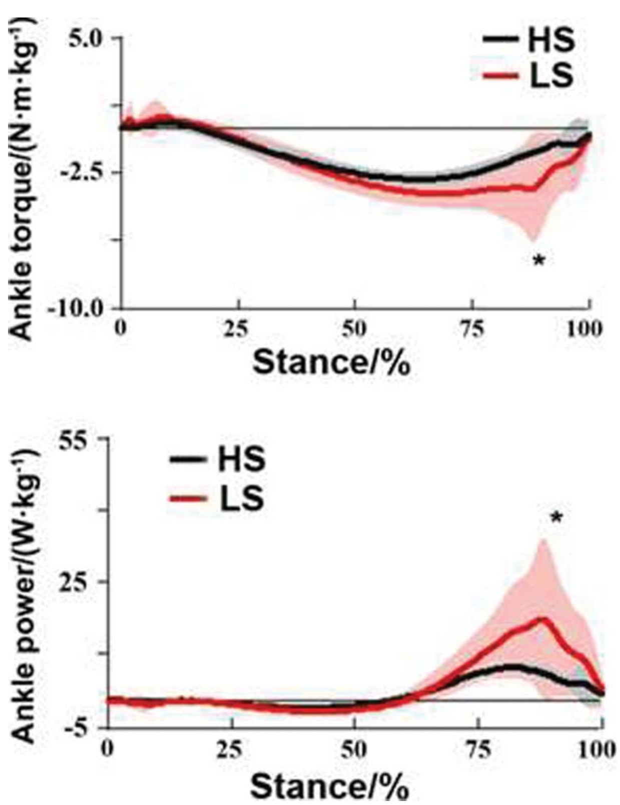

Figure 1. Comparison of the high-top (HS) and low-top (LS) shoes on the ankle torque (upper) and the ankle power (lower) during the stance phase of the lay-up jump.

while no significant differences were found in the k between the shoe conditions.

Discussion and conclusion

In WB-DF test, high-top shoes can effectively reduce ankle joint excursion in the sagittal plane. However, in LU, shoe collar height did not affect ankle kinematic characteristics in the sagittal plane. Therefore, the high-top shoes in this study do not restrict the flexion/extension performance of the ankle joint during actual jumping. On the other hand, different shoe collar heights did not alter jump performance, but can influence the peak plantarflexion torque and power of during push-off of LU. In most running and jumping manoeuvres, the performance depends on the total power output of hip, knee and ankle joints (Stefanyshyn & Nigg, 2000). Therefore, an increase in single ankle plantarflexion power does not necessarily lead to an improvement in jumping height. But it is suggested that the greater power in LS during LU may have a greater potential in the actual game with different conditions, such as the distance of the shot and opposition (Miller et al., 1993). Thus, high shoe collar height would be applied to practical with the caution of affecting the partial kinetic characteristics of the ankle joint in the sagittal plane.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The effect of prolonged standing on the body and the impact of footwear hardness

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Keywords: biomechanics; work shoes; discomfort; musculoskeletal; plantar pressure

Introduction

Prolonged standing is an occupational feature for around half of all workers. Risk of musculoskeletal disorders of the lower back, lower extremities and feet is increased in this population (Andersen, Haahr, & Frost, 2007).

Softer flooring with greater elasticity and decreased energy absorption has been associated with decreases in subjective discomfort (Cham & Redfern, 2001). Footwear and insoles have also been shown to influence discomfort during prolonged standing (Orlando & King, 2004) and have the advantage of being individual and portable.

However, the effect of prolonged standing and interventions on lower limb biomechanical parameters is not fully understood (Waters & Dick, 2015). Further, no research has investigated the effect of altering individual footwear parameters on the biomechanics of prolonged standing.

Purpose of the study

The purpose of this study was to investigate the impact of prolonged standing on the body and determine the impact of altering footwear density.

Methods

Twelve participants (*male*: 6, *female*: 6, *age*: 28 ± 5 years; *weight* 68 ± 11 kg; *height*: 1.7 ± 0.1 m) attended two sessions, 24 hours apart in which they stood for 3 hours whilst completing a series of continuous simulated stationary work tasks. These required standing and only minor shuffling of feet. Two versions (hard/soft) of surgical footwear were tested, manufactured from the same molds (Table 1). Participants were blind to this difference.

Every 30 minutes within the 3 hour task, subjective discomfort on a visual analogue scale and calf circumference were measured. According to a pre-determined schedule, the same tasks were completed every 30 minutes and biomechanical data was collected during a stationary manual task and a task that involved weight shifting as objects were moved, but without stepping.

Kinematic (100 Hz, Vicon), force plate (one foot/plate) (1500 Hz, Kistler) and in-shoe pressure data (50 Hz, Pedar) were measured.

A two-way within subject ANOVA with Bonferroni *post hoc* was used, with a significance level of 0.05 (SPSS).

Results

Over time, increases in *discomfort* for all regions (whole body, lower back, upper leg, knee, calf, ankle and foot) were found for both shoes ($p < 0.05$). Ratings of *shoe sole hardness* increased over time ($p < 0.05$) but did not differ between shoes. The increase in *low back discomfort* was greater in the harder shoe ($p = 0.047$).

Calf circumference also increased throughout ($p < 0.001$), the *centre of pressure* shifted laterally and internal ankle *inversion moment* increased ($p < 0.001$). Whole foot and heel plantar pressure increased over time (*mean pressure*, *max pressure, contact area* and *PTI* ($p < 0.05$)).

There was an interaction (i.e. shoe) effect for the *mean pressure* of the whole foot over time ($p = 0.047$), with the softer shoe increasing pressure at a greater rate. However, the absolute values did not differ between shoes as the softer shoe started with a lower mean pressure value.

Eight of the participants identified a preferred shoe (five softer, three harder). The preferred shoe had a greater medial midfoot *contact area* ($p = 0.04$) and a greater *internal inversion moment* ($p = 0.004$).

Discussion and conclusion

Prolonged standing increases numerous variables related to musculoskeletal disorders.

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Table 1. Difference in tested footwear.

Criteria		Softer shoe	Harder shoe
Hardness (cellular – Shore A/3 s)		34	38
Hardness (compact – Shore A/3 s)		76	69
Density (cellular g/cm^3)		0.22	0.20
Energy absorption (J)		31.2	34.2
Shock absorption	Deceleration (m/s^2)	100	120
	Penetration (mm)	8.0	6.5
	% energy return	32	29

Despite only small variances between the tested footwear, associated lower back discomfort and in-shoe pressure changes were observed between the shoes.

Large standard deviations indicate that individuals react differently over time and between shoes. Despite no subjective differences in the shoe hardness, two-thirds of participants stated a shoe preference with preferred shoes having a greater medial foot contact area and an increased internal inversion moment.

This study reinforces the need to use strategies to support musculoskeletal performance in workplaces that demand prolonged standing. It has demonstrated footwear hardness affects biomechanical variables and appears to be related to footwear comfort and preferences. Future research must investigate the effect of footwear design parameters on standing as well as to focus on the effect of individual differences.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Footwear-induced changes in ankle biomechanics during unanticipated side-step cutting in female soccer players

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Keywords: sex-specific cleat; lower extremity biomechanics; cleats; soccer; side-step cut

Introduction

Sex-specific soccer cleats have recently been supported in part due to the overall popularity of soccer and growth of female players worldwide (Althoff & Hennig, 2014). Additionally, sex differences in foot morphology exist, warranting the development of female-specific shoe lasts

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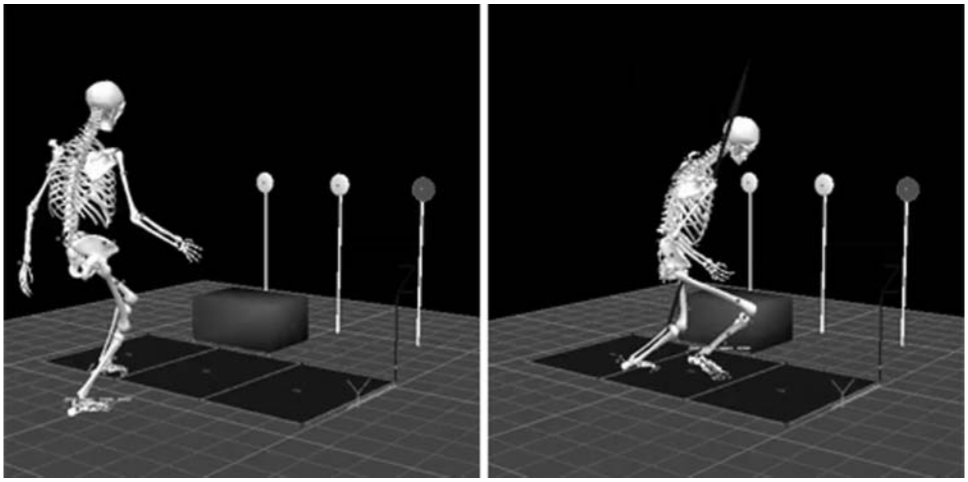

Figure 1. Side-step cut with light stimulus that provided a cue to either plant and back pedal or cut at a 90-degree angle.

rather than a scaled-down male shoe, as inappropriate fit may lead to higher injury risk (Krauss, Grau, Maiwald, & Horstmann, 2008; Kulessa, Golhofer & Gehring, 2017). Footwear should balance performance enhancement and reduction of injury risk. The most commonly injured body site in soccer is the ankle followed by the knee (Fong et al., 2007). Eighty-five percentage of ankle sprains are considered lateral ankle sprains with most common mechanism of injury involving ankle inversion (Ferran & Maffulli, 2006). In addition, a specifically debilitating injury in female soccer is a ruptured anterior cruciate ligament (ACL). We have previously shown that high-risk biomechanics, specifically larger knee abduction loads, predict future ACL injury in young female athletes (Hewett et al., 2005). Therefore, in order to decrease the risk of injury without compromising performance, research of female-specific soccer cleats during dynamic movements is necessary.

Purpose of the study

The purpose of this study was to determine the effects of sex-specific footwear on the biomechanics soccer players during unanticipated side-step cutting.

Methods

Twenty-four high school and collegiate female soccer players (height 165.2 ± 6.4 cm; mass 61.3 ± 9.5 kg; age 17.6 ± 2.9 years) were fitted with two pairs of soccer-specific cleats (adidas X 15.1 (control, CT) and female-specific prototype adidas ACE 17.1 (prototype, PT)). Unanticipated side-step cuts (Figure 1) with each cleat were performed on an artificial turf surface with data collected from 3D motion analysis cameras (Motion Analysis Corp), force platforms (AMTI), and in-cleat pressure distribution (Novel). Statistical comparisons between cleats were made using a repeated measures ANOVA ($p < 0.05$).

Results

Knee abduction moment was not statistically different between cleats during the deceleration phase of the side-step cut (CT: −61 ± 29.6 N m; PT: −57.7 ± 27 N m; $p = 0.53$). However, maximum ankle inversion moment (Figure 2) was significantly reduced in the PT compared to CT cleat (CT: 42.9 ± 18.2 N m; PT: 32.8 ± 14.7 N m; $p < 0.001$). Higher peak pressure was found in the lateral forefoot in the CT compared to PT cleat ($p = 0.017$). In contrast, higher peak pressure existed in the PT cleat compared to CT cleat in the medial midfoot ($p = 0.007$), lateral midfoot ($p = 0.01$), and central forefoot ($p < 0.001$). There were no differences in stance time ($p = 0.6$) or approach velocity ($p = 0.5$) between cleats.

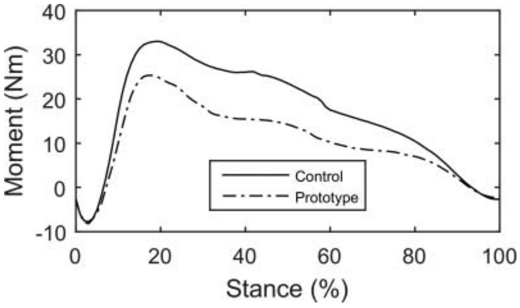

Figure 2. External ankle inversion (+) moment during stance phase of side-step cut.

Discussion and conclusion

The findings of this study indicate that while wearing the PT cleat, biomechanical differences exist during side-step cutting without increasing risk of knee injury (as evidenced by no change in knee abduction moment between cleats) and without sacrificing performance (as stance time and approach velocity were not different between cleats). The most salient and promising finding was decreased ankle inversion while wearing the new PT compared to standard footwear. This may indicate that risk for ankle sprains could be reduced and may relate to the female-specific last and stiffness properties used in the PT design (i.e. the last was narrower in the heel of the PT compared to CT condition). There was also a design change to the stud configuration between cleats, a property that can influence the contact area and traction between the cleat and surface (Kulessa et al., 2017). Within the central forefoot, the PT cleat had a more prominent stud which likely relates to greater peak plantar pressure, a measure that is widely used to evaluate comfort in shoes (Henning & Sterzing, 2010). Modification of the length or position of central forefoot stud should be considered. Overall, the findings of this study show potential, and encourage the continued research and design of sex-specific footwear for female soccer players.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Does heel offset alter tensile load in the Achilles tendon during treadmill walking?

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Keywords: heel height; tendon; speed of sound; pitch; toe spring

Introduction

Footwear remains a prime candidate for the prevention and rehabilitation of Achilles tendon injuries. Elevation of the heel with footwear has been suggested to decrease tensile load in the Achilles tendon (Reinschmidt & Nigg, 1995). Recently, however, standard running shoes with an inherent 10-mm heel lift were shown to increase tensile load in the Achilles tendon compared to barefoot (BF) walking (Wearing et al., 2014); hence, questioning the role of standard shoes in the prevention and management of tendinopathy.

Purpose of the study

This study used transmission-mode ultra-sound (US) to evaluate the effect of footwear, with differing heel elevation, on Achilles tendon loading and basic gait parameters. Given that the transmission speed of US is

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governed by the density and elastic modulus of tendon and is proportional to the tensile load to which it is exposed (Pourcelot, Defontaine, Ravary, Lemâtre, & Crevier-Denoix, 2005), it was hypothesized that the peak transmission velocity of US in the Achilles tendon would be lower in footwear with greater heel elevation.

Methods

Four pairs of prototype athletic shoes (S1–S4) were tested. Shoes were of identical size (US male 9) and assembled from the same last and constructed from the same materials (Figure 1(a)).

Shoes differed in heel offset, the differential in height between the rearfoot and forefoot, which was achieved by varying the graded thickness of the midsole beneath the rearfoot (heel offset [mm]: 0.4, 5.0, 9.9, 14.8; toe spring [°]: 22, 20, 19, 17). As a consequence, the mass also differed between shoes (weight [g]: 151, 162, 173, 181).

Axial transmission velocity of US was measured in the right Achilles tendon of 20 healthy adults (10 male; age, 31 ± 9 years; height, 1.72 ± 0.04 m; weight, 67.8 ± 14.2 kg) during shod and unshod walking on an instrumented treadmill (FDM–THM–S, Zebris Medical, Isny, Germany, Figure 1(b)). The treadmill measured vertical ground reaction force and spatiotemporal gait data (Figure 1(c)).

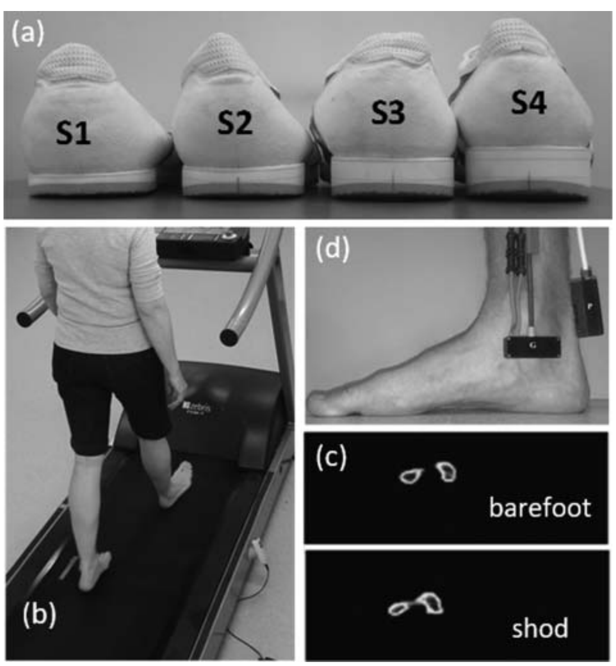

Figure 1. Four prototype shoes with heel offsets ranged from 0.4 mm (S1) though to 14.8 mm (S4). (a) An instrumented treadmill incorporating a pressure platform (b) measured spatiotemporal gait parameters and vertical ground reaction force during barefoot and shod walking (c). Placement of the electrogoniometer (G) and US probe (P) (d).

A custom-built ultrasonic device incorporating a probe with a 1 MHz emitter and four collinear, regularly spaced receivers was positioned over the midline of the posterior aspect of the right Achilles tendon, 1 cm above the calcaneus. An electrogoniometer (SG110A, Biometrics, Gwent, UK) quantified sagittal ankle movement (Figure 1(d)).

Following acclimatization, each participant's self-selected comfortable walking speed (1.02 ± 0.14 m/s) was used for BF and shod (S1–S4) conditions. After 6 minutes of steady-state gait, vertical ground reaction force, spatiotemporal gait data sagittal ankle movement and US velocity in the Achilles tendon were synchronously recorded for 60 seconds at 120 Hz. The order of each gait condition was randomized for each participant. Statistical comparisons were made using repeated measure ANOVAs ($\alpha = .05$).

Results

Compared to BF, shod walking conditions were characterized by significantly lower cadence, longer step length, greater ankle dorsiflexion, larger peak vertical ground reaction force and higher US velocity in the tendon (Table 1). Increasing heel offset resulted in a progressively lowered peak US velocity in the tendon but increased vertical ground reaction force peak one (Fz1, $P < .05$).

Table 1. Mean (SD) gait parameters.

	BF	S1	S2	S3	S4
Cadence (Hz)	0.91 (0.05)	0.87 (0.05)	0.87 (0.05)	0.87 (0.05)	0.87 (0.05)
Step length (m)	0.74 (0.06)	0.76 (0.05)	0.75 (0.05)	0.74 (0.05)	0.77 (0.05)
Fz1 (N)	741 (146)	783 (163)	790 (166)	810 (163)	800 (166)
Dorsiflexion (°)	6.3 (1.8)	7.1 (2.7)	7.1 (2.8)	7.3 (2.8)	7.3 (2.1)
US velocity(m/s)	2201 (99)	2248 (115)	2246 (120)	2240 (113)	2237 (114)

All conditions significantly different ($P < .05$).
Barefoot significantly different from all shod ($P < .05$).
‡Barefoot significantly different from S1 and S4 ($P < .05$).

Discussion and conclusion

Peak US velocity, and hence tensile load, in the Achilles tendon were higher during shod than BF walking. Although this effect was partially countered by a heel offset, an elevation in excess of 14.8 mm is required to return peak tendon loads to BF levels.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Calcaneal movement measured by skin versus shoe-mounted markers

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Keywords: calcaneus; running shoe; biomechanics; athletic footwear; gait analysis

Introduction

Rearfoot motion has been studied extensively in running and is frequently linked with injury due to its coupling with more proximal segments of the lower extremity. Rearfoot eversion is the common focus of investigation while movement in other planes is often neglected. Recent work has found that while eversion is linked with tibial internal rotation, a stronger coupling relationship occurs between tibial internal rotation and calcaneal movement in the transverse plane (Fischer, Willwacher, Hamill, & Brüggemann, 2017). Calcaneal movement in shod gait is frequently estimated using heel markers placed on the shoe heel counter, not accounting for potential heel movement within the shoe. Few studies have accounted for actual calcaneal movement using bone-pin markers; however, due to the invasive nature of these studies, the number of participants and the running velocity are low (Arndt, Westbald, Winson, Hashimoto, & Lundberg, 2004; Arndt et al., 2007). There is a need to quantify transverse plane in-shoe movement of the calcaneus using non-invasive methods while running at higher speeds.

Purpose of the study

The purpose of this study was to use non-invasive methods to evaluate the transverse and frontal plane movement of the calcaneus within shoe relative to the movement of the shoe heel counter.

Methods

Ten healthy recreational runners participated in this study (29.9 ± 6.8 years). Three-dimensional kinematic data were collected in a neutral shoe (Brooks Glycerin 13). To quantify heel motion, three markers were placed on the heel counter of each shoe. In addition, three wand markers were affixed to the skin of the calcaneus through three 25-mm holes cut into the heel counter (Figure 1). Previous work has established that holes of this size provide adequate space for marker movement without significantly affecting shoe integrity (Bishop, Arnold, Fraysse, & Thewlis, 2014).

Each participant completed five over-ground running trials at 3.5 m·s^{-1}. Kinematic data were filtered using a low-pass Butterworth filter with a cut-off frequency of 12 Hz. Rearfoot segment angles were calculated using a Cardan Xyz rotation-sequence in Visual 3D (C-Motion Inc., Germantown, MD) and were interpolated, normalized to 101 data points across stance, and averaged over five trials for each participant. For statistical analysis, paired *t*-tests were used to determine the significance of

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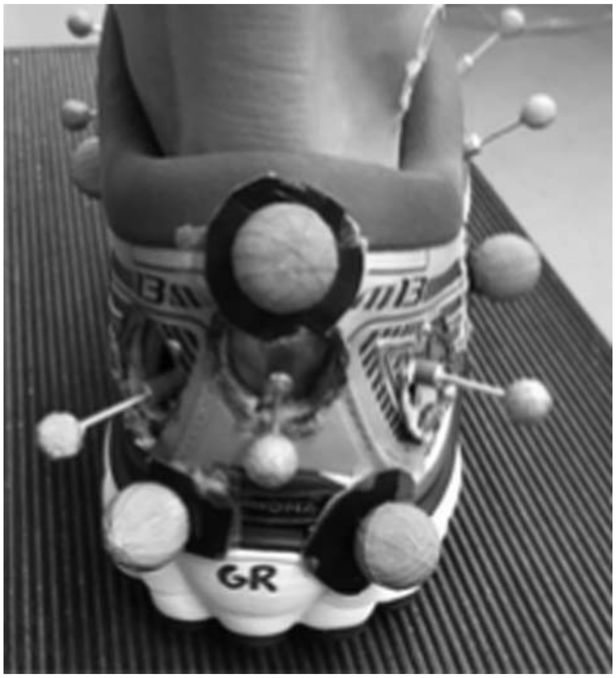

Figure 1. Marker set-up with three markers attached to heel counter and three calcaneal markers.

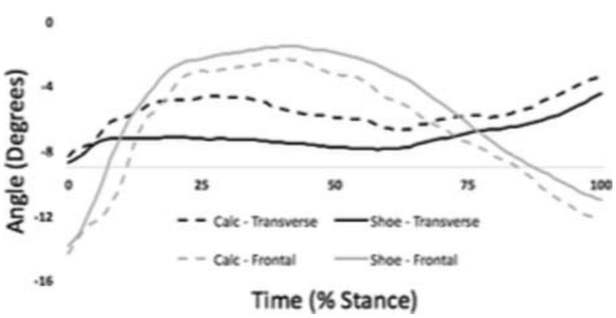

Figure 2. Transverse and frontal plane foot motion over stance for the calcaneus and shoe coordinate systems. +, adduction/inversion; −, abduction/eversion.

range of motion (ROM) in the first 50% of stance ($\alpha = 0.05$).

Results

For the segment range of motion in the transverse plane, there was a significant increase in ROM of the calcaneus relative to the shoe ($5.53 \pm 2.26°$, $2.76 \pm 2.14°$, $p = 0.010$), indicating greater calcaneal movement compared to the shoe heel counter. There were no frontal plane differences ($p = 0.187$).

Discussion and conclusions

The differences in transverse plane ROM between the calcaneus and the shoe heel counter indicate a lack of coupling between the foot and shoe. The stiffness of the heel counter in the transverse plane may resist this motion in the shoe itself. These results also indicate external shoe markers are insufficient to measure the underlying in-shoe foot motion in the transverse plane and that shoe movement may not be representative of the actual, in-shoe foot motion in that plane. Interestingly, in the frontal plane, the calcaneus and shoe heel were found to be coupled with each other. Thus, markers placed on the heel counter would appear sufficient to measure the actual calcaneal movement in the frontal plane without the need to place markers directly on the foot.

Further investigation on the effects of materials used for the heel counter and an effort to improve coupling between foot and shoe through new support technologies is warranted. By improving this coupling, it may be possible to reduce excessive movement at the knee as rearfoot and knee movements are coupled, particularly in the transverse plane.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

Brooks Sports, Inc.

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A proposed experiment: to assess the effect of football boot design on agile movement

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Keywords: stiffness, torsional; sports biomechanics; motion control; athletic footwear; joint coordination

Introduction

In field sports, agility is essential for performance and injury avoidance. It is partly determined by mechanical interactions of the shoe/boot and turf. Research has evaluated the effect of footwear on movement agility by typically performing a biomechanical analysis of cutting manoeuvre tasks (Kulessa, 2017). This approach incurs limitations from a motor control perspective. During movement control tasks, the nervous system regulates the effective limb length (vertical component) and orientation (horizontal component) rather than element details (Auyang & Chang, 2013; Bosco, Eian, & Poppele, 2006). Quantifying limb effector control requires a sufficient set of trials, underscored by consistent initial conditions and a prescribed performance outcome (Latash, 2008). Achieving limb effector repeatability is difficult using pre-planned cutting manoeuvres and context validity is also limited. We sought to develop a novel test method to overcome these limitations.

Purpose of the study

To assess the effect of football boots on limb coordination, by employing the Uncontrolled Manifold Hypothesis (UCM, Scholz & Schoner, 1999).

Methods

A healthy male adult gave informed consent. Five different pairs of boots were selected based on differences in their combined outsole stiffness and stud geometry. The movement protocol included a motorized treadmill, where the participant stepped onto a forwards moving belt (0.7 m/s), then stepped forwards again from a height of 0.2 m onto a ground-embedded force platform. A custom-designed tray was screwed directly into the platform's shell, filled with a secured bed of artificial turf (75 mm, Revolution360, FieldTurf Pty, USA). Three randomly cued target lights projected onto the ground at 10, 12 and 2 o'clock with respect to the treadmill direction. The target was triggered just prior to free fall. The participant's instruction was to quickly and accurately hop to target. Each target was presented 30 times per limb, for a total of 180 trials per session (limb $\times$ target $\times$ trial), with sessions completed on separate days. The pairing of unlike boot model combinations was different between test days. Effect of fatigue on target accuracy was reduced by alternate left and right landings and a random target sequence. 3D kinetic and kinematic data were collected by 1000 Hz force platform (AMTI, USA) and 500 Hz VICON camera system, synchronized and captured by Nexus software (Oxford Metrics Ltd, UK). A biomechanical model and segment angles were processed (C-Motion, USA). The movement task was subdivided into deceleration and acceleration phases. Boot effect on inter-segment coordination was assessed using custom script (MathWorks Inc, USA). Segment length and segment angle data were input variables to compute three main parameters from the UCM method: (1) goal-equivalent variability (GEV, no change to limb length), (2) non-goal equivalent variability (NGEV, changes limb length), and (3) the normalized difference between GEV and NGEV (i.e., IMA, index of motor abundance).

Results

Support time was 0.9 $\pm$ 0.12 s; body centre of mass (BCOM) downwards and forwards velocity at impact was -1.70 ± 0.05 and 1.51 ± 0.07 m/s. Peak GRFz was $3.7 \pm 0.2 \times$ BWT, occurring at 0.05 ± 0.004 s after initial contact. Anterior flight length of BCOM was 0.35 ± 0.02 m. While there was no boot effect on the UCM parameters, there were relative differences in GEV and NGEV profiles when comparing between vertical and horizontal components of the limb effector (Figure 1). There was a

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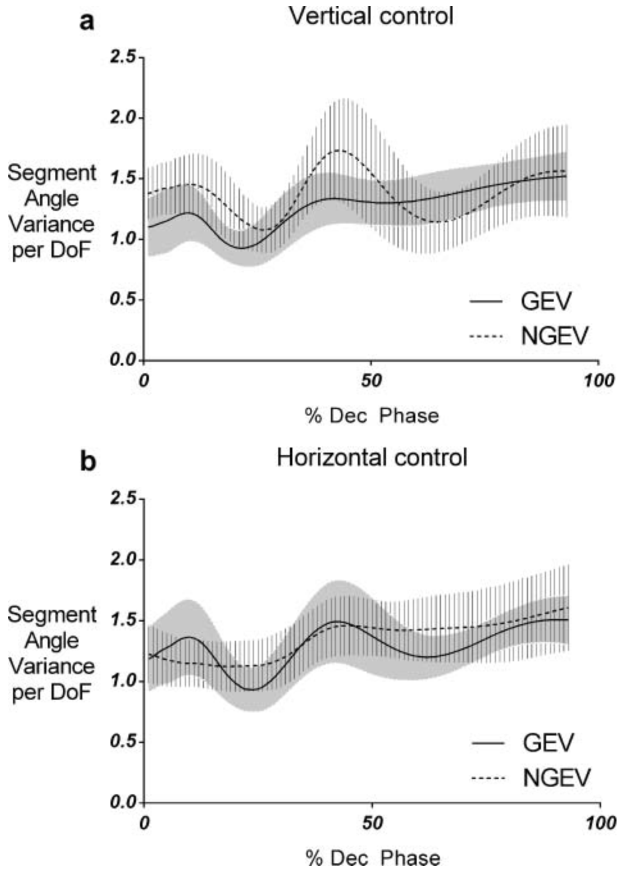

Figure 1. Representative GEV and NGEV variance for vertical (a) and horizontal (b) control of the body centre of mass position during the deceleration phase.

significant difference in IMA during acceleration phase ($P < 0.05$, Figure 2).

Discussion and conclusion

The test produced challenging biomechanical conditions appropriate for testing boot design. While the case study did not reveal an effect of boot design on UCM parameters, there were findings that are consistent with outcomes

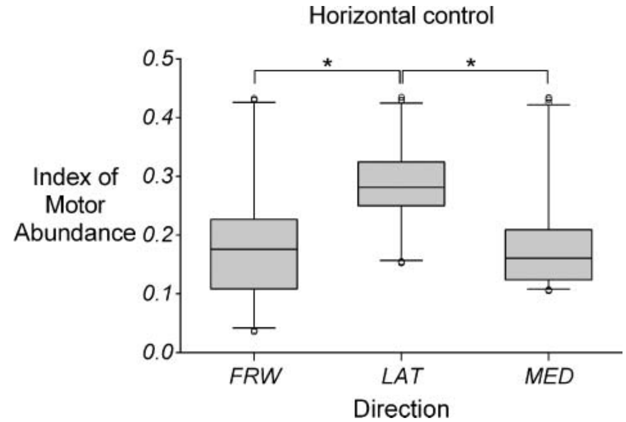

Figure 2. Average IMA for horizontal stabilization, comparing forward, lateral and medial hop direction during the acceleration phase. * denotes significant difference ($P < 0.05$).

of hop task with target constraints (Auyang, 2013). Acceleration to a lateral target suggests significantly greater stability of horizontal position of BCOM compared with the less challenging task of forward or medial acceleration. This method will be applied to a broad sample of athletes.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The influence of different aspects of sport shoes on the comfort

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Keywords: sport shoe; comfort; sole comfort; wrapping; stiffness

Introduction

Comfort has been the most desired footwear feature by consumers. Shi Kai's investigation about 35,000 families indicates that 54% of the investigated consumers regard the comfort as the most important index (Shi et al., 2009). The comfort may relate to fatigue, hurt and the performance of athletes (Miller et al., 2000; Mündermann et al., 2002). 'The seminar about the footwear comfort' held in March 2011 has defined the comfort of shoes. It is people's comprehensive psychological and physiological evaluation while they wear the shoes. The comfort consists of the fit, the functionality and the circumstance in the shoes.

Purpose of the study

The purpose of this study was to examine the importance of difference aspects on the comfort of footwear.

Methods

Ten healthy adult males (25 ± 1.5 years old; 67 ± 5.6 kg; 171 ± 2.5 cm; shoe size 8) participated in this study. All subjects' feet must (1) without high arches or flat foot. (2) Be in good physical condition. Twenty-five pairs of shoes from different sports brands are used in this study.

Eye patch was used and the brand's logo was covered to avoid the bias of brands. All subjects should try on the shoes within 10 min. The questionnaire was introduced to the subjects before the study. Based on previous studies (Jordan et al., 1997; Mündermann et al., 2001), a 13-item perception questionnaire was designed. Data collected was analysed by SPSS 20.0. Correlation analysis was used between the shoe wrapping, fit, sole comfort, air permeability, support, and overall comfort ($p < 0.05$).

Results

After a correlation analysis between different aspects and the overall comfort (Table 1), wrapping, fit, sole comfort, and support were significantly correlated with overall comfort.

The features shown in significant correlation with overall comfort were analysed by stepwise multiple regression. The result is shown in Table 1. Ultimately, the formula including two variables (R2 = 0.833), sole comfort and wrapping. The multiple linear regression result was shown in Table 2. The formula including the sole comfort and the wrapping can be defined as: overall comfort = 0.426 (sole comfort) + 0.607 (wrapping).

Discussion

People's feet are sensitive to the shoes while running or walking. Not only because they are abundant of sensors but also for high impact and pressure. The material and the structure make great influence to the comfort.

A pair of shoes possessing functions like cushion, energy storage, and feedback is based on the material and structure of the sole (Mündermann et al., 2001). The stiffness of material is one of the physical properties. The feet are moderate and touch the shoe material directly. So the material must be moderate enough to satisfy the foot's sense. Goonetilleke found that high stiffness sole is more uncomfortable (Goonetilleke et al., 1999). Some studies indicate that if the material is stiffer than the skin, the touching area have to be considered when evaluating the comfort of shoes (Hohmann. et al., 2016). In this study, different parts were included in wrapping, like the metatarsophalangeal part, the instep part and the heel collar part.

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Table 1. Correlation analysis and relationship between different aspects and the overall comfort.

	r	P	Standard coefficient	P
Wrapping	0.84	0.00*	0.43	0.023*
Fit	0.79	0.00*	0.228	0.167
Sole comfort	0.76	0.00*	0.417	0.001*
Air permeability	0.2	0.35	—	—
Support	0.56	0.00*	−0.14	0.907

*means the statistically significance.

Table 2. The multiple regression of sole comfort and the wrapping.

Index	Standard coefficient	P
Sole comfort	0.426	0.00*
Wrapping	0.607	0.00*

*means the significant correlation.

Conclusion

(1) The wrapping, the shoe fit, the sole comfort, and the support made great influence to the overall comfort when people evaluate the sport shoes.
(2) The relationship of the shoe wrapping, the sole comfort, and the overall comfort can be described by a multiple regression equation: overall comfort $= 0.426 \times$ (sole comfort) $+ 0.607 \times$ (wrapping).

Disclosure statement

No potential conflict of interest was reported by the authors.

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How do we fit underground coal mining work boots?

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Keywords: 3D scanning; work boots; fit; underground coal mining; comfort

Introduction

Well-fitted footwear provides an appropriate level of protection, support and comfort during walking (de Castro et al., 2010), and reduces the potential for foot problems and foot pain (Manna et al., 2001). To fit properly, the internal footwear shape should match the shape of a wearer's foot. In underground coal miners, however, there are mismatches between the shape of their feet and the internal work boot dimensions. The impact these boot-foot mismatches have on work footwear satisfaction remains unclear (Dobson et al., 2017). Uncomfortable footwear does not have poor fit ratings at every point on a shoe. This indicates that work boot fit might be more important at some areas of the foot rather than others (Au & Goonetilleke, 2007), although this notion remains unexplored.

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Purpose of the study

The purpose of this study was to determine the association between the internal work boot shape–foot shape match and work boot satisfaction in underground coal miners.

Methods

Three-dimensional foot scans (INFOOT; I-Ware, Japan) were collected for 197 underground coal miners (39.2 ± 9.6 years of age; 178.7 ± 5.8 cm; 92.8 ± 12.6 kg). Boot moulds representing the internal dimensions of the standard safety footwear worn by underground coal miners in the Illawarra Region (Aus; gumboot and leather lace-up) were constructed out of Plaster of Paris (Uni-PRO, Australia). These moulds were scanned using the same procedure. The following dimensions of each foot and boot mould were measured: length, ball girth circumference, breadth, instep circumference, heel breadth, height of the instep, ball girth height and heel girth circumference. Differences between these measurements were calculated and grouped into 12 categories. Categories depended on the difference value; 0–10, 10–20, 20–30, 30–40, 40–50, >50 mm, and whether the miner's feet were smaller (−) or larger (+) than the internal dimensions of their work boots.

The participants also completed a survey, which sought information on the their incidence of foot problems, lower limb and lower back pain history and ratings of work footwear fit and comfort.

To assess mining work boot fit relative to underground coal miner boot satisfaction, cross tabulations with a Pearson's Chi-squared test were applied to the survey data (foot problems, lower limb and lower back pain history, and work footwear fit and comfort) and the difference in values between the miner's feet and their internal boot dimensions (SPSS Version 21, USA). This design determined whether the position of a miner's foot inside their work boot was significantly associated ($p < 0.05$) with their incidence of foot problems, lower limb and lower back pain history, and ratings of work footwear fit and comfort.

Results

Lower back pain incidence reported by the coal miners was significantly related to heel breadth ($\chi^2 = 8.1$, $p = 0.015$) and heel girth circumference difference values ($\chi^2 = 15.4$, $p = 0.038$). That is, a gap of 40–50 mm at the heel girth circumference and 10–20 mm at the heel breadth led to an increased incidence of lower back pain. Of the miners who reported having foot pain, heel girth circumference deviations significantly affected this occurrence ($\chi^2 = 45.7$, $p = 0.005$). Comfort ratings were significantly affected by heel girth circumference

Table 1. Significant ($p \leq 0.05$) relationships for the variables instep height, ball girth height and heel girth circumference based on the difference between the dimensions of underground coal miner's feet and their internal work boot dimensions.

	Difference	Instep Height	Ball Girth Height	Heel Girth Circumference
Boot Bigger	-20-30mm	Poor fit	Very comfortable Very good fit	Very comfortable
	-10-20mm	Less likely poor fit	Good fit Less likely indifferent comfort Less likely reasonable fit	Indifferent comfort
	-0-10mm		Uncomfortable - indifferent Poor - reasonable fit	Very uncomfortable

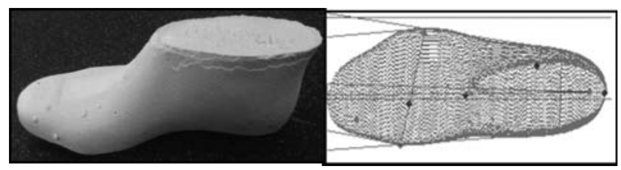

Figure 1. An example mould representing the internal shape of the gumboot and the associated 3D scanned image.

($\chi^2 = 75.6$, $p = 0.001$) and ball girth height ($\chi^2 = 46.4$, $p = 0.000$) deviations (see Table 1). Whereas fit ratings were significantly affected by deviations in instep height ($\chi^2 = 39.8$, $p = 0.001$; see Table 1) and ball girth height ($\chi^2 = 32.2$, $p = 0.009$) (see Table 1). Finally, instep height deviations significantly affected hip pain incidence ($\chi^2 = 12.7$, $p = 0.019$). No significant relationships were found in regards to length or foot breadth.

Discussion and conclusion

Whether the shape of a work boot matches a miner's foot at the heel, ball girth and in-step appears to be more important than the traditional measurements of length and width. Gaps of 0–10 mm between a miner's foot and the edge of their work boots in terms of width were insufficient for a boot to be deemed comfortable. A gap of 10–20 mm between the foot and boot appeared to be the minimum at the in-step and ball girth, whereas 20–30 mm at the heel, to ensure the workers deemed their footwear as satisfactory. This gap dimension may be required to allow for foot changes during work. There is a tendency for the miner's feet to become hot and sweaty over time, leading to swelling inside their boots.

The results of the present study have important implications for the fit of work boots for underground coal miners.

Disclosure statement

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Effect of fore-medial-side thin insole on lower extremities biomechanics in college male basketball players

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Keywords: insoles; metatarsal heads; kinetics; kinematics; athletic performance

Introduction

Injuries most often seen in male and female basketball players were the ankle (male 20.3%, female 17.9%) and knee (male 12.3%, female 16.4%) (Zelisko, Noble, & Porter,. 1982). More than 60% injuries happened in games and practices were ankle ligament sprains, knee injuries (internal derangements and patellar conditions) and upper leg muscle-tendon strains in the lower extremity (Agel *et al.* 2007).

Foot pressure distribution pattern starts from heel to toe with pronation movement. Peak pressure at the hallux increases by 40%, while the lateral forefoot undergoes a 54% decrease during cutting movements compared to running straight (Ellis *et al.* 2004). The main influence of shoes is modifying the behavior of the forefoot by changing the pressure distribution across the metatarsal heads and increasing the contact times for the toes (Soames, 1985). Providing more space for the first metatarsal part in the shoes may curtail first metatarsal stress.

The authors of the present study supposed that thinning the fore-medial side of the insole would allow of more space for the medial forefoot and the incline as resistance on the medial metatarsal would help the cutting of the basketball movement.

Purpose of the study

The purpose of this study was to investigate the effect of fore-medial-side thin insole (TI) on lower extremities kinematics and kinetics in college male basketball players.

Methods

Seven male college basketball players voluntarily participated in the study (heights = 173.1 ± 3.1 cm; weights = 68.6 ± 5.7 kg; age = 21.1 ± 2.0 years).

They wore the same basketball shoes (Nike Zoom Hyperfuse Low X) with two types of insoles (Footdisc Proactive Med Arch): one type was original insole (OI); another one was fore-medial-side TI (Figure 1).

Subjects were asked to perform L-cut (L), V-cut (V), shuttle run (SR) of basketball movement in a 5-metre running way with their maximum speed after 10-min warm-

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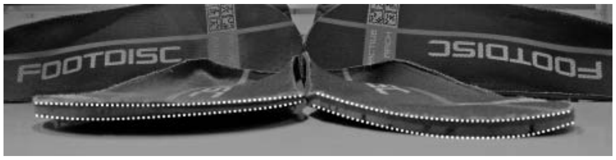

Figure 1. Fore-medial-side thin insole (left) and original insole (right).

up. They were instructed to step on a force platform with their right leg when they changed direction. Three trials were performed for each condition with their best efforts. There was a 30-sec rest between trials.

Eleven cameras (200 Hz, Motion Analysis Corporation, Santa Rosa, CA, USA) and one force platform (2000 Hz, AMTI Inc., Watertown, MA, USA) were used in synchronization for data collection. Kinematic and ground reaction force (GRF) data were analysed using the MotionMonitor software (Innovative Sports Training Inc., Chicago, IL, USA) during the support phase of the changing direction movement which was determined by 30 N GRF thresholds for foot contact and takeoff of the force platform. A modified Helen Hays marker set was used to identify thigh, shank and foot segments of the right lower extremity. The excursion of the heel during support phase was calculated.

Non-parametric Wilcoxon signed rank test was used to determine the difference of the variables of interest between the original and fore-medial-side TIs. *P* value was set at 0.05.

Results

The fore-medial-side TI showed less foot inversion angle at contact (TI: 25.3 ± 53.0, OI: 34.8 ± 56.3 deg), peak ankle lateral shear force (TI: 0.17 ± 0.06, OI: 0.23 ± 0.05 BW) and ankle dosiflexion moment (OI: 2.29 ± 0.48, TI: 1.63 ± 0.63 Nm/kg) than the OI in L-cut.

In V-cut, the fore-medial-side TI showed less hip internal rotation at contact (TI: 9.1 ± 7.5, OI: 14.4 ±6.±7 deg) than the OI.

In shuttle run, the fore-medial-side TI showed less hip abduction at contact (TI: 18.5 ± 13.1, OI: 25.4 ± 9.7 deg) than the OI.

Discussion and conclusion

The major finding of present study was that the fore-medial-side TI changed the ankle and hip kinematics and ankle kinetics of the lower extremity.

There was a considerable amount of dorsiflexion in the ankle-sprain injured subject compared to the non-injured one (Kristianslund, Bahr, & Krosshaug, 2011). Inversion sprains of the ankle have traditionally been described as resulting from a combination of foot inversion and plantar flexion (Andersen, Floerenes, Arnason, & Bahr, 2004). Less foot inversion and lateral shear force of the fore-medial-side TI found in the present study could imply that the fore-medial-side TI helps to reduce the risk of ankle sprain during the changing direction movement.

Previous researchers indicated that the potential for increased ligament loading during cutting maneuvers is a result of the large increase in varus/valgus and internal/external rotation moments (Besier, Lloyd, Cochrane, & Ackland, 2001). The present study did not measure the moment, but the results showed decrease in hip internal rotation and abduction implying that less valgus would be found in the fore-medial-side TI. This may reduce the loading at the ankle during the changing direction movement of basketball.

In conclusion, the fore-medial-side TI can help to decrease the risk of injuries of lower extremity.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The influence of arch support insole on table tennis forehand stroke using Falkenberg footwork

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Keywords: kinetics; kinematics; insoles; arch support; athletic performance

Introduction

Among various techniques in table tennis, forehand loop is considered as one of the most frequently used attacking strokes in competitions (Qian et al. 2016). The footwork movement and conversion centre of gravity are crucial for the forehand loop. Falkenberg footwork is one of the most often used footworks. The performance of Falkenberg footwork requires the stability of two feet and centre of mass during the forehand striking. There is a conversion process of centre of mass from right to left take the right-handed subject as an example. The range of movement of Falkenberg footwork is great, so the stability and coordination of the lower extremity is crucial during the striking.

Previous study indicated that the arch support insole can improve dynamic stability and attenuate the impact force (Arastoo, Aghdam, Habibi, & Zahednejad, 2014; Mulford, Taggart, Nivens, & Payrie, 2008). The authors of the present study supposed that the arch support insole may change the kinematics of table tennis players and help them to stabilize the lower extremity during Falkenberg footwork.

Purpose of the study

The purpose of this study was to examine the effects of arch support insoles on kinematics during table tennis forehand stroke using Falkenberg footwork.

Methods

Twenty male Division I right-handed table tennis players (age = 23.1 ± 2.6 years, height = 172.3 ± 5.4 cm, weight = 68.0 ± 7.5 kg) participated in this study.

After 10-min warm-up, participants performed three times of forehand stroke using Falkenberg footwork with arch support insoles (Footdisc, Inc., Taipei, Taiwan) and flat insoles (Waldner Flex 3, DONIC, Inc., China) in a randomized order, respectively.

Kinematic data were collected with a motion analysis system (Motion Analysis Corporation, Santa Rosa, CA, USA) at 120-Hz sampling rate. Ground reaction force (GRF) was collected with one AMTI force platforms (AMTI Inc., Watertown, MA, USA) at 1200-Hz sampling rate. The cameras were synchronized to the force platform. Kinematic and GRF data were analysed using the MotionMonitor software (Innovative Sports Training Inc., Chicago, IL, USA).

The support phase of Falkenberg footwork was determined by 30-N GRF thresholds for foot contact and take-off of the force platform. A modified Helen Hays marker set was used to identify thigh, shank and foot segments of the right lower extremity. The excursion of the heel during support phase was calculated.

Pair-sample *t*-test was performed using SPSS 18.0 software (SPSS, Inc., Chicago, IL, USA) to compare differences between the foot arch support insole and flat insole on variables. The level of significance was set at $p < .05$.

Results

The HF^{d}, FA^{d} and HE^{ml} of the arch support insole was smaller than those of the flat insole ($p = 0.040$, 0.022 and 0.047, respectively). The KF^{d} and GRF^{ap} of the arch

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Table 1. Analysed variables.

	Arch support insoles	Flat insoles
ST (ms)	372.8 22.0	378.3 ± 21.9
HE^{ap} (cm)	0.3 ± 1.0	0.1 ± 0.4
HE^{ml} (cm)*	20 ± 7	24 ± 10
GRF^{ap} (N)*	−0.42 ± 0.15	−0.37 ± 0.13
GRF^{ml} (N)	1.18 ± 0.25	1.18 ± 0.26
GRF^{v} (N)	1.70 ± 0.25	1.71 ± 0.26
HF^{m} (deg)	30.59 ± 9.94	33.64 ± 11.71
HF^{c} (deg)	29.15 ± 9.64	31.75 ± 11.52
HF^{d} (deg)*	1.44 ± 2.2	1.90 ± 2.72
HR^{m} (deg)	0.33 ± 13.78	1.18 ± 12.08
HR^{c} (deg)	−5.97 ± 13.21	−6.62 ± 11.09
HR^{d} (deg)	6.29 ± 5.21	7.8 ± 5.58
HA^{m} (deg)	36.32 ± 8.56	35.55 ± 8.66
HA^{c} (deg)	32.21 ± 7.9	31.41 ± 7.7
HA^{d} (deg)	5.82 ± 9.47	5.96 ± 10.08
KF^{m} (deg)	60.65 ± 8.49	59.27 ± 9.62
KF^{c} (deg)	43.24 ± 13.77	45.71 ± 13.6
KF^{d} (deg) *	17.41 ± 11.36	13.56 ± 10.77
KR^{m} (deg)	26.65 ± 18.91	25.99 ± 13.92
KR^{c} (deg)	14.40 ± 15.41	14.44 ± 16.3
KR^{d} (deg)	12.75 ± 7.78	12.2 ± 8.12
AF^{m} (deg)	15.15 ± 5.01	14.33 ± 4.89
AF^{c} (deg)	−3.49 ± 15.15	−3.07 ± 15.5
AF^{d} (deg)	21.16 ± 8.99	20.23 ± 7.99
FI^{m} (deg)	12.13 ± 8.89	11.23 ± 8.68
FI^{c} (deg)	4.27 ± 7.54	4.39 ± 8.84
FI^{d} (deg)	8.40 ± 5.49	8.17 ± 6.17
FA^{m} (deg)	29.48 ± 10.04	30.23 ± 11.72
FA^{c} (deg)	22.91 ± 10.03	20.96 ± 12.14
FA^{d} (deg) *	6.58 ± 3.82	9.26 ± 5.55

*Significant between arch support and flat insoles ($p < .05$).
[ap] Anterior-posterior (minus value denotes posterior).
[ml] Medial-lateral.
[v] Vertical.
[m] Maximum.
[c] At contact.
[d] Difference from the contact to maximum.
ST, support time;
HE, heel excursion;
HF, hip flexion;
GRF, ground reaction force;
HR, hip internal rotation (minus value denotes hip external rotation);
HA, hip abduction;
KF, knee flexion;
KR, knee internal rotation;
AF, ankle dorsiflexion (minus value denotes ankle plantar flexion);
FI, foot inversion;
FA, foot abduction.

support insole was greater than those of the flat insole ($p = 0.011$ and 0.001, respectively).

Discussion and conclusion

The major finding of the present study was that wearing arch support insoles showed less excursion and abduction of the foot, which meant that the foot with arch support could be stabilized without sliding during Falkenberg footwork.

The great posterior GRF in arch support insoles indicated the table tennis players got more support from the ground, in the meantime, players bended their knee more and stiffened their hip and ankle either for the impact absorption or later propulsion. These minor changes in arch support insoles could greatly affect the finely crafted movements of table tennis that was occurred quickly and needed a precise execution of forehand strike using Falkenberg footwork (Marion *et al*., 2009).

Disclosure statement

No potential conflict of interest was reported by the authors.

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Is immediate comfort while running in cushioned versus minimal footwear related to plantar foot sensitivity?

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Keywords: comfort; running shoe; tactile sensation; running; minimal footwear

Introduction

The variety of footwear available requires careful consideration regarding if and when to run in specific designs. Traditional cushioned running shoes attenuate the perception of loads applied to the lower extremity (Robbins, Hanna, & Jones, 1988). Similarly, walking and running on hard surfaces or in minimal shoes is associated with increases in foot pressures similar to those observed in people with reduced plantar sensitivity (Lane, Landrof, Bonanno, Raspovic, & Menz, 2014; Warne et al., 2014). Altered sensory experiences may have detrimental effects on local foot loading characteristics, as well as kinematics and neuromuscular patterns (Fiolkowski, Bishop, Brunt, & Williams, 2005; Nurse & Nigg 2001; Wakeling, Nigg, & Rozitis, 2002). Interference with the body's sensory system could result in a runner being inadequately prepared to absorb impact forces (Robbins et al. 1988).

Comfort may provide insight into the transmission of information through the sensory system, acting as a surrogate for plantar foot sensitivity. However, the relationship between foot sensitivity and comfort of cushioned shoes compared to shoes with minimal cushioning is unclear.

Purpose of the study

We studied the relationship between foot sensitivity and comfort in cushioned shoes compared to running shoes with minimal cushioning.

Methods

Seventy-five healthy runners (39 (52%) females, mean ± SD age: 35 ± 9 years) were recruited from the community. Volunteers were aged between 18 and 60 years, and had a current running load of ≥2 sessions per week with a minimum distance of 10 km/week. We excluded runners who: (i) were habitual barefoot or minimal shoe runners; (ii) had previously attempted and abandoned barefoot running due to lower limb injury or discomfort; (iii) had current lower limb or spinal pain/injury or a history of lower limb tendinopathy; (iv) had rheumatological, metabolic or neuropathic conditions that could alter planter foot sensation.

Participants underwent a battery of quantitative sensory testing (QST) on the plantar surface of their dominant foot. Four sites on the plantar surface of the foot were tested in a random order: 1st metatarsophalangeal joint (MTPJ), medial midfoot (at 50% foot length), base 5th metatarsal (MT) and midpoint of heel. QST was conducted in a standardized order. Mechanical detection threshold was tested using a standardized set of Von Frey Monofilaments. Mechanical pain threshold was evaluated using seven weighted pinprick stimulators. A 64-Hz tuning fork was used to measure vibration detection threshold, and recorded as time to cessation of vibration sensation. Pressure pain threshold was measured using a pressure algometer.

Participants then completed two five-minute running trials at a self-selected speed. The different footwear conditions were applied in a random order: (i) cushioned neutral shoe (New Balance 1030v3); (ii) minimal shoe (New Balance Minimus zero). Participants then selected their most comfortable footwear condition as per previous protocol (Mills, Peter, & Bill, 2010). Participants were grouped based on the footwear condition that they rated as most comfortable. Footwear sensitivity measures were compared between groups using one-way analysis of variance.

Results

Forty-seven individuals ranked the cushioned shoe as their most comfortable and 28 ranked the minimal shoe as most comfortable. With respect to pain sensitivity, individuals

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who ranked the cushioned shoe as most comfortable were significantly more sensitive to mechanical pain detection at their heel (109.2 mN, 9.6–208.7; $p = 0.032$) and 5th MT head (87.7 mN, 13.2–162.2; $p = 0.022$) than those who ranked the minimal shoes as most comfortable. There were no differences in pressure pain thresholds. With respect to mechanical detection, individuals who ranked cushioned shoes most comfortable had significantly lower sensitivity to vibration at their 1st MTPJ (mean difference −2.33 s, 95% CI −4.43 to −0.23; $p = 0.03$) and 5th MT base (−1.76 s, −3.51 to −0.2; $p = 0.048$). There were no significant differences between groups for mechanical detection threshold.

Discussion and conclusion

Participants who rank cushioned running shoes as most comfortable have plantar foot sensitivity profiles that were more sensitive to mechanical pain, but less sensitive to vibration, compared to those who rank minimal running shoes as most comfortable. This suggests that cushioning may be preferred to avoid pain, or because it offers a vibration damping that reduces the need for a muscle tuning response. This may have implications when advising runners on wearing a cushioned or a minimal running shoe, as comfort preference may be an indication of intrinsic capacity to tune impact force. For example, advising that transitioning to less cushioned running footwear may expose the runner to detrimental effects on soft tissue, potentially being painful.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effect of fore-medially pitted high-heeled shoes modification on foot pressure during standing and walking

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Keywords: high-heeled shoes; hallux valgus; shoe-last; foot pressure; gait

Introduction

Women are usually wearing high-heeled shoes (HHS) for pretty posture demonstration. Nevertheless, HHS interrupts the natural foot position, resulting in discomfort and pain around toes, ball of the foot, heel and arch, and produces a chain reaction of negative effects up to the lower limb and the spine (Cronin, 2014).

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HHS would displace the COP locus anteromedially during the stance phase of high-heeled gait (Han, Park, & Im, 1999) and lead to the increase in confining pressure caused by high-pressure distribution over the forefoot and increase in the pressure on the hind foot, which may cause deformation of toes such as hallux valgus and heel pain over a long period (Park, 2009).

To make fashion and comfort in the same shoes, the authors of the present study proposed a new concept on HHS design by pitting the fore-medial part of HHS around ball of foot area. It was supposed that the modification would in term allow of more space for the medial forefoot, and the pit as resistance on the medial metatarsal head area would prevent the forward sliding of the foot during gait.

Table 1. Foot pressure variables (mean ± SD).

	Non-pitted	Pitted
Standing		
DPCP (cm) [M]	2.9 ± 1.3	3.6 ± 0.1
FCA (cm^2) [L]	37.4 ± 5.7	41.8 ± 4.0
PFCP (KPa)	33.6 ± 5.4	35.4 ± 2.9
Walking		
DPCP (cm) * [L]	1.8 ± 1.36	3.5 ± 0.1 * [L]
FCA (cm^2)	48.8 ± 3.7	48.7 ± 3.0
PFCP (KPa)	65.9 ± 14.6	67.7 ± 15.1

*Significant difference between non-pitted and pitted HHS ($p < .05$).
[L] Large effect size.
[M] Medium effect size between non-pitted and pitted HHS.

Purpose of the study

The purpose of this study was to investigate the effect of the fore-medially pitted high-heeled shoes modification on foot pressure during standing and walking.

Methods

Seven females who work in clothing stores voluntarily participated in the study (height = 160.4 ± 3.9 cm; weight = 47.4 ± 4.1 kg; age = 31.3 ± 11.1 years; duration of HHS use = 8.0 ± 6.5 years).

Non-pitted and pitted HHS (Twu Huolong Precision Lasts Co., LTD) were used in the study. Both shoes were characterized by 2-cm-height forefoot outsole and 9-cm-height heel with 1.5-cm^2 heel base. The pitted HHS has been molded with 1.5 cm radius and 0.2 cm depth around the first metatarsal head area, fore-medial part of the shoes, using the fore-medially embossed shoe-last.

Subjects were asked to walk on a treadmill at a constant speed of 1.5 km/hr for 15 min to mimic prolonged HHS wearing with non-pitted and pitted HHS, randomly. Then they did an additional 30-sec walking on a treadmill followed by a 10-sec standing for foot pressure measurement. The foot pressure data were collected using F-scan in-shoe system (Tekscan Inc., Boston, MA). The distance from the point of peak contact pressure to the medial edge of the first metatarsal (DPCP), contact area (FCA), and peak contact pressure (PFCP) of forefoot of the dominant leg were analysed using the F-scan software (F-Scan Research 7.0, Tekscan Inc., Boston, MA, USA).

Non-parametric Wilcoxon-signed rank test and effect size (Cohen's d) were used to determine the difference of the variables of interest between the non-pitted and pitted HHS. P value was set at 0.05. The effect size was classified as small ($d = 0.2$), medium ($d = 0.5$), and large ($d \geq 0.8$).

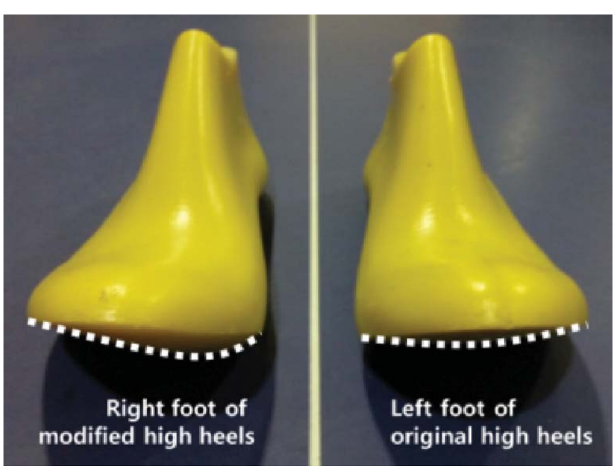

Figure 1. Front view of the shoe-last.

Results

The DPCP was significantly increased in pitted HHS during walking with large effect size. The DPCP was also increased with medium effect size in pitted HHS during standing. The FCA increased with large effect size in pitted HHS during standing.

Discussion and conclusion

Wearing the pitted HHS resulted in centred shift of peak contact pressure of forefoot during standing and walking and altered the trajectory of COP. The tight-fitting toe box of HHS can drastically stress the medial forefoot due to high-pressure distribution on the first metatarsal head and big toe, which could lead to pain and foot deformation such as hallux valgus (Bae, Ko, & Lee, 2015; Gu et al., 2014). The finding of the present study indicated that pitting the fore-medial part of HHS can reduce the stress in the first metatarsal head and big toe area during standing and walking.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Twu Huolong Precision Lasts Co., LTD

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A new approach to quantify the centre of pressure (COP) trajectory using a shoelace formula as a potential measure of movement control during walking and running

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Keywords: foot movement (in shoe); foot pressure; COP trajectory; walking; running

Introduction

The centre of pressure (COP) is an instantaneous point location of the ground reaction force. COP trajectory can provide useful information to characterize movement patterns or detect changes in foot and shoe function (Chiu, Wu, & Chang, 2013). However, most studies have used the COP trajectory as a more descriptive and qualitative form of information regarding human movement because of the difficulties in standardization and quantification of the data (Han, Paik, & Im, 1999).

The shoelace formula, a mathematical algorithm used to calculate the area of the simple polygon (Bart, 1986), also could be applied to calculate the area of the scattered COP points from a force plate or plantar pressure insole. Thus, successful differentiation of changes in movement patterns using this new approach can potentially suggest COP trajectory as more objective and quantitative data to determine the characteristics of movement under different conditions (i.e. foot and shoe function).

Purpose of the study

The purpose of this study was to determine movement and speed effects on the area of COP trajectory calculated by using a shoelace formula.

Methods

Thirty-six male heel-toe runners (mean weight: 71.17 ± 6.03 kg, mean height: 176.08 ± 3.98 cm, mean age: 22.02 ± 1.62 years) who had no medical history of lower extremity injuries, who ran for more than an hour per week and who wore the same size shoes were recruited for this study. Each subject performed treadmill walking and running at three different speeds (walking at 1.3 m/s, running at 3.5 and 4.5m/s) while wearing the same shoes (US9) provided by the company (FILA, Korea). Each subject was given 5 minutes of free walking time to adjust, and then data were collected with 2 minutes of walking or running, and a 5-minute break between each speed. The

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COP coordinates were obtained at each stance using the Pedar-X system (Novel, Munich, Germany) with a sampling of 100 Hz, and the total area of the COP point was calculated with Matlab2014. The area of all COP points from 10 strides was calculated for each subject and, then, averaged for each speed.

The method to calculate COP trajectory area for each subject is as follows:

(1) Calculation of the COP at 1 frame.

$$\mathrm{copX} = \frac{\sum_{i=1}^{n} x_i p_i}{\sum_{i=1}^{n} p_i}, \quad \mathrm{copY} = \frac{\sum_{i=1}^{n} y_i p_i}{\sum_{i=1}^{n} p_i}$$

where i is the position where cells of the insole are pressed upon application of weight load, x, y are the x- and y-coordinates of activated cells (mm), and p is the pressure of activated cells (kPa).

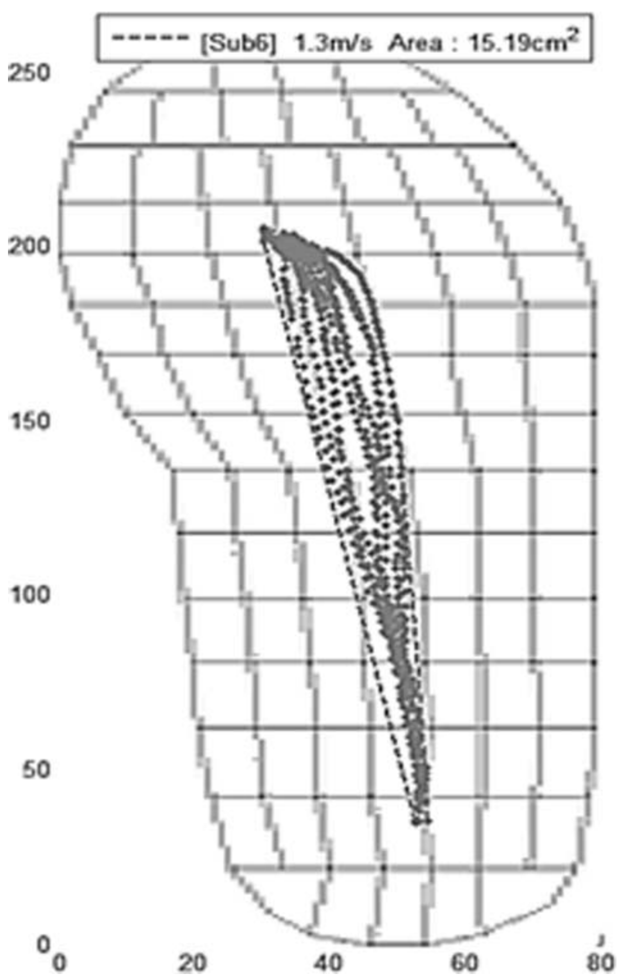

Figure 1. Calculated COP area.

Table 1. Average of COP area by each speed (unit: cm^2).

Value	1.3 m/s	3.5 m/s	4.5 m/s
M	23.78	14.38	12.74
(SD)	(5.80)	(4.15)	(2.31)

*Indicates significant difference compared to 1.3 m/s.

(1) Calculation of the area connecting the outside of total COP points.

$$\mathrm{Area} = \frac{1}{2}\left|\sum_{i=1}^{n-1} x_i y_{i+1} + x_n y_1 - \sum_{i=1}^{n-1} x_{i+1} y_i - x_1 y_n\right|$$

where i is the number of x, y points (total number within 10 stances of a subject) and x, y are the COP coordinates of each frame (Figure 1).

For statistical analysis, the one-way repeated ANOVA ($\alpha = 0.05$) was performed to examine the changes between the three different speeds.

Results

The total trajectory area of COP at each speed was 23.78 $\pm$ 5.80 cm^2 at 1.3 m/s, 14.38 $\pm$ 4.15 cm^2 at 3.5 m/s, and 12.74 $\pm$ 2.31 cm^2 at 4.5 m/s (see Table 1). There were significant differences in the area between 1.3 and 3.5 m/s ($p = 0.000$), and between 1.3 and 4.5 m/s ($p = 0.000$).

However, no statistically significant difference was observed between 3.5 and 4.5 m/s ($p = 0.078$).

Discussion and conclusion

To our knowledge, this is the first study to quantify the changes in movement patterns by the area of COP trajectory in human locomotion. The findings indicate that a new algorithm successfully differentiates the changes in the area of COP trajectory between walking and running. There was a tendency to show an increased area of COP trajectory at a slower running speed compared with a faster speed but it was not statistically significant. Finally, this study guardedly suggests that the area of COP trajectory can be used as an indication of stability or posture control under different conditions (i.e. running style and speed or changes in foot and shoe function). However, some improvements in the calculation process using optimized feature points and data accuracy with a proper reliability test are still required for future applications.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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The influence of foot strike and speed on the validity of inertial sensors to analyse rearfoot kinematics during running

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Keywords: inertial sensors; validity; rearfoot kinematics; foot strike; running speed

Introduction

Quantifying foot kinematics is decisive for characterizing running patterns, as well as for developing and producing running shoes (Walther & Mayer, 2008). 3D motion capture systems (MoCap) are the 'gold standard' for analysing foot kinematics, since they enable a high degree of standardization and measurement precision.

However, Oriwol, Dannemann, Gaudel, and Maiwald (2015) reported general differences in running gait variability in the field compared to laboratory environments. Therefore, adequate and valid measurement systems for field tests are needed.

Inertial measurement units (IMUs) are considered suitable for field tests, because they are small, lightweight and wireless.

IMUs have already been used to characterize spatio-temporal parameters for gait and running (Norris et al., 2014). They have also been found to have high validity (Donath et al., 2016) compared to MoCap.

However, previous validation studies of rearfoot kinematics in running reported a high measurement error for IMUs (Lederer et al., 2011; Mifsud, Kristensen, Villumsen, Hansen, & Kersting, 2014).

It is currently unclear which factors influence the measurement error of IMUs.

Purpose of the study

The aim of the study was to compare rearfoot kinematics collected using MoCap and IMU during running, and to quantify the effect of foot strike pattern and speed on MoCap–IMU agreement.

Methods

Fifty recreational runners (37 ♂, 13 ♀; age: 34 ± 8.5 years; height: 178 ± 10 cm; weight: 71 ± 10 kg) with different foot strike patterns performed one running trial consisting of running three minutes each at 10, 12 and 15 km/h on a treadmill. IMUs (aims datalogger DX3.2, Xybermind GmbH, Tübingen, Germany; 3-axis acc; gyro; 16 g; 2000°/s, 400 Hz) on the right heel cap and MoCap with seven markers on the right shoe (*Qualisys AG*, Gothenburg, Sweden; 400 Hz) simultaneously acquired kinematic raw data. For each speed, maximum eversion velocity (EVvel [°/s]) and eversion range of motion (EVrom [°]) were calculated from MoCap and IMU data using *MATLAB* (MathWorks, Inc., Natick, MA, USA).

Data were pooled across 100 strides for each subject to calculate mean differences (bias) and limits of agreement (LoA, [lower limit, upper limit], absolute and relative to the mean) between MoCap and IMU. The relationship between foot strike angle of MoCap (SA) and bias was assessed using Spearman's Rho (ρ).

Results

The results are presented in Table 1. Negative values indicate an overestimation of IMU. We found systematic bias

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Table 1. Bias, LoA (absolute and relative) and Spearman's Rho.

	EVvel			EVrom		
Speed	Bias ± LoA (°/s)	Bias ± LoA (%)	SA and bias (ρ)	Bias ± LoA (°/s)	Bias ± LoA (%)	SA and bias (ρ)
10 km/h	−78.4 ± 113	−20.1 ± 29.0	0.07	−2.28 ± 4.1	−18.3 ± 32.8	−0.58
12 km/h	−94.2 ± 121.8	−21.4 ± 27.7	0.02	−2.8 ± 4.1	−19.8 ± 29.3	−0.52
15 km/h	−114.9 ± 142.8	−21.4 ± 26.6	−0.07	−2.5 ± 5.3	−16.4 ± 34.5	−0.59

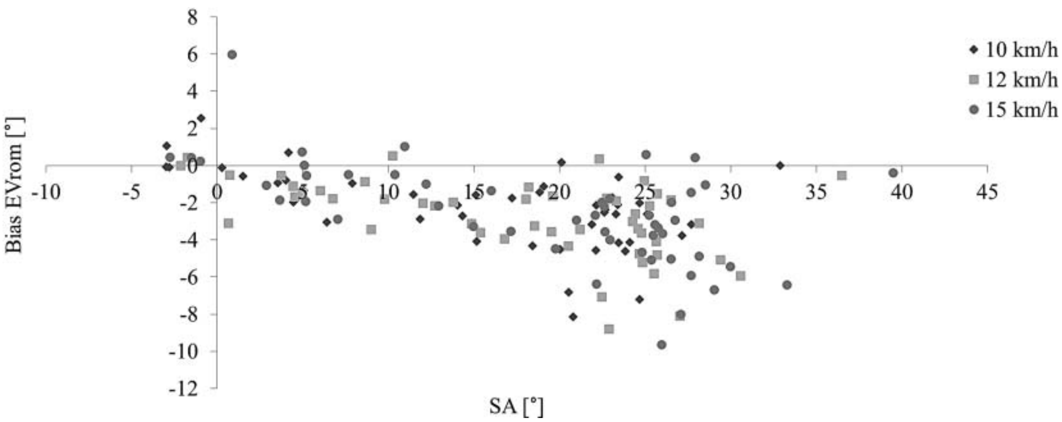

Figure 1. Correlation between bias EVrom and SA (10, 12 and 15 km/h).

for EVvel (~20%) and EVrom (~18%) independently of speed. A moderate relationship between bias and SA of EVrom for all speeds (Figure 1) was also found. Conversely, EVvel showed no correlation between bias and SA.

Discussion and conclusion

Eversion variables differed substantially between MoCap and IMU, although less than previously reported (Lederer et al., 2011; Mifsud et al., 2014). Possible sources of this discrepancy include cumulative integration error and the susceptibility of the sensors to external vibrations (Yoon Lee, & Najafi, 2012), as through impact. Furthermore, foot strike characteristics appear to affect the validity of IMU kinematics. Otherwise, the investigated running speeds show no influence on relative measurement error.

To reduce measurement error in future studies, IMU hardware restrictions must be eliminated, adequate sensor locations examined and valid procedures to calculate eversion variables analysed.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Walking and running speed effects on plantar pressure distribution

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Keywords: speed; walking; running; plantar pressure; pressure distribution

Introduction

The influence of velocity on plantar pressure measurements have been reported in several previous studies. Kernozek et al. (2000) reported that all plantar loading variables increased with the exception of contact area when treadmill running speed was increased from 2.24 to 3.13 m/s; however, the insole data were only divided into four regions in the investigation. Additionally, little attention has been given to the potential that different shoes may have effect on plantar pressure distribution.

Purpose of the study

The purpose of this study was to identify the influence of walking and running velocity on plantar pressure variables in healthy adults aged 18–26 years, and to investigate the differences on plantar loading patterns between walking and running at the same speed.

Methods

In total, 49 healthy males, aged between 18 and 26 years, without known neurologic or lower extremity orthopaedic pathology participated in this study. Plantar pressure factors were recorded at a treadmill as subjects wore the same socks and shoe at 14 different speeds (2, 3, 4, 5, 6, 7 km/h in walking and 5, 6, 7, 8, 9, 10, 11, 12 km/h in running). Data of pressure parameters, including max force, peak pressure, contact area (CA), force–time integral (FTI), and pressure–time integral (PTI), were recorded by Pedar-X insole plantar pressure measurement system. The insole data were divided into eight anatomical regions (Figure 1).The areas designated by the masks were the great toe (GT), little toes (LT), medial metatarsal (MM), central metatarsals (CM), lateral metatarsals (LM), medial arch (MA), lateral arch (LA), and heel (H). Statistical analysis was conducted by SPSS 17.0. The level of $\alpha = 0.05$ ($p \leqslant 0.05$) was perceived as significant for statistical analyses.

Results

Maximum force and peak pressure significantly increased on GT, LT, MM, CM, and H when treadmill running speed was increased from 2 to 7 km/h. Peak pressure also increased on MA. FTI and PTI significantly decreased on all the metatarsal, arch, and heel when treadmill running speed was increased from 2 to 7 km/h. Faster running speed (from 5 to 12 km/h) resulted in significantly higher

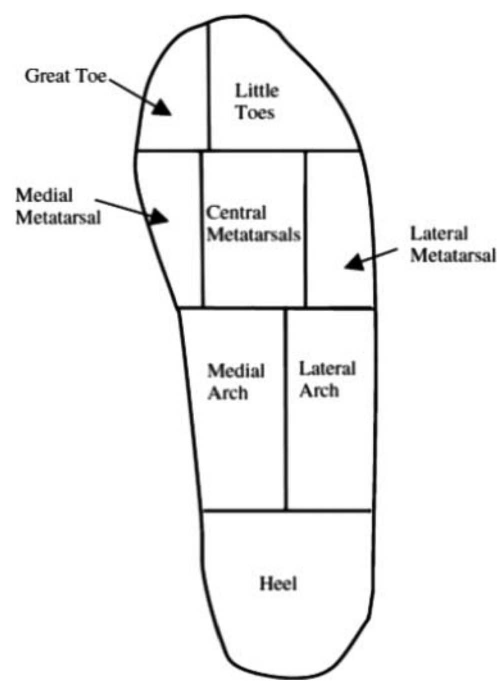

Figure 1. Foot masks.

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Table 1. Peak pressure (kPa) in different speeds of walking and running.

	Speed (km/h)					
Masks	Walk-5	Run-5	Walk-6	Run-6	Walk-7	Run-7
GT	$212.64 \pm 76.02^*$	$162.85 \pm 67.65^*$	$232.22 \pm 80.22^*$	$162.81 \pm 69.91^*$	$241.20 \pm 90.38^*$	$160.97 \pm 72.05^*$
LT	$100.36 \pm 43.12^*$	$78.26 \pm 25.14^*$	$114.75 \pm 45.83^*$	$80.98 \pm 26.51^*$	$128.16 \pm 48.81^*$	$82.74 \pm 26.69^*$
MM	192.94 ± 68.28	171.20 ± 51.80	200.33 ± 73.69	177.61 ± 48.06	191.94 ± 67.44	192.08 ± 53.33
CM	$165.16 \pm 38.27^*$	$149.34 \pm 31.94^*$	167.99 ± 45.01	158.67 ± 36.61	159.11 ± 47.44	171.82 ± 35.15
LM	99.86 ± 32.62	103.44 ± 26.14	97.80 ± 31.09	89.18 ± 26.82	$89.18 \pm 26.82^*$	$114.66 \pm 30.08^*$
MA	$57.66 \pm 11.43^*$	$77.86 \pm 15.78^*$	$66.00 \pm 13.27^*$	$81.01 \pm 14.24^*$	$74.86 \pm 15.40^*$	$84.03 \pm 14.58^*$
LA	$74.09 \pm 13.97^*$	$92.84 \pm 24.55^*$	$79.63 \pm 14.72^*$	$96.00 \pm 24.15^*$	$87.43 \pm 15.33^*$	$98.36 \pm 23.52^*$
H	$137.50 \pm 18.64^*$	$106.98 \pm 27.33^*$	$163.99 \pm 22.40^*$	$113.47 \pm 27.26^*$	$196.81 \pm 27.78^*$	$118.35 \pm 28.04^*$

Note: $^*p \leqslant 0.05$, between the same speed.

values for maximum force under all the plantar regions expect for GT, but significantly lower values for FTI and PTI under all the plantar regions expect for the heel. Comparing with walking, running took lower maximum force and peak pressure on toes and heel, higher maximum force on CM, LM, all the arch, and higher peak pressure on arch. Furthermore, running resulted in lower FTI and PTI on all the plantar regions expect for MM. Speed did not have a significant effect on CA.

Discussion and conclusion

Walking and running speed was significantly associated with plantar pressure measurements in different regions of the foot. Metatarsal and heel presented increased impact with the increase of velocity. The results of this study indicate that control of walking and running speed is essential in obtaining related foot loading data. These findings reveal the need for special consideration in sports shoes' design to increase comfort and improve gait performance. This study also provided data that can explore the reason of foot discomfort in at-risk populations.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Effect of shoe type on rearfoot motion

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Keywords: foot movement (in shoe); rearfoot stability; rearfoot control; kinematics; gait

Introduction

Rearfoot motion during gait has been linked to various musculo-skeletal problems, including inversion sprains (Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005), knee, hip and lower back pain. This has led to considerable interest in designing shoes to promote optimal

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rearfoot motion, particularly in the frontal plane. However, to date, there has been relatively little research investigating the effect of different shoe designs on rearfoot motion during gait. It has been found that static measures of foot posture are poor predictors of dynamic function during gait (McPoil & Cornwall, 1996). It is therefore necessary to measure rearfoot motion in a dynamic context. In addition, it is assumed that controlling rearfoot motion in the frontal plane impacts proximal factors such as the knee adduction moment. Research investigating this phenomenon has produced mixed results, partly due to the heterogeneous nature of study populations included.

Purpose of the study

The purpose of this study was to investigate the effect of two different types of shoe designs on frontal plane rearfoot motion during gait. The secondary aim was to determine the effect of changes in rearfoot motion on the knee adduction moment.

Methods

Sixteen healthy adult subjects participated in this study (5 female, 11 male; average age 32 years). To minimise subject heterogeneity, all subjects wore size 9 shoes. Data were collected during level walking at self-selected speed using a 10-camera system (Vicon, UK), collecting at 100 Hz. Four force plates (AMTI, USA) collected ground reaction force data at 1000 Hz. A modified version of the Oxford Foot Model (Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006) was used to measure rearfoot motion, with markers placed on the outside of the shoe in the shod conditions.

Three different walking conditions were assessed including walking barefoot, in a minimalist shoe (Nike Free) and a supportive shoe (Ascent Sustain). Maximum and minimum rearfoot motions during different phases of the stance phase were measured, along with the range of motion. In addition, maximum knee adduction moment during stance was also calculated.

A minimum of eight trials were averaged together for each subject in each walking condition. Differences between walking conditions were assessed using repeated measures ANOVA with *post hoc* analysis (alpha = 0.05) for each variable.

Results

Frontal plane rearfoot motion during barefoot walking was found to be similar to that reported previously (Stebbins et al., 2006). When compared to the shod conditions, the minimalist shoe displayed significantly more rearfoot valgus and less varus motion than barefoot, whilst there

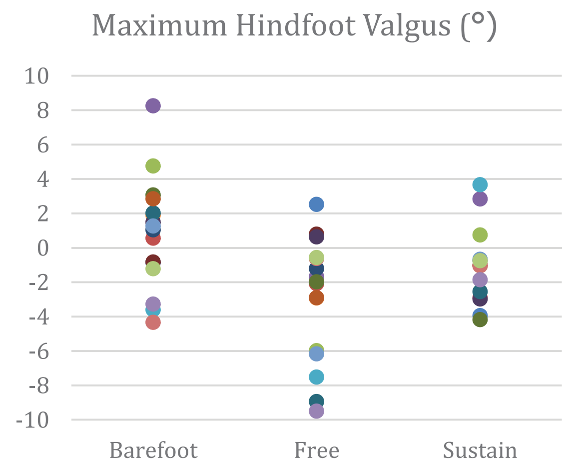

Figure 1. Individual averages from each subject for maximum hindfoot values during gait.

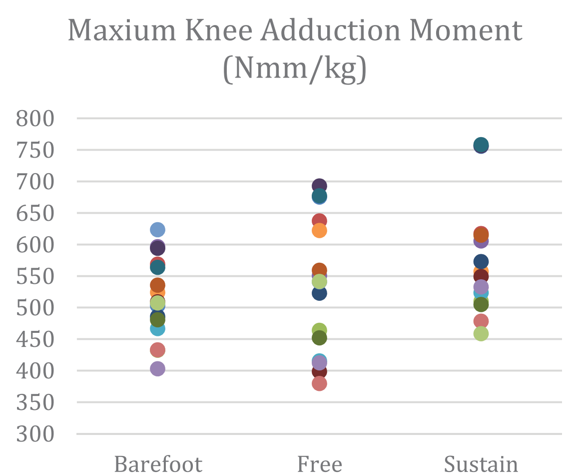

Figure 2. Individual averages from each subject for maximum knee adduction moment.

was no significant difference between barefoot motion and that displayed when walking in the supportive shoe (Figure 1).

Conversely, knee adduction moment was similar to barefoot when walking with the minimalist shoe, but tended to be greater when wearing the supportive shoe (Figure 2).

Discussion and conclusion

The results of this research suggest that the supportive shoe was better at replicating rearfoot motion than the minimalist shoe. The increased valgus in the minimalist shoe may be attributable to the restricted ability of the foot to accommodate ground terrain. While the supportive shoe also restricts motion, it appears better at holding the foot in a more physiological position. However, this did not translate to more proximal effects as the knee adduction moment was similar to barefoot when wearing the minimalist shoe, and tended to be greater in the supportive shoe. It is therefore necessary to take both local and proximal shoe effects into account when assessing shoe efficacy. Interestingly, the overall spread in the data was reduced in the supportive shoe, even within this homogenous population, meaning shoe effects may be more predictable in supportive shoes. One limitation of this study was that markers were placed on the outside of the shoe, meaning shoe, rather than true foot motion, was assessed. Overall, the results suggest there were differences in shoe effects on foot motion dependent on shoe design. Further research in this area is warranted.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Different mechanisms and effects of wedged and dual hardness insoles on foot pronation during human running

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Keywords: pronation; sole wedge; insoles; running shoe; ground reaction forces; centre of pressure

Introduction

Foot pronation has been known to cause severe pain on ankle and/or knee during running (Shih, Wen, & Chen, 2011). Functional shoes and insoles, therefore, were suggested to ease the level of pronation and relieve the pain in lower limbs.

Few studies reported that the functional insoles, including wedged insoles and insoles with dual hardness in medial–lateral (M-L) directions, significantly reduced the pain in lower limbs (Baker et al., 2007, Kirby, 2010). The wedged insoles support the medial foot arch with the additional wedge inserted below, and thus reduces the eversion angle of the foot. Dual hardness insoles provide soft and hard cushioning in lateral and medial sides of foot, respectively. This leads to a larger compression in the lateral than medial side of the insole and induces a decrease in foot eversion angle. Both types of insoles are known to reduce foot pronation and related pain, but the differences in the mechanisms and effects on foot pronation remain unknown.

Purpose of the study

The aim of this study is to compare and find the different effects of medially wedged and dual hardness insoles on the human running, especially on the foot pronation.

Methods

Five healthy male subjects (38.2 ± 12.1 years old) participated in this study under the consent form approved by Korea National Institute for Bioethics Policy prior to the experiments.

The medially wedged insoles (Scott, Inc., Switzerland) with three levels of wedge heights (control (without wedge) – 7–10 mm in heights) were used in the experiments. The three types of dual hardness insoles (composed of EVA of 55–55, 55–45, 55–35 shore hardness in M-L direction with a thickness of 5 mm) were made.

Subjects ran on the force treadmill with a force plate (Bertec, USA) at 3.0 m/s during 1 minute for each insole

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with the same shoes (ZeeQuick, Reebok, USA). During running, kinematic and kinetic data including ground reaction force (GRF) were collected using motion capture system (Motion Analysis, USA) and force treadmill with a force plate (Bertec, USA), with sampling frequencies of 200 and 600 Hz, respectively, and a cut-off frequency of 15 Hz. Joint torque and power were calculated using inverse dynamics method.

Results

M-L GRF in a lateral direction increased during mid-stance phase with the dual hardness insoles. In addition, anterior–posterior (A-P) GRF showed no significant difference; however, vertical GRF during heel transient was increased by approximately 0.25 BW as the lateral hardness increased.

Both medially wedged and dual hardness insoles reduced the pronation angle and also the pronation moment by approximately 13%. The mechanisms of such reduced pronation moment, however, were quite different from each other.

The medially wedged insoles shifted centre of pressure (COP) to the medial side (approximately 1.6 mm) without a significant change in vertical, A-P and M-L GRF. On the other hand, the dual hardness insoles shifted the COP to the lateral side (approximately 3.2 mm) (Figure 1) with increased (cutting) GRF in a lateral direction (approximately 0.1 BW).

The decreases of the moment arms from the COP to the axis of a subtalar joint were observed in both cases. The medially shifted COP for the medially wedged insoles and the increased GRF in a lateral direction for the dual hardness insoles contributed to these results, respectively (Figure 2).

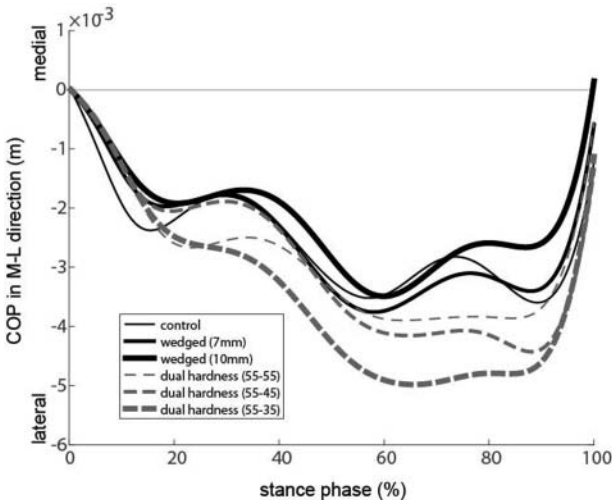

Figure 1. Trajectory of COP in M-L direction for a representative subject during running with the medially wedged (black lines) and the dual hardness insoles (grey dashed lines). Thicker lines indicate higher wedge and softer dual insoles.

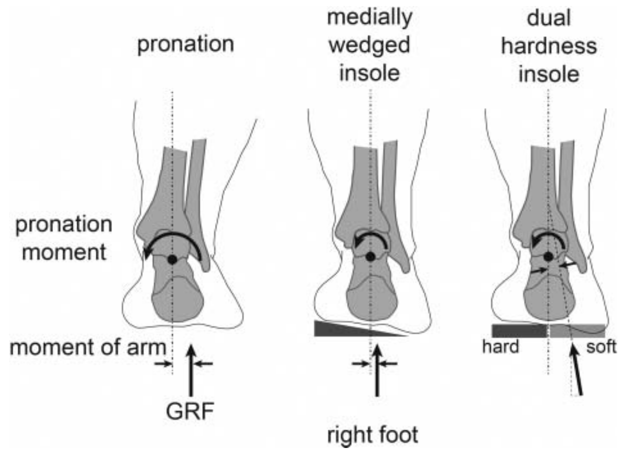

Figure 2. Comparison of mechanism of action of the medially wedged and the dual hardness insoles on the pronation.

Discussion and conclusion

According to the results, both the medially wedged and the dual hardness insoles led to a decrease in pronation angle and moment. The biggest differences in those insoles were the shifted directions of M-L COP, which are opposite. The medially wedged insoles shifted M-L COP in a medial direction. However, the dual hardness insoles shifted M-L COP in lateral direction with increasing lateral GRF.

Placing the soft cushioning at the lateral side of the dual hardness insoles changed the pronation moment and the direction of GRF. This may induce the change in the moment of knee varus/valgus and hip abduction/adduction. Thus, the levels of injury to knee and hip determine which insoles to be used.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Establishment of the estimated technique for joint loads in foot during running

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Keywords: foot movement (in shoe); joint loading; stress fractures; measurement systems; functional footwear

Introduction

Recently, long distance running is getting a popular leisure with health conscious and popularity of running events. Meanwhile, it has been said that risk of lower extremity injuries such as tibial and metatarsal bones stress fractures is increased with repeated loads during long distance running (Van Gent et al., 2007). To clarify the loads subjected to foot joints, previous studies mainly discuss them based on plantar pressure distribution and ground reaction force (GRF) (Amir et al., 2011; Beragstra et al., 2015). However, in order to precisely calculate foot joint loads, it is important to consider three-dimensional (3D) GRF distribution at foot–ground interface.

Purpose of the study

The purpose of this study is to establish an estimated method for joint loads in foot during running. The 3D force distribution was measured by using special shoes with miniature force sensors (Moriyasu et al., 2010). The joints torques were estimated by a new foot simplified model with 3D force distribution on foot planter.

Methods

Foot model

To estimate joint torques in a foot, a simplified foot model is defined as a link segment model composed of toe, metatarsal, and heel region as shown in Figure 1. Joints which linked each segment are corresponding to metatarsalphalangeal (MP) joint (toe-metatarsal), and Tarsalmetatarsal (TM) joint (metatarsal-heel) region.

Estimated method of joint torque in foot

For quantification of foot joint loads, 3D force distribution was measured by a shoe with 3D miniature force sensors (USL06-H5, TecGihan Co. Ltd.). By changing the sensors arrangement, tri-axial forces at 19 positions of the sole were collected. Each joint torque components $\tau^i_{\mathrm{MP}}, \tau^i_{\mathrm{TM}}, \tau^i_{\mathrm{Ankle}}$ were obtained from forces at each sensor positions and length between each joint to each axis of sensor which located bottom of each segments as shown in Figure 1. MP, TM, and ankle joint torques $T_{\mathrm{MP}}, T_{\mathrm{TM}}, T_{\mathrm{ankle}}$ were calculated by sum of joint torque components $\tau^i_{\mathrm{MP}}, \tau^i_{\mathrm{TM}}, \tau^i_{\mathrm{ankle}}$, respectively.

Measuremental methods

A healthy male subject (31 years; 1.70 m; 70.0 kg) participated in this study. The running experiment was conducted at running speed of 3.3 m/sec on all-weather track. The measurement method of sensor shoe was the same

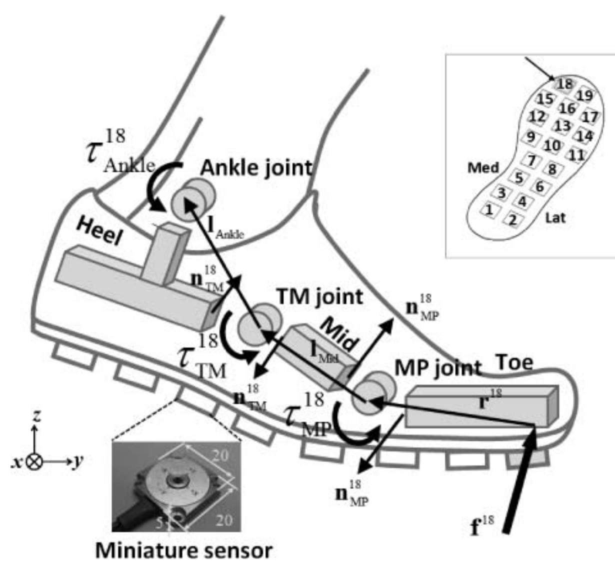

Figure 1. Image of foot model and number of sensor position. Allows show force (*f, n*), torque (?) and length (*l*) vectors (*l*, *r*) at $i = 18$.

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procedure used in the previous research (Moriyasu et. al., 2010). At the same time, a motion capture system (VICON-T; Vicon Motion Systems Ltd.) was used for measuring trajectories of foot characteristic points. GRF was recorded by using a force plate (Model 9287A; Kistler Insrument Corp).

Results

Figure 2 shows time histories of TM joint torque components τ^i_{TM}, horizontal, and vertical components $\tau^i_{\mathrm{TMF_{xy}}}$, $\tau^i_{\mathrm{TMF_z}}$ in mid to fore regions. Position where contributes to joint torque is changing from heel to toe during stance phase. It is recognized that torque components calculated by the conventional force plate did not estimate at the beginning of stance phase. The result explained that TM torque calculated by the proposed method was in close agreement with the torque obtained from conventional one from 76% to 100% of stance phase. In comparison of joint torque components, τ^i_{TM} located at the MP joint (i = 12, 13, 14) are larger than the other areas at TM joint, and their peak values appeared at approximately 50% stance phase. In order to clarify influence of $\tau^i_{\mathrm{TMF_{xy}}}$, $\tau^i_{\mathrm{TMF_z}}$ on τ^i_{TM}, $\tau^i_{\mathrm{TMF_{xy}}}$ at $i = 12, 13, 14$ under MP joint increased

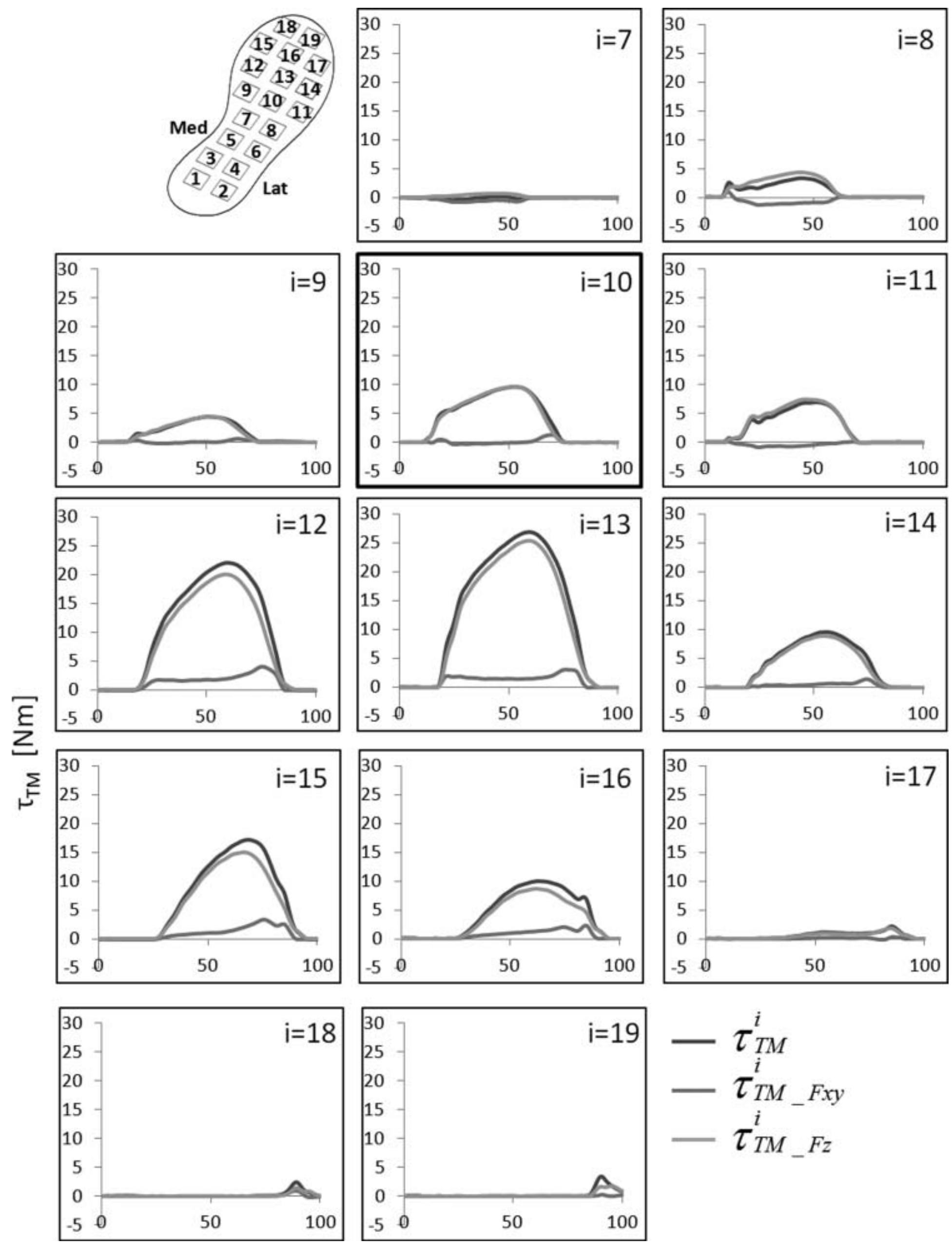

Figure 2. Time histories of TM joint torque components at $i = 7$–19 during stance phase.

by 80% of stance phase. After 80% stance phase (toe off), $\tau^{i}_{\mathrm{TMF}_z}$ is a dominant component in τ.

Discussion and conclusion

It was founded that τ^{i}_{TM} and the appearance peak time vary depending on sensor position. According to van Gent et al., metatarsal bones stress fractures are increased with repeated loads during long distance running. Therefore, given the behaviours of τ^{i}_{TM}, it is expected that mechanism of bone and ligament stress is clarified. It is concluded that the proposed technique to estimate foot joints torques helps a functional sole design for injury prevention.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Walking in minimal shoes and standard hiking boots on smooth and rough surfaces

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Keywords: minimal footwear; boots; kinematics; substrate; ankle

Introduction

Minimal shoes, lacking a firm heel cup, a rigid sole or longitudinal arch support, involve less mechanical constrictions at the level of the foot and ankle during walking and offer increased extero- and proprioception. It is unclear, however, whether minimal footwear, when compared to conventional hiking boots, involve kinematics changes at the ankle and at more proximal joints, and how this is affected, or not, by walking on a rough rather than a smooth substrate.

Purpose of the study

The purpose of this study is to quantify differences in gait (if any) between walking in a minimal shoe and a conventional hiking boot on two different hard substrates: smooth and rough.

Methods

Ten healthy subjects (5/5 male/female, age 27 ± 7 years, height 1.76 ± 0.12 m, mass 67 ± 9 kg) walked at preferred speed over two 14.4-m long walkways: one smooth and one complex with vertical variation up to 27 mm. Subjects were marked using a 67-marker set consisting of anatomical and tracking markers.

Kinematics were recorded using a 12 IR-camera system (Qualisys, Oqus-7, 200 fps) and analysed in Visual3D 6.0 and MatLab 2016b (Figure 1). Two footwear conditions were tested: the subjects' own walking boots and a standard minimal shoe (Vivobarefoot "The One"). Five trials were analysed per condition (total 200 trials). For every trial, we calculated basic spatiotemporal gait parameters and angular displacement of the knee and ankle joints.

Results

Subjects walked significantly slower (ANOVA, $P < 0.001$) on the rough substrate (mean: 1.48 m/s) than on the smooth substrate (mean: 1.56 m/s) but there was no difference between footwear conditions.

Ankle and knee displacements are overall similar but there are striking differences in variance. Focusing on the ankle, variance is significantly higher (Levene's test, $P < 0.001$) in boots (6.77°) versus minimal shoes (2.87°) on

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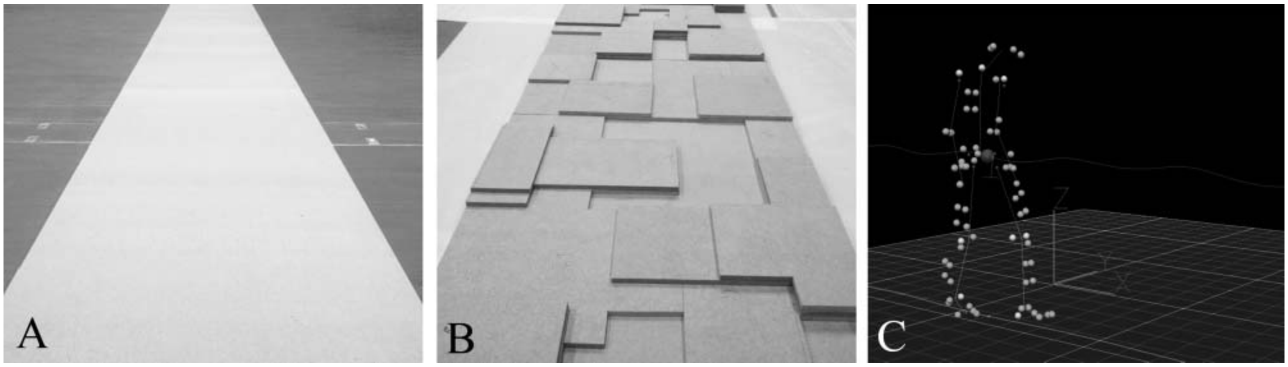

Figure 1. (a) Smooth and (b) rough substrate. Width of the substrates is 0.61 m. (c) Biomechanical model is used.

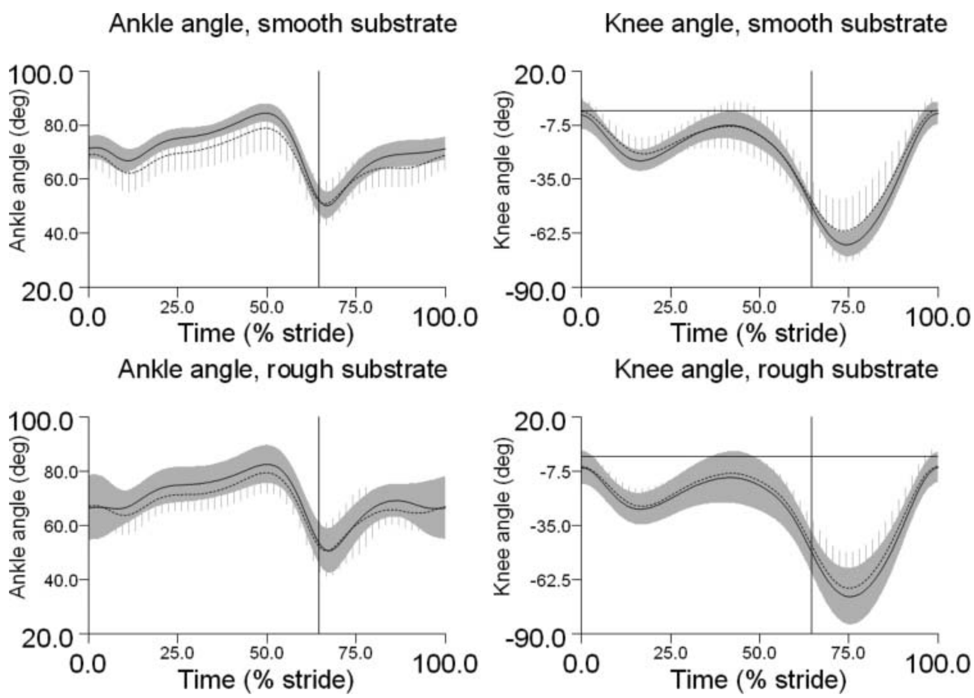

Figure 2. Average joint angles curves (ankle plantar/dorsiflexion, knee flexion/extension) normalized to stride duration. The vertical line is toe-off. Solid line: minimal shoe; dashed line: boot. Areas show standard deviation around the mean (solid: minimal shoe; vertical lines: boot).

the smooth substrate. On the rough substrate, this difference is very small albeit significant ($P = 0.005$) (boots: 6.45°, minimal shoes: 6.20°). Focusing on between-substrate comparisons, variance is significantly lower on the smooth substrate for minimal shoes, ($P < 0.001$) but similar for boots ($P = 0.457$) (Figure 2).

Discussion and conclusion

Substrate and footwear (most clearly on the smooth substrate) seems to influence movement variability, rather than average movement patterns per se.

Ankle variance is substantially smaller in minimal shoes on the smooth substrate compared to any other condition. These preliminary results will be further explored by more fine grained analyses on more subjects and on very highly variable natural substrates. Moreover, electromyography should inform whether the observed kinematic variability reflects variation in underlying motor patterns or whether these are merely a result of different mechanical boundary conditions.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Shod versus barefoot effects on force and power development during a conventional deadlift

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Keywords: athlete; barefoot; footwear; weight lifting; running shoe; foot-to-ground interaction

Introduction

The strength and conditioning and rehabilitation sectors use the deadlift as an integral part of their programming (Escamilla et al., 2000). Furthermore, what shoes to wear while deadlifting is a regular topic of weightlifting equipment forums. Strength and conditioning researchers and a range of industry practitioners suggest that weightlifting shoes, non-compressive soled shoes such as Converse™ 'Chuck Taylors' or unshod (socks only or barefoot) conditions are essential in providing a stable platform and effective force transfer from the ground to the bar (Cressey, 2008). In contrast it is claimed that soft soled shoes produce instability and indirect ground reaction force (GRF) transmission, thus compromising lifting performance.

Purpose of the study

The purpose of this study was to identify some possible kinetic differences during the concentric phase of a conventional deadlift between shod and unshod conditions. It was hypothesised that the unshod condition would result in higher peak GRF, higher rates of force development and therefore higher peak lifting power, and more stable centre of pressure excursion (COP).

Methods

Ten males and four females (age = 28.8 ± 5 years, weight = 76.5 ± 10.7 kg, height = 174.1 ± 7.6 cm and 1RM deadlift = 133.6 ± 40.2 kg) of various athletic and training backgrounds performed four repetitions of a conventional deadlift at 60% and 80% of 1RM in shod (S) and unshod (US) conditions for two sessions in a counterbalanced, crossover experimental design. Peak force (PF), rate of force development (RFD), COP for anterior-posterior (COP-AP) and medio-lateral (COP-ML) kinetic data were recorded using a single Kistler 9287 force plate. Data for peak power (PP) was collected using a Gym Aware Power Tool 5 attached to the barbell. After agreeing to participate, each participant gave informed consent and completed a general health screen. All the testing was preceded by a standardised warm up.

Results

There were significant main effects of load for PF, COP-AP and COP-ML with pair-wise *post hoc* comparisons indicating a significant increase in the magnitude and displacement of these variables at 80% of 1RM when compared to 60% of 1RM. In relation to shoe condition main effects, all variables were non-significant except for RFD ($F = 4.93$; $p = 0.045$; $\eta p^2 = 0.28$) which indicated that the unshod condition produced a greater RFD than a shod condition, respectively (2046.21 vs. −1956.39 $N{\cdot}s^{-1}$). There were no significant interactions for any of the other variables assessed.

Discussion and conclusion

As expected, there was a clear effect of load on the kinetic variables of PP, PF, COP-AP and COP-ML. However, the main focus of this investigation was on any possible shoe effects, and then whether there were any shoe-load interactions.

The rate of vertical ground reaction force development (RDF) showed a significant main effect of shoe with unshod lifters able to apply more force at a faster rate. Interestingly though, given the strength of this statistic and the fact that there were no main effects of shoe condition for PF and PP, this result needs to be interpreted cautiously. Further investigation would be needed to confirm this result and to determine any plausible causes that may be of interest to those engaged in this practice. Additionally, there were no shoe × load interactions and so it must be acknowledged that this study did not provide

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conclusive evidence to support our hypotheses, or indeed, anecdote and common practice.

Although there were no other significant main effects of shoe condition, it is worth noting that main effects of shoe condition for COP-AP approached significance ($F = 3.91$, $p = 0.07$, $\eta p^2 = 0.23$) with the unshod condition showing greater COP excursion than the shod condition. With this pattern emerging in both deadlift and back squat studies (Whitting et al., 2016), it may indicate that the hypothesis that regular training shoes provide less stability during the conventional deadlift is incorrect within the current parameters of COP testing (Lafond et al., 2004). It is also important to note that Whitting et al. (2016) mentioned a difference in perception of stability between shoe conditions and that motor control strategies to manipulate the load lifted may be a factor in determining the limits of COP excursion. Nonetheless, given the findings of this preliminary investigation it would appear that unshod conditions may not have a substantial effect on kinetic parameters during the conventional deadlift.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The impact of current footwear technology on free moment application in running

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Keywords: running mechanics; running shoe; running injury; frictional moment; locomotion; transverse plane

Introduction

In running, the free moment (FM) is considered a key variable in intra- and inter-segmental transverse plane loading of the support leg (Willwacher, Goetze, Fischer, & Brüggemann, 2016, Yang et al., 2014). The effect of footwear technology on FM application has not been sufficiently studied despite evidence to suggest that high FM amplitudes may be related to common lower extremity overuse injuries (Milner, Davis, & Hamill, 2006, Willwacher et al., 2006).

Purpose of the study

The purpose of this study was to determine the effect of current footwear on FM application to the ground in straight running.

Methods

FM and lower extremity kinematics data of 103 male and female recreational runners (age: 38 ± 13 years, mass: 70 ± 10 kg, height: 1.75 ± 0.08 m) were collected in two labs using 0.9 × 0.6 m force plates.

The participants were advised to run at a speed of 3.5 m/s in six different footwear conditions of the same brand (Brooks Sports Inc., Seattle, WA, USA). Footwear ranged from a well-cushioned, medially supported model (Adrenaline) to an experimental prototype shoe representing an extremely minimal running shoe without a noteworthy amount of cushioning or support elements ('sock' shoe). For each condition five trials per runner were time-normalized to the stance phase and subsequently averaged. All averaged curves were pooled together and assigned to three FM pattern groups by using functional

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principal component (fPCA) and cluster analysis techniques (Willwacher et al., 2016). The effects of footwear and pattern membership on the following parameters were assessed using a two-factor repeated measures ANOVA: fPCA Eigenfunction scores 1–4; minimum and maximum FM amplitudes as well as negative, positive and net FM angular impulse (alpha = 0.05). To compare the strength of footwear effects, effect sizes (partial η^2) were calculated.

Results

The fPCA analysis of the FM data-set identified the same basic FM moments as in a previous publication, which are considered to reflect differences in the general running style adopted by runners (Willwacher et al., 2016).

Significant footwear main effects ($p < 0.001$) were detected for every parameter under investigation. Footwear effects were strongest in parameters related to the amount of FM internal rotation amplitudes (fPCA scores on FM Eigenfunction 1, max. FM, positive and net FM impulse; partial η^2: 0.16–0.32). More minimal footwear (Sock, Pure Connect, Pure Cadence) increased internal rotation FMs in all three FM pattern groups (Figure 1). Footwear effects were smaller than differences due to the general running style adopted. This was reflected in much greater differences in parameters between FM pattern groups compared to differences imposed by footwear conditions (Figure 1). The average difference between the shoe with the highest and lowest maximum FM amplitude ranged from 0.018 to 0.023 Nm/kg within the FM pattern groups, while differences of 0.115 were observed between the averages of maximum FM between FM pattern groups 1 and 3.

For some parameters, significant interaction effects were obtained, indicating a different response to footwear

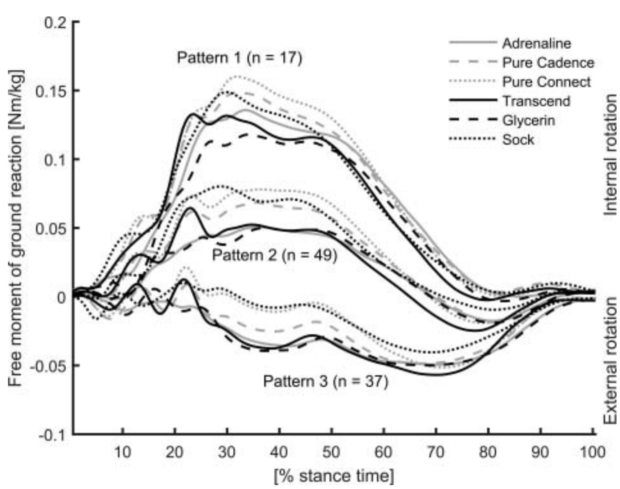

Figure 1. Ensemble averages of time-normalized FM curves for each condition.

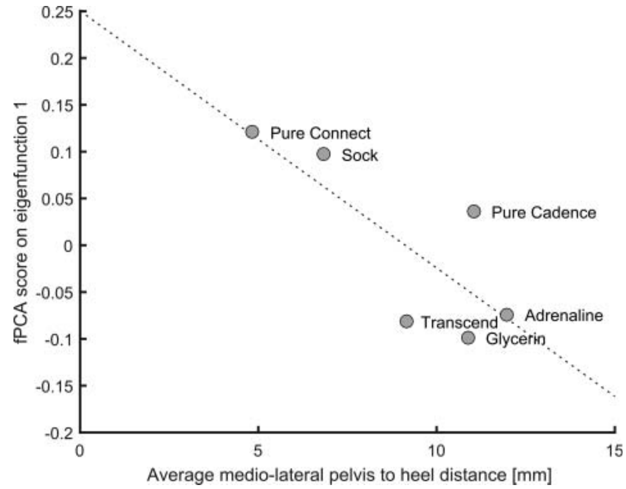

Figure 2. Relationship between mean crossover (MLdist) and mean scores on the first fPCA Eigenfunction of the FM data-set.

technology for runners with different FM application patterns. Small, but convincing, correlations between the stance phase averaged medio-lateral distance between the midpoint of the pelvis and a marker on the posterior aspect of the heel (MLdist) and fPCA score 1 were obtained ($r = 0.26$–0.32; $p < 0.01$ for different footwear conditions). A significant footwear main effect was observed for MLdist ($p < 0.001$, partial η^2: 0.14).

Discussion and conclusion

Current running footwear affects the FM application on the ground to a smaller extent than the general running style. Recently, FM variability has been related to the amount of crossover between legs in running (Meardon & Derrick, 2008). In the present study, correlations between a parameter that indirectly measures crossover (MLdist) and FM parameters were found.

Footwear conditions were also systematically affecting MLdist which might explain one mechanism how footwear can affect FM amplitudes (Figure 2). Nonetheless, future studies need to explore the relationship between footwear technology and FM application in greater detail.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

Brooks Sports Inc., Seattle, WA, USA.

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Influences of heel gradient on functional roles of the support leg muscles in running

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Keywords: running shoe; midsole thickness; muscle force contribution; ground reaction forces; induced force analysis; functional role

Introduction

Heel gradient (HG), the midsole thickness change along the foot longitudinal direction, is one of the most important parameters for the design of running shoe, and many researchers have investigated the biomechanical effects of HG on running motion (e.g. Reinschmidt & Nigg, 1995). Since human movement is mainly driven by joint torques originating in muscle contractions, induced-acceleration analyses have been implemented to quantify functional roles of joint torques and muscle forces (e.g. Hamner, Seth, & Delp, 2010; Koike, Mori, & Ae, 2007). However, effects of the HG on the roles of muscles have not been investigated.

Purpose of the study

The purpose of this study is to investigate the influences of the HG on the contributions of support leg muscle forces to generating ground reaction force (GRF) in running.

Methods

Five running shoes with different HGs, as shown in Table 1, were used. Seven healthy male heel strikers performed constant speed running of 3.3 m/s. Coordinate data of 47 reflective markers attached on subject's body and GRF were measured with a motion capture system (VICON-MX, Vicon Motion Systems Ltd.) and a force plate (9287B, Kistlar Inc.), respectively. The coordinate data of the centre of pressure (COP) were approximated by fifth-order polynomials to obtain stable values in the second-order differentiation. The joint torques about individual joint axes were calculated through inverse dynamics.

The whole body was modelled as a system of 15 rigid-linked segments. The linear and angular equations of motion for individual segments can be expressed in a matrix form as follows:

$$\boldsymbol{M}\dot{\mathbf{V}} = \boldsymbol{PF} + \boldsymbol{Q}_{\mathrm{a}}\boldsymbol{T}_{\mathrm{a}} + \boldsymbol{Q}_{\mathrm{p}}\boldsymbol{T}_{\mathrm{p}} + \boldsymbol{N} + \boldsymbol{H} + \boldsymbol{G} \qquad (1)$$

where $\boldsymbol{M}$ is the inertia matrix and $\boldsymbol{V}$ is the generalized velocity vector consisting of linear and angular velocity vectors with respect to the centre of gravity for all the segments. $\boldsymbol{P}$ is the coefficient matrix for vector $\boldsymbol{F}$ which contains all joint force vectors and the GRF vectors. $\boldsymbol{Q}_{\mathrm{a}}$ and $\boldsymbol{Q}_{\mathrm{p}}$ are the coefficient matrices for vectors $\boldsymbol{T}_{\mathrm{a}}$ and $\boldsymbol{T}_{\mathrm{p}}$ which contain active and passive joint torque vectors, respectively. $\boldsymbol{H}$ is the gyroscopic effective moment vector, and $\boldsymbol{G}$ is the vector due to the gravitational force.

The equation for constraint conditions in which adjacent segments are connected by joint, and the support leg is connected with ground at COP by a virtual joint is expressed as follows:

$$\boldsymbol{C}\mathbf{V} = \dot{\eta} \qquad (2)$$

where $\boldsymbol{C}$ is the coefficient matrix for vector $\boldsymbol{V}$, and $\eta\cdot$ is the vector consisting of the velocity of COP.

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Table 1. Midsole thickness in forefoot and rearfoot areas in each shoe.

Heel gradient	−10 mm	−5 mm	0 mm	+5 mm	+10 mm
Rearfoot	5 mm	5 mm	5 mm	10 mm	15 mm
Forefoot	15 mm	10 mm	5 mm	5 mm	5 mm

The equations for anatomical constraint axes (e.g. varus/valgus axis at elbow and knee joint), along which the joints cannot rotate freely, can be expressed as follows:

$$\boldsymbol{A}\boldsymbol{V} = \boldsymbol{0} \tag{3}$$

where $\boldsymbol{A}$ is the coefficient matrix for vector $\boldsymbol{V}$.

Substituting Equations (2) and (3) into Equation (1), a dynamic equation for joint force vector $\boldsymbol{F}$, which includes GRF, can be obtained as follows:

$$\boldsymbol{F} = \boldsymbol{A}_{\mathrm{F,Ta}}\boldsymbol{T}_{\mathrm{a}} + \boldsymbol{A}_{\mathrm{F,V}} + \boldsymbol{A}_{\mathrm{F,G}}\boldsymbol{G} + \boldsymbol{A}_{\mathrm{F},\eta}\ddot{\eta} \tag{4}$$

A musculoskeletal system of the support leg consisting of 33 Hill-type muscles was developed by using SIMM software (MusculoGraphics, Inc.), and muscle forces were estimated by using the optimization approach which minimizes cubed activation of muscles (Yokozawa, Fujii, & Ae, 2007). The estimated muscle torques on the support leg joints were substituted into $\boldsymbol{T}_{\mathrm{a}}$ of Equation (4), and then the contributions of muscles to the GRF were calculated.

Results

Figure 1 illustrates relationships between HG and muscle contributions of Vasti (VAS), gastrocnemius (GAS), and soleus (SOL) to GRF impulse during the support phase. With the increase in rearfoot thickness relative to that in forefoot, (1) braking impulse (BI) induced by VAS decreased, (2) vertical impulse (VI) by VAS increased, (3) propulsion impulse (PI) by GAS decreased, (4) VI by GAS increased, and (5) VI by SOL decreased. There is no tendency about PI induced by SOL.

Discussion and conclusion

Vasti has roles in the body support and braking, and triceps surae has roles in body support and propulsion during the support phase; these roles are the same as observed in the previous report (e.g. Hamner et al., 2010). The increase in midsole thickness in the heel region makes the direction of GRF vector generated by VAS upward and that by SOL forward. These results indicate that the HG would affect muscle force efficiency of the running motion.

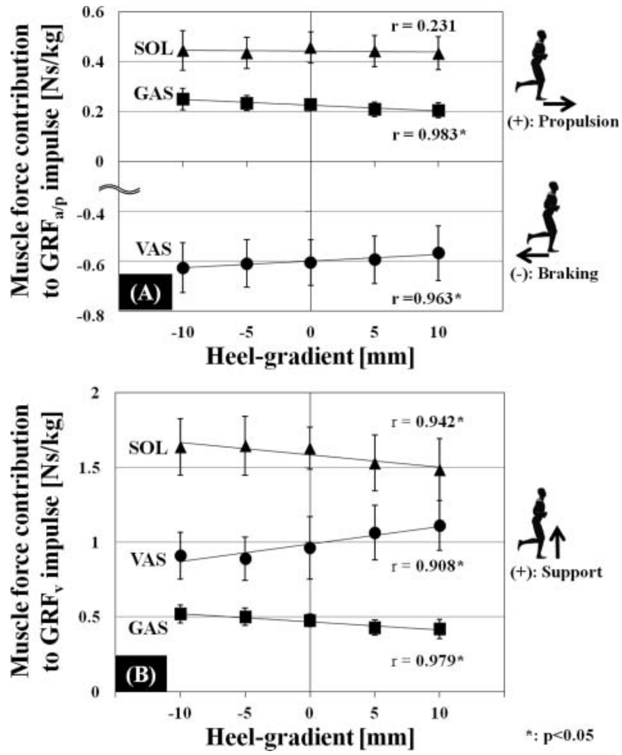

Figure 1. Relationships between HG and contributions of muscle forces to (A) anterior/posterior and (B) vertical GRF impulses during support phase for each shoe.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effects of the toe spring angle of bobsleigh shoes on bobsleigh start time and forefoot bending angle in preparation for the 2018 Pyeongchang Winter Olympics

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Keywords: toe spring angle; bobsleigh; high-speed camera; Winter Sports; biomechanics

Introduction

The bobsleigh event is one of the fastest sports in that the outcome of an event is decided by shortening the time record by just 0.01 s. Many studies have been conducted on the aerodynamic factors of the sled and athletes' posture and steering ability during the bobsleigh run for setting new time records (Chowdhury, Loganathan, Alam, & Moria, 2015; Ubbens, Dwight, Sciacchitano, & Timmer, 2016). However, recently, the importance of the start stage is being emphasized (Dabnichki & Avital, 2006). In the start stage, the explosive power of the athletes for pushing the sled is considered important, but the role of a shoe that can thrust the explosive power of the athletes to the slippery ice surface is equally important. In general, regarding the selection and development of bobsleigh shoes, the toe spring angle (TSA) of the forefoot is related to the forefoot bending angle (FBA) during walking or running, which is linked to sports performance.

Purpose of the study

The objective of the present study was to provide basic data for the development of bobsleigh shoes exclusive for Korean bobsleigh athletes by identifying the relationship between FBA and bobsleigh start stage based on various TSA.

Methods

The participants of the present study consisted of six reserve members of the Korean national bobsleigh team who received sufficient explanation on the objective of the study and volunteered to participate. The bobsleigh shoes used in the experiment were prototypes developed by 'T' Corporation, which consisted of types A, B, and C with TSAs of 30°, 35°, and 40°, respectively (Figure 1). To analyse the time taken during the start, digital run-time metres (SR-500SP, Seed Tech, Korea) were set up at intervals of 0–5, 5–10, and 0–10 m for data collection. Moreover, to measure the FBA during the bobsleigh start motion, a high-speed camera (S-pri, AOS, Switzerland) was set up on the outside of the track, approximately 5 m from the start line, for collection of data at a rate of 1000 frames/s. For the FBA measurements, seven markers (points 1–7) were attached on the sides of the outsole in the forefoot and midfoot areas. The internal angle created by lines connecting three points in the central area of the seven attached markers was defined as a single angle and analysed accordingly (Figure 1).

Results

The results of the analysis of the time required in the bobsleigh start stage showed significant differences between the shoes at the interval of 5–10 m, with *post hoc* test results indicating a statistically significant difference between the type C and A shoes ($^{*}p < .05$). Moreover, the results of the analysis of FBA in the bobsleigh start stage showed significant differences between the shoes at angles 2, 3, 4, and 5. The *post hoc* test results indicated significant differences between types B and C at angle 2 ($^{**}p < .01$); types A and C at angle 3 ($^{**}p < .01$); types A and C, and types B and C at angle 4 ($^{***}p < .001$); and types A and B, and types B and C at angle 5 ($^{**}p < .01$; Table 1).

Discussion and conclusion

The findings of the present study showed that bobsleigh start while wearing type C shoes shortened the time

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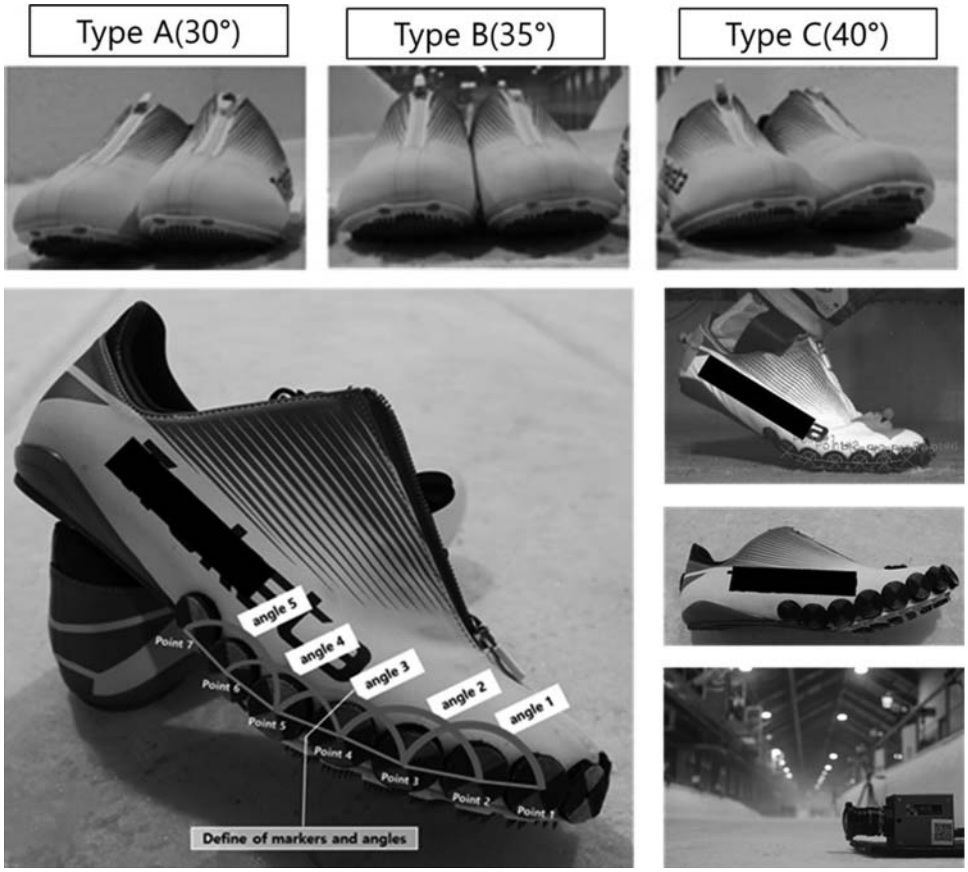

Figure 1. Shoes used in the experiment and the definitions of the markers attached to the sides of the experimental shoes used for measuring forefoot bending angle.

record. Anatomically, forefoot bending occurs in the metatarsophalangeal joint and is controlled by the foot intrinsic muscle and calf muscles (Goldmann, Sanno, Willwacher, Heinrich, & Brüggemann, 2013). As running speed increases, FBA should also increase, but excessive forefoot bending can lead to plantar fasciitis injury and increased fatigue. Meanwhile, excessive bending restriction can reduce the efficiency of using the ground reaction force and cause limited range of motion, which act as obstacles to sports performance. Type C shoes were designed to have a TSA of 40°; as such, they induced the most suitable FBA at angles 3 and 4, which correspond to

Table 1. Analysis results of forefoot bending angle and time required during the bobsleigh start time according to toe sprint angle (mean ± SD).

Shoe type		Type A	Type B	Type C
Lap time	0–5 m	1.41 ± 0.45	1.41 ± 0.33	0.41 ± 0.69
	5–10 m*	1.00 ± 0.01	1.00 ± 0.01	0.99 ± 0.03[b]
	0–10 m	2.42 ± 0.59	2.41 ± 0.04	2.40 ± 0.10
Forefoot bending angle	Angle 1	13.55 ± 5.62	12.49 ± 4.25	13.01 ± 1.18
	Angle 2**	12.37 ± 3.84	13.93 ± 3.40	10.06 ± 1.18[b]
	Angle 3**	11.21 ± 4.54	13.00 ± 2.59	14.71 ± 1.18[c]
	Angle 4***	12.88 ± 7.02	9.60 ± 3.34	18.62 ± 1.18[b,c]
	Angle 5**	13.65 ± 3.41	20.46 ± 11.68[a]	13.38 ± 1.18[b]

$^{*}p < 0.05$; $^{**}p < 0.01$; $^{***}p < 0.001$.
[a]Significantly different between type A and type B.
[b]Significantly different between type B and type C.
[c]Significantly different between type A and type C.

the metatarsal joint, in providing the bending moment for the driving force of the bobsleigh athletes, which contributed to the shortening of the time record in the start stage.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Reproducibility of rearfoot kinematics measured with inertial sensors during outdoor running on different surfaces

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Keywords: reliability; inertial sensors; rearfoot kinematics; outdoor running; field testing

Introduction

Characterizing stride dynamics during endurance running requires analysing a variety of strides (Oriwol, Milani, & Maiwald, 2012). Spatial restrictions of biomechanic laboratories often limit the number of strides to be recorded.

Inertial measurement units (IMUs) appear to be a suitable measuring instrument to collect data during outdoor running. They can record biomechanic characteristics of a large number of consecutive strides performed in a realistic running environment.

In the previous studies, we found good reproducibility of IMU data from treadmill and outdoor runs carried out on a uniform running surface (Dannemann et al., 2015; Gaudel et al., 2015). However, it is questionable whether different and more variable surfaces in outdoor runs and different time gaps between measurement days (MD) influence the reliability of rearfoot kinematics collected by IMU.

Purpose of the study

The purpose of the study was to analyse the reproducibility of rearfoot kinematics using IMU during outdoor running on different surfaces.

Methods

Twenty-two well-trained runners (14♂, 8♀; age: 31 ± 11 years; height: 178 ± 9 cm; weight: 72 ± 11 kg) performed three outdoor running trials over 45 min. There was a resting period of one week between M1 and M2 and three weeks between M2 and M3. Based on their self-reported running habits, subjects ran at a self-selected pace in their own running shoes on their surface of preference (uneven surface: $n = 15$, asphalt: $n = 7$). Running speed was monitored in all runs and variation was limited to ±1 km/h between runs. Maximum eversion velocity (EVvel [°/s]) and stride duration (StrD [ms]) were collected using IMUs (aims datalogger DX3.2, Xybermind GmbH, Tübingen, Germany; 3-axis acc; gyro; 16 g; 2000°/s, 400 Hz) mounted at the left heel cap. Data were averaged for each subject and MD across 2000 consecutive strides from the middle of each run to calculate mean differences (bias) and limits of agreement (LoA) for two MD and absolute (abs) measurement error (RMSE) across all three MD (Atkinson & Nevill, 1998). Bias, LoA and RMSE were normalized to the means of the included MD.

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Table 1. Relative parameters for StrD and EVvel on uneven surface, asphalt and overall.

	StrD		EVvel	
	Uneven	Asphalt	Uneven	Asphalt
$Bias_{M1-M2} \pm$ LoA (%)	−0.3 ± 3.6	0.3 ± 3.5	−0.8 ± 13.8	−0.9 ± 15.7
$Bias_{M2-M3} \pm$ LoA (%)	0.5 ± 3.7	0.2 ± 2.6	−7.7 ± 19.3	7.9 ± 23.3
$Bias_{M1-M3} \pm$ LoA (%)	0.2 ± 4.4	0.5 ± 5.6	−8.5 ± 19.9	7.0 ± 17.8
RMSE*2.77 (%) (abs)	3.9 (29.2 ms)	4.1 (29.3 ms)	18.0 (105.5°/s)	19.2 (118.8°/s)
	Overall		Overall	
RMSE*2.77 (%) (abs)	3.9 (28.7 ms)		21.6 (128.8°/s)	

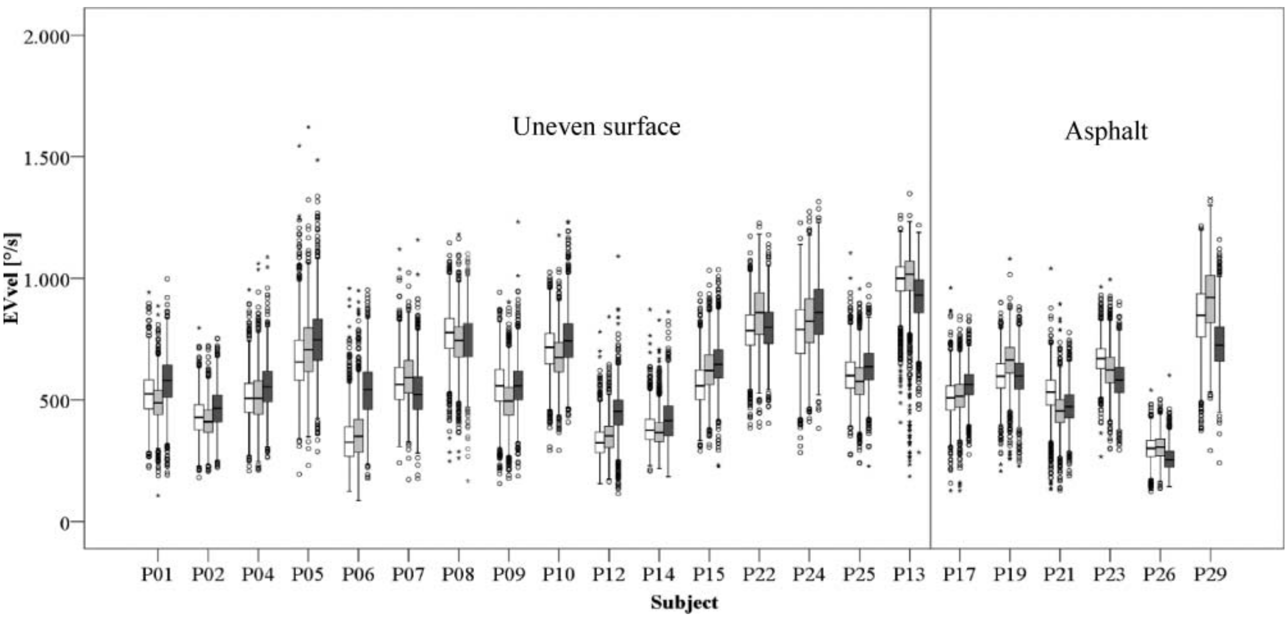

Figure 1. Boxplots EVvel across all subjects and all MD. (M1 – white, M2 – light grey, M3 – dark grey).

Results

For pooled surface data, the LoA interval for EVvel (±14.2%–24.9%) was larger than for StrD (±3.4%–4.7%). Splitting data into subgroups for surface did not influence the LoA and reliability (Table 1).

However, EVvel reliability decreased with increasing time gap between MD (Table 1). Figure 1 illustrates high inter- and intra-subjective variability between and within MD for EVvel.

Discussion and conclusion

The data of our study suggests that the type of running surface does not influence the reliability and measurement error associated with IMU-based rearfoot kinematics. However, the measurement error between the paired comparisons of the MD indicates a time-dependent increase of error for EVvel. It points at an unexplained source of variability of running behaviour over time, which cannot be accounted for in the current study framework.

StrD appears to be more reliable than EVvel in outdoor running, which is consistent with results reported by Dannemann et al., 2015. Moreover, LoA and measurement error were found to be comparable to those obtained from treadmill running (Gaudel et al., 2015).

Future studies should examine whether the assessment of means is beneficial to the analysis of time series of kinematic data of running, or whether non-linear methods can uncover information inaccessible through linear statistics.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The influence of rocker outsole design on the ground reaction force alignment during walking

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Keywords: footwear; rocker; design; ground reaction forces; diabetes

Introduction

Rocker outsole shoes are commonly prescribed to diabetic patients to offload a particular plantar area to reduce the risk of diabetic foot ulceration (Hutchins, Bowker, Geary, & Richards, 2009). Besides plantar pressures, shear stresses have been previously affiliated to the aetiology of diabetic foot ulcers (Wrobel & Najafi, 2010). However, previous studies mostly concentrated on the effect of different rocker outsole designs on plantar pressure reduction. It can be envisaged that the rocker outsole design can affect shear stresses during walking by changing the contact angles and forces during the rocking phase of shoe–ground interaction.

Purpose of the study

This study aims to investigate if the ground reaction force alignment is influences by the design of three different rocker outsoles.

Methods

One female participant with diabetic neuropathy, aged 50 years and body mass index of 30.1 kg/m^2, with no history of previous ulceration was recruited after ethical approval was granted by the local university. Three different (A, B and C) designs of toe-only rocker outsoles (with the rocker angle, apex angle, apex position as the % of the shoe length from heel, as A: 10°, 80°, 60%; B: 15°, 95°, 52%; C: 20°, 95°, 60%, respectively) were used. These designs were previously shown to be most effective in plantar pressure reduction in diabetic patients (Chapman et al., 2013). Kinematic and kinetic data were synchronously collected while walking either barefoot or with each of the rocker outsole shoes with a controlled self-selected speed. Qualisys motion capture system synchronized with a Kistler force plate with a sampling frequency of 100 Hz were used to measure GRF and marker positions, respectively. Instantaneous contact angle using GRF vectors was calculated according to Equation (1).

$$\text{GRF alignment angle } (^\circ) = \text{arc tan}\,(F_z/F_y)*180/\pi \qquad (1)$$

The greater the contact angles, the closer GRF vectors to normal indicating that lower shear forces are produced. In order to measure the rocking angle, three reflective markers were placed over the posteroinferior aspect of heel (a), 1st metatarsal head (MTH$_1$) (b), and anterior tip of the shoe (c). The instantaneous rocking angles were calculated separately for two intervals during the stance phase gait. Heel rise to the instant when the MTH$_1$ vertical position is minimal determines the first rocking phase over the rocker apex, and minimum MTH$_1$ vertical position to toe off determines the second rocking phase over the anterior aspect of the rocker. These were calculated using Equation (2).

$$\text{Rocking angle } (^\circ) = \text{arc tan}\,((x_{zb} - x_{za})/(y_{zb} - y_{za}))*180/\pi \qquad (2)$$

Results

The mean GRF alignment angle during the rocking phase (from heel rise to toe off) were 73.3° for barefoot, 71.96° for rocker A, 70.19° for rocker B, and 75.6° for rocker C.

The graphs (Figure 1) show a similar trend of contact angle changes while rocking in all four conditions that is descending gradually from heel rise to almost toe off with the greatest values over the apex position and the least ones over the anterior aspect of the rocker. Pearson

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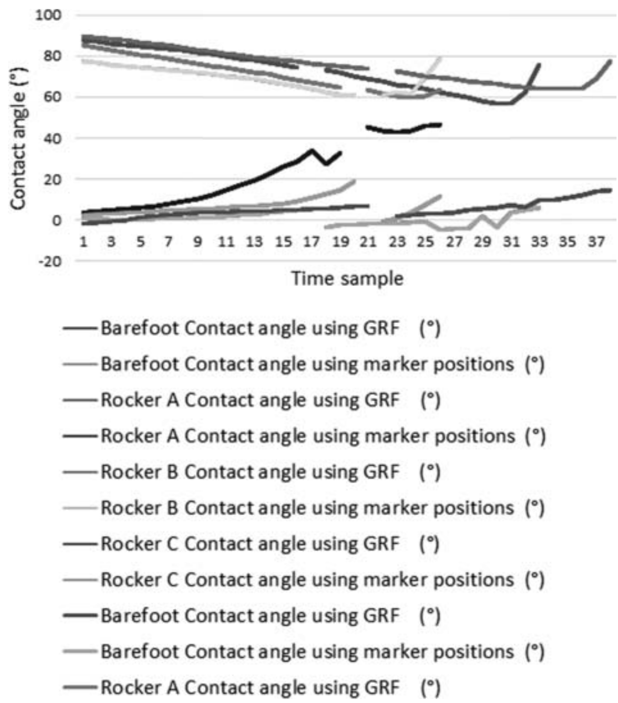

Figure 1. Contact angles (interruption points represent the time of minimum MTH_1 vertical position).

correlation analysis using SPSS software (Version 22, 2014) showed that there are significant negative correlations between the GRF alignment angle and rocking angles in rocker shoes A ($r = -0.9$) and C ($r = -0.6$) which means as the rocking angles increase, contact angles decrease and shear stresses increase.

Discussion and conclusion

The results show that the tested rocker shoes influence the GRF alignment angle. The lowest average GRF alignment angle was found for rocker B which can be related to the different rocker position of the rocker B that is placed in almost mid-length of the shoe outsole (52%). Although it has been previously well demonstrated that the rocker outsoles are the most effective shoe modification to decrease plantar pressures, they can increase shear stresses compared to barefoot. Our results from one subject indicate that the shoe design influences the GRF alignment and rocking angles. Further studies with a larger participant number are needed to investigate the effect of the rocker outsole design on the GRF alignment and rocking angles between individuals.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The potential of foot mounted 3D accelerometers to predict lower extremity loading in running

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Keywords: smart footwear; running mechanics; locomotion; pattern recognition; machine learning

Introduction

Sensor equipped footwear (SEF) becomes more and more popular within the running population. SEF aims at providing useful information for runners regarding their running style. From a scientific perspective, SEF offers the potential to acquire valuable data for longitudinal studies, e.g. with respect to running injury development.

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Accelerometers (ACCs) are the most commonly used sensors in SEF. Currently, relatively basic parameters (e.g. step frequency, step length and impact acceleration) are calculated by SEF. Nonetheless, in order to aid runners in improving their running biomechanics and to gain valuable (big data) information on injury development, parameters describing the loading of the lower extremities might be more relevant. As force platforms and motion capture equipment are not available outside of the laboratory, an alternative approach could be to predict loading parameters using machine learning algorithms based on reasonably big training data-set.

Purpose of the study

The purpose of this study was to estimate the potential of SEF using three-dimensional (3D) ACCs to predict basic loading variables during constant speed, straight running.

Methods

Complex motion analysis, including force plate measurements (1000 Hz, Kistler AG, Winterthur, Switzerland) and 3D opto-electronic motion capturing (200 Hz, Vicon, Oxford, UK) were performed on 221 male and female subjects at a speed of 3.5 ± 0.2 m/s.

Marker trajectories and ground reaction force (GRF) data were filtered using a recursive fourth-order Butterworth filter (cut-off frequency: 20 and 100 Hz, respectively). Ankle and knee joint moments were determined using standard inverse dynamics procedures. 3D acceleration signals were calculated for the markers placed on the posterior aspect of the heel and the tip of the first toe by two-fold numerical differentiation of the position data. ACC signals were expressed within the anatomical coordinate systems of the rearfoot and forefoot, respectively, in order to replicate a shoe mounted acceleration measurement using 3D ACCs. The following features were extracted from the ACC signals for the antero-posterior, medio-lateral and vertical direction during the stance phase: Negative and positive peaks for the entire stance phase and for the first and second half of stance separately; average acceleration for the entire stance phase and for each 10% interval separately; positive, negative and net integrals of acceleration with respect to time. Furthermore, body height, body mass, age and gender were included in the feature vector. This resulted in a total of 64 features for the single marker (heel and toe, respectively) ACC signals-based and 124 features for the two markers (heel plus toe) ACC signal-based analyses.

We implemented least absolute shrinkage and selection operator regression (Tibshirani, 1996) to predict peak positive and negative 3D GRFs and free moments, as well as peak external sagittal and frontal plane ankle and knee joint moments. For each prediction, the model resulting in the lowest mean squared error after fivefold cross validation was chosen. As most loading parameters scale with body height and mass, the performance of the ACC-based models was compared against a model including only the anthropometric data of the subject as predictor variables. To evaluate the comparability of the predicted with respect to the measured data, we used Bland Altman limits of agreement.

Results

Adding features extracted from footwear acceleration signals improved the predictions made by the regression models compared to a model using only body mass, height, age and gender (Table 1).

Best agreement between predicted and measured values was found for sagittal plane parameters, while loading parameters in the other planes showed lower agreement. Combining the feature sets from heel and toe ACCs

Table 1. Limits of agreement of the predictions made by the regression models using four different predictor feature sets. As a reference, the measured means and standard deviations for the respective parameters are presented.

	Limits of agreement of prediction				Measured
Parameter	Heel	Toe	Heel + Toe	Anthro only	Mean ± SD
Peak vertical force (N)	170	180	150	270	1758 ± 327
Peak propulsive force (N)	34	43	30	60	213 ± 41
Peak braking force (N)	66	58	57	82	272 ± 61
Peak medial force (N)	59	56	52	78	76 ± 42
Peak lateral force (N)	49	58	44	74	73 ± 40
Peak internal free moment (Nm)	4.6	5.1	4.3	6.1	4 ± 3
Peak ankle extension moment (Nm)	30	34	28	40	185 ± 40
Peak ankle eversion moment (Nm)	18	19	17	22	22 ± 12
Peak knee flexion moment (Nm)	48	49	43	58	223 ± 51
Peak knee adduction moment (Nm)	35	35	33	41	62 ± 26

resulted in a better predictive performance compared to predictions made from ACC features extracted from only one marker (Table 1).

Discussion and conclusion

Gaining better insight into the loading volume and intensity of runners outside of the lab will allow progress in academia and industry. Within academia, monitoring the load experienced during every day activities will improve the quality of longitudinal research studies addressing overuse injury development. This is because real world load patterns prior to the onset of the injury could be analysed and the influence of activity related fatigue could be considered. The footwear industry could use SEF information to optimize footwear selection or customization for an individual. SEF might also give feedback to the customer with respect to a less injury-risky movement behaviour.

The results of the present study provide evidence that the prediction based estimation of loading parameters based on machine learning algorithms might be a feasible approach for the prediction of loading parameters. Nonetheless, prediction quality needs to be further improved, in particular for loading parameters outside of the sagittal plane. This might be achieved by using more sophisticated machine learning techniques and/or by adding sensor information from other locations or different sensors, e.g. gyroscopes or pressure sensors.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Analysis of plantar pressure during climbing for the development of sports climbing shoes

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Keywords: climbing; foot pressure; pedar-X system; footwear; biomechanics

Introduction

In rock climbing and sports climbing activities, the use of varied equipment is needed. Consequently, there have been studies on chocks, nuts, and tension in the rope used during sports climbing. However, most of the precedent studies have been on equipment improvement, and examined the design and materials used to minimize the risk factors associated with sports climbing (Vogwell & Jinguez, 2007). Climbing shoes represent one of the most important basic equipment needed during sports climbing. In particular, compared with other leisure sports, the importance of shoes during sports climbing is emphasized the most (McHenry, Arnold, Wang, & Abboud, 2015). This is because climbing shoes are the most important basic equipment to increase the rock-climbing abilities of climbers. Climbing shoes stably support the feet on the foot holders attached to the artificial rock-climbing wall, during various positions and surface conditions. Among the literature related to sports climbing, there have been studies on the improvement of shoes according to the types of pain or injuries in the feet or ankles during climbing (Killian, Nishimoto, & Page, 1998); however, most of these precedent studies were based on the physique and shoe conditions of Westerners.

Purpose of the study

The objective of the present study was to present basic data for the future development of climbing shoes appropriate for Koreans, by analysing the plantar pressure in

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Table 1. Analysis of plantar pressure during sports climbing by each shoe type (mean ±SD).

Foot pressure	Area	Type A	Type B
Contact area (cm^2)	M1	13.21 ± 5.70	9.10 ± 2.50
	M2	14.20 ± 3.61	13.66 ± 3.52
	M3	7.02 ± 3.59	4.98 ± 4.11
	M4**	13.98 ± 4.70	3.40 ± 4.36
Maximum force (N)	M1*	71.29 ± 41.48	151.51 ± 98.67
	M2**/	110.15 ± 65.32	257.23 ± 106.14
	M3**	13.58 ± 14.11	49.70 ± 31.01
	M4	71.79 ± 43.11	119.50 ± 56.61
Maximum mean pressure (kPa)	M1	71.29 ± 41.48	151.51 ± 98.67
	M2**	110.15 ± 65.32	257.23 ± 106.14
	M3	33.58 ± 14.11	49.70 ± 31.01
	M4	71.79 ± 43.11	119.50 ± 56.61

Note: M1: lateral forefoot regions, M2: medial forefoot regions, M3: midfoot regions, and M4: heel region.
$^{*}p < 0.05$.
$^{**}p < 0.01$.

prototypes designed with different outsole hardness and structure to be suitable for the climbing environment in Korea.

Methods

The participants in the present study consisted of nine climbers who received sufficient explanation on the objective of the study and volunteered to participate. Type A, the prototype for the present experiment, was made based on the advantages of European and American climbing shoes, and it was structured with hardness in the centre of the forefoot and heel of Shore A 55 for better grip and shock absorption, as well as hardness in the edges of the forefoot and heel of Shore A 75 for improving support. Type B, the control, is an American climbing shoe that is the most popular climbing shoe in Korea among novice and intermediate climbers. It is also the best-selling shoe among those new to climbing, as it is sold in climbing gyms. Moreover, it has a single outsole structure with hardness of Shore A 63. The participants were instructed to climb by executing predetermined warm-up and cool-down motions, as well as four climbing motions in the order of inside step, rock over, counterbalance, and outside step. Each participant wore randomly assigned shoes. For plantar pressure, the pedar-X system (Germany, Novel) was used and data were collected with each foot divided into four separate regions: lateral forefoot region (M1), medial forefoot region (M2), single midfoot region (M3), and heel region (M4).

Results

The results of plantar pressure analysis during sports climbing showed that Type A had a statistically significantly wider contact area than Type B ($^{*}p < .05$). Moreover, Type A had a statistically significantly lower maximum force than Type B in M2 ($^{*}p < .01$), M1 ($^{**}p < .05$), and M3 ($^{*}p < .05$), whereas Type A also had a statistically significantly lower maximum mean pressure than Type B in M2 ($^{*}p < .05$).

Discussion and conclusion

In general, physical balance in an upright position is maintained by using the entire foot to widen the basal plane; however, during sports climbing, balance is maintained by different parts of the foot depending on the condition of the foot holder and the climber's balance (Phillips et al. 2012). This activity maximizes the point of action of the force that makes close contract to or accumulates in the feet, and thus shoes with an ergonomic design for distributing such force are needed. The findings in the present study showed that for improving shock absorption and grip, Type A, designed to have a different hardness in the centre of the forefoot and heel and the edges of the forefoot and heel, was more effective than Type B in reducing the maximum force and pressure based on superior contact area, which may be a localized basal surface during climbing. Therefore, future studies should conduct tests on slipping that can occur during stepping on holders, as well as on more detailed kinematic effects that appear from stable support.

Disclosure statement

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Effects of midsole design on extrinsic foot muscles metabolism during running

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Keywords: preferred motion path; midsole posts; mfMRI; muscle metabolism; extrinsic foot muscles

Introduction

Midsole constructions of running shoes use materials of different hardness or posts to counteract rearfoot eversion and/or lateral posts to control adduction moments at the knee. It has been proposed that the extrinsic foot muscles, particularly the inverter and everter muscles, act to control the amount of rear motion. It has also been proposed that these muscles alter their activation in order to maintain a preferred motion path. This concept becomes more relevant when revealing that rearfoot frontal and transverse plane motions are strongly coupled with the tibia internal rotation. Deviations from the preferred movement path in the frontal and the transverse planes of the knee may increase discomfort, may lead to additional mechanical loading and intuitively may increase the risk of overuse injury. The role of posts to control the foot and therefore knee biomechanics is not well understood. The active role of the inverter and the everter muscles is still unclear.

A complete picture of muscle activation of the extrinsic foot muscles is difficult to detect using surface electromyography. Muscle functional magnetic resonance imaging (mfMRI) has been established as a tool to detect metabolic activity in muscle with an excellent spatial resolution.

Purpose of the study

The purpose of this study was to examine the relative metabolic activity using mfMRI of the extrinsic foot muscles when running in shoes with experimentally varied midsole properties resulting in altered deviations from the preferred motion path of the ankle and knee joints.

Methods

Ten young males and females participated in this study. Two pairs of running shoes were used with one of the shoe midsoles experimentally modified with cylinder-like posts on the lateral aspect. Subjects completed three test sessions on three days. On one day, kinematic and force platform data were recorded. On days two and three, mfMRI data were recorded. One day the participants ran on a treadmill at their preferred training speed for 15 minutes in the baseline shoe (baseline), the other day in lateral post shoe. Within the 15 minutes of running, participants performed two 2.5-minute bouts followed by a 5-minute bout in each of the footwear conditions. MRI scans were performed before and after each running bout. Due to the treadmill in close neighbourhood of MRI scanner, the participants entered the magnet within 10 seconds of finishing the run. Multiple slice MRI was performed on a 3-T Philips Ingenia Scanner. T2 mapping was taken and T2 values were calculated of medial (GM) and lateral gastrocnemius (GL), soleus (SO), peroneus longus (PER), tibialis anterior (TA) and posterior (TP), and extensor digitorum (EDL).

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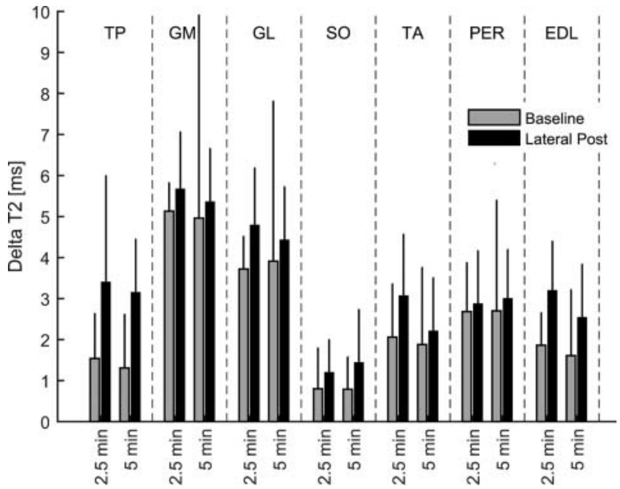

Figure 1. T2 changes between pre- and post-exercise conditions of bouts 1 (2.5 minutes) and 2 (2.5 minutes) (means, SD). All data in ms.

Results

During the first 2.5-minute bout, T2 values increased significantly for all of the analysed muscles. The experimental intervention with lateral posting significantly altered the muscle metabolism for the TP (120%), the EDL (72%), the SO (50%), the TA (49%), the GL (29%), and not significantly for the GM (11%) and the PER (7%). However, there were no significant changes in the T2 values in the two following bouts in all muscles.

The intervention affected knee joint fontal plane movement and increased peak abduction angle from 3.8° ± 4.0° to 5.1° ± 4.3° ($p < 0.05$). In the transverse plane peak internal tibia rotation changed from 6.0° ± 3.9° to 7.5° ± 5.0°. Rearfoot eversion range of motion increased in lp-shoe from 8° to 14° ($p < 0.01$).

Discussion and conclusion

The preferred movement path shown in running with the baseline footwear was significantly affected by the laterally posted midsole. A deviation was identified for the ankle and knee joints in the frontal and transverse planes of motion. These deviations appear to have been counteracted by increased muscle metabolism in the extrinsic foot muscles and especially the inverter muscles. Increased muscle may lead to higher joint loading and an increased risk of overuse injuries. These results indicate that footwear effects are easier to detect by monitoring the activity from muscles involved in counteracting the externally applied perturbations. This suggests that future footwear research must consider how muscle metabolism is affected by different perturbations in addition to measuring kinematics. In doing so, fatiguing protocols may provide valuable information regarding the interaction between muscle metabolism and kinematic deviations from the preferred movement path.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effects of midsole density distribution on kinematics and kinetics in running

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Keywords: running shoe; injuries; locomotion; midsole construction; footwear

Introduction

The running shoe industry offers a broad spectrum of footwear designs to maintain biomechanical loading in a desired range to prevent the risk of running-related injuries. Medial or lateral posts are intended to affect lower limb kinematics, or aim at controlling the point of force

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application and therefore loading at the proximal joints (Lewison, Fukuchi, Worobets, & Stefanyshyn, 2013).

Traditionally, medial posts are positioned in the mid- and rearfoot region to avoid a collapse of the medial arch. Nonetheless, the highest ground reaction forces occur after the heel has left the ground, raising the question whether posts implemented within the forefoot region would affect joint loading to a greater extent compared to posterior posts.

Purpose of the study

The purpose of this study was to examine changes in lower limb running kinematics and kinetics that occur when systematically varying midsole density distribution in the forefoot and rearfoot region.

Methods

Four healthy men (age: 27 ± 2.2 years; height: 184.8 ± 2.2 cm; body mass: 80.8 ± 9.0 kg) were equipped with retroreflective markers attached to the pelvis, thigh, shank, rearfoot and forefoot of the right leg. Data were recorded by means of a three-dimensional motion capture system (Vicon Motion Systems, Oxford, UK, 200 Hz) while running over a force plate (Kistler Instrumente AG, Winterthur, Switzerland, 1000Hz) at 2.5 m/s. Eight different footwear conditions, varying in their midsole density distribution, were analysed. The different conditions were created by cutting out holes into the consistent EVA sole of a neutral sneaker (Chariot Heritage, Brooks Sports Inc., Seattle, WA, USA) (Figure 1).

To modify the density distribution of the midsole, hard plastic pins were inserted into the holes at different areas of the midsole (Figure 1 and 2). The following midsole density distributions were analysed: medial anterior (MA, pins along the anterior part of the medial side), medial

Figure 1. Midsole modification: medial posterior.

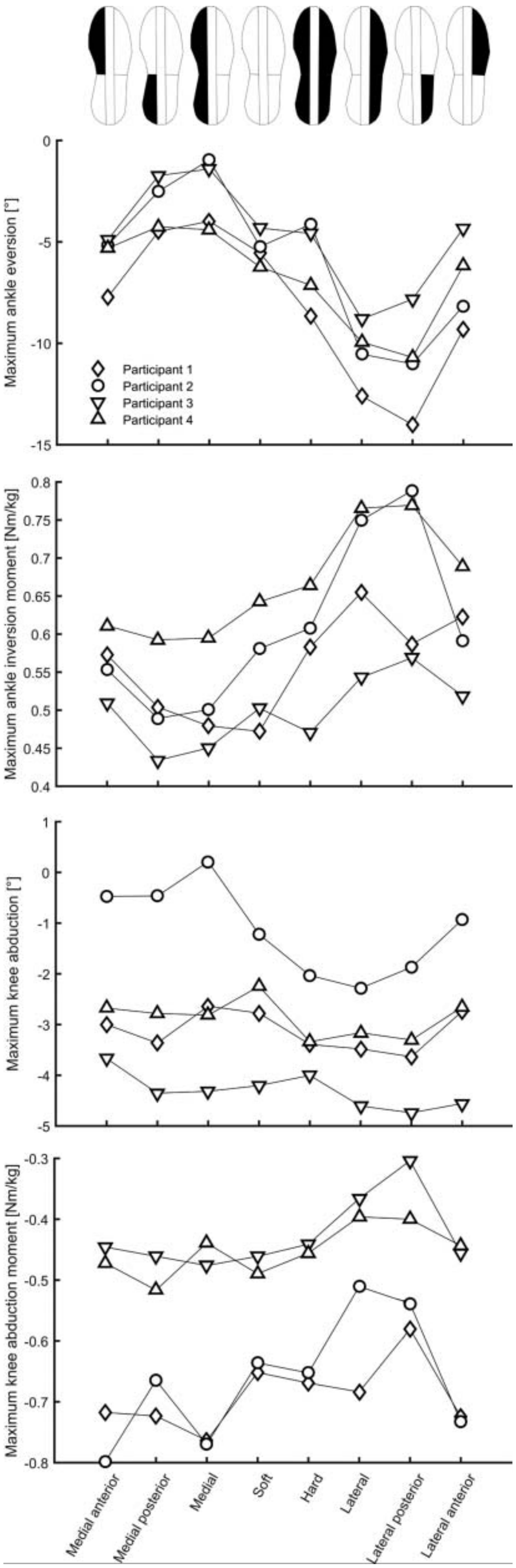

Figure 2. Effects of different midsole density distributions on kinematic and kinetic variables.

posterior (MP, pins along the posterior part of the medial side), medial (M, pins along the medial side), soft (S, no pins), hard (H, all pins), lateral (L, pins along the lateral side), lateral posterior (LP, pins along the posterior part of the lateral side) and lateral anterior (LA, pins along the anterior part of the lateral side) (Figure 2). Joint kinematics and internal joint moments were calculated using a five segment Newton Euler inverse dynamics model of the right lower extremity.

Due to the low sample size, only qualitative comparisons between conditions were performed.

Results

Overall, medial and lateral posts had a stronger effect on ankle than on knee joint kinematics and kinetics. Compared with medial posts, lateral posts led to an increase of maximum ankle eversion and maximum ankle inversion moment. Furthermore, lateral posts had an effect on reducing the maximum knee abduction moment unlike medial posts. No systematic effect of both lateral and medial posts on maximum knee abduction could be observed. Posts in the anterior region of the foot had no systematic effect on the analysed parameters.

Discussion and conclusion

The hypothesis that changing the midsole density in the forefoot region would affect joint loading was not observed.

We suggest that, during the time when the heel left the ground, changes of lower limb and foot alignment already occurred. Therefore, posting underneath the forefoot does not appear to affect the realignment of the medial arch.

The effects found for medial and lateral posts correspond to results found in the literature and led to kinematic and kinetic changes (Lewinson et al., 2013), especially at the ankle joint.

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Disclosure statement

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Effects of footwear design on rearfoot adduction in running

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Keywords: rearfoot adduction; foot kinematics; ankle joint; running kinematics; locomotion; joint kinetics

Introduction

Numerous studies have examined the mechanical coupling between foot and shank (e.g. Stacoff, Nigg, Reinschmidt, van den Bogert, Lundberg, 2000). These studies have focused on the coupling between rearfoot eversion and tibial rotation neglecting potential coupling mechanisms of rearfoot adduction to tibial rotation. Recent research suggests a strong positive correlation between these two variables in barefoot running (Fischer, Willwacher, Hamill & Brüggemann, 2017). However, the effect of footwear on global rearfoot adduction amplitudes is currently unknown. Descriptions of coupling

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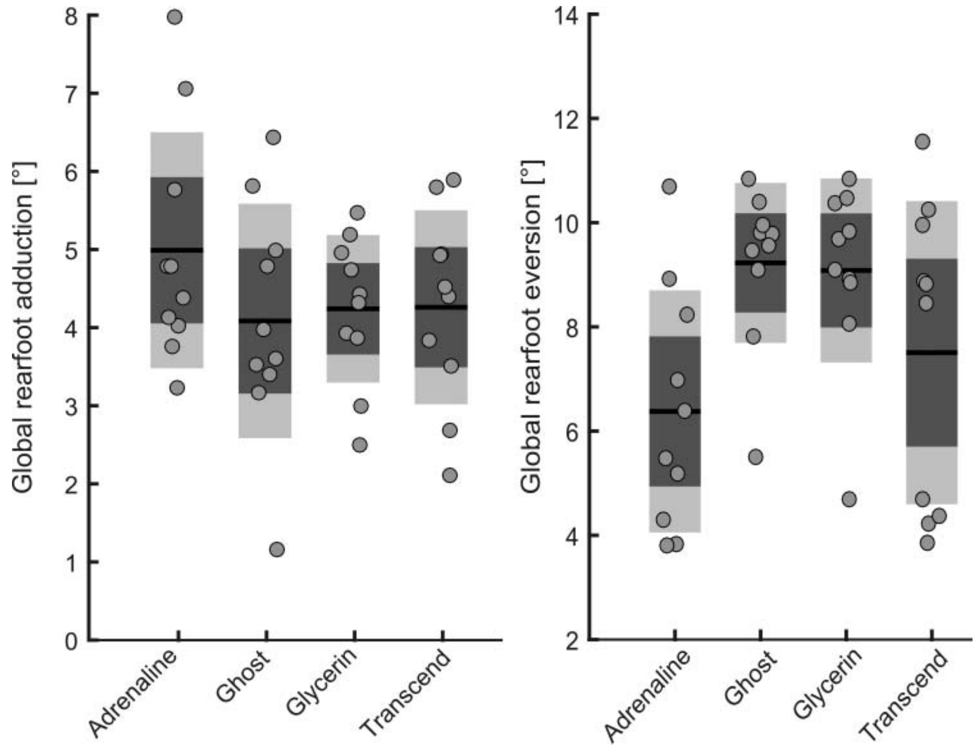

Figure 1. Comparison of global rearfoot motion for all tested conditions.

mechanisms at the ankle joint may, therefore, be incomplete and the influence of rearfoot adduction on tibial rotation remains unclear.

Purpose of the study

The purpose of this study was to investigate the influence of four different footwear midsole designs on global rearfoot adduction and eversion amplitudes and their relationship to tibial rotation during the stance phase of running.

Methods

Ten young males (age: 24.6 ± 2.0 years, body mass: 77.0 ± 7.7 kg and standing height: 184.2 ± 3.7 cm) participated in this study. Each participant was equipped with retroreflective markers attached to the shank and the rearfoot. Specifically, three markers were placed directly on the skin of the rearfoot through circular holes within the heel cap. Trajectory data were recorded by means of a three-dimensional motion capture system (Vicon Motion Systems, Oxford, UK, 250 Hz), while running over a force plate (Kistler Instrumente AG, Winterthur, Switzerland, 1000Hz) at 2.5 m/s. This task was performed in four different footwear designs (Ghost, Glycerin, Adrenaline GTS, Transcend, Brooks Sports Inc., Seattle, WA, USA). Global segment orientations within the laboratory coordinate system were determined for tibial rotation, rearfoot adduction and rearfoot eversion. During each stance phase, the time of maximum internal tibial rotation was determined and range of motion values from touch down to the time of maximum internal tibial rotation were calculated. A single factor repeated measures ANOVA combined with Bonferroni post hoc tests (alpha = 0.05) was applied to evaluate the observed effects.

Results

Significant footwear main effects for rearfoot eversion (F (3,27) = 6.3; p = 0.002; η^2 = 0.410) and for rearfoot adduction (F (3,27) = 3.1; p = 0.042; η^2 = 0.258) were found.

The Adrenaline GTS was associated with greater rearfoot adduction than the other shoes, with rearfoot eversion significantly decreased compared to the Ghost (Figure 1). No significant footwear effects for tibial rotation were found (F (3,27) = 1.2; p = 0.341; η^2 = 0.115)). When pooling all analysed trials together, global rearfoot eversion and adduction correlated significantly (p < 0.001) with tibial rotation in a partial correlation analysis (r = −0.28 and r = 0.30, respectively).

Discussion and conclusions

The support shoe (Adrenaline GTS) reduced rearfoot eversion and was associated with greater rearfoot adduction compared to the other shoes. This result indicates that the medial support was effective at reducing rearfoot eversion, as was intended by the shoe manufacturer but that further footwear technologies must be developed in order to guide calcaneal adduction. The present study found a correlation between rearfoot adduction and tibial rotation with similar strength as a previously published study on the barefoot running (Fischer et al., 2017). This indicates the relevance of rearfoot adduction control.

The missing influence of footwear on tibial rotation could be explained by the contradictory results found for the effect of footwear on rearfoot eversion compared to rearfoot adduction. On average, the contrary effects would indicate no effect on tibial rotation.

Future motion control footwear design, for runners suffering from problems related to tibial rotation, should take rearfoot adduction into consideration.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Foot kinematics in the hypermobility type of Ehlers–Danlos syndrome using the Ghent Foot Model

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Keywords: Foot kinematics; gait; Ehlers–Danlos Syndrome; foot and ankle; clinical gait analysis

Introduction

The Ehlers–Danlos syndrome (EDS) comprises a heterogeneous group of inherited connective tissue disorders. The main clinical manifestations of the hypermobility type (hEDS), the most common subtype, include mild skin involvement, severe generalized joint hypermobility, recurrent joint dislocations and chronic joint and limb pain.

A significant part of these musculoskeletal complaints is located in the lower extremity, with a 77.8% prevalence of ankle and foot pain (Rombaut, Malfait, Cools, De Paepe, & Calders, 2010). On a structural level, hEDS patients seem to have more foot anomalies such as pes planus, hallux valgus and claw toe. These foot alterations together with joint and muscle pain in the lower limb and difficulties with transmitting muscle force result in an insufficient functioning of the ankle–foot complex.

Until now, only a few studies investigated gait in patients with hEDS (Cimolin et al., 2011; Galli et al., 2011; Rigoldi et al., 2012 Celletti et al., 2013; Rombaut et al., 2011). They walk with a reduced gait velocity, step length and stride length. hEDS show altered kinematics of the pelvis, hip, knee and ankle during the different phases of gait. However, the results for the foot are limited to single segment models and remain poorly understood. Therefore, there is a need to get a better insight into the biomechanics of the ankle–foot complex of hEDS, to develop evidence-based therapy guidelines for this patient group.

Purpose of the study

The purpose of this pilot study was to investigate the foot kinematics of hEDS patients in comparison with healthy controls, using the six-segment Ghent Foot Model (GFM).

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Methods

After ethical approval, 23 female patients participated in the study (41 ± 11.0 years, 72 ± 19.3 kg, 165 ± 9.0 cm). Twenty-three healthy females, without foot complaints or injuries to the lower limb, formed the control group (41 ± 11.5 years, 66 ± 10.8 kg, 166 ± 5.4 cm). The marker configuration was in accordance with De Mits, Segers, Woodbury, Elewaut, De Clercq & Roosen, (2012). All subjects walked barefoot over a 12-m long instrumented walkway. Kinematic data were collected with an 8-camera opto-electronic system (Oqus 3, Qualysis, Sweden) at 500 Hz. A plantar pressure platform (Footscan®, RSscan, Belgium) was mounted on top of a force platform (Accugait, AMTI, USA) and both were built into the middle of the walkway. Force data were collected at 1000 Hz and pressure data at 500 Hz. A midgait protocol was used and

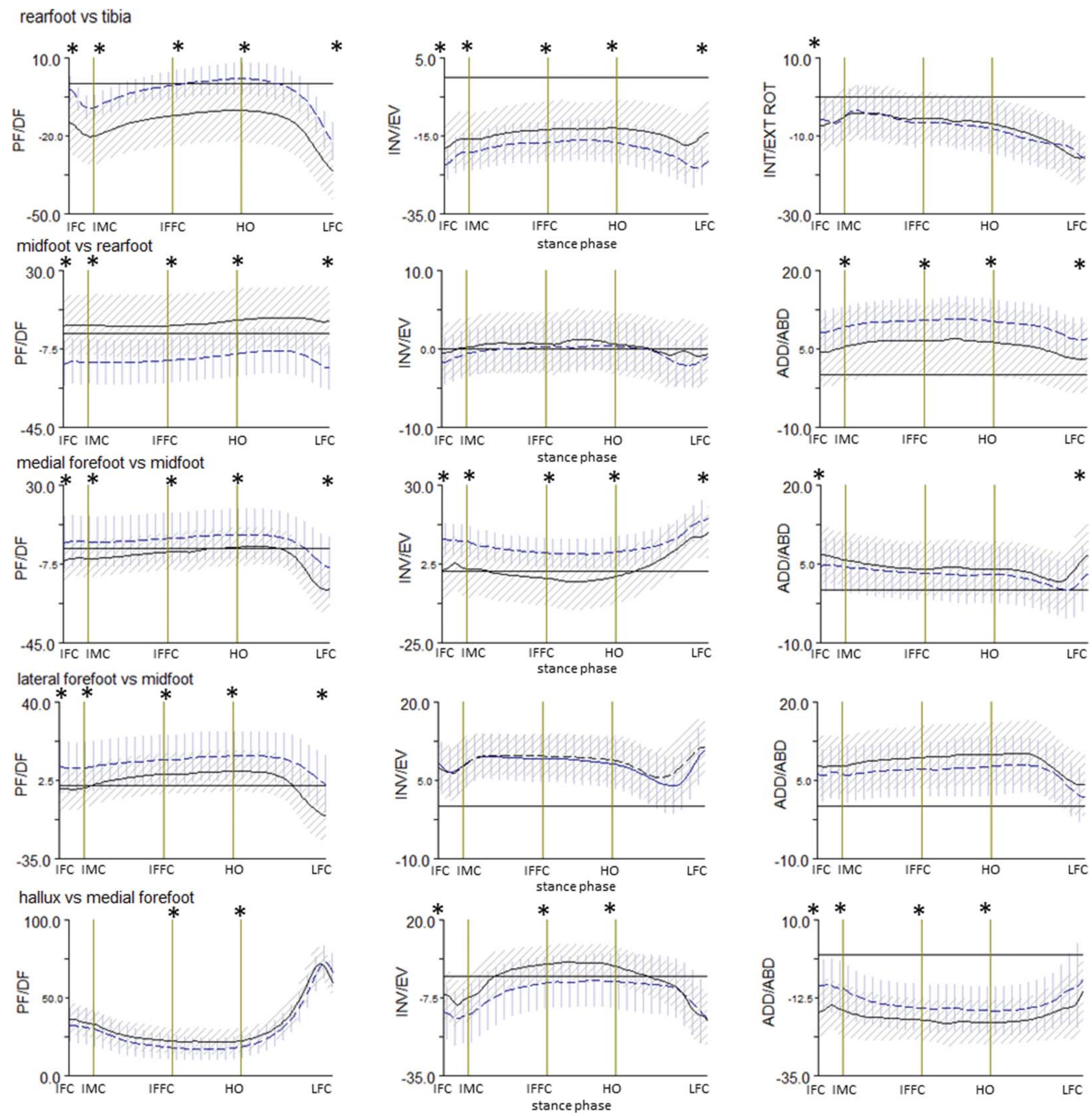

Figure 1. Foot kinematics (mean + standard deviation).
Notes: hEDS, dashed line; healthy, solid line. $^{*}p \leq 0.05$. PF/DF, plantar–dorsiflexion; INV/EV, inversion–eversion; INT/EXT ROT, internal–external rotation; ADD/ABD, adduction–abduction; IFC, initial foot contact; IMC, initial metatarsal contact; IFFC, initial forefoot flat contact; HO, heel off; LFC, last foot contact.

trials with obvious targeting for the platform were excluded from further analysis. The subjects used a self-selected comfortable walking speed.

The captured data were analyzed with Visual 3D software (C-motion, Germantown, MD, USA) using standard Euler rotations.

Linear Mixed Models (IBM SPSS 24, Chicago, IL) were used for the statistical analysis on five discrete moments in time, derived from the kinetic data (De Cock, De Clercq, Willems, & Witvrouw, 2005). A Bonferroni correction was applied and $p \leq 0.05$ was considered to be statistically significant.

Results

Preliminary results showed some significant differences between the healthy control group and the patients for the ROM of the different foot segments (see Figure 1). All segments showed significant differences in the sagittal plane. For inversion and eversion differences were found in the rearfoot, medial forefoot and hallux and for rotation the differences are mainly located in the midfoot and the hallux.

Discussion and conclusion

The foot kinematics show some significant differences between the hEDS and the healthy controls. As shown earlier, this might be related to joint hyperlaxity, reduced muscle strength, reduced endurance capacity, hypertonia of certain muscles, structural foot deformities, all clinical manifestations for the hEDS. The impact of our findings on foot function and the clinical relevance has yet to be determined in order to develop evidence-based therapy guidelines for this patient group, especially towards providing them with the correct braces, shoes and/or orthoses/insoles.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Coordination pattern between the forefoot and rearfoot during walking on an inclined surface

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Keywords: walking; inclined surface; segment coordination; multi segment foot; vector coding

Introduction

Vector coding (VC) is a nonlinear data analysis technique that provides a quantitative measure of coordination and coordination variability. The outcome measure derived from the VC technique is referred to as the coupling angle. Chang, Emmerik, and Hamill (2008) whilst reporting

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these coupling angles introduced the terms ‘in-phase’, ‘anti-phase’, ‘proximal phase’ and ‘distal phase’. Although this approach provided a fresh outlook to the coordination pattern between rearfoot and forefoot, a later study (Needham, Naemi, & Chockalingam, 2015a) built on this approach to provide a novel technique that incorporated information on segment and phase dominance. This approach enabled us to offer further understanding of the coupling angle to contain in-phase or anti-phase coordination along with segmental dominancy within 45° coordination pattern bins.

Whilst Needham, Naemi, and Chockalingam (2015b) looked at inter-segmental foot movement during level walking, one needs to understand how this movement and the coordination pattern changes whilst walking on an incline. We negotiate surfaces of varying incline during activities of daily living. An understanding of the inter-segmental movement and the coordination patterns will influence the way we design our functional footwear and foot orthotics.

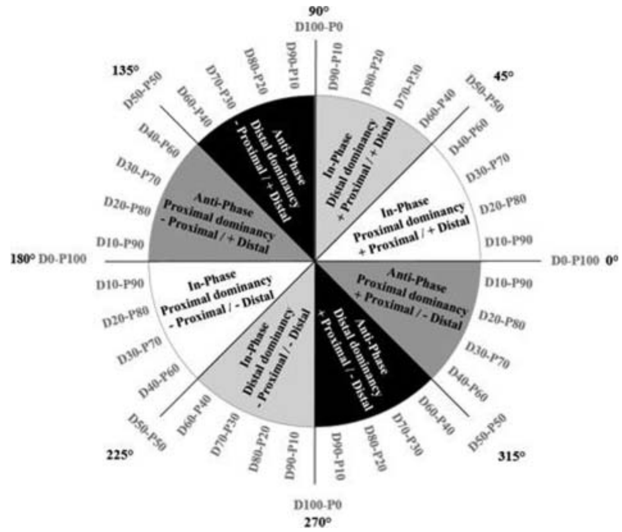

Figure 1. Coordination pattern classification. Segmental dominancy is shown around the circumference of the polar plot (grey text).

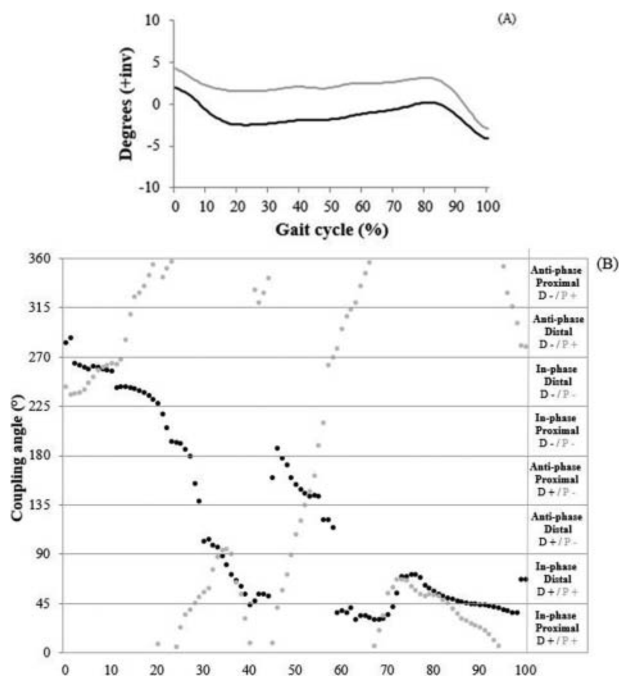

Figure 2. Mean relative kinematic waveforms (A) and coupling angle (B) for forefoot–rearfoot coordination in the frontal plane during level ground (black line/dots) and incline (grey line/dots) slope walking.

Purpose of the study

The purpose of this study is to employ the previously reported technique (Needham et al., 2015a) to examine the relationship between forefoot and rearfoot during walking on an inclined surface.

Methods

Eight male participants (mean ± SD: age: 21 ± 2.83 years, height: 180.75 ± 7.96 cm, body mass: 72.86 ± 10.57 kg) with no history of musculoskeletal impairments took part in this study. Ethical approval was sought and received from the University Research Ethics Committee. An 8-camera motion capture system (Vicon, Oxford, UK) collected rearfoot/medial forefoot segmental angle data in accordance with a marker configuration defined by Chang, Rodrigues, and Van Emmerik (2014). Participants were required to walk at a preferred speed on level ground and on an incline slope of 13°.

Segment angle data over five trials were processed in Visual3D (C-Motion, USA) and normalised for time to 100% of stance. Information on procedures and calculations regarding the VC technique can be found elsewhere (Needham, Naemi, & Chockalingam, 2014). The coupling angle was assigned to a coordination pattern classification proposed by Needham et al. (2015a) (Figure 1).

Results

Discussion and conclusion

Our results indicate that during level walking, the coordination pattern in the frontal plane was mostly in-phase with forefoot dominancy. Although these results are generally comparable to Chang et al. (2008), the minimal variation observed within the coordination pattern could be attributed to kinematic modelling of forefoot (entire metatarsals in the previously published paper versus medial forefoot in the current investigation). During walking on an inclined surface, there is still a forefoot dominance. However, there is a clear change in the coordination pattern which appears to be in anti-phase at certain marked sections of the stance phase (Figure 2(a)).

Whilst there is a similarity in the pattern of the waveforms between level and incline walking (Figure 2(a)), they do not provide detailed information on the coordination pattern. However, as indicated in Figure 2(b), the new coordination classification technique identifies subtle changes in the segmental data. These changes provide an insight into the complexity of multi-segment foot kinematics both in level and in incline walking, which has not been observed or reported in traditional approaches. The results of the current investigation and the techniques reported within this study have the potential not only to inform functional footwear design but also a wider clinical and scientific community to understand foot function.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Bipedal in-shoe kinetics of skateboarding – the ollie

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Keywords: impact forces; footwear; skateboarding; biomechanics; bipedal

Introduction

Skateboarding is a globally participated and popular sport with a reported participation of over 11 million in the United States alone (SGMA, 2007). Published epidemiological studies have stated the significant incidence of musculoskeletal injuries associated with skateboarding (Frederick, Determan, Whittlesey, & Hamill, 2006). The inherent nature of skateboarding makes it difficult to quantify using standard laboratory methods. In-field protocols have been attempted to quantify the metabolic demands of the activity, while only partial segments of a given movement have been quantified from a biomechanics perspective (Frederick et al., 2006; Hetzler, Hunt, Stickley, & Kimura, 2011). Therefore, unique methods of testing are necessary to develop a complete understanding behind the basic movements of skateboarding.

Purpose of the study

The purpose of this pilot study was to quantify the basic skateboarding manoeuvre of the ollie using novel methods and technology in the athlete’s own environment.

Methods

Four experienced male skateboarders participated in this pilot study. Each subject wore identical Vans Authentic footwear, but used their own skateboard throughout the

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data collection. Each subject was tasked with performing an ollie under three randomized conditions: (1) standing ollie (SO); (2) rolling ollie (RO); (3) ollie down (OD). The OD manoeuvre was performed utilizing a 36.0-cm platform. OpenGo (Moticon GmbH, Munich, Germany) wireless sensor insoles were utilized to continuously record underfoot forces at 50 Hz. Each insole contains 13 pressure sensors with a specified load range of 0–40 N cm^{-2} along with an accelerometer. To ensure a similar fit in the footwear, the production insoles were replaced with the OpenGo insoles during the data collection. Statistical comparisons were made using a single-factor ANOVA ($\alpha = 0.05$).

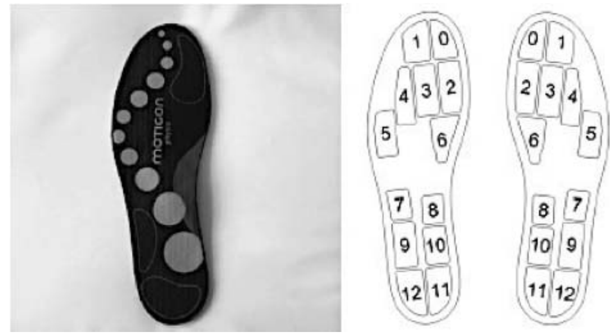

Figure 1. OpenGo wireless pressure sensing insoles with a diagram of sensor placement.

Results

The average peak take-off forces during an SO, RO and OD were 2.47 ± 0.38 BWs, 2.55 ± 0.51 BWs and 2.34 ± 0.32 BWs, respectively. Average peak landing forces of the SO, RO and OD were 2.40 ± 0.33 BWs, 2.71 ± 0.23 BWs and 3.15 ± 0.51 BWs, respectively. Pressure distribution during take-off and landing was centred around the medial forefoot in sensors 0, 2 and 3 shown in Figure 1.

Discussion and conclusion

The measured take-off forces were similar to previous studies that evaluated the impact forces from an ollie movement (Frederick et al., 2006; Nevitt, Determan, Cox, & Frederick, 2008). The landing forces in our findings were different when compared to previous literature (Nevitt et al., 2008). Nevitt et al. compared landing forces from various platform heights for the OD manoeuvre, while forces of 3.15 ± 0.51 BWs when landing from a height of 36.0 cm we found in this pilot study, they do not compare to the 4.61 ± 0.80 BWs from a platform of 22.9 cm. Previous literature on skateboarding forces utilized force plates to determine the force measurements for the whole system. It is critical to keep in mind that the board, bushings and wheels of the skateboard, as well as the footwear may provide for shock attenuation. As shown in Figure 2, the OpenGo system differentiates itself by providing for force measurements underneath each foot without the need for a tethered data logger on either the distal limb or waist of the subject (Stöggl & Mariner, 2016)

The unique methods of data collection in this pilot study allow the subject to perform in his or her given environment without external influence from testing equipment. This pilot study reveals that skaters experience significant forces underfoot while having the various factors of shock attenuation. Future investigation in the role

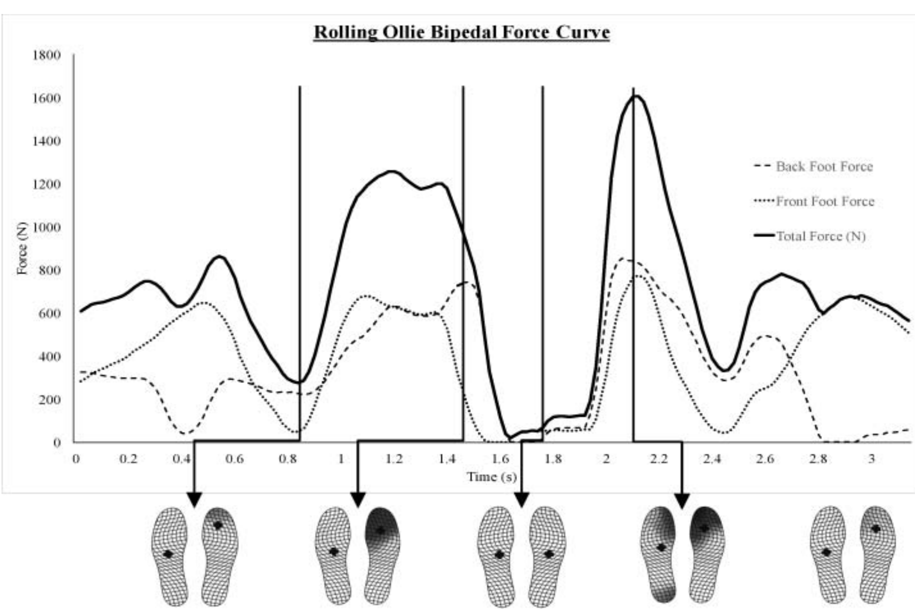

Figure 2. Forces and underfoot pressures from a rolling ollie trial.

of shock attenuation of the skateboard and footwear would be useful to footwear and skateboard manufacturers.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Effects of cleat stiffness on footwear comfort and performance in American football: A randomized control trial

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Keywords: athletic footwear; American football; cleated shoe; stiffness (bending or compressive); field testing

Introduction

Comfort and performance are the most desirable features that athletes consider when selecting footwear (Hennig & Sterzing, 2010). While footwear stiffness modifications aim to reduce the number of foot and ankle injuries during activity, the interaction between the foot and shoe surface can directly influence the perceived comfort, and therefore have functional performance implications (Wintana, Goonetilleke, Xiong, & Au, 2009). The effects of varying levels of shoe plate stiffness on comfort and functional performance are not well understood, particularly across an athletic season.

Purpose of the study

To determine the effects of footwear stiffness modification on subject-specific assessments in a large cohort of high school American football players prior to and during a football season.

Methods

Ninety-seven high-school-aged American football players (age 16.1 ± 1.2 years, height 175.3 ± 7.2 cm, mass 80.0 ± 21.3 kg) were fitted with two identical football specific cleats (CrazyQuick 2.0, adidas) with different magnitudes of plate stiffness measured by three-point bend testing (stiff: 61.2 N/mm, moderate stiffness: 47.3 N/mm). Subjects completed field-based performance assessments (Pro-agility test, T-test, vertical jump) during the pre-season. The order of testing for each cleat was randomized for each subject. Following the field-based assessments for each footwear condition; the athletes' perception of the footwear was assessed via a computer-based visual analogue scale (VAS), Likert questions and regions of discomfort chart. Following the field-based assessments; subjects' cleat preference was recorded. Each subject was then randomly assigned one of the two cleat conditions to wear throughout the season. A weekly questionnaire assessed player wear rate, perceived comfort and perceived performance. Logistic regression determined the primary predictors of athlete cleat preference. Linear mixed effects models determined differences throughout the 12-week season between cleats.

Results

During the pre-season testing, there were no statistical differences between cleats (stiff vs. moderately stiff) in the

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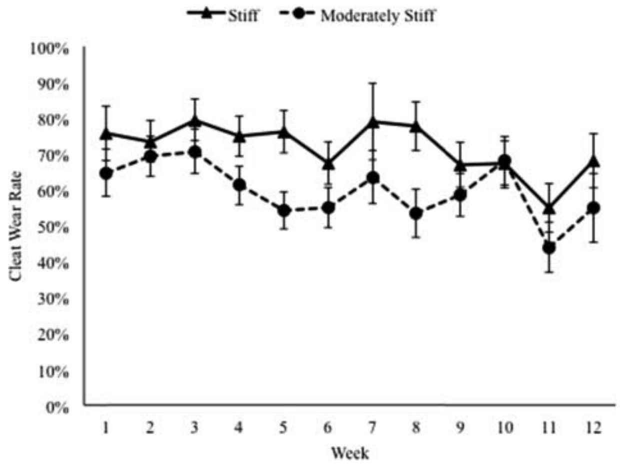

Figure 1. Wear rate during 12-week football season.

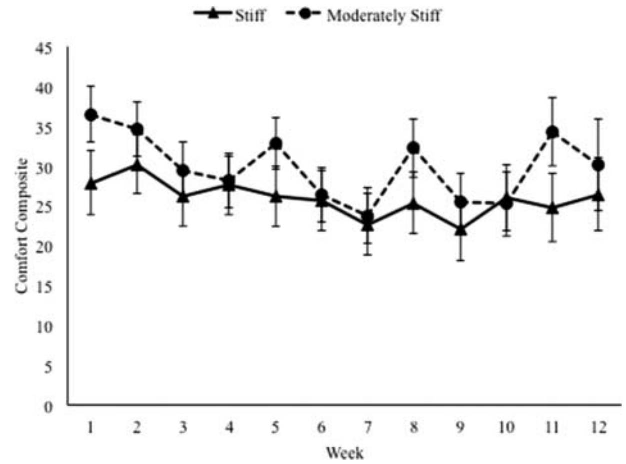

Figure 2. Composite discomfort rating during 12-week football season.

pro-agility (5.62 s (95% CI = 5.76–5.47 s) vs. 5.63 s (95% CI = 5.78–5.48 s), $p = 0.520$), vertical jump (53.94 cm (95% CI = 51.75–56.13 cm) vs. 53.95 cm (95% CI = 51.77–56.12 cm), $p = 0.900$), or T-test ((12.75 s (95% CI = 13.19–12.32 s) vs. 12.69 s (95% CI = 13.09–12.29 s), $p = 0.103$). Immediately following the performance testing, discomfort ratings were not statistically different between conditions for whole shoe ((19.5 (95% CI = 15.3–23.8) vs. 18.6 (95% CI = 14.3–22.9), $p = 0.721$), insole ((24.2 (95% CI = 19.4–28.9) vs. 20.4 (95% CI = 16.0–24.9), $p = 0.112$), shoe upper ((22.4 (95% CI = 18.1–26.8) vs. 22.1 (95% CI = 17.2–27.1), $p = 0.898$), or composite score ((22.2 (95% CI = 18.3–26.1) vs. 20.1 (95% CI = 16.1–24.2), $p = 0.268$). Overall, 58.9% of the players preferred the moderately stiff footwear compared to 41.1% that preferred the stiffer footwear. Logistic regression identified that the difference score of composite visual analogue discomfort rating ($p < 0.001$) and difference score in T-test performance ($p = 0.011$) were the two primary predictors of cleat preference, correctly classifying 87.5% of the cases. Figure 1 illustrates the wear rate of the footwear during the 12-week football season. The wear rate for stiff footwear (71.8% (95% CI = 63.1%–80.5%)) was significantly greater ($p = 0.049$) compared to moderately stiff (59% (95% CI = 51.7%–67.9%)). There was a main effect for week, where the cleats were worn less near the end of the season ($p = 0.002$). Composite discomfort rating (average of three VAS ratings) was not statistically different during the 12-week season ((25.9 (95% CI = 19.7–32.1) vs. 29.9 (95% CI = 24.2–35.6), $p = 0.349$) (Figure 2). Self-reported rating of performance was not different between cleats ($p = 0.3$). The overall injury severity score over the 12-week season was not different between footwear ($p = 0.93$; stiff: 10.2 (5.9–14.5); moderately stiff: 10.3 (6.3–14.3)).

Discussion and conclusion

Difference scores between cleats in a rating of discomfort and an agility T-test performance correctly predicted initial cleat preference (stiff or moderately stiff). Therefore, comfort (initial and in-season) and performance are likely important factors that influence the overall evaluation of footwear preference in high school football players.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

adidas International, inc

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The effect of shoe sole stiffness on plantar pressure and patient satisfaction in patients with diabetes at high risk of foot ulceration

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Keywords: footwear; pressure distribution; forefoot (foot); biomechanics; plantar pressure

Introduction

High plantar peak pressure is an important risk factor of foot ulceration in diabetes. Custom-made footwear aims to redistribute high plantar pressures from plantar foot regions at high risk for ulceration to regions at less risk for ulceration. Different aspects of the shoe can contribute to the relief of plantar peak pressure, including a rocker bottom outsole configuration of the shoe (Hutchins, Bowker, Geary, & Richards, 2009; Chapman et al., 2013). Shoe outsole stiffness has also been suggested to affect forefoot peak pressure, but little is known about this relationship in at-risk diabetic patients.

Purpose of the study

The purpose of this study was to assess the effect of shoe outsole stiffness on forefoot plantar peak pressures during gait and patient satisfaction in patients with diabetes who are at high risk of foot ulceration.

Methods

Twenty-four diabetic patients (16 male, mean age 67 years) at high ulcer risk were included in this cross-sectional study. A custom-made insole that was designed based on the barefoot peak pressures and three-dimensional foot shape data of the patient was tested in an extra-depth diabetes shoe with either a tough 1.8 cm thick rubber outsole with 18 degrees rocker configuration (Careline tough, Choose Your Shoes, Heythuysen, the Netherlands) or the same shoe with same rubber outsole and rocker configuration but with a 3 mm carbon reinforcement of the outsole (Careline stiff). The shoe conditions were evaluated in random order and pressures were evaluated with Pedar-X (Novel, Munich) while the patient walked at a comfortable speed that was standardized between conditions. Peak pressures were calculated for the metatarsal heads, hallux, midfoot and heel regions. Patient satisfaction was assessed using a Visual Analogue Scale (score 0–10, 10 representing best perception) for walking comfort, shoe fit, shoe weight and appearance. Outcomes were compared using repeated measures ANOVA ($P < 0.05$).

Results

Significantly lower metatarsal head peak pressures were present with the Careline stiff outsole compared to the Careline tough outsole. Mean peak pressures were lower by 9%–12% (Table 1). In >83% of patients with the Careline stiff and in >71% of cases with the Careline tough, metatarsal head pressures were <200 kPa, a level found to indicate some protection against foot ulceration (Waaijman et al., 2014). No significant effect of shoe outsole stiffness was found in the hallux region. Patient satisfaction generally received moderate scores (Table 2). Higher satisfaction scores on walking comfort, shoe fit, shoe

Table 1. Mean peak pressures and standard deviation per condition per region (in kPa).

	MTH1	MTH2-3	MTH4-5	Hallux
Careline stiff	146 ± 52	158 ± 44	107 ± 36	164 ± 37
Careline tough	163 ± 59	180 ± 41	117 ± 38	171 ± 48
P-value	<.001	<.001	<.001	0.141

Note: Data are mean ± SD ($n = 24$).

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Table 2. Patient satisfaction scores per condition.

	Walking comfort	Shoe fit	Shoe weight	Shoe appearance
Careline stiff	5.8 ± 2.3	5.7 ± 2.3	7.3 ± 2.3	4.7 ± 3.1
Careline tough	5.9 ± 2.2	6.1 ± 2.7	7.8 ± 1.8	5.1 ± 2.9
P-value	0.871	0.499	0.069	0.101

Note: Values are mean ± SD ($n = 24$), between 0 and 10 (10 is highest possible satisfaction).

weight and shoe appearance were found with the Careline tough outsole, but with small non-significant differences present between the two outsole conditions.

Discussion and conclusions

A stiff shoe outsole led to significantly lower peak pressures at the metatarsal heads than a less stiff, tough shoe outsole. Shoe sole stiffness appears to have only small non-significant effects on items of patient satisfaction that are considered relevant in diabetic patients who are at risk for foot ulceration. Therefore, a reinforced stiff shoe outsole is recommended to relief forefoot peak pressures in high risk patients. A reinforced stiff outsole has been suggested to potentially cause issues with gait stability in diabetic patients who have severe loss of proprioception, and this should be taken into account as a possible contraindication. Further investigation is required to assess the effect of range of outsole stiffnesses on plantar peak pressures and walking comfort to assist in defining the most optimal outsole stiffness for shoes for high-risk diabetic patients.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Check for updates

Effects of minimalist footwear and stride length reduction on the probability of metatarsal stress fracture: a weibull analysis with bone repair

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Keywords: minimal footwear; stress fractures; running; biomechanics; finite element models (FEM)

Introduction

Physical activity such as distance running exposes bones of the lower extremity to lengthy bouts of repetitive loading, which causes mechanical fatigue characterized by the manifestation of tissue microdamage. If the rate of microdamage accumulation is greater than the rate of bone repair, tissue quality may be diminished resulting in an increased risk of stress fracture. The metatarsal bones are highly susceptible to this injury, encompassing 10% of all stress fractures in active populations (Iwamoto et al., 2003). Microdamage formation in bone is a strong function of strain magnitude; therefore, mechanisms that reduce bone strain may also reduce stress fracture risk.

Purpose of the study

The purpose of this research was to determine the effect of minimalist footwear and stride length reduction on the probability of metatarsal bone stress fracture in running.

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Table 1. Description of running conditions.

Condition	Shoe type	Stride length
1	Traditional	PSL
2	Traditional	90% PSL
3	Minimalist	PSL
4	Minimalist	90% PSL

Note: PSL = preferred stride length.

Methods

Fourteen male recreational runners (age: 26.2 ± 4.2 y; height: 178.4 ± 5.4 cm; mass: 75.6 ± 5.6 kg) participated in this study. Subjects ran more than 10 km/week, had no lower limb injuries within three months prior to testing, were rearfoot strikers, and had no prior experience running in minimalist footwear.

Computed tomography (CT) scans were obtained for each subject's right foot in both a traditional (cushioned) and minimalist (uncushioned) shoe. Subjects subsequently ran overground at their preferred speed for a 5-km run (±5%) in four conditions (Table 1) while motion capture and plantar pressure data were collected.

Seven retro-reflective markers were adhered to each shoe to create a rearfoot and a forefoot segment. The sagittal angle of the rearfoot segment was added to the neutral metatarsal angles from the CT scans to obtain sagittal-plane metatarsal angles throughout ground contact. These metatarsal angles were then combined with plantar pressure data to calculate forces acting on the metatarsal head using a musculoskeletal (MSK) model of the metatarsophalangeal joint (Stokes, Hutton, & Stott, 1979).

Metatarsal bones were segmented from CT scans to create subject-specific finite element (FE) models. Elements were assigned isotropic, linear-elastic material properties as a function of bone apparent density (Rho, Hobatho, & Ashman, 1995). The proximal end of the FE models was constrained and forces from the MSK model were applied distally at the metatarsal heads. Von Mises equivalent strains were computed from the FE model to obtain a single scalar representation of strain magnitude for each element. A Weibull analysis that incorporated the effects of strain magnitude and stressed volume on damage accumulation as well as damage repair by basic multicellular units was used to calculate a subject-specific probability of failure (P_{fr}) for a running volume of 20 km/week over 100 days (Edwards et al., 2009).

Repeated measures ANOVAs were used to test for differences in P_{fr} between conditions for each metatarsal ($p \leq 0.05$). A Bonferroni post-hoc correction was used to account for experimentwise error.

Results

No significant interactions were observed between shoe type and stride length. The P_{fr} was significantly greater in the minimalist shoe for metatarsals 1, 2, and 4 (Table 2), however no significant differences in P_{fr} were observed between stride length conditions.

Discussion and conclusion

The minimalist shoe was associated with a higher risk of P_{fr}, or stress fracture when compared to running in a traditional shoe. A post hoc examination revealed smaller metatarsal angles at the timepoint of peak plantar force in the minimalist shoe, causing a greater applied bending moment to the bone and a larger strain. While small reductions in strain were observed at 90% PSL, these were not large enough to overcome the additional damage that occurred from an increased number of loading cycles for the 20 km/week distance.

These results suggest that running in minimalist shoes significantly increases one's risk of developing a metatarsal stress fracture, while running at a reduced stride length is not an effective technique to mitigate metatarsal stress fracture risk. Further research is required to determine if the increased bending moment observed in minimalist shoe conditions was a function of cushioning, heel-to-toe drop, or some combination of the two.

Table 2. Estimated marginal means for peak Pfr (%) as a function of shoe type and stride length (SD).

Met	1	2	3	4	5
TS	0.004 (0.007)	0.56 (0.26)	0.22 (0.70)	0.02 (0.02)	0.004 (0.007)
MS	0.050 (0.080)	6.53 (2.53)	0.86 (1.94)	0.15 (0.18)	0.037 (0.080)
p	**0.036**	**0.024**	0.081	**0.008**	0.115
PSL	0.027 (0.054)	4.01 (6.47)	0.49 (1.10)	0.09 (0.11)	0.018 (0.030)
90% PSL	0.026 (0.037)	3.08 (3.86)	0.59 (1.55)	0.08 (0.09)	0.022 (0.060)
p	0.908	0.262	0.480	0.558	0.672

Note: TS = traditional shoe, MS = minimalist shoe, and met = metatarsal.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Foot strike classification: a comparison of methodologies

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Keywords: biomechanics; forefoot strike; gait analysis; joint moments; foot and ankle

Introduction

Classification between forefoot (FFS) and rearfoot (RFS) strike running patterns are important for identifying different types of running styles and the functional role of the foot and ankle (Altman & Davis, 2012). However, there is yet to be a classification method that represents the musculoskeletal function of the foot. For example, the foot strike index (FSI) identifies where the centre of pressure is relative to the proximal end point of the plantar surface (Cavanagh & Lafortune, 1980). However, this does not explain the turning effect about the ankle because both the position of the ankle and orientation of the ground reaction force (GRF) vector are not considered. Other studies consider a foot progression angle at foot strike (FSA) (Altman & Davis, 2012). To our knowledge, no studies have evaluated foot strike classification methods against muscular activity about the ankle, or rotational kinetic energy, even though these contribute to global effect of loading rate (Lieberman et al., 2010).

Purpose of the study

This study evaluates the classification of foot strike between traditional variables and a more functional measure of ankle angular momentum. Further, we investigate correlations between local foot strike measures and the global variable of loading rate.

Methods

Thirty healthy male elite athletes ran barefoot (BF) and shod (SH) across a 30-m force-instrumented runway at self-selected speed, completing 10 valid running trials per condition. 3D kinetic and kinematic data was synchronized and captured by Nexus software (Oxford Metrics Ltd, UK). Biomechanical model and parameters were processed in Visual3D (C-Motion, USA). Foot strike index, foot progression angle, and ankle joint moment were derived from kinetic and kinematic data; average loading rate (ALR) was derived from analogue signals. The functional foot strike classifier was based on ankle angular impulse, the sum of the time-integrated ankle moment signal during the period when the vertical ground reaction force (VGRF) ascended from 20 to 100% of body weight. Two FSA methods were used: FSA_0 relative to 0 degrees (Altman & Davis, 2012), and FSA_L relative to 5.6 degrees (Lieberman et al., 2015). Two FSI methods were used: relative to proximal foot end point (FSI_C, Cavanagh & Lafortune, 1980), and a new FSI based on centre of pressure with respect to ankle and metatarsal anatomical landmarks (FSI_A). The average loading rate was

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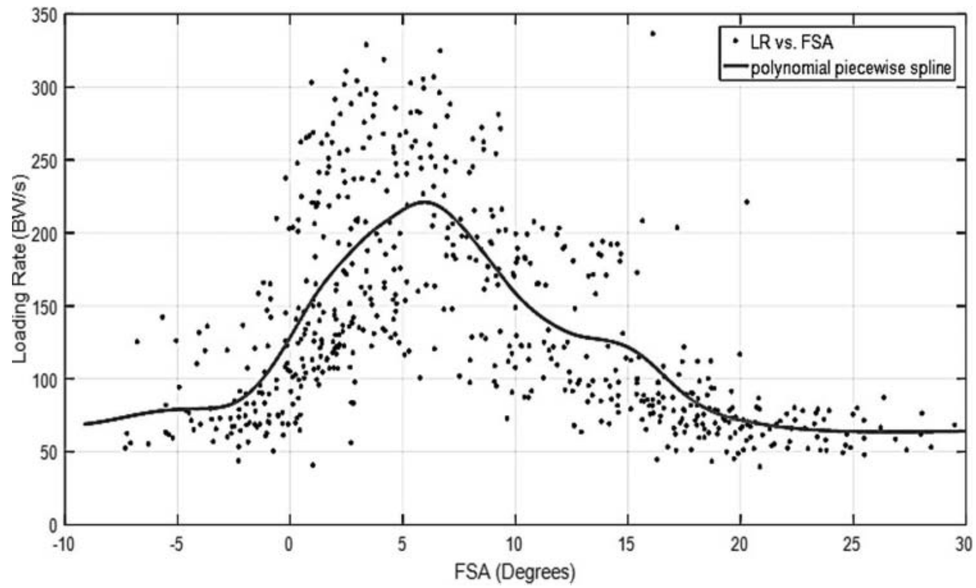

Figure 1. Average loading rate and foot strike angle. A multi-polynomial smoothing spline, $f(x)$ = smoothing spline ($p = 0.04$), $R^2 = 0.49$. FSA below 5.9 degrees are FFS.

calculated according to conventional methods, except we chose the period when VGRF ascended from 20 to 100% body weight. Linear and nonlinear regression analysis was performed in MATLAB (MathWorks Inc, USA).

Results

Table 1 presents the results of the classification parameters. The conventional foot strike index method (FSI_C), and the zero reference foot progression angle method (FSA_0), demonstrated the largest difference when compared against AnkleH_{ratio}. Using McNemar's test for change in classification, the parameters that were not statistically different were FSA_L and FSI_A.

Figure 1 shows a plot between FSA and average loading rate for all foot strike landings ($n = 600$). A polynomial regression reveals a maximum for ALR at a FSA of 5.9 degrees. Linear regression revealed significant correlations between ALR and FSA for three of the four running conditions: (SH & RFS, −0.72), (SH & FFS, +0.52), (BF & FFS, +0.74).

Table 1. Percentage of cases that reported FFS classification when comparing against FFS classified by AnkleHratio.

Variable	%FFS by Var	%FFS by H_{ratio}	%Diff	P value
FSA_L	47.5	51.7	4.2	0.23
FSA_0	12.5	51.7	39.2	<0.001
FSI_A	44.2	51.7	7.3	0.08
FSI_C	37.5	51.7	14.2	<0.001

Discussion and conclusion

This study showed that when employing an FSA reference value of 5.9 degrees, or ankle joint defined FSI, we find a foot strike classification method that is more closely aligned with the ankle function compared to conventional methods. Further, FSA of 6 degrees was associated with the highest loading rates, but these occur least frequently. Avoiding midfoot strike may be evidence of a kinematic cost policy that serves to reduce high loading rates; however, this requires landing strategies that are at opposite ends of a FSA continuum: either rely upon shoe cushioning for large dorsiflexion, or rely upon intrinsic foot structure and function for large plantarflexion.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The effect of Spraino® slide patches on muscle activity and ankle joint loading during a turning manoeuvre

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Keywords: ankle sprain; athletic footwear; foot and ankle; turning; prevention

Introduction

Ankle injuries are among the most frequent injuries sustained in sports with approximately 75% being lateral sprains (Fong, Hong, Chan, Yung, & Chan, 2007). Prevention of ankle sprains is a broad field of study and several studies have been conducted trying to assess how to minimize the chance of getting an ankle sprain with a previous sprain being the most important risk factor (Beynnon, Murphy, & Alosa, 2002). Therefore, avoidance of first-time sprains would be the most effective intervention.

It has been suggested that the positioning of the foot and activation of the muscles stabilizing the ankle affect the occurrence of ankle sprains (Wright, Neptune, van den Bogert, & Nigg, 2000). Spraino® slide patches (Spraino ApS, Copenhagen, DK) claim to reduce the friction between the foot and the surface when the foot makes initial contact in extreme positions. This causes the foot edge to slide, thereby turning into a stable position before being fully loaded.

Purpose of the study

The purpose of this study was to investigate the effect of Spraino® slide patches on ankle joint moments and muscle activation during a 180-degree turning manoeuvre.

Methods

Eleven practitioners (9 females) of team sports participated in the study (age: 22.0 ± 1.0 years; height: 173 ± 5 cm, weight: 69.0 ± 9.5 kg). They were injury free for six months prior to the experiment and did not show any signs of an instable ankle joint.

Each subject carried out five trials with and without the slip patches (10 mm wide, Figure 1) in random order, with 5-m run-up at approximately 80% of maximum intensity with sufficient rest periods to minimise fatigue. Ground reaction forces (GRFs) were collected a force plate (AMTI OR6, USA) at 1000 Hz. Kinematic data were recorded from 26 retroreflective markers placed on anatomical landmarks of the lower extremities and shoes with an eight-camera motion capture system (Qualisys AB, Gothenburg, Sweden, 500 Hz). Data were low-pass-filtered with cut-off frequencies of 100 and 14 Hz, respectively. A custom model was generated in Visual 3D (C-Motion, Germantown, USA). GRF, ankle kinematics and ankle moments were extracted for the whole stance phase as well as the first and last 30 ms; early contact (EC) or late contact (LC), respectively. Electromyographic data were collected from five leg muscles using biovision (Wehrheim, Germany) amplifiers and a data-logger, sampling at 2000 Hz. EMG data were band-pass-filtered (20-500 Hz) and rectified. Mean muscle activity was calculated from 50 ms prior to contact to TD (PRE), within the first 120 ms (MEC) and from 120 to 220 ms after contact (MCO).

A paired *t*-test with Bonferroni adjustment was applied to compare between footwear conditions ($P < 0.05$).

Results

The patches did not affect the contact time, vertical or horizontal GRF of the turning front foot. There was no significant effect of the slip patches on ankle kinematics or inversion ankle moments during EC or LC. Further, none of the subjects reported any experience of slipping episodes during all tests.

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Figure 1. Spraino® slide patch (10 mm) on the lateral forefoot.

The medial gastrocnemius and soleus muscle showed a significantly increased ($P < 0.05$) activity during MLC.

Discussion and conclusion

The main result of this study was that Spraino® slide patches did not alter ground contact mechanics during a 180-degree change of direction task. In particular, the lack of a difference in horizontal GRF indicates that the use of the patches does not reduce friction and is therefore not detrimental to performance or mechanical safety.

The observation that the EMG signals did not change during pre-contact as well as in a time window representing potential reflex-mediated activity (MEC; <120 ms) further supports the notion that stability was not compromised and that no slipping occurred. A manual screening of all trials confirmed this observation and showed that all subjects made initial contact with the medial side of the shoe.

A significant increase in plantar flexor activity during late stance may be related to the push-off which may require higher activity of these muscles due to a different foot alignment.

It can be concluded that the use of lateral slip patches may not compromise performance and ankle safety in 180-degree change of direction tasks. It cannot be extrapolated if the patches would work in a case where the lateral sole of the shoe catches the ground (Wright et al., 2000) while it remains plausible that a protective effect exists.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Influence of rugby surface type and boot stud length on lower limb biomechanical variables associated with Achilles tendinopathy during Rugby Union specific movements

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Keywords: sports biomechanics; rugby; cleated shoe; artificial turf; Achilles tendinopathy

Introduction

Achilles tendinopathy poses a challenge to medical practitioners within Rugby Union, especially during the competitive season (Cook & Purdam, 2013), with injury resulting in time loss occurring during matches (0.78 per 1000 hours) and training (1.7 per 1000 hours) (Sankey, Brooks, Kemp, & Haddad, 2008). Reported biomechanical variables associated with an increased risk of

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developing Achilles tendinopathy include large magnitudes of rear foot movement, tibial rotation and dorsiflexion range of motion (Lorimer & Hume, 2014). Changes in sports playing surfaces and footwear are also cited as a contributory risk factor (Knobloch, Yoon, & Vogt, 2008).

Three playing surfaces are used in Rugby Union: natural turf (NAT), third-generation artificial turf (3G) and more recently combination pitches composed of natural turf augmented with artificial fibres (AUG). Players will often alter rugby boot stud/cleat length in an attempt to mitigate the mechanical characteristics of the different playing surfaces.

Purpose of the study

The purpose of this study was to determine if biomechanical risk factors for Achilles tendinopathy are modified by playing surface and footwear when running and performing rugby-specific movements, to help identify appropriate footwear characteristics for the different playing surfaces.

Methods

Fourteen injury-free male senior academy players from an Aviva Premiership Rugby Union team participated in the study (97.43 ± 6.80 kg, 1.83 ± 0.04 m, 20.87 ± 1.07 years). During data collection participants wore standardized footwear with a six forefoot stud and two heel stud configuration (Waitangi, Mizuno). Three-dimensional motion of the lower limbs and trunk was recorded at 200 Hz using a four-sensor active marker opto-electronic motion capture system (Coda, Charnwood Dynamics, UK). Plantar pressure data were synchronously collected at 100 Hz using pressure insoles (Pedar, Novel GmbH). Participants performed straight line running (RUN), ninety degree change of direction (90CoD) and 95 kg resisted sled pushing tasks (PUSH; Figure 1), over three surface conditions (NAT, 3G and AUG) and three stud length conditions (13 mm, 16 mm and 21 mm length studs; Figure 2). Statistical comparisons of the left leg

Figure 1. PUSH task.

Figure 2. Stud conditions: 13, 16 and 21 mm.

(change of direction leg) were made using a two-way (Surface × Stud Length) ANOVA with repeated measures (alpha = 0.05).

Results

RUN: Lower limb joint angles were not influenced by changes in playing surface ($P > 0.05$). However, the 21-mm stud condition resulted in increases in peak rearfoot eversion angle, tibial internal rotation range of motion compared to the 13 mm condition as well as increased peak pressures at the medial and lateral forefoot ($P < 0.001$). **90CoD**: The 3G surface demonstrated increased peak medial heel plantar pressures when compared to the NAT surface ($P < 0.001$). When compared with 13 mm, 21 mm resulted in increases in peak rear foot inversion angle and increased medial forefoot, lateral midfoot and lateral heel peak pressures ($P < 0.001$). **PUSH**: The 21-mm stud condition resulted in significantly higher peak dorsiflexion angle, peak pressures at the toes, medial and lateral forefoot compared to the 13 mm condition ($P < 0.001$). An interaction effect was observed between surface and stud length for peak dorsiflexion angle and peak pressures at the medial and lateral heel ($P < 0.001$).

Discussion and conclusion

Acute alterations in playing surface did not influence lower limb motion; however, alterations in plantar pressure distribution were observed between 3G and NAT during the 90CoD and PUSH tasks. This suggests that plantar pressure distribution was altered, this was not sufficient to influence Achilles tendinopathy risk factors. The longest stud condition altered ankle joint and rear foot kinematics in addition to plantar pressure distribution when compared with the smallest stud condition, regardless of task. These alterations in joint kinematics are suggested to be influenced by an increased amount of surface penetration afforded by the 21-mm studs. The observed increases in rear foot motion during the 90CoD task and tibial rotation during the RUN task suggest that the 21-mm stud condition may increase risk of Achilles tendinopathy (Lorimer & Hume, 2014). Further prospective research is needed to determine if stud length is an external risk factor for Achilles tendinopathy development.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Changes in lower limb biomechanics and metatarsal stress fracture with different military boots

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Keywords: stress fractures; work boots; plantar pressure; metatarsals; heel loading

Introduction

For Royal Marines (RM) recruits, metatarsal stress fractures have been highlighted as a particular concern (Ross & Allsopp, 2002). The footwear worn has been identified as one possible factor contributing to this injury.

RM recruits are issued with boots at the start of training. Following several years with a consistent standard boot (Boot A), two new boot models were introduced (Boot B and Boot C). The maintenance of an injury database throughout this period identified a marked reduction in metatarsal stress fracture incidence, leading to suggestions that the change in footwear might have contributed.

Suggested biomechanical risk factors for metatarsal stress fractures include high forefoot loading and ankle plantar-flexor moment, and restricted ankle dorsi-flexion (Hughes, 1985; Nunns, Stiles, & Dixon, 2012).

Purpose of the study

The purpose of this study was to investigate the influence of military boots on peak forefoot pressure, ankle dorsi-flexion angle and ankle plantar-flexor moment. It was hypothesised that peak forefoot pressure and ankle moment would be reduced for the new boot models, whilst ankle dorsi-flexion angle would be increased. Peak rate of loading of impact force was also compared for the boot types.

Methods

Thirty RM recruits performed five walking and five running trials in each of the test footwear. Synchronised force plate (AMTI, 800 Hz) and kinematic data (Codamotion, 400 Hz) were collected. Pressure insole data were collected at 100 Hz (Novel, Germany). For walking trials at 1.4 m s^{-1} (±5%), volunteers carried a bergen (35 kg) and weapon. For running trials, a speed of 3.6 m s^{-1} (±5%) was used.

Peak forefoot pressure, ankle dorsi-flexion angle, ankle plantar-flexor moment and loading rate of impact force were determined. Group mean values were compared using one-way ANOVAs with repeated measures ($P < 0.05$).

Results

Results for loaded walking and for unloaded running are presented in Tables 1 and 2, respectively.

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Table 1. Mean (SD) for peak forefoot (FF) pressure, ankle dorsi-flexion (DF), ankle plantar-flexor (PF) moment and loading rate during loaded walking (*indicates different from Boot A).

	Boot A	Boot B	Boot C
Peak FF pressure (MPa)	125.0 (49.8)	103.1 (45.0)	83.9* (38.4)
Ankle DF (°)	9.1 (4.3)	11.1 (4.2)	10.8 (3.5)
PF moment (N m/kg)	2.08 (0.19)	2.06 (0.18)	2.08 (0.28)
Loading rate (BW/s)	50.9 (17.8)	60.2 (28.6)	63.6* (25.5)

Table 2. Mean (SD) for peak forefoot (FF) pressure, ankle dorsi-flexion (DF), ankle plantar-flexor (PF) moment and loading rate during running (*indicates different from Boot A).

	Boot A	Boot B	Boot C
Peak FF pressure (MPa)	123.4 (50.0)	100.6* (39.9)	86.5* (40.2)
Ankle DF (°)	21.7 (6.6)	22.8 (6.8)	23.0 (3.4)
PF moment (N.m/kg)	2.35 (0.65)	2.25 (0.76)	2.54 (0.24)
Loading rate (BW/s)	176.8 (76.6)	159.4 (66.4)	181.0 (83.4)

Compared with Boot A, peak forefoot pressure was lower in Boot C during walking, and lower in Boot B and Boot C during running ($P < 0.05$). Peak ankle dorsi-flexion angle and ankle plantar-flexor moment were not influenced by the footwear worn. Compared with Boot A, peak loading rate was greater in Boot C ($P < 0.05$).

Discussion and conclusion

The greater forefoot pressure for Boot A compared with Boot B (running) and Boot C (walking and running) suggests that training in Boot A presents a greater risk of developing metatarsal stress fractures. The observed reduction in metatarsal stress fracture incidence with the move away from this boot supports this suggestion.

The similar peak ankle dorsi-flexion angle and ankle plantar-flexor moment for all boot conditions suggests no advantage of either of the new boot models with regard to these specific variables. Since peak loading rate has been associated with tibial stress fracture risk (Zadpoor & Nikooyan, 2011), the greater rate for Boot C compared with Boot A (walking) suggests aspects of this boot design may increase injury.

In conclusion, the reduction in forefoot loading provides a plausible explanation for the lower metatarsal stress fracture incidence with the change in standard issue boots.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The effect of insole hardness distribution on calf muscle loading and energy return during a forward badminton lunge

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Keywords: badminton; insoles; hardness; modelling; lunge

Introduction

With 200 million practitioners, badminton may be considered as one of the world's most popular sports. The lunge is the most fundamental badminton manoeuvre and accounts for approximately 15% of all movements (Hong, Wang, Lam, & Cheung, 2014). Wearing appropriate footwear may reduce the risk of injury through shock attenuation and stabilization, and potentially improve performance. Energy return concepts have repeatedly been applied to footwear with inconsistent results (Stefanyshyn & Nigg, 2000). It appears conceivable that foot positioning within a shoe and ankle movement during ground contact may affect the energy storage and return within the musculoskeletal system. This potential interaction between footwear and biological system has not yet been investigated for a badminton-specific lunge movement.

Purpose of the study

The aim of this study was to determine the effect of insoles with varying segmented hardness in a badminton shoe on elastic energy utilization within the gastrocnemius muscle-tendon complex.

Methods

Fourteen male badminton players (age: 24.8 ± 7.7 years, weight: 72.6 ± 7.0 kg, height: 178 ± 5cm) were recruited for the study. The subjects wore a badminton shoe (Li-Ning AYAK23-2, Beijing, China), with holes cut for marker placements to allow direct measurements of the foot kinematics. Insoles incorporating different hardness in the rearfoot, midfoot and forefoot regions were designed (Table 1). The insoles were built by a Milling machine (Sensor Medica, Rome, Italy). The hardness were categorized as Soft (35 Asker C), Medium (48 Asker C) and Hard (60 Asker C).

The right forward lunge protocol was adapted from the previous studies (e.g. Hong et al., 2014). All trials were conducted on a standard badminton court surface with two embedded 0.9 × 0.9 m force platforms (AMTI, Watertown, MA, USA, 1000 Hz). A 12-camera motion analysis system (Vicon, Oxford Metrics, UK) sampling at 200 Hz was used. A shuttlecock was suspended 0.6 m above the ground, at a distance of 0.6 m from the centre of the force platform as specified by Lam, Ding, and Qu (2017). A total of 50 right forward lunge trials (10 trials × 5 insoles) were measured allowing for sufficient rest periods. During rests the insoles of the shoes were changed, and markers were reattached onto the foot. A blinded and counterbalanced protocol across insole conditions was adopted to avoid the fatigue effect of doing multiple lunges.

Kinematic and kinetic data were input to a musculoskeletal model (AnyBody Technology A/S, Aalborg, Denmark), composed of 9 segments and 15 degrees of freedom (Lund, Andersen, de Zee, & Rasmussen 2015). and a 'AnyMuscleModel3E' advanced muscle model employed. Component power was calculated by multiplying the passive element, tendon and muscle forces by their

Table 1. The categorization of the hardness of each of the five tested insoles.

Insole	Rearfoot	Midfoot	Forefoot
1	Hard	Medium	Hard
2	Hard	Medium	Soft
3	Medium	Medium	Medium
4	Soft	Medium	Hard
5	Soft	Medium	Soft

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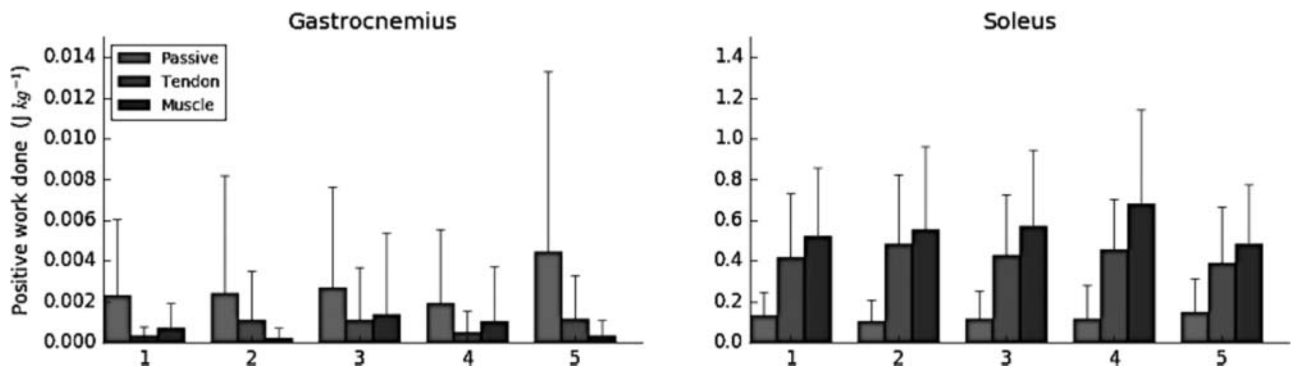

Figure 1. Positive work for the passive, tendon and muscle elements for the five insoles (Table 1).

respective contraction velocities. Hereafter, mechanical work was determined by integrating the corresponding power curves (Lai, Schache, Lin, & Pandy, 2014).

A one-way ANOVA was used to statistically compare the effect of the elastic energy within the five different insole conditions in a right forward badminton lunge. A P-value of <0.05 was considered significant.

Results

No significant differences were found between insoles for the three elements of the gastrocnemius (passive element $P = 0.300$, tendon $P = 0.384$, muscle $P = 0.353$) or the soleus (passive element $P = 0.988$, tendon $P = 0.795$, muscle $P = 0.686$).

Discussion and conclusion

Results showed that different insole hardness distribution did not systematically alter the elastic energy return during a lunge. This might be due to the insoles in the present study being relatively thin compared with midsole of the badminton shoe while inter-individual variability points at subject-specific alterations in movement technique. This variability requires further investigation.

One important finding in this study was that the soleus showed a substantially greater work contribution than the gastrocnemius (Figure 1) suggesting that elastic energy storage mainly occurs in the soleus MTU during a maximal badminton lunge. Future research should include investigations of elastic energy return in different lunge techniques as observed here.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The effect of sex on measures of foot mobility and stiffness in children and adolescents

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Keywords: foot measures; children; foot mobility; foot stiffness; footwear

Introduction

Foot mobility has been suggested to be an important consideration in the selection of footwear (Farndon, Robinson, Nicholls, & Vernon, 2016). In contrast to measures of static measures of foot structure, quasi-static measures of foot mobility and stiffness have also been reported to be useful predictors of injury and dynamic foot posture in adults (McPoil, Vicenzino, Cornwall, Collins, & Warren, 2009; Zifchock, Davis, Hillstrom, & Song, 2006). However, relatively little is known about foot mobility and stiffness in children and whether the moderating effect of sex reported in adults (Zifchock et al., 2006) is also evident in children.

Purpose of the study

This study investigated the effect of sex on measures of foot mobility and foot stiffness (FS) in children and adolescents, aged 8–14 years. Consistent with research in adults (Zifchock et al., 2006), it was hypothesized that females would present with a more compliant (less stiff) and mobile foot than males.

Methods

Clinical measures of foot length (FL), midfoot width (MFW) and dorsal arch height (AH) of the right foot (Williams, & McClay, 2000) were made in 13 male (age 10.5 ± 1.8 years, height 147.9 ± 11.3 cm, mass 43.9 ± 10.2 kg) and 7 female children (age 10.9 ± 1.7 years, height 147.1 ± 16.9 cm, mass 45.4 ± 17.7 kg) using a modified foot assessment platform (McPoil et al., 2009). The platform (Figure 1) incorporates a portable force plate (PASPORT 2-Axis Force Platform PS-2142, PASCO, Roseville, CA), which allowed quantification of vertical force (resolution, 0.1N, full scale of 1100 N) during three quasi-static loading conditions: non-weight bearing (NWB), seated-weight bearing (SWB), double limb stance (DLS).

Foot mobility (FM) was calculated according to the equation outlined by McPoil et al. (2009):

$$FM = \sqrt{dAH^2 + dMFW^2},$$

where *dAH* and *dMFW* reflect the change in arch height and midfoot width between DLS and NWB, respectively.

FS was calculated according to the equation (Zifchock et al., 2006):

$$FS = \frac{dF}{BW \times dAHI},$$

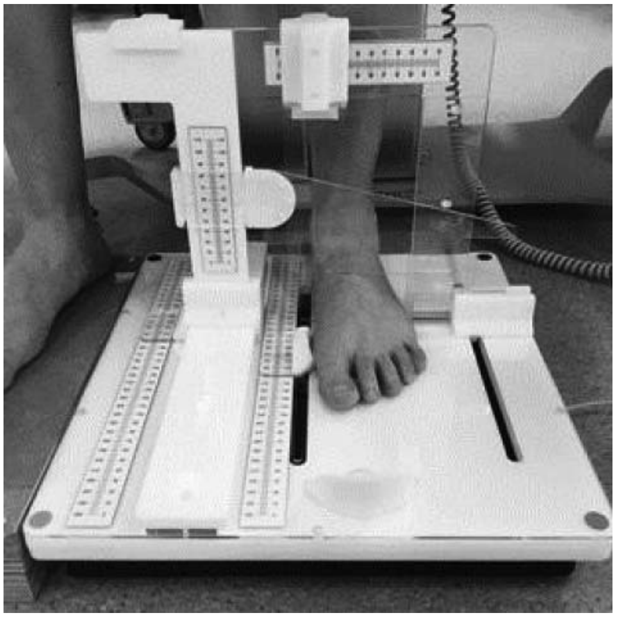

Figure 1. Platform used to measure FM and FS.

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Table 1. Mean (SD) FM and FS measures.

Variables	Boys	Girls
n	13	7
Foot mobility (mm)	7.1 (1.1)	6.3 (1.8)
Foot stiffness (BW/AHI)	2031 (1576)	1502 (900)

where *dF* represents the change in measured force between DLS and SWB, normalized to bodyweight (BW) and *dAHI* represents the corresponding change in arch height index (AHI), the ratio of DH to FL.

Statistical comparisons between males and females were made using student's *t*-tests ($\alpha = 0.05$). Scatter plots and Pearson product–moment correlations were used to investigate potential relationships among FM, FS and measures of body anthropometry, such as height, weight and body mass index.

Results

There was no statistically significant difference in FM or FS between the male and female children (Table 1). FS was moderately correlated with age (r^2, 0.27), height (r^2, 0.31), weight (r^2, 0.30) and BMI (r^2, 0.30). There were no significant correlations among FM, AHI, and measures of body anthropometry (r^2, 0.01–0.04).

Discussion and conclusion

Contrary to our hypothesis, measures of foot mobility and stiffness were not significantly different between male and female children. It should be noted, however, that measures of FM in the current study were approximately three times lower than that reported in healthy adults (McPoil et al., 2009), suggesting that FM may be subject to developmental, scalar or maturational factors. FS, in contrast, was similar to published values in adults (age 18–65 years), and fell between those reported for young adult males (≈2180 ± 660 BW/AHI units) and females (≈1240 ± 300 BW/AHI units). Given the limited sample size of this pilot study, further research on the effect of sex on measures of foot mobility in children seems warranted.

Disclosure statement

No potential conflict of interest was reported by the authors.

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An optimal bending stiffness of running shoes to improve running efficiency

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Keywords: running shoe; efficiency; bending; metatarsals; athletic performance

Introduction

Since a report on improved running economy with running shoes with larger longitudinal bending stiffness was introduced (Roy & Stefanyshyn, 2006), a number of research has been conducted to find out how stiffened shoes help runners reduce metabolic cost. Moreover, possibility that there can be an optimal shoe stiffness was also raised because shoes with high stiffness lowered the running efficiency.

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The mechanism of improved running economy with the optimal shoe stiffness, however, is yet to be discovered. Possible explanations for those are that the stiffened shoes assist propulsion of body in a form of elastic energy stored during shoes flexion (Oleson Adler, & Goldsmith, 2005) and that the optimal stiffness might depend on muscular capacity of each runner.

Purpose of the study

In this study, we studied how stiffened shoes improve the running efficiency. We raised two hypotheses in this research.

The first hypothesis was that the elastic energy stored during the stiffened shoes flexed could help runners propel the body.

The second one was that there would be an optimal stiffness for running economy. A high level of stiffness in shoes could disturb natural joint flexion of the runners and make them feel discomfort or induce overwork at other joints. We tried to find out the optimal shoe bending stiffness satisfying the maximum elastic energy with the lowest discomfort and overwork for runners.

Methods

Twenty-one healthy male subjects (26.3 years old) participated in this research and signed a consent approved by KAIST IRB prior to experiments. Shoe stiffness was controlled by inserting thin carbon plates under insoles. Total five stiffnesses of carbon plate (stiffness of 1.5, 10.0, 24.5, 32.1 and 42.1 Nm/rad with heights of 0 (control), 0.8, 1.2, 1.5 and 1.8 mm, respectively) were used in the experiments.

During five weeks, maximal VO2 test, three weeks of adaptation sessions, biomechanical test and metabolic test were conducted. Subjects visited laboratory two times a week and ran under their submaximal speed during 25 minutes a day. To eliminate error due to experimental order, metabolic tests were repeated two times in a different order of shoe stiffness. Kinematic, kinetic and metabolic data were collected using Hawk cameras (Motion Analysis, US), force treadmill with force plate (Bertec, US) and Quark (Cosmed, Italy), respectively.

Results

As hypothesized, elastic energy was stored during metatarsophalangeal joint (MTPJ) flexion and returned as a form of ground reaction force (GRF) during push-off phase. Therefore, muscular effort to make propulsion force was decreased as the shoes stiffness increased (Figure 1).

The shoes with higher stiffness than the critical stiffness induced uncomfortable MTPJ movements and inefficiency. The higher resisting torque given by stiffer shoes resulted in the reduced maximal MTPJ flexion.

The reduced flexion lengthened moment of arm from the ankle joint to centre of pressure, and resulted in the reduced GRF with the nearly same amount of ankle torque.

To maintain a constant running speed, overwork at other lower limb joints was required. Rather than exerting larger push-off from plantar flexors, subjects showed longer stance and push-off duration to compensate such decreased propulsion force. With the shoes with the higher stiffness than the critical stiffness, there was sharp and significant increased overwork at other joints (Figure 2).

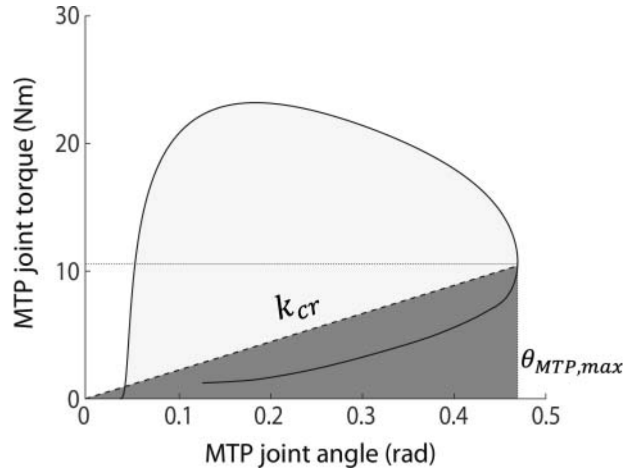

Figure 1. MTPJ flexion angle and torque during running and definition of the critical shoe stiffness (kcr). Joint work done by muscular-tendon system (lightly shaded area) and stiffened shoes using elastic energy (darkly shaded area).

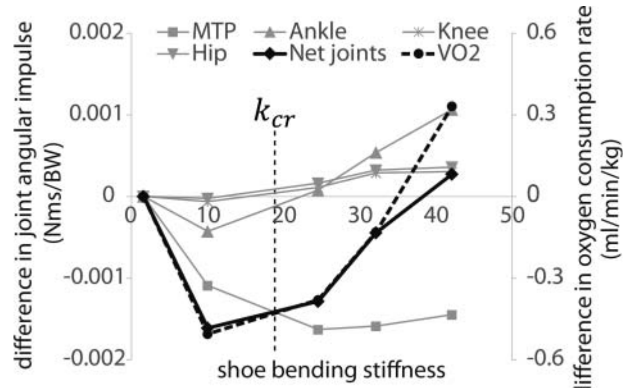

Figure 2. Relative lower limbs joint (gray lines) and net angular impulse (black solid line) in left axis. Difference in oxygen consumption rate (black dashed line) on right axis for various shoe stiffnesses.

Collectively, it seems that the critical shoe stiffness could utilize storage and return of the elastic energy as a propulsion force, and thus, it could save muscular efforts especially in the MTPJ. The critical shoe stiffness could also avoid the disturbed MTPJ flexion and resulted overwork, and therefore, it may affect optimal shoe stiffness. The oxygen consumption rate also showed a minimum value (−1.0 ± 2.1%) near at the critical shoe stiffness (Figure 2).

Discussion and conclusion

As we hypothesized, carbon insoles could store and return the elastic energy as propulsion force. The muscular effort of the subjects decreased as the shoe stiffness increased. The shoe stiffness greater than the critical stiffness, however, disturbed the natural maximal MTPJ flexion and reduced propulsion with lengthened ankle moment arm. Therefore, the optimal shoe stiffness could be obtained by maximizing the elastic energy and minimizing the overwork at other joints. Such criterion could be applied to other assisting devices to improve the locomotion efficiency.

In this study, five weeks of adaptation was conducted but a longer duration of adaptation can be used to study the time-dependent effects of stiffened shoes.

Moreover, the same study on elite athletes, who are possibly with higher muscular capacity and the critical stiffness values than non-athletes, could help study the relationship between the critical stiffness values and optimal stiffness.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

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Comparison of zig-zag run performance, surface hardness and foot motion

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Keywords: natural turf; 3 g artificial turf; shock absorption; cutting; soccer

Introduction

Recently, the use of artificial turf for soccer has become very popular. Although several cohort studies reported no clear differences for injury risk between artificial turfs and natural turf (Fuller, Dick, Corlette, & Schmalz, 2007a; 2007b), soccer players still had an overall negative impression of playing on artificial turfs and felt they required greater physical effort (Andersson, Ekblom, & Krustrup, 2008). There is a possibility that some specific property of artificial turfs might be linked to that impression from players.

Purpose of the study

The present study was designed to examine the effect of surface differences (natural and artificial turfs) on soccer-specific zig-zag run performance and foot inversion/eversion motion during cutting.

Methods

Five experienced university soccer players (height = 171.5 ± 4.5 cm; weight = 65.0 ± 4.0 kg) volunteered to participate in the present study. They were asked to com-

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Figure 1. Soccer boot with a gyro sensor instrumented to measure foot inversion/eversion angular velocity during cutting.

plete a zig-zag run on four different surfaces: natural turf (NT), two types of artificial turf with shock pad (ATP1 and ATP2) and artificial turf without shock pad (AT). All the players used the same type of soccer boot with different sizes with a gyro sensor instrumented to measure ankle inversion/eversion angular velocity (Figure 1). Shock absorbency of each surface was also measured according to the procedure approved by FIFA Quality Programme for Football Turf. For each player, the running time was measured twice for each surface using a pair of photocells.

Results and discussion

The shortest running time was recorded on AT, followed by NT (+0.04 s), ATP1 (+0.26 s) and ATP2 (+0.30 s). This ranking of the time corresponded with that of surface hardness (AT > NT > ATP1 > ATP2), indicating that more shock absorption of the surface tended to restrain player's zig-zag run performance.

This agrees with the findings of the study by Andersson, Ekblom, and Krustrup (2008) who reported that soccer players felt greater physical efforts were required

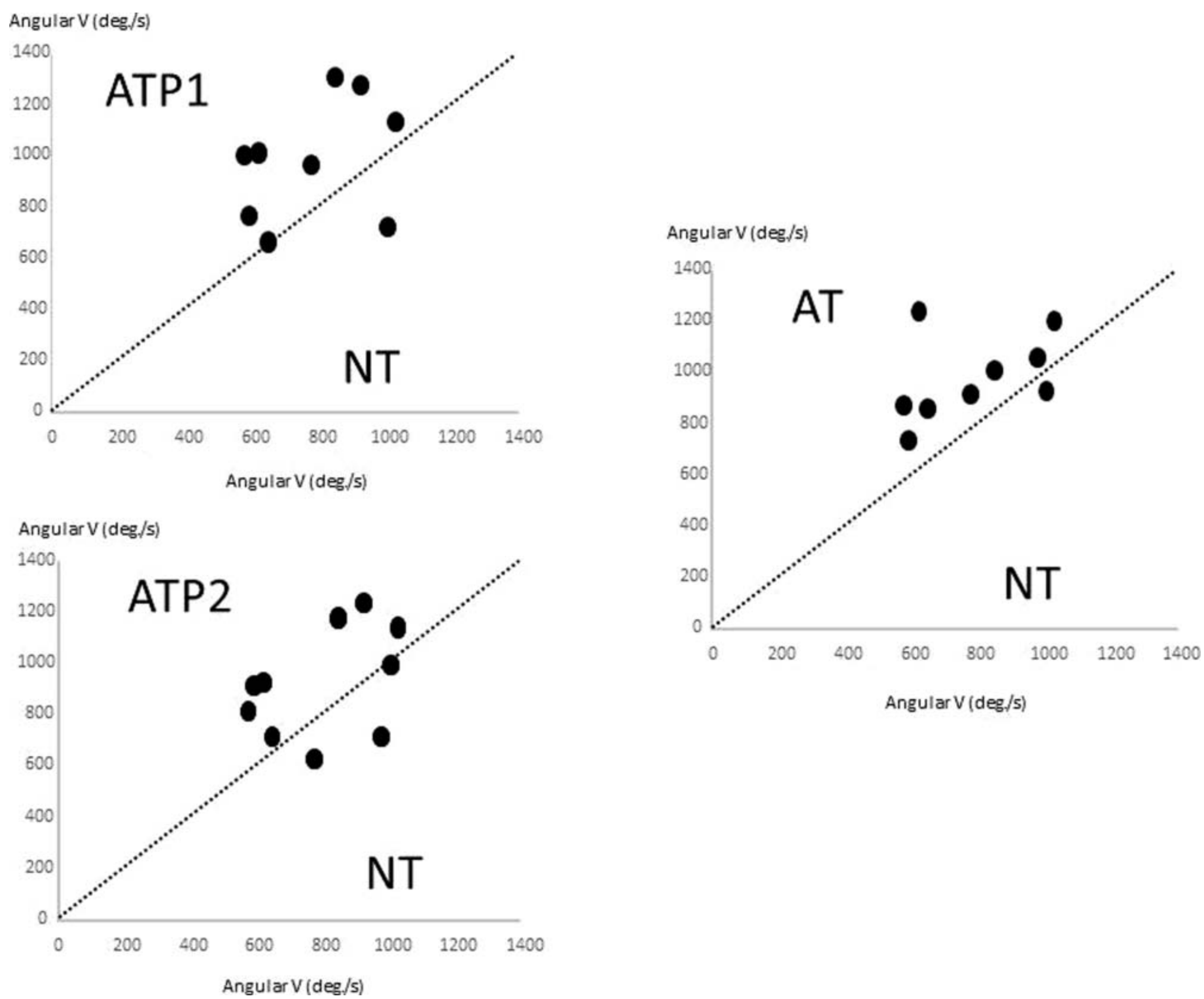

Figure 2. Comparisons of peak foot inversion angular velocities between natural turf and artificial turfs. Top panel (NT vs. ATP1), middle panel (NT vs. AT) and bottom panel (NT vs. ATP2).

when they played on artificial turfs. From the comparison of the foot inversion angular velocities between NT and the three artificial turfs (ATP1, ATP2, AT), a clear trend was observed for the right foot motion during the first left turn. In most trials, players exhibited a higher foot inversion angular velocity when they ran on the artificial turfs (ATP1 = 9/10 cases; ATP2 = 7/10 cases; AT = 9/10 cases) (Figure 2). Different surface friction and deformability might account for it.

Conclusion

Our results highlighted two trends: (1) softer artificial turf (typically with shock pad) surface hardness likely restrain zig zag run performance and (2) foot inversion angular velocity appears to be greater when players cut on artificial turfs.

Disclosure statement

No potential conflict of interest was reported by the authors.

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The biomechanical effects of split-sole shoes on normal walking

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Keywords: functional footwear; foot and ankle; gait analysis; biomechanics; injuries

Introduction

Split-sole shoe consists of shoe outsole that is not connected at the midfoot region, affording higher foot flexibility. This outsole design that permits a larger range of joint motion also allows easy transition of foot during stance. It is hypothesized that the flexibility provided by split-sole shoes may result in several kinematics and kinetics changes in the lower extremity joints.

Methods

Twenty healthy subjects (10 males, age 24.6 ± 1.3 years) were recruited. Subjects were required to perform trials in split-sole shoes and control shoes with single rigid outsole. Force plates (Kistler, Amherst, New York, USA) and motion capture system (Vicon, Oxford Metric, UK) were used in the study (Malerba et al., 2014). In order to better capture the foot motion, Oxford Foot Model (OFM) (Mindler, Kranzl, Lipkowski, Ganger, & Radler, 2014) was employed (Figure 1). The subjects were required to execute at least five successful walking trials. Post-processing of data was performed using Nexus 1.8.3 and Polygon 3.5 (Vicon, Oxford Metric, UK) software.

Walking efficiency was analysed with the use of an oxygen consumption test (VO_2) (METALYZER 3B, Cortex, Germany) while walking on treadmill at 5 km/hr with the treadmill set at 0% grade for a period of 10 min per trial (Figure 2). A total of three trials each for both shoe conditions were performed.

Results

Significant differences were identified particularly in the ankle joint. Ankle angle was found to be significantly higher ($p < 0.05$) in the frontal plane during propulsive stage of stance phase when subjects walked in control shoes. Ankle force and moment in frontal plane increased significantly ($p < 0.05$) in control shoe condition. Ankle power was also reported to be elevated significantly ($p < 0.05$) especially during toe-off. VO_{2max} reading for split-sole shoes was 14.89 ± 0.34 mL/kg/min, which was significantly lower ($p < 0.05$) than the oxygen demand in

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Figure 1. Front, side and rear views of subjects with Oxford Foot Marker (OFM) placement.

control shoes which achieved an average value of 15.09 ± 0.30 mL/kg/min.

Discussion and conclusion

The split-sole shoe is specially fabricated with independent forefoot and rearfoot outsoles. The absence of midfoot outsole imposes less restriction and allows more flexibility to the midfoot joint movement.

In terms of kinematics and kinetics, ankle joint variables in frontal plane show significant differences between control and split-sole shoe conditions. The elevation of ankle angle results in increased ankle force and moment acting on the ankle in control shoes. The finding is in agreement with previous studies that have successfully demonstrated the increased ankle inversion and moment in stiffer footwear during running and cutting movements (Graf & Stefanyshyn, 2013). The increased torsional stiffness of the control shoes is associated with its increased resistance against torsion about the longitudinal axis (Stacoff, Kaelin, Stuessi, & Segesser, 1989). In order to compensate the limited foot flexibility due to the sturdy outsole construction, ankle tends to invert more and this overdone ankle motion in frontal plane raises the external inversion moment adversely (Stacoff et al., 1989). With reduced inversion, split-sole shoe wearers have increased ankle stability and hence lower probability of excessive ankle motion and moment (Snedeker, Wirth, & Espinosa, 2012). This decreases the likelihood of overstretching the ankle ligaments and subsequently protects the ankle from injuries (Gigi et al., 2015).

The ankle power is reduced significantly with the use of split-sole shoes. The positive peak of ankle power, the instance when plantar flexors undergo concentric contraction to propel the body forward, decreases by 14% ($p < 0.05$) in split-sole shoe condition. The decline in ankle power implies less power output by the muscles and decreased the likelihood of fatigue and overuse injuries.

The use of split-sole shoes imposes less restriction on midfoot flexibility and thus helps to reduce the risk of ankle injuries. The walking performance can also be

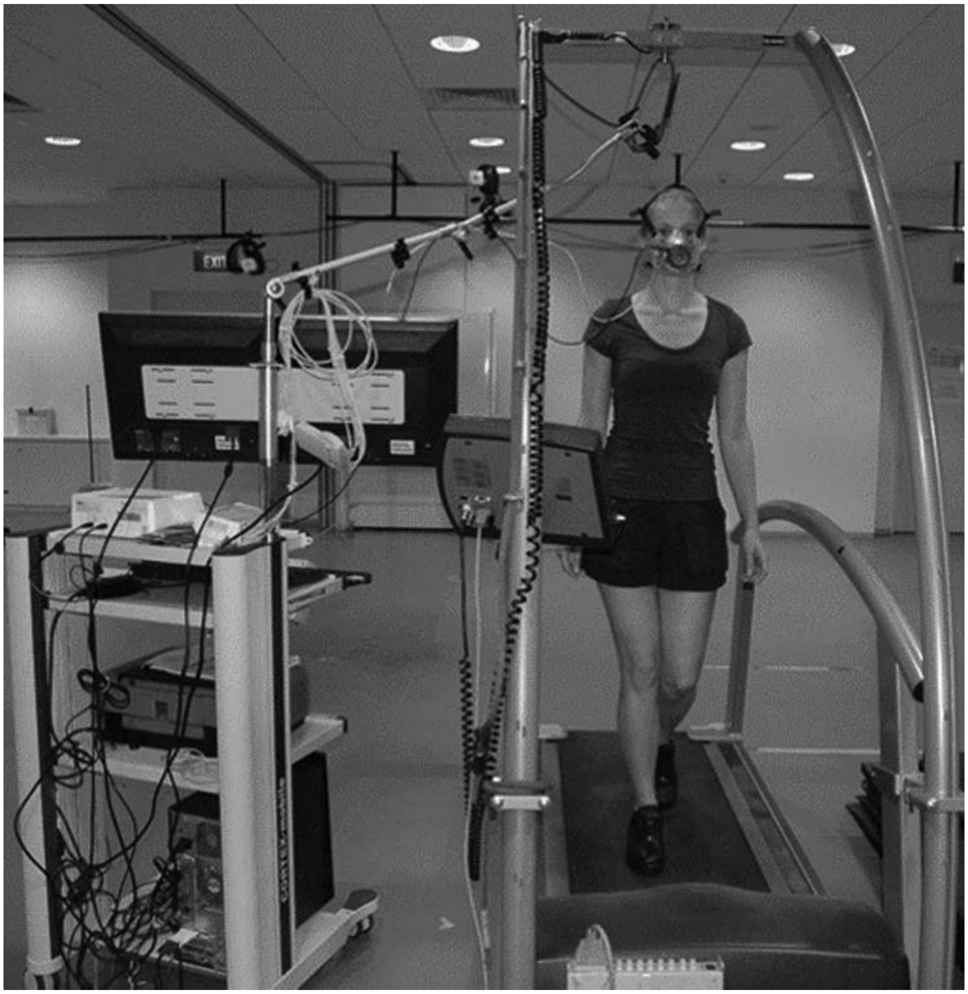

Figure 2. Subject walking on treadmill with the VO_2 max set-up.

enhanced with the use of split-sole shoes by reducing the energy demand.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Biomechanical effects of variable stiffness shoes in normal walking after 60-minute adaptation

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Keywords: knee injury; functional footwear; gait analysis; adaptation; joint moments

Introduction

The use of variable stiffness shoes (VSS) has been found to be conducive to reducing the external knee adduction moment (EKAM) (Teoh et al., 2013). However, it is still unknown whether the claimed biomechanical effects of VSS remain after the subjects have adapted to the footwear. The objective of the study is to examine the effect of VSS immediately upon usage and its effect after 60-minute of adaptation. It is hypothesized that the lowering of EKAM is not just the short-term response due to the immediate use of VSS.

Methods

Twenty healthy subjects (10 males, age 24.6 ± 1.3 years) were recruited. Subjects were instructed to walk in three shod conditions, e.g. control, VSS and VSS after 60-minute of adaptation. Force plates (AMTI, Watertown, MA) and motion capture system (Vicon, Oxford Metric, UK) were used. Plug-in-gait marker set was applied as shown in Figure 1. The subjects were required to execute at least five successful walking trials.

During the 60 minutes of adaptation period, participants were encouraged to stroll or walk briskly around the school compound. Any case of fatigue or discomfort had to be reported and the subjects would be excluded from the study.

Results

The results showed that the EKAM was significantly reduced with the immediate use of VSS (8.43%, $p < 0.05$) as shown in Table 1. The reduction remained significant even after the given 60-minute adaptation time. No significant difference was identified between the two VSS conditions of immediate use and usage after 60 minutes.

Figure 1. Front and side views of subjects with Plug-in-Gait (PIG) placement.

Discussion and conclusion

Assumption is made that participants will learn and adapt to the stiffness changes reflexively within the adaptation time given. Sixty minutes is believed to be sufficient as studies have shown that adaptation time needed for the

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Table 1. EKAM profiles of the three shod conditions examined.

EKAM (N·mm/kg)	Control	VSS Immediate	VSS 60 minutes
1st peak	712.93 (145.22)	652.82 (111.21) *	658.24 (96.26) *
2nd peak	438.52 (130.47)	431.82 (145.81)	418.63 (138.64)

*p-value < 0.05 as compared to the control shoe.

subjects to familiarize with the new minimalist running shoe is about 5–8 minutes (Hardin, van den Bogert, & Hamill, 2004). The experimental condition of VSS after 60-minute adaptation has shown significantly lowered EKAM as compared to control ($p < 0.05$). The lowering of EKAM is observed to persist after the 60-minute usage although the % reduction has gone down slightly from 8.4% to 7.7%. The results support the hypothesis that VSS can effectively reduce the EKAM even subjects have familiarized and adapted to the footwear. The finding also clears the doubt on the reduction of EKAM upon immediate usage that might be just a temporary compensation strategy in response to the variable stiffness design. In fact, based on the experimental outcome, the decreasing effect on EKAM is found to be sustainable.

No significant change is identified between the two cases of VSS indicating the overtime gait modifications by the participant do not reduce the effect of VSS in lowering the medial loading on knee joint. The results coincide with several studies that focus on the long-term effect of VSS that have shown sustainable biomechanical changes identified from the immediate use (Erhart, Dyrby, D'Lima, Colwell, & Andriarcchi, 2010; Erhart-Hledik, Elspas, Giori, & Andriarcchi, 2012). However, these research works only focus on the patients with knee OA. Till date, there is no study conducted to investigate the time effect of VSS on individuals with no knee OA. The relevant studies conducted have only examined the immediate effect of VSS (Fisher, Dyrby, Mündermann, Morag, & Andriacchi, 2007; Teoh et al., 2013). Previous study by Teoh et al. (2013) commented that the lack of adaptation period for the subjects to familiarize themselves with the usage of VSS before test may compromise the accuracy of findings. The biomechanical changes observed may be just a temporary gait response to the unfamiliar footwear.

Based on the findings, the effect of VSS on the reduction of EKAM is found to be consistent even after familiarization. The influence of the variable stiffness sole design is inferred to be helpful in alleviating the loading across the medial compartment of knee joint and hence prevent the onset of medial knee OA.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Check for updates

Three-dimensional measurement of the human cadaver foot bone kinematics under axial loading condition using biplane X-ray fluoroscopy

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Keywords: bone; 3D deformation; kinematics; arch (foot); biomechanics

Introduction

The anatomical design of the human foot is considered to facilitate generation of bipedal walking. To clarify the morphofunctional mechanisms of the human foot, efforts have been made to directly capture the three-dimensional (3D) movements of the foot bones during walking using bone pins (Lundgren et al., 2008) and X-ray fluoroscopy (Wang et al., 2016). However, such observed foot bone movements during walking were generated as a consequence of the neural control of forces generated by the extrinsic and intrinsic muscles of the foot. Therefore, it is difficult to infer the innate mobility of the foot from the *in vivo* kinematics of the foot bone during walking. On the other hand, if we use cadaver feet, such factors can be better controlled, possibly leading to a deeper understanding of the innate patterns of the foot bone and hence the causal relationship between foot bone kinematics and the functional adaptations of the foot bone morphology and structure for generation of bipedal locomotion.

Purpose of the study

The purpose of this study was to investigate the 3D kinematics of the foot bones under a weight-bearing condition using cadaver specimens, to characterize the innate mobility of the human foot inherently prescribed in its morphology and structures.

Methods

Figure 1 shows our experimental set-up. In the present study, five fresh frozen cadaver feet (average age at death, 80 years; range, 68–90 years; two females and three males; one right and four left feet) were loaded vertically by putting weight and the 3D movements of the foot bones due to the vertical compression were quantified using biplane fluoroscopy. The specimens were axially loaded up to 600 N. We did not apply forces to the muscle tendons to observe innate pure bony movements just produced by the morphology and structure of the human foot. Paired radiographic images of the foot under axial loading conditions were obtained using the biplanar fluoroscopic system (Shimadzu, Kyoto, Japan) and we determined the motion of the tarsal and metatarsal bones by a model-based registration technique (Ito et al., 2015). Briefly, 3D surface models of foot bones were generated prior to motion measurement based on computed tomography, and the bone models were then registered to biplane fluoroscopic images such as to maximize similarity measures between occluding contours of the bone surface models with edge-enhanced fluoroscopic images. To describe the change in the relative positions and orientations of each of the bones due to axial loading, a local coordinate system was defined.

Results

Figure 2 shows the 3D skeletal movements of the foot bones under static axial loading. All bones were observed to move inferiorly, medially and anteriorly. We also observed that the medial translation of the talus and navicular was larger than the calcaneus and cuboid, indicating that the talus slid medially with respect to the calcaneus. The talus was also adducted due to internal rotation of the tibia and metatarsal bones were abducted. The subtalar, talonavicular and calcaneocuboid joints were all everted, dorsiflexed and abducted, respectively, with increasing load, but inversion/eversion angle of the calcaneocuboid joint was very small.

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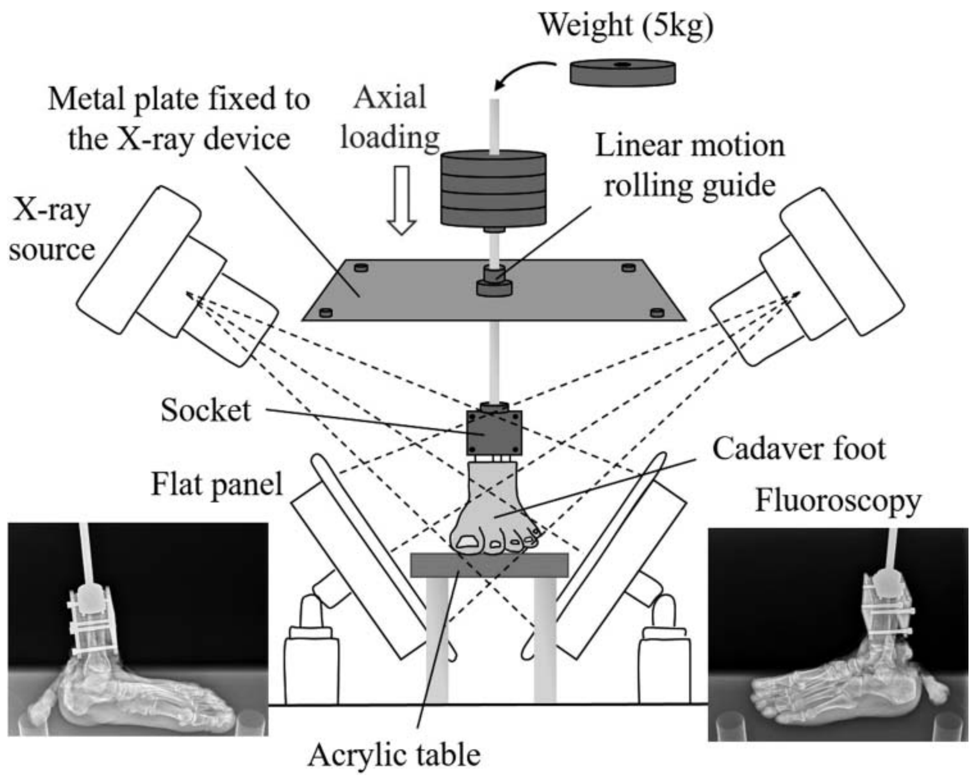

Figure 1. Experimental set-up.

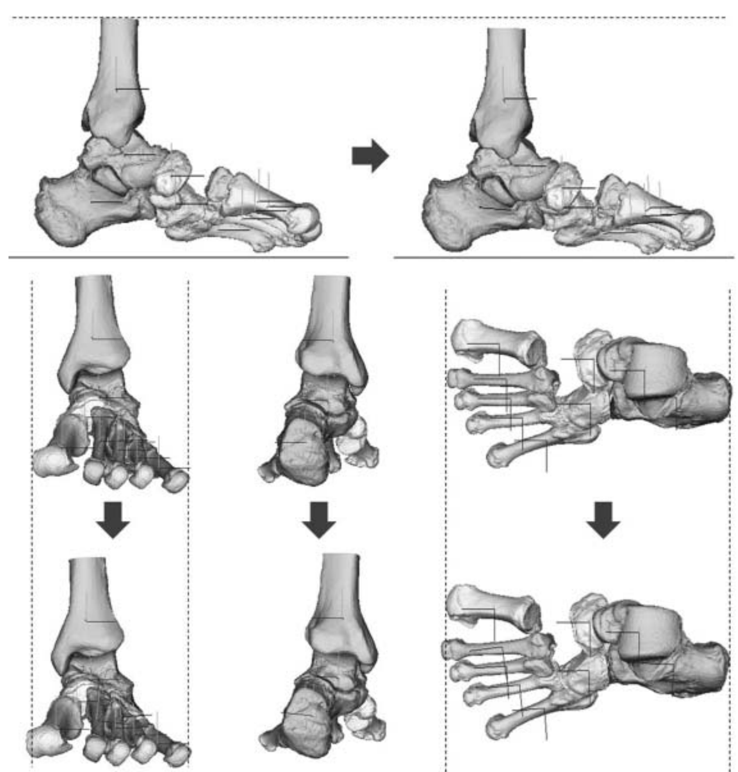

Figure 2. Reconstructed three-dimensional foot bone movement.

Discussion and conclusion

The present study demonstrated that the talus is medioinferiorly translated and adducted as the calcaneus is everted due to axial loading, causing internal rotation of the tibia and flattening of the medial longitudinal arch in the foot. Furthermore, as the talus is adducted, the talar head moves medially with respect to the navicular, inducing abduction of the metatarsals. This mechanism possibly contributes to balance the moment actions to the body around the vertical axis during walking.

The present study also found that the cuboid everted concomitantly with the eversion of the calcaneus. This is possibly due to a locking mechanism of the calcaneocuboid joint (Bojsen-Møller, 1979). The human cuboid possesses a prominent medial calcanean process; therefore, the human calcaneocuboid joint is structurally more constrained than those of the non-human apes.

Such detailed understanding about the innate mobility of the human foot will contribute to clarifying functional adaptations as well as pathogenic mechanism of the human foot.

Disclosure statement

No potential conflict of interest was reported by the authors.

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Science, pseudoscience and footwear science

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Keywords: footwear; athletic footwear; functional footwear; review; zippers

Introduction

In 2011, the ISB Footwear Symposium in Tübingen (Grau, 2011) featured special interest sessions on ‘barefoot’ and ‘unstable’ footwear. Ultimately, important questions about these classes of products were resolved by lawyers, not scientists, and at enormous cost. Was the footwear science community relevant during these product cycles, or merely following (at some distance) consumer marketing trends?

More generally, when popular media are bloated by ‘science denial’, ‘alternative facts’ and conspiracy theories, bloat popular media, I believe it is time to examine the role of footwear scientists and their responsibilities to both manufacturers and consumers.

The context

Footwear biomechanists, even specialists, must communicate across many domains. Biomechanics itself is not the domain of scientists alone, but encompasses the work of engineers, material scientists, medical practitioners, mathematicians and others who are less constrained by the scientific method. Footwear science, too, is multidisciplinary. An inclusive definition would embrace psychology, social and behavioural sciences, anthropology, in addition to the biological and physical sciences.

The footwear industry provides much of the context, both the context and the funding for most of our work, bringing designers, developers, material scientists, marketers and other professionals into the community.

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Potential *c*onflicts

The major footwear brands value science as an important product development tool and rely on the integrity of research outcomes. They are, nonetheless, sellers of consumer products. They are legally constrained by the financial interests of their shareholders and must respond quickly to competitive trends. Engineers and other 'problem solvers' often fit more easily into this world, while the scientist must resolve hypothetical conflicts among academic freedom, impartiality and financial dependence.

Real conflicts

In my experience, the most common and very real conflicts we face are not related to integrity but to process. The industry's product cycle is driven by short timelines, often too short for meaningful research into new concepts. Marketers need short and definitive sound bites, not nuanced opinions. These snippets of information are distributed via press releases, sales meetings and social media rather than peer-reviewed journal – outlets in which plausibility often suffices as truth. What we see as best practices, others working on the same project may as view as restrictions on creativity – often justifiably.

Of all the potential risks, perhaps the most challenging is the attention drawn by sensational and novel findings. The consumer product industry thrives on these headline events – and scientists do too, it appears. Higginson and Munafò (2016) modelled the ways in which scientists can maximize their career success and found that in an environment that rewards fast turnover and high-profile claims, it would be rationale to cut corners. A related study (Smaldino & McElreath, 2016) also suggests a trade-off between success and scientific rigor. Under such conditions, methodology may be compromised and findings less likely to be reproducible.

Pseudoscience and footwear science

'Pseudoscience' (see Shermer, 2002) can be described a collection of beliefs that are presented as science-based knowledge but are actually hypotheses that have not been tested by the scientific method or have failed objective testing. Full-blown cases of pseudoscience are fortunately rare in the established footwear biomechanics community, but footwear-related examples from other sources are easily found.

A claim of 'scientifically proven' is a clear sign of pseudoscience. Mathematicians can prove things but scientists cannot. Scientists create knowledge not by proving things but by invalidating alternative hypotheses and by gradually shrinking the extent of our ignorance.

More troubling in my view are the many examples of logical errors that support overstated conclusions and somehow escape peer review. 'No significant differences' does not imply similarity. 'Significant trends' are not significant. A comparison of production shoe models is not a controlled experiment. Observation and *Everriculum Trinus* can yield testable hypotheses but are not experimental designs. 'May', 'might' and 'could' are not conclusions. The most important attributes of an experiment are its repeatability and reproducibility, not its results.

Discussion

The broad community in which footwear scientists work and especially the commercial goals of our sponsors can present challenges. Dealing with them is our responsibility and ours alone. My suggestions would include the following:

- Acknowledge that commercial pressures exist and develop plans for managing them.
- Recognize that both we and our industry partners benefit from good science and when the needs of consumers are prioritized.
- Make reproducibility the cornerstone of research programs.
- Encourage students to repeat another's experiment before designing their own. Review 'cognitive bias' and other pitfalls as part of their training.
- Provide and expect more rigorous peer review; giving weight to negative findings repeated experiments.
- Be prepared to identify and challenge pseudoscience, wherever it emerges.
- Be reminded that the 'scientific method' remains, both practically and philosophically, the best available tool for answering real-world questions and distinguishing fact from fiction.

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Zeitgeist: the good, the bad, the ugly???

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Anthroposociology

No one can be 100% sure when our ancestors started wearing shoes. We are the only species to walk in a bipedal manner, and to date, no other mammal has evolved, to make shoes. Anthropologists inform us the development of the human foot ran in tandem with the evolution of the human brain. Psychologists confirm shoes are more than simple foot attire: they serve a social significance which indicates the personality of the wearer.

Anthroposociology affirms the three important functions of shoes are decoration, modesty, and protection.

Long been debated, as to what came first, with the common consensus it was protection, yet the evidence to support this hypothesis is scant. Modesty as a concept, and comparatively new, leaves the primary function of footwear as decoration. Decoration as we know is to beautify bodily appearance in order to attract admiring glances which in turn fortifies our self-esteem. Simply put shoes, outwardly represent a very important non-verbal sign of gender, presence, and personality. This truly makes the shoemaker's role very important.

Throughout history, many factors have influenced the development of shoes and their design including raw materials, status, disease, wars, bitter rivalry, and the Space Age. In summary, what has driven humans and their preoccupation with their shoes and what they look like, throughout history, are down to two factors: craftsmanship and available technologies. I would argue that being able to deport from A to B in a manner and style appropriate to civilized human beings is a basic human right.

The focus of this conference is to examine how best to continue this evolution and pursue how scientists can help. To facilitate this, I hope a brief, if somewhat irreverent, review of the history of footwear may help in this worthwhile quest.

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