Full length article

Motion-control shoes help maintaining low loading rate levels during fatiguing running in pronated female runners

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ABSTRACT

Background: The use of motion-control shoes may assist pronated runners to maintain their stability throughout a fatiguing running. However, there are no studies describing the effects of fatigue on running biomechanics of runners with pronated feet.
Research question: Whether motion-control shoes can assist pronated recreational female runners to maintain impact loading patterns following a fatiguing protocol?
Methods: Twenty-two female rearfoot runners with foot pronation were asked to perform a fatiguing treadmill running protocol using a neutral shoe or a motion-control shoe in two separate occasions. Before (Pre-fatigue) and after the fatiguing protocol (Post-fatigue), participants were asked to run overground on a track that contained two force platforms to record ground reaction forces and moments. Running speed were $3.3\,\mathrm{m\,s^{-1}}$ ($\pm$ 2.5% variability). The effects of shoe type and fatigue were investigated on the peak vertical impact ground reaction force (pvIGRF), time to reach pvIGRF, vertical loading rate (LR) and peak negative foot free moments (FM).
Results: Pronated runners presented lower LR with motion-control shoes compared to neutral shoes Pre- ($p < 0.005$; $-18 \pm 25\%$) and Post-fatigue ($p < 0.001$; $-27 \pm 15\%$). This change in LR was predominantly driven by a longer time to reach pvIGRF with motion-control shoes ($p < 0.001$, 39%). The pvIGRF and LR increased after fatiguing running with neutral shoes (pvIGRF: $p < 0.05$; $18 \pm 28\%$; LR: $p < 0.05$; $15 \pm 22\%$), but not with motion-control shoes. Furthermore, there were strong correlations between FM and LR for both Pre-fatigue ($r = -0.61$, $p < 0.005$) and Post-fatigue measurements ($r = -0.66$, $p < 0.01$), but only for the motion-control shoes.
Significance: These results suggest that motion-control shoes prevent exacerbated fatigue-related increases in mechanical loading following initial contact in pronated female runners.

1. Introduction

The incidence of lower extremity running injuries ranged from 19.4% to 79.3% [1], and one injury occurs every 100 training hours [2]. Several factors may cause running-related injuries, such as (among others) training routines, running technique and overweight [2,3]. Recently, foot alignment has been associated to running-related injuries, especially for people with pronated feet running wearing standard (neutral) shoes [4]. Foot pronation contributes to lower extremity misalignment, being related to tibial and hip internal rotation during walking and running [5]. Consequently, foot pronation may lead to medial tibial stress syndrome, patellofemoral pain, and low back pain during running [6,7].

Also, increased foot pronation causes increased external knee adduction moment (KEAM) [8], which has been strongly associated with musculoskeletal disorders such as the development of knee osteoarthritis [9–11]. Therefore, foot-pronated runners must take precautions to reduce undesired loading during running and attenuate injury risks. Female runners in special present specific lower limb kinetic profile, which naturally generates greater hip and knee moments in the frontal plane during walking and running [12]. Distinct lower limb muscle recruitment, passive and dynamic knee stiffness, neuromuscular control, femoral notch width and wider pelvis (i.e, greater Q angle) are commonly associated to the greater moments generated by females during locomotion [13–16]. Therefore, there are several factors associated uniquely to females – especially those presenting pronated feet – that require further research towards minimizing their risk of injuries.

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The peak vertical impact ground reaction force (pvIGRF) and vertical loading rates (LR) have been defined as running-related injuries predictors [17]. A greater LR may be related to a shorter time to reach the peak force, which has been also related to running injuries [18]. Recently, the foot free moments were considered an index of torsional stress on the lower limbs [18,19]. Free moments describe the vertical moment applied in the center of pressure, and have been associated with tibial stress fracture in distance runners, becoming an alternative predictor of running injuries [18]. These biomechanical variables have been important to define the etiology of running-related injuries etiology and should be explored to describe potential instruments and/or devices to reduce running-related injuries.

Motion-control shoes can counteract the mechanical misalignment caused by foot pronation during running [20]. This type of shoe present unique deformation rates and material geometry between medial and lateral sides, reducing excessive rearfoot pronation [20]. However, the effectiveness of motion-control shoes in assisting pronated runners to maintain impact loading when fatigued has not been established to date. Fatigue can drastically influence lower limb muscle activation during running, potentially reducing the effectiveness of stabilization mechanisms. Motion-control shoes were shown to limit knee internal rotation in runners with over-pronation, regardless whether fresh or fatigued [20,21]. In addition, some studies describing peak ground reaction forces following fatiguing running and landing have shown reductions in pvIGRF [22] during running [23]. Conversely, other studies found increases in pvIGRF and LR [22,24] following fatigue. Such discrepancies may be related to different strategies runners may use to cope with fatigue. Moreover, foot pronation may require larger activation of knee and hip stabilizers during locomotion [6], increasing the complexity of the neuromuscular control during running. The use of motion-control shoes may assist pronated runners to maintain their stability throughout a fatiguing running [20]. However, there are no studies describing the effects of fatigue on running biomechanics of runners with pronated feet.

The aim of the present study was to determine whether motion-control shoes could be more effective in assisting pronated recreational female runners to maintain their running kinetic patterns following a fatiguing protocol when compared to neutral shoes. It was hypothesized that motion-control shoes would induce no changes in pvIGRF, LR and free moments following fatigue, whereas these variables would be increased following fatigue when pronated runners used neutral shoes.

2. Methods

2.1. Participants

Twenty-two female recreational runners (age: 24.1 ± 5.6 years; height: 165.6 ± 10.4 cm; mass: 63.21 ± 12.05 kg) volunteered to participate in this study. Inclusion criteria were the presence of excessive foot pronation, running training volume of 2–3 times/week for 45 min or 10 km each workout, diagnosed foot pronation and consistent heel strike running pattern [25]. Foot pronation was defined as a navicular drop > 10 mm [6] (14.2 ± 2.5 mm) and a foot posture index above 10 (11.2 ± 0.5). Exclusion criteria were the occurrence of any musculoskeletal lower limb injury in the previous six months from the experiment, any type cardiopulmonary disorders and performance of systematic running training program for competitive performance. Prior to participation, written informed consent was obtained from all participants. Ethics approval was obtained from the Research Ethics Board of the Medical Sciences, University of Mohaghegh Ardabili (IR-ARUMS-REC-1396-135), and registered with the Iranian Registry of Clinical Trials (IRCT20170806035517N2).

2.2. Experimental design

The experiment was conducted throughout two sessions occurring at least seven days apart. In each day, participants were asked to run while wearing either motion-control shoes designed for over-pronators (ASICS GEL-Kayano 24; Fig. 1) or neutral running shoes (ASICS GEL-Nimbus 19; Fig. 1). The order of the shoes across days was randomized. A short familiarization consisting of 10 min treadmill running with the selected shoe was provided prior to data recordings in each day. This protocol has been proven to induce short-term adaptation to shoe type [26]. The running speed during familiarization was 3.3 m s^{-1} (± 2.5% variability).

Retro-reflexive markers were placed on specific participant's landmarks and segments for motion capture data collection. In each day, participants performed an overground running (Pre-fatigue measurement) followed by a fatiguing treadmill running protocol. A second overground running protocol was conducted immediately after the fatiguing running task (Post-fatigue measurement, Fig. 1).
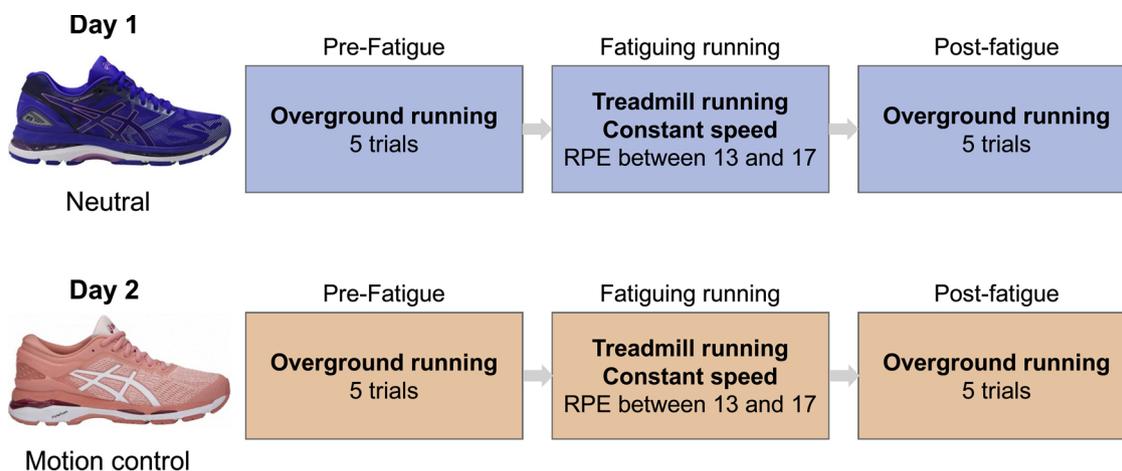

Fig. 1. Neutral shoe (*top row*) and motion-control shoe (*bottom row*) experimental design. The two sessions were separated by at least seven days. For the fatiguing running protocol, a constant speed was set when the rate of perceived exertion (RPE) was between 13 and 17. The order of the testing (neutral x motion-control) was randomized across days.

2.3. Overground and treadmill fatiguing running protocols

The overground running protocols were conducted using an 18-m walkway containing two ground reaction force platforms embedded in the middle of the walkway. Participants were instructed to reach the platforms with each foot. Objective criteria to discard a trial were: 1) the dominant foot did not land on the force platforms; 2) the participant lost balance during the trial; 3) participant ran with midfoot or forefoot strike pattern. From each participant, at least five trials meeting the criteria above and in which the running speed was $3.3 \, \mathrm{m \, s^{-1}}$ ($\pm 2.5\%$ variability) were recorded.

The fatiguing protocol consisted of running on a motorized treadmill (Horizon Fitness, Omega GT, USA), set with no inclination. Participants started the protocol at 6 km/h, and the treadmill speed was increased 1 km/h every 2 min [27]. Rate of perceived exertion was collected at the end of each stage through a 15-point Borg scale [28]. A steady-state running period was initiated when the participant reported a perceived exertion of 13 or higher. Throughout this steady-state period, perceived exertion and heart rate (Polar RS100, Polar Electro Oy, Woodbury, NY) were collected at every 30 s. The fatiguing protocol was terminated after two minutes from the instant the participant reported perceived exertion equal or above 17, or if the heart rate was above 80% maximum heart rate (defined by the formula maximum heart rate = 220 minus participant's age). The average heart rate within the last two minutes of the fatiguing protocol was 177 ± 7 (beats per minute).

2.4. Running shoes

These shoes were selected based on their availability in the local market and comparable design. According to the manufacturer website (https://www.asics.com/us/en-us/gel-kayano-24/p/0010298530.9016; https://www.asics.com/us/en-us/gel-nimbus-19/p/0010291326.9701), main characteristics of both running shoes were: 25 mm heel height, 12 mm forefoot height, 13 mm heel to toe drop, and the use of FlyteFoam material in the midsole. The FlyteFoam midsole is a light and durable compound made from Kevlar and polyester PET (organic) fibres (https://intraining.com.au/new-asics-flytefoam/). In addition, the motion-control shoe includes rearfoot and forefoot GEL Technology, which may reduce impact forces at contact. The major structural difference between the two selected shoes was the composite materials in the midsole.

2.5. Ground reaction forces

Ground reaction forces and moments were recorded using two force platforms (Kistler AG, Winterthur, Switzerland). The force platforms were connected to a Vicon MX system (Vicon Motion Systems, Oxford, UK) that recorded data at a 1000 Hz sampling frequency. Kinetic data were analyzed during the stance phase of running, defined as the interval from ground contact (vertical GRF > 10 N) to toe off (vertical GRF < 10 N)). Kinetic data were filtered using a fourth-order low-pass Butterworth filter with a cutoff frequency of 20 Hz (determined by residual analysis [29]). The first (impact) peak on the vertical force, related to heel contact, was defined as the pvIGRF. LR was computed as the average slope from 20% to 80% of the pvIGRF. Also, free moments of the foot were computed [30]. The peak negative and peak positive of free moments were recorded for the statistical analysis. All ground reaction forces and free moments were normalized with respect to the body weight (BW) and BW × height, respectively.

2.6. Optical motion capture

An optical motion capture system (Vicon Nexus, Oxford Metrics, UK) was used to record the tri-dimensional position of retro-reflexive markers on relevant body segments at a 200 Hz sampling frequency. A set of 18 markers (8 mm diameter) positioned following the CAST

marker set technique [31] was used to extract segmental kinematics in six degrees of freedom. Two rigid clusters of four non-orthogonal markers (*e.g.*, tracking markers) were placed on the lateral shank and thigh to define CAST marker clusters [31]. In addition, tracking markers were placed on the calcaneus, first and fifth metatarsophalangeal joints. The running shoes were modified by opening holes on the material on top of the bone landmarks of interest with holes smaller than $1.6 \, \mathrm{cm} \times 2.0 \, \mathrm{cm}$. Previous studies have been demonstrated that holes smaller than $1.7 \, \mathrm{cm} \times 2.1 \, \mathrm{cm}$ would have no significant influence on the deformation of the shoe or the kinematics of the foot in the shoe during level walking [32,33]. The calcaneus bone was tracked using a wand marker (10 mm wand length), whereas the markers on the first and fifth metatarsophalangeal joints were tracked using markers placed directly on the skin. Anatomical markers were placed over the iliac crests, L5-S1 joint, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, calcaneus, first and fifth metatarsophalangeal joints.

A static calibration trial was recorded with the participant in anatomical position with all retro-reflective markers (anatomical and tracking) in place. This calibration trial was used to determine joint centers and segment coordinate systems. Subsequently, anatomical markers were removed for motion capture recordings during running. The centers of rotation for the knee and ankle joints were defined as the midpoint between the medial and lateral femoral condyle and malleolus markers [34]. The center of the hip joint was calculated using a geometrical prediction method (Davis method) that optimized the joint center by using the distance markers placed on the anterior superior iliac spine and the major trochanter [34]. Kinematic data was filtered using a fourth-order low-pass Butterworth filter with a cutoff frequency of 10 Hz. Lower limb joint kinematics were extracted using tri-dimensional Euler rotation sequences, while lower limb joint moments were calculated using three-dimensional inverse dynamics. The software visual 3D (C-Motion, Rockville, Maryland) was used for all data analysis following recommendations provided by the International Society of Biomechanics, especially to report ankle inversion/eversion angles [35]. Inertial parameters were estimated from established anthropometric data [36]. The peak ankle eversion angle and the first peak KEAM were used for statistical analysis, respectively.

2.7. Statistical analysis

Data normality was examined and confirmed using Shapiro-Wilk-Tests for all investigated variables. The effects of fatigue (pre-fatigue *vs* post-fatigue) and shoe type (neutral *vs* motion-control) were assessed by a 2-way ANOVA with repeated measures for the running speed, pvIGRF, LR, time to reach the pvIGRF, peak KEAM, peak negative free moments and peak ankle eversion angle during running. Group-specific and Bonferroni corrected pre-post changes were calculated if necessary. Effect sizes (partial ETA-square, ηp^2) were displayed for all statistical tests ($0.2 < \eta p^2 < 0.5$ = small effect, $0.5 < \eta p^2 < 0.8$ = medium effect, $\eta p^2 > 0.8$ = large effect). Pearson correlation coefficients were used to assess the relationship between peak negative free moments, KEAM, and LR variables. Correlation coefficients were defined as very weak (0 to 0.19), weak (0.2 to 0.39), moderate (0.4 to 0.59), strong (0.6 to 0.79) and very strong (0.8–1). The significance level was set at $p < 0.05$. All analyses were performed using SPSS (version 22.0).

3. Results

No significant differences were found between running speed across conditions (neutral shoe pre-fatigue: 3.28 ± 0.04 m/s; neutral shoe post-fatigue: 3.27 ± 0.04 m/s; motion-control shoe pre-fatigue: 3.28 ± 0.03 m/s; motion-control shoe post-fatigue: 3.28 ± 0.04 m/s) (p > 0.05). The average time in steady state running was 11 ± 3 min, and the average running speed was 3.8 ± 0.33 m/s.
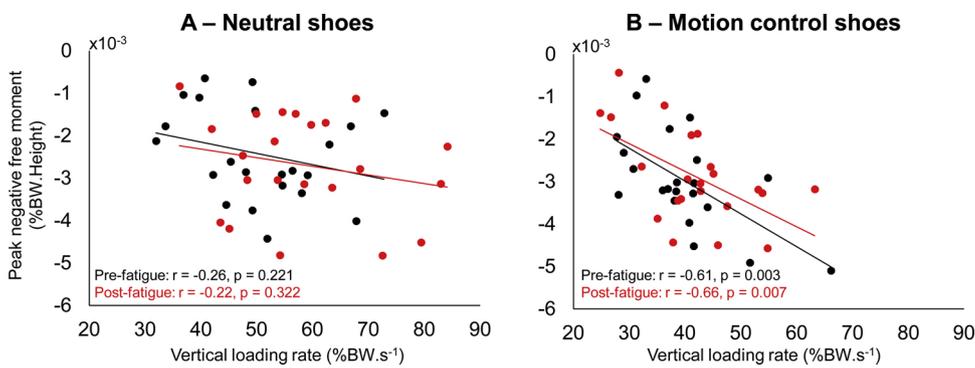
Fig. 2. Vertical loading rate (*panel A*), Peak vertical ground reaction force (GRF, *panel B*), Time to peak vertical GRF (*panel C*), Peak negative foot free moment (*panel D*), Peak knee adduction moment (*panel E*), and peak ankle adduction angle (*panel F*) when pronated runners used the neutral shoes (N) or the motion-control shoes (M) before (PRE) and after a fatiguing running exercise (POST). Small gray circles indicate individual data, which are connected from pre- to post-fatigue for all participants. The large circles indicate average values, whereas the thick vertical lines indicate ± 1 standard deviation for the control shoes (*in black*) and the motion-control shoe (*in blue*). * denotes significant difference between the pre and post-fatigue for the neutral shoe (p < 0.005). # denotes significant difference between the neutral and motion-control shoes (p < 0.05).

3.1. Peak forces and LR

There was a significant effect of shoe type for the LR (p < 0.0001; F = 36.323; $\eta p^2 = 0.634$). Post-hoc analysis did not reveal any pairwise statistical difference between shoes in both pre- and post-fatigue measurements for pvIGRF (p > 0.05, Fig. 2A). However, pronated runners presented lower LR with the motion-control shoes when compared to neutral shoes during both pre- (p = 0.002; -18 ± 25%) and post-fatigue measurements (p < 0.001; -27 ± 15%).

In addition, there was a significant main effect of fatigue for both pvIGRF (p = 0.005; F = 10.911; $\eta p^2 = 0.342$, Fig. 2A) and LR (p < 0.01; F = 7.708; $\eta p^2 = 0.268$, Fig. 2B). Post-hoc analyses revealed that the pvIGRF and LR increased from pre- to post-fatigue when pronated runners used the neutral shoe (pvIGRF: p < 0.05; 18 ± 28%; LR: p < 0.05; 15 ± 22%). However, no significant fatigue-related changes were found when pronated runners used the motion-control shoes. The pvIGRF increased 7.6 ± 19% and the LR increased 7.5 ± 23% from Pre- to Post-fatigue (p > 0.05).

3.2. Time to reach peak force and free moments

There was a significant main effect of shoes on the time to reach the pvIGRF (p < 0.0001, F = 62.500, $\eta p^2 = 0.749$, Fig. 2C). It was found that runners using motion-control shoes presented a longer time to reach the pvIGRF in both Pre- and Post-fatigue measurements (p = 0.005). There was no effect of fatigue on the time to reach the pvIGRF (p > 0.05). There was no main effect of either shoes or fatigue on foot peak negative free moments (Fig. 2D), as well as on peak KEAM from this sample of pronated runners (p > 0.05, Fig. 2E).

3.3. Ankle eversion

The statistical analyses indicated significant main effects of shoes for peak ankle eversion angle (p = 0.008; F = 8.500; $\eta p^2 = 0.288$). Pronated runners presented lower peak ankle eversion angle with the motion-control shoes (pre: 3.42 ± 5.85°; post: 6.44 ± 5.77°, Fig. 2F) when compared to neutral shoes (pre: 6.46 ± 7.40°; post: 9.39 ± 6.13°) during both pre- (p = 0.044) and post-fatigue measurements (p = 0.012). In addition, there was a significant main effect of fatigue for peak ankle eversion angle (p = 0.016; F = 6.797; $\eta p^2 = 0.245$). Post-hoc analyses revealed that the peak ankle eversion angle increased from pre- to post-fatigue when pronated runners used both the neutral shoes (p = 0.024) and motion-control shoes (p = 0.021).

3.4. Peak negative free moment vs LR relationship

There were no significant correlations between peak negative free moments and LR Pre-(r = − 0.26, p > 0.05) and Post-fatigue

(r = − 0.22, p > 0.05, Fig. 3A) when runners used the neutral shoe. However, there were strong correlations between peak negative free moments and LR for both Pre-fatigue (r = − 0.61, p < 0.005, Fig. 3B) and Post-fatigue measurements (r = − 0.66, p < 0.01) when pronated female runners used the motion-control shoes.

3.5. KEAM vs LR relationship

There were non-significant correlations between LR and peak KEAM for the neutral shoes Pre- (r = 0.03, p > 0.05) and Post-fatigue (r = 0.09, p > 0.05). Similarly, very weak and non-significant correlations were found between these variables for the motion-control shoe Pre- (r = 0.009, p > 0.05) and Post-fatigue (r = 0.13, p > 0.05).

4. Discussion

The main findings of this study were that motion-control shoes attenuated the rate of increase in pvIGRF and LR following fatiguing running when compared to neutral shoes. Motion control shoes can also reduce the peak ankle eversion angle when compared to neutral shoes, but ankle eversion will increase following fatigue regardless the type of shoe used. Moreover, there was a stronger association between peak negative free moments and LR when runners used motion-control shoes, which was preserved following a fatiguing running protocol. These results suggest that motion-control shoes prevent exacerbated fatigue-related increases in early mechanical loading in pronated female runners. Moreover, motion-control shoes seem to provide a robust mechanism for runners to control the relationship between impact loading and medio-lateral foot loading in fatiguing conditions.

As hypothesized, pronated runner using neutral shoes presented greater pvIGRF and LR following fatiguing running. Literature shows agreement [37] and disagreement [38] with respect to our findings of increased impact forces with fatigue during running with neutral shoes. An increased pvIGRF has linked to increased pre-activation of the stabilizing musculature (such as the knee extensors) following fatigue [37], which increases joint stiffness [22,24]. Previous studies have demonstrated an association between increased LR and the development of knee osteoarthritis [39], as well as the occurrence of running-related injuries [40]. In addition, previously injured female runners are more susceptible to increases in LR following a fatiguing running session LR [41]. Such results suggest that increases in LR may represent excessive loading that can lead to musculoskeletal disorders. Increased in LR have been partially explained by insufficient lower extremity strength relative to body weight, low quality of footwear as well as inappropriate running technique [42–44]. Our results contribute to the current knowledge by demonstrating that fatigue can be a relevant factor to increase LR in female pronated runners that do not wear shoes appropriated for their foot alignment.

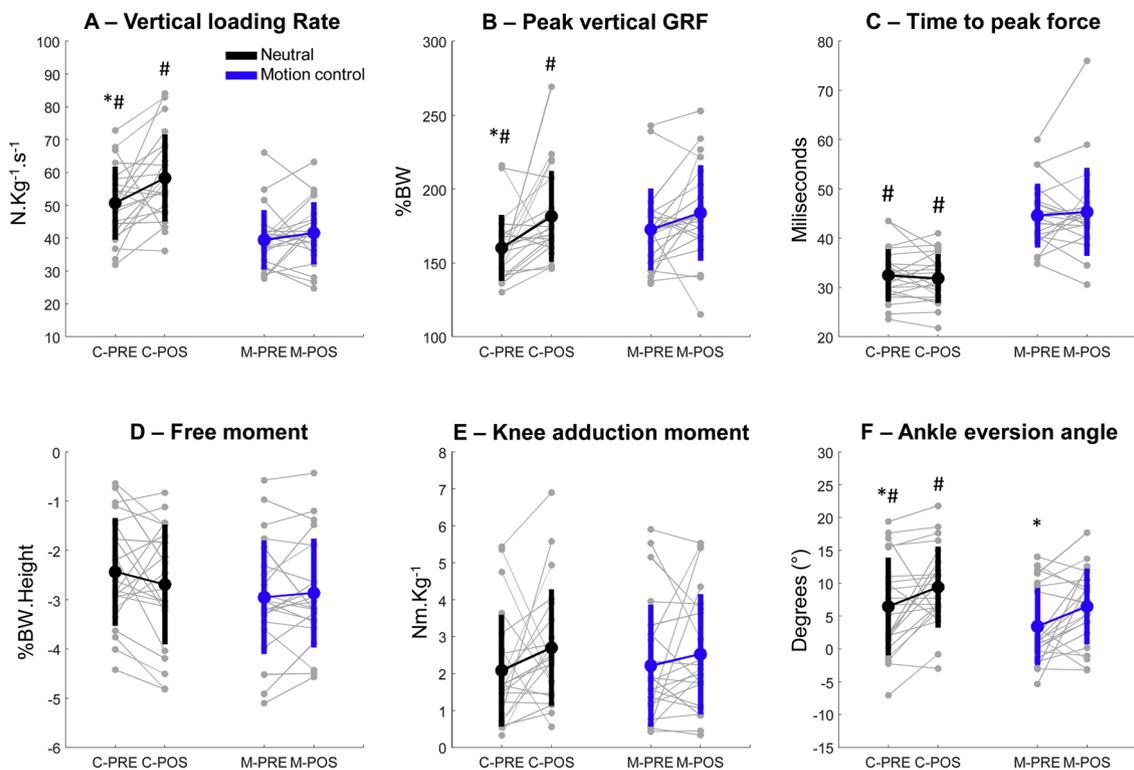In this study, motion-control shoes allow pronated runners to

Fig. 3. Pearson correlation coefficient between vertical loading rate (*x axis*) and peak negative free moments (*y axis*) Pre- (*black dots and trend lines*) and Post-fatigue measurements (*red dots and trend lines*) for the neutral shoes (*panel A*) and the motion-control shoes (*panel B*). Pearson correlation coefficient (r) and significance level (p) are shown for each comparison.

maintain their levels of LR and pvIGRF following fatiguing running. To our knowledge, this is the first study demonstrating that motion-control shoes can assist pronated runners to control pvIGRF during a fatiguing workout. It is expected that fatigue might reduce neuromuscular control, reducing the capacity to control impact loading. The lower LR during running with motion-control shoes compared with neutral shoes was related to an extended period to reach pvIGRF. Such timing adaptation may be related to specific changes in running kinematics, such as the reduced peak ankle eversion angle described in our results. The specific midsole design and material of the motion-control shoe may be directly related to the changes in foot kinematics [20]. The midsole construction is the only relevant difference between the motion-control and neutral shoes used in this study, as they were made by the same manufacturer with highly similar properties and materials. Therefore, our results demonstrate that motion-control shoes were efficient to assist runners in maintaining their early vertical loading levels, potentially reducing the risk of injuries related to instantaneous impact loads.

Our results indicated an inverse association between LR and peak negative free moments for female pronated runners using the motion-control shoes. Interestingly, such association was present for both fresh and fatiguing conditions. Conversely, there was no such association between LR and peak negative free moments when runners used neutral shoes. Excessive ankle eversion can significantly increase both peak negative free moments and LR during walking and running [30]. Clinically, ankle eversion and arch collapse are assumed to be biomechanical factors that increase plantar fascia tension [45]. Since our results demonstrated that motion-control shoes reduced peak ankle eversion compared with neutral shoes during both pre and post-fatigued conditions, reductions in both peak negative free moments and LR were expected during running with motion-control shoes. However, only LR measured from motion-control shoes was reduced when compared to control shoes. It has been suggested that negative free moments may be indicative of the torsional stress exerted on the lower limbs [46]. Therefore, the inverse correlation between peak negative

free moments and LR suggest that runners presenting the lowest LR were also those presenting peak negative free moments closer to zero. In other words, motion-control shoes allowed runners to minimize torsional forces when the LR was reduced. It is noteworthy that the reduced LR was caused by an extended time to reach the pvIGRF, rather than reduction in force magnitudes. Therefore, our results suggest that pronated runners adapt their foot kinematics (by reducing ankle eversion) when using motion-control shoes. Consequently, this kinematic adaptation leads to a prolonged time to reach pvIGRF, while concomitantly reducing their torsional forces during stance. More importantly, reduced ankle eversion and torsional forces were still present under fatiguing conditions, representing an important mechanism to avoid exacerbated impact loading during running. A previous study demonstrated that impact loading was increased when normal ankle eversion in healthy recreational runners was prevented during running [47]. Also, there was no alteration in impact loading when normal ankle eversion was exaggerated [47].

The KEAM has been traditionally used to describe knee loading on the frontal plane, being related to running-related injuries [48]. It has been shown that increased medial support in shoes may move the ground reaction force line of action medially, thereby increasing the moment arm at the knee [49]. Such change in running mechanics might increase loading in the medial knee compartment in pronated runners using motion-control shoes. In the present study, we assessed the KEAM during running with neutral and motion-control shoes, and there were no shoe-related differences either before or after fatigue. The lack of fatigue-related differences in KEAM corroborates the study from Longpre et al. who induced lower limb muscle fatigue using a knee flexion/extension resistance exercise at 50% maximum isometric contraction [50]. These authors also found no changes in KEAM during running following the fatiguing protocol. It is noteworthy the scarce number of studies reporting changes in knee mechanics in fatiguing conditions during running. Therefore, our study adds substantial contributions to the field by reporting fatigue-related changes for pronated female runners using neutral or motion-control shoes.

The lack of shoe-related effects on KEAM in our study corroborates a previous investigation that reported similar KEAM in mature and young female runners using neutral and motion-control shoes [21]. Combined, this evidence suggest that motion-control shoes may not necessarily and/or directly increase knee medial compartment loading in pronated female runners. However, it is noteworthy that the presented results are limited to describing changes in running kinetics in female runners with pronated, therefore our findings may not be applicable to male runners, as gender influences lower extremity movement patterns during running [12]. Future studies replicating the proposed design in male pronated runners are necessary for establishing the effects of motion-control shoes on this population.

5. Conclusion

In summary, our results suggest that motion-control shoes prevent exacerbated fatigue-related increases in mechanical loading following heel strike in pronated female rearfoot runners. Moreover, motion-control shoes seem to provide a robust mechanism for runners to control the relationship between impact loading and foot torsion in fatiguing conditions.

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References

[1] R. Van Gent, D. Siem, M. van Middelkoop, A. Van Os, S. Bierma-Zeinstra, B. Koes, Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review, Br. J. Sports Med. 41 (8) (2007) 469–480.

[2] L. Linton, S. Valentin, Running with injury: a study of UK novice and recreational runners and factors associated with running related injury, J. Sci. Med. Sport 21 (12) (2018) 1221–1225.

[3] B.T. Saragiotto, T.P. Yamato, L.C.H. Junior, M.J. Rainbow, I.S. Davis, A.D. Lopes, What are the main risk factors for running-related injuries? Sport. Med. 44 (8) (2014) 1153–1163.

[4] L. Malisoux, N. Chambon, N. Delattre, N. Gueguen, A. Urhausen, D. Theisen, Injury risk in runners using standard or motion control shoes: a randomised controlled trial with participant and assessor blinding, Br. J. Sports Med. 50 (8) (2016) 481–487.

[5] T.R. Souza, R.Z. Pinto, R.G. Trede, R.N. Kirkwood, S.T. Fonseca, Temporal couplings between rearfoot–shank complex and hip joint during walking, Clin. Biomech. 25 (7) (2010) 745–748.

[6] N. Farahpour, A. Jafarnezhadgero, P. Allard, M. Majlesi, Muscle activity and kinetics of lower limbs during walking in pronated feet individuals with and without low back pain, J. Electromyogr. Kinesiol. 39 (2018) 35–41.

[7] V.H. Chuter, X.A.J. de Jonge, Proximal and distal contributions to lower extremity injury: a review of the literature, Gait Posture 36 (1) (2012) 7–15.

[8] R.A. Resende, R.N. Kirkwood, K.J. Deluzio, E.A. Hassan, S.T. Fonseca, Ipsilateral and contralateral foot pronation affect lower limb and trunk biomechanics of individuals with knee osteoarthritis during gait, Clin. Biomech. 34 (2016) 30–37.

[9] K.L. Bennell, K.-A. Bowles, Y. Wang, F. Cicuttini, M. Davies-Tuck, R.S. Hinman, Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis, Ann. Rheum. Dis. 70 (10) (2011) 1770–1774.

[10] A. Chang, K. Moisio, A. Guermazi, J. Chmiel, O. Almagor, P. Prasad, et al., The relationship between the external knee adduction moment during gait and progression of knee tissue damage over two years in persons with knee osteoarthritis, Osteoarthr. Cartil. 22 (2014) S44–S45.

[11] E.F. Chehab, J. Favre, J.C. Erhart-Hledik, T.P. Andriacchi, Baseline knee adduction and flexion moments during walking are both associated with 5 year cartilage changes in patients with medial knee osteoarthritis, Osteoarthr. Cartil. 22 (11) (2014) 1833–1839.

[12] R. Ferber, I.M. Davis, D.S. Williams Iii, Gender differences in lower extremity mechanics during running, Clin. Biomech. 18 (4) (2003) 350–357.

[13] K.D. Shelbourne, T.J. Davis, T.E. Klootwyk, The relationship between intercondylar notch width of the femur and the incidence of anterior cruciate ligament tears, Am. J. Sports Med. 26 (3) (1998) 402–408.

[14] E.M. Wojtys, J.A. Ashton-Miller, L.J. Huston, A gender-related difference in the contribution of the knee musculature to sagittal-plane shear stiffness in subjects with similar knee laxity, J. Bone Jt. Surg. 84 (1) (2002) 10–16.

[15] M.R. Hutchinson, M.L. Ireland, Knee injuries in female athletes, Sport. Med. 19 (4)

[16] C.A. Oatis, Kinesiology: The Mechanics and Pathomechanics of Human Movement, Lippincott Williams & Wilkins, 2009.

[17] A.A. Zadpoor, A.A. Nikooyan, The relationship between lower-extremity stress fractures and the ground reaction force: a systematic review, Clin. Biomech. 26 (1) (2011) 23–28.

[18] C.E. Milner, I.S. Davis, J. Hamill, Free moment as a predictor of tibial stress fracture in distance runners, J. Biomech. 39 (15) (2006) 2819–2825.

[19] T. Ohkawa, T. Atomi, K. Hasegawa, Y. Atomi, The free moment is associated with torsion between the pelvis and the foot during gait, Gait Posture 58 (2017) 415–420.

[20] R.T. Cheung, G.Y. Ng, Efficacy of motion control shoes for reducing excessive rearfoot motion in fatigued runners, Phys. Ther. Sport. 8 (2) (2007) 75–81.

[21] K. Lilley, V. Stiles, S. Dixon, The influence of motion control shoes on the running gait of mature and young females, Gait Posture 37 (3) (2013) 331–335.

[22] C.R. James, B.W. Scheuermann, M.P. Smith, Effects of two neuromuscular fatigue protocols on landing performance, J. Electromyogr. Kinesiol. 20 (4) (2010) 667–675.

[23] A. Nummela, H. Rusko, A. Mero, EMG activities and ground reaction forces during fatigued and nonfatigued sprinting, Med. Sci. Sports Exerc. 26 (5) (1994) 605–609.

[24] R.C. James, J.S. Dufek, B.T. Bates, Effects of stretch shortening cycle exercise fatigue on stress fracture injury risk during landing, Res. Q. Exerc. Sport 77 (1) (2006) 1–13.

[25] A.A. Jafarnezhadgero, M. Majlesi, H. Etemadi, D. Robertson, Rehabilitation improves walking kinematics in children with a knee varus: randomized controlled trial, Ann. Phys. Rehabil. Med. 61 (3) (2018) 125–134.

[26] N. Delattre, N. Chambon, E. Berton, N. Gueguen, G. Rao, Effect of time during a running session with minimal footwear, Comput. Methods Biomech. Biomed. Eng. 16 (sup1) (2013) 104–105.

[27] M. Anbarian, H. Esmaeili, Effects of running-induced fatigue on plantar pressure distribution in novice runners with different foot types, Gait Posture 48 (2016) 52–56.

[28] G. Borg, Borg's perceived exertion and pain scales, Human kinetics (1998).

[29] D.A. Winter, Biomechanics and Motor Control of Human Movement, John Wiley & Sons, 2009.

[30] S. Almosnino, T. Kajaks, P.A. Costigan, The free moment in walking and its change with foot rotation angle, Sports Med. Arthrosc. Rehabil. Ther. Technol. 1 (1) (2009) 19.

[31] A. Cappozzo, F. Catani, U. Della Croce, A. Leardini, Position and orientation in space of bones during movement: anatomical frame definition and determination, Clin. Biomech. 10 (4) (1995) 171–178.

[32] R. Shultz, T. Jenkyn, Determining the maximum diameter for holes in the shoe without compromising shoe integrity when using a multi-segment foot model, Med. Eng. Phys. 34 (1) (2012) 118–122.

[33] A. Stacoff, C. Reinschmidt, E. Stüssi, The movement of the heel within a running shoe, Med. Sci. Sports Exerc. 24 (6) (1992) 695–701.

[34] R.B. Davis III, S. Ounpuu, D. Tyburski, J.R. Gage, A gait analysis data collection and reduction technique, Hum. Mov. Sci. 10 (5) (1991) 575–587.

[35] T.F. Besier, D.L. Sturnieks, J.A. Alderson, D.G. Lloyd, Repeatability of gait data using a functional hip joint centre and a mean helical knee axis, J. Biomech. 36 (8) (2003) 1159–1168.

[36] W.T. Dempster, W.C. Gabel, W.J. Felts, The anthropometry of the manual work space for the seated subject, Am. J. Phys. Anthropol. 17 (4) (1959) 289–317.

[37] C. Nicol, P. Komi, P. Marconnet, Fatigue effects of marathon running on neuromuscular performance, Scand. J. Med. Sci. Sports 1 (1) (1991) 10–17.

[38] J.D. Willson, T.W. Kernozek, Plantar loading and cadence alterations with fatigue, Med. Sci. Sports Exerc. 31 (12) (1999) 1828–1833.

[39] A.A. Jafarnezhadgero, M.M. Shad, M. Majlesi, U. Granacher, A comparison of running kinetics in children with and without genu varus: a cross sectional study, PLoS One 12 (9) (2017) e0185057.

[40] H. van der Worp, J.W. Vrielink, S.W. Bredeweg, Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis, Br. J. Sports Med. 50 (8) (2016) 450–457.

[41] K.E. Gerlach, S.C. White, H.W. Burton, J.M. Dorn, J.J. Leddy, P.J. Horvath, Kinetic changes with fatigue and relationship to injury in female runners, Med. Sci. Sports Exerc. 37 (4) (2005) 657–663.

[42] A. Herbaut, P. Chavet, M. Roux, N. Guéguen, F. Barbier, E. Simoneau-Buessinger, The influence of shoe aging on children running biomechanics, Gait Posture 56 (2017) 123–128.

[43] B.J. Bowser, R. Fellin, C.E. Milner, M.B. Pohl, I.S. Davis, Reducing impact loading in runners: a one-year follow-up, Med. Sci. Sports Exerc. 50 (12) (2018) 2500–2506.

[44] J.J. Tate, C.E. Milner, Sound-intensity feedback during running reduces loading rates and impact peak, J. Orthop. Sports Phys. Ther. 47 (8) (2017) 565–569.

[45] S.Y. Lee, J. Hertel, S.C. Lee, Rearfoot eversion has indirect effects on plantar fascia tension by changing the amount of arch collapse, Foot (Edinb.) 20 (2-3) (2010) 64–70.

[46] P.-F. Yang, M. Sanno, B. Ganse, T. Koy, G.-P. Brüggemann, L.P. Müller, et al., Torsion and antero-posterior bending in the in vivo human tibia loading regimes during walking and running, PLoS One 9 (4) (2014) e94525.

[47] S. Perry, M. Lafortune, Influences of inversion/eversion of the foot upon impact loading during locomotion, Clin. Biomech. 10 (5) (1995) 253–257.

[48] I. Fong, W. Li, W. Tai, T. Tsang, J. Zhang, T. Chen, et al., Effect of foot progression angle adjustment on the knee adduction moment and knee joint contact force in runners with and without knee osteoarthritis, Gait Posture 61 (2018) 34–39.

[49] K.B. Shelburne, M.R. Torry, J.R. Steadman, M.G. Pandy, Effects of foot orthoses and valgus bracing on the knee adduction moment and medial joint load during gait, Clin. Biomech. 23 (6) (2008) 814–821.

[50] H.S. Longpré, J.R. Potvin, M.R. Maly, Biomechanical changes at the knee after lower limb fatigue in healthy young women, Clin. Biomech. 28 (2013) 441–447.