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Research article

Biomechanical Analysis of Running Foot Strike in Shoes of Different Mass

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Abstract

Different shoes and strike patterns produce different biomechanical characteristics that can affect injury risk. Running shoes are mainly designed as lightweight, minimal, or traditional cushioned types. Previous research on different shoes utilized shoes of not only different mass but also different shoe structures. However, it is unclear whether biomechanical changes during running in different shoe types with differing mass are the result of the structural design or the mass of the shoe. Thus, the purpose of this study was to investigate the effect of shoes of different mass on running gait biomechanics. Twenty male runners participated in this study. The experimental shoe masses used in this study were 175, 255, 335 and 415 g. The peak vertical ground reaction force increased with shoe mass ($p < 0.05$), but the strike index, ankle plantarflexion at initial contact, peak moment of the ankle during the stance phase, and initial contact angles of the lower extremity joints did not change. During the pre-activation phase, the integrated EMG data showed that the tibialis anterior muscle was the most activated with the 175 g and 415 g shoes ($p < 0.05$). During the push-off phase, the semitendinosus, lateral gastrocnemius and soleus muscles displayed higher activation with the heavier shoes ($p < 0.05$). The center of pressure also moves forward; resulting in mid foot striking. The lightest shoes might increase gastrocnemius muscle fatigue during the braking phase. The heaviest shoes could cause semitendinosus and triceps surae muscle fatigue during the push-off phase. Therefore, runners should consider their lower extremity joints, muscle adaptation and cushioning to remain in their preferred movement path.

Key words: Running shoes, injury prevention, electromyography.

Introduction

Recently, in the field of sports medicine, there has been a large amount of research conducted on the biomechanical adaptations to running in modified footwear. Specifically, there has been an increased focus on barefoot running (Becker et al., 2014; Cheung and Rainbow, 2014; Powell et al., 2014; Rao et al., 2015; Strauts et al., 2015; Thompson et al., 2015), running strike patterns (Ahn et al., 2014; Lieberman et al., 2010; Shih et al., 2013), and various running shoe styles (Barnes et al., 2010; Bonacci et al., 2013; Hollander et al., 2014; Squadrone and Gallozzi, 2009). Researchers have found that different shoes and different strike patterns produce different biomechanical characteristics that can affect injury risk. Previous research comparing barefoot and shod running has demonstrated that barefoot running can change the landing strategy through adaptation to a forefoot strike (FFS) as opposed to a rearfoot strike (RFS), thereby reducing landing impact transients (Lieberman et al., 2010). Thus, the collision force is one of the main factors causing lower extremity running injuries during the running braking phase (Lieberman et al., 2010; Shih et al., 2013). Another comparison between barefoot running and shod running found that the barefoot running stride was shorter, with decreased contact time and higher stride frequency (De Wit et al., 2000; Divert et al., 2005). Shod running attenuates the foot-ground impact by adding damping material to avoid direct contact with the ground. Therefore, shod running may lead to a decrease in the storage and restitution of elastic energy and lower net efficiency (Divert et al., 2008). Furthermore, a comparison of the effects of running barefoot, shod and in Vibram Fivefingers (Vibram SpA, Albizzate, Italy), a lightweight minimalist shoe, revealed that runners in Vibram Fivefingers demonstrated longer strides and lower stride frequency relative to barefoot conditions and decreased contact time relative to shod conditions (Squadrone and Gallozzi, 2009).

Forefoot or midfoot strikes (MFS) are a major component of barefoot running, while most shod runners use a heel strike pattern (De Wit et al., 2000; Hasegawa et al., 2007; Lieberman et al., 2010). The cushioning capabilities of some minimalist shoes allow the runner to adopt a heel strike pattern and avoid painful heel contact with the ground (Bonacci et al., 2013; Willy and Davis, 2014). As the heel strike pattern is associated with greater dorsiflexion of the ankle joint, this pattern can reduce the ability of the ankle to attenuate impact forces (Hollander et al., 2014; Willy and Davis, 2014). However, shod running prevents a direct heel to running-surface impact during landing, leading to longer strides through modification of the contact geometry (De Wit et al., 2000). It is essential to understand how foot strike strategies are modulated by different footwear, as it has been established that different strike patterns can affect the risk of lower extremity injuries (Hamill et al., 1999; Wang et al., 2018). Therefore, if we understand how these factors change foot strike patterns, we can prescribe footwear more appropriately to individuals.

Researchers have also previously found that different types of running shoes (e.g., Vibram Fivefingers, lightweight shoes, and minimalist shoes) can affect running biomechanics and, therefore, influence the risk of injury (Bonacci et al., 2013; Sinclair, 2014; Squadrone and Gallozzi, 2009; Willy and Davis, 2014). Along with changes in design, there are also differences in the mass of

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running shoes: approximately 150-200 g for marathon running and Vibram Fivefingers shoes, 200-300 g for a neutral running shoe and 360 g for a traditional running shoe (Mizuno, 2018). In the past, few have explored the independent effect of mass, and its influence on running biomechanics is not well understood. Related studies have observed that minimalist shoes have moderate beneficial effects on running economy (Perl et al., 2012). It has been demonstrated that there is no detrimental effect of shoe mass on metabolic cost if the combined mass of both shoes is less than 440 g. However, if the combined mass of the shoes exceeds 440 g, there is a positive correlation with metabolic cost (Fuller et al., 2015).

It is well established that the human body will change its foot strike pattern during running in response to different designs of running shoes. The majority of shod runners use a RFS, and minimal shoes or barefoot runners use FFS and MFS, respectively (Hasegawa et al., 2007; Larson, 2014). Currently, running shoes are mainly designed as lightweight shoes, minimal shoes, and traditional cushioned running shoes. Therefore, it is unclear whether the biomechanical changes during running in different shoe types of differing mass are the result of the structural design or mass of the shoe. To address this, the current study was conducted using four identical types of shoes that differed only in mass to study the effect of shoe mass on running biomechanics in isolation. A standard, traditional running shoe was selected for this study as it provided an average amount of cushioning that would not influence running mechanics due to excessive or inadequate cushioning. It was hypothesized that greater shoe mass would result in increased vertical ground reaction force (VGRF) and a change to MFS running patterns.

Methods

Participants

Twenty collegiate male physical education students volunteered for this study and provided written informed consent. The participants' mean age, height and mass were 21.8 years (SD=1.2), 1.72 m (SD=0.03) and 68.00 kg (SD=4.32), respectively. None of the participants had a history of lower extremity injuries during the six months prior to the experiment. The runners were previously familiarized with shoes of different mass a minimum of four times at the test site, where they ran one time along the 10 meter runway in each shoe on four different days in one week. The study was approved by the Antai Medical Care Corporation Memorial Hospital (No. 15-066-B1).

Protocol

Four shoe mass conditions were used in this study (Figure 1); for each condition, a single shoe was 175 g, 255 g (4x20 g lead mass), 335 g (4×40 g lead mass), or 415 g (4×60 g lead mass). Lead mass was attached to the shoes to achieve the required total mass and were evenly distributed on the four sides of the shoe. As a warm-up, participants were instructed to run back and forth for 20 minutes at a comfortable self-selected pace around the corridor, as well as perform static stretching. Following warm-up and prior to data collection, participants were asked to habituate to the runway at a comfortable self-selected pace and were given 5 practice trials in each footwear condition to become familiarized with the shoe mass. Allowing the participants to self-select the running pace facilitated a more natural response to running with additional mass on the shoes. Following familiarization, participants were instructed to run down the runway, in which two force plates were positioned to capture a right and left foot impact. Participants completed five trials for each mass condition, and three trials with stable foot contact were selected and averaged to increase the stability of measurements. Data utilized in the analysis were recorded from the participant's right foot contact with the first force plate to left foot toe-off on the second force plate (Figure 2).

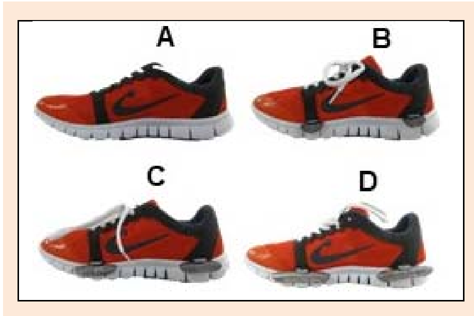

Figure 1. Footwear used for each condition. A) 175g condition, B) 255g condition, C) 335g condition, and D) 415g condition.

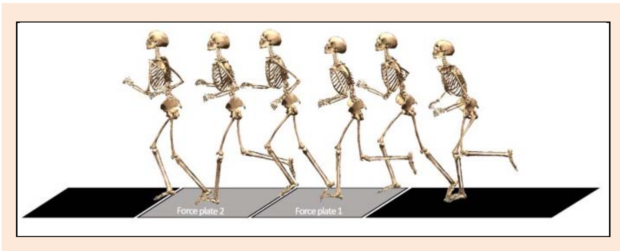

Figure 2. Experimental set up.

Biomechanical measurements

An eight-camera motion analysis system (Qualisys Track Manager [QTM], Oqus 100, Gothenburg, Sweden) was used to collect whole-body motion data with a total of 40 reflective markers placed on bony landmarks (19 mm in diameter) according to the Helen Hayes Marker set. Marker trajectories were sampled at 200 Hz. Two AMTI force plates (BP600900, AMTI Inc., Watertown, MA, USA) were used to collect the GRF data at 1000 Hz. All GRF data were normalized to participant body mass. A surface EMG system (Myomonitor IV, Delsys Inc., Natick, MA, USA) was used to measure muscle activation of the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), semitendinosus, tibialis anterior (TA), gastrocnemius lateralis (GAS lat), gastrocnemius medialis (GAS med), and soleus (SO) muscles at a 1000-Hz sampling rate. All electrodes were placed in the lengthwise direction of the muscle on the right leg. The electrodes and electrode wires were wrapped on the thigh and shank with an elastic bandage to prevent dislocation during running. Right foot strike kinematic, kinetic and EMG data were synchronized with a Qualisys 64-channel analog-to-digital interface.

Data analysis

Three-dimensional coordinates of the reflective markers and the force data were imported into MATLAB software (version 7.0; The MathWorks Inc., Natick, MA, USA) from QTM software for data reduction and analysis. The measurement of VGRF during the stance phase was used for identification of initial contact (IC) and toe-off (TO). The braking phase was identified using the anterior-posterior GRF, and the stance phase was further divided into the braking phase and the push-off phase. The braking phase was identified by negative anterior-posterior values from IC to zero, and the push-off phase was identified by positive anterior-posterior values from zero to TO (Wang et al., 2018).

The pre-activation phase was defined as the time from 50 ms before foot IC (Shih et al., 2013). A VGRF threshold of 20 N was used to determine IC and TO. The kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz. The kinetic data were then low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz. A kinematic model was generated by defining the skeletal segment (foot, shank). The right-handed orthogonal Cardan Z-Y-X rotation sequence was used to define the relative joint between the foot and shank segments. Joint angles were normalized to the bilateral static stance measurement. Ankle plantarflexion was calculated at IC.

Right second metatarsal head (RTOE), right calcaneus (RHEE) and right lateral malleolus (RANK) markers established the foot local coordinate system. The center of pressure at IC along the longitudinal axis of the foot coordinate system. Right foot length multiplied by 100 was normalized to obtain the strike index (SI) (Altman and Davis, 2012). Only the right foot SI was used to characterize the foot strike patterns. The raw EMG signals from each muscle were corrected for DC offset, rectified and low-pass filtered at 10 Hz with a zero-phase second-order Butterworth filter (Hamner and Delp, 2013). Root mean square (RMS) values for the pre-activation phase, braking phase, push-off phase and stance phase were used to compare each muscle activation level among each of the shoe conditions. Integrated EMG (IEMG) was defined as the summation of each RMS value normalized to the maximum voluntary contraction recorded across all trials for each subject (Shih et al., 2013).

Statistical analysis

All dependent variables were statistically analyzed using SPSS 14.0 for Windows (IBM Corporation, Armonk, NY, USA). Statistical comparisons were made using multiple one-way, repeated measures ANOVAs to compare SI, joint angles, peak VGRF and IEMG among shoe mass conditions. The chi-square test was used to compare shoe mass condition SI numbers. Fisher's least significant difference (LSD) test was used for the pairwise comparisons. The significance level was set at α=0.05 for all statistical tests.

Results

According to the SI, the foot strike patterns were divided into RFS, MFS and FFS (Figure 3). For the 175 g condition, there were six subjects with RFS (30%), five subjects with MFS (25%) and nine subjects with FFS (45%). For the 225 g condition, there were four subjects with RFS (20%), seven subjects with MFS (35%), and nine subjects with FFS (45%). For the 335 g condition, there were three subjects with RFS (15%), eight subjects with MFS (40%), and nine subjects with FFS (45%). For the 415 g condition, there was one subject with RFS (5%), eleven subjects with MFS (55%) and eight subjects with FFS (40%). The result of the chi-square test was not significant for SI change (p = 0.399), as shown in Figure 3.

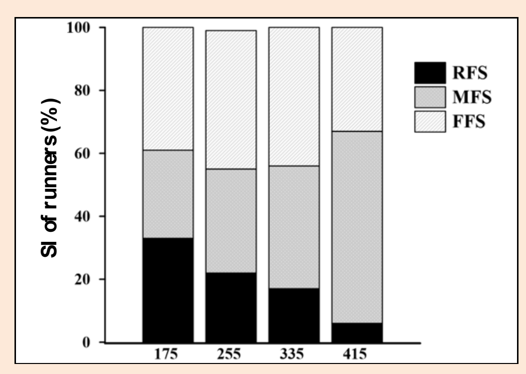

Figure 3. SI of runners for the different shoe weights.

The results for the SI, peak VGRF, and ankle plantarflexion angle and of the moment calculations are presented in Table 1. The peak VGRF was significantly different in the stance phase (p = 0.043, η^2 = 0.13). Post hoc testing revealed that the 415 g condition led to greater peak VGRF than the 175 g condition (p = 0.044) and 255 g condition (p = 0.013).

The EMG results from each muscle in the pre-activation phase are presented in Table 2. Significant differences in the TA during running were observed among the different shoe masses (p = 0.008). The greatest TA pre- activation phase was observed for the 175 g condition. Post hoc testing revealed that the 175 g condition led to greater activation than the 255 g condition (p = 0.022) and the 335 g condition (p = 0.010). It was also determined that the TA

Table 1. Effects of different shoe mass on biomechanical parameters during running.

	175g	255g	335g	415g	p-value
SI (%)	73.75±2.08	68.46±11.54	67.77±11.83	67.03±19.37	0.661
Ankle plantarflexion at initial contact (deg)	23.94±12.74	18.93±13.21	19.42±11.39	19.63±10.35	0.066
Peak moment of ankle plantarflexion during the stance phase (Nm/BW*%)	2.94±0.66	2.71±0.64	2.59±0.65	2.66±0.68	0.094
Peak VGRF (N/BW)	27.82 ±3.79	27.67 ±3.52	28.23 ±4.31	28.63 ±3.93	0.043*[a, b]

* Significant differences ($p < 0.05$) between shoes of different masses. a: Significant difference between 175g and 255g. b: Significant difference between 225g and 315g.

Table 2. IEMG during the pre-activation phase (unit: %MVC).

	175g	255g	335g	415g	p-value
Vastus lateralis	5±0.01	4±0.01	4±0.02	4±0.01	0.738
Vastus medialis	5±0.02	4±0.01	4±0.01	5±0.01	0.302
Rectus femoris	8±0.04	8±0.03	9±0.04	9±0.03	0.339
Biceps femoris	21±0.04	22±0.04	23±0.07	20±0.04	0.126
Semitendinosus	20±0.03	20±0.04	22±0.06	21±0.05	0.362
Tibialis anterior	28±0.05	20±0.08	24±0.10	24±0.08	0.008*
Gastrocnemius lateralis	8±0.04	8±0.04	8±0.04	9±0.04	0.932
Gastrocnemius medialis	7±0.03	7±0.05	7±0.06	8±0.08	0.987
Soleus muscles	7±0.03	5±0.04	6±0.04	6±0.04	0.312

* Significant differences ($p<0.05$) between shoes of different mass during the pre-activation phase.

Table 3. IEMG during the braking phase (unit: %MVC).

	175g	255g	335g	415g	p-value
Vastus lateralis	0.5±0.14	0.49±0.14	0.60±0.25	0.52±0.23	0.173
Vastus medialis	0.49±0.08	0.45±0.07	0.46±0.07	0.48±0.10	0.211
Rectus femoris	0.36±0.16	0.36±0.15	0.36±0.17	0.39±0.19	0.458
Biceps femoris	0.47±0.17	0.44±0.22	0.42±0.16	0.47±0.22	0.632
Semitendinosus	0.43±0.10	0.35±0.11	0.38±0.12	0.46±0.07	0.179
Tibialis anterior	0.32±0.08	0.27±0.12	0.28±0.15	0.31±0.12	0.344
Gastrocnemius lateralis	0.51±0.14	0.51±0.14	0.53±0.11	0.59±0.10	0.243
Gastrocnemius medialis	0.43±0.16	0.49±0.19	0.50±0.13	0.50±0.13	0.468
Soleus muscles	0.55±0.09	0.53±0.10	0.57±0.08	0.59±0.10	0.099

* Significant differences ($p<0.05$) between shoes of different masses during the braking phase.

demonstrated higher activity in the pre-activation phase in the 415 g condition than in the 255 g condition ($p = 0.008$). There were no significant differences in IEMG during the braking phase, as shown in Table 3.

The sEMG results from each muscle in the push-off phase are presented in Table 4. Significant differences for the semitendinosus ($p = 0.04$), GAS lat ($p = 0.014$) and SO muscles ($p = 0.041$) during running were observed among the different shoe masses. Post hoc testing revealed that the semitendinosus activation was significantly greater for the 415 g condition than for the 175 g condition ($p = 0.018$) and 255 g condition ($p = 0.001$). Post hoc testing also revealed that GAS lat activation was significantly greater for the 415 g condition than for the 255 g condition ($p = 0.006$). Additionally, SO activation was significantly greater for the 335 g condition than for the 255 g condition ($p = 0.001$) and for the 415 g condition than for the 255 g condition ($p < 0.001$).

The sEMG results from each muscle in the stance phase are presented in Table 5. Significant differences were observed for the GAS lat ($p = 0.005$) and SO muscles ($p = 0.028$) during running in shoes of different mass. Post hoc testing revealed that the GAS lat activation was significantly greater for the 415 g condition than for the 175 g condition ($p = 0.001$) and 255 g condition ($p = 0.001$). Additionally, SO activation was significantly greater for the 335 g condition than for the 255 g condition ($p = 0.042$), and SO activation for the 415 g condition was significantly greater than that for the 255 g condition ($p < 0.001$).

Table 4. IEMG during the push-off phase (unit: %MVC).

	175g	255g	335g	415g	p-value
Vastus lateralis	11±0.06	10±0.03	15±0.10	16±0.15	0.349
Vastus medialis	10±0.07	8±0.02	11±0.02	10±0.04	0.573
Rectus femoris	9±0.05	9±0.04	9±0.03	9±0.03	0.605
Biceps femoris	47±0.17	54±0.29	57±0.26	67±0.35	0.380
Semitendinosus	37±0.18	39±0.17	44±0.21	50±0.11	0.040*
Tibialis anterior	10±0.06	11±0.07	13±0.09	13±0.09	0.050
Gastrocnemius lateralis	38±0.13	39±0.09	45±0.12	48±0.11	0.014*
Gastrocnemius medialis	45±0.06	48±0.09	48±0.09	51±0.07	0.326
Soleus muscles	40±0.08	38±0.07	49±0.08	47±0.09	0.041*

* Significant differences ($p<0.05$) between shoes of different mass during the push-off phase.

Table 5. IEMG during the stance phase (unit: %MVC).

	175g	255g	335g	415g	p-value
Vastus lateralis	31±0.09	30±0.08	37±0.17	34±0.19	0.315
Vastus medialis	31±0.05	27±0.03	28±0.04	30±0.06	0.114
Rectus femoris	23±0.09	22±0.08	22±0.08	24±0.10	0.206
Biceps femoris	47±0.14	49±0.25	50±0.21	57±0.26	0.458
Semitendinosus	40±0.10	37±0.12	41±0.12	48±0.08	0.073
Tibialis anterior	22±0.06	19±0.08	21±0.11	22±0.09	0.383
Gastrocnemius lateralis	45±0.05	45±0.07	49±0.07	53±0.06	0.005*
Gastrocnemius medialis	44±0.10	49±0.12	49±0.11	50±0.07	0.230
Soleus muscles	48±0.05	46±0.08	53±0.07	53±0.08	0.028*

* Significant differences ($p<0.05$) between shoes of different mass during the stance phase.

Discussion

Different styles of running shoes (i.e., Vibram Fivefingers, lightweight shoes, and minimalist shoes) will change the lower extremity loading patterns of running (Bonacci et al., 2013; Sinclair, 2014; Squadrone and Gallozzi, 2009; Willy and Davis, 2014). The results of this study show that the mass of running shoes will change the peak VGRF and the activation of lower extremity muscles during running, independent of shoe type.

Previous studies found that the varied structures of different types of shoes alter the running strategy. A comparison of Vibram Fivefingers (148 g) and standard running shoes (341 g) demonstrated that the SI of the Fivefingers is significantly larger than that of standard running shoes as the runner's center of pressure was posterior from the heel at IC. Fivefingers and standard running shoes differ in terms of mass and structure. Specifically, the soles of Fivefingers are thin and thus are unable to absorb the landing. The SI parameters of the Fivefingers are similar to those of barefoot running (Squadrone and Gallozzi, 2009). In this study, the shoe design in each condition was identical, and only the mass of the shoe was modified. We found that the foot strike pattern was RFS in 30% of runners wearing the 175 g shoes. As the mass of the shoes increased to 415 g, some subjects changed to an MFS pattern (55%). However, wearing the heaviest shoes tended to move the center of pressure forward but did not change the ankle angle. Runners who used the FFS may not alter their SI when wearing shoes with differing mass. This is similar to previous literature demonstrating that running with minimalist shoes was similar to habitual running shoes (Squadrone et al., 2015). However, when the shoe mass increased, larger collision forces may have caused runners to adopt a MFS pattern. This supports the hypothesis that greater shoe mass would result in increased vertical ground reaction force (VGRF) and change to MFS running patterns.

A comparison of the 255 g condition and the 415 g condition revealed that the max VGRF did not change during the stance phase. However, past research comparing Vibram Fivefingers with general running shoes found that lighter shoes produce a smaller maximum VGRF (Squadrone and Gallozzi, 2009). The results of this study show that the peak VGRF of the 255 g condition decreased when the sole structure was the same as that of a standard running shoe. Previous research has shown that runners will change their foot strike pattern to FFS when wearing Vibram Fivefingers. The FFS pattern has the benefit of reducing collision forces and may protect against impact-related injuries in the lower extremities (Lieberman et al., 2010). In addition, this study found that the maximum VGRF in the stance phase was significantly increased when the runner was running in the 415 g shoes. The generation of transient impact force increases with larger GRFs and may increase the risk of injury (Shih et al., 2013). Nigg et al (2015) asserted that changes in the VGRF do not have a direct relationship with the occurrence of injuries. This belief is supported by the paradigm stating that allowing runners to select shoes that maintain their optimal movement path (based on their comfort filter) can effectively reduce the risk of injury. Regardless of the point of view, the GRF is an external force that directly flows into proprioception to stimulate the musculoskeletal system (Khassetarash et al., 2015). These forces and the way that a runner attenuates them can affect the human muscle mechanisms and the preferred movement path. The results of the current study demonstrate that heavier shoes lead to greater peak GRF. While it is unclear whether this will lead to injury, it is true that we may change our foot strike pattern to attenuate greater forces.

When running in the 415 g shoes, 61% of runners adopted an MFS pattern. When running, the foot strike pattern may be the main factor that influences the ground impact force. Compared with the RFS landing strategy, the MFS landing strategy has been demonstrated to reduce the loading rate during running (Giandolini et al., 2013). Other studies have noted that while most shod runners use the RFS strategy (De Wit et al., 2000; Hasegawa et al., 2007; Lieberman et al., 2010), the impact force is lower with FFS barefoot running than with RFS barefoot and RFS shod running (Lieberman et al., 2010). This may be due to the lack of cushioning provided by the suspension structure in the sole of the shoe; when runners experience increased impact loads, the FFS strategy is adopted to increase the attenuation of collision forces (Lieberman et al., 2010). Increased impact force may not be directly related to skeletal tissue damage in the lower extremities; however, muscle vibration caused by impact with the ground may cause muscle injuries or discomfort (Nigg, 2010). As suggested by (Nigg et al., 2015), the optimal movement path is unique to individual runners and running conditions. Therefore, this study infers that participants running in different shoe mass may experience changes in muscular demand to maintain a desired movement path. To reduce the discomfort caused by these changes, the runners may adopt the MFS strategy to effectively accommodate the changing task demands.

This study found no difference in ankle plantarflexion at IC in runners wearing shoes of different mass. Previously, it was shown that there is greater plantarflexion, as well as a lower maximum GRF, when running in lighter Vibram Fivefingers than when running in standard shoes (Squadrone and Gallozzi, 2009). Another study that compared minimalist, racing flat and regular shoes also found no difference in the contact angle among the three types of running shoes (Bonacci et al., 2013), which is consistent with the results for the four different shoe masses used in this study. However, the running shoes used in this study had the same material structure but different shoe mass. Previous research on different shoes utilized shoes of not only different mass but also different shoe structures, especially in terms of the sole characteristics. A different shoe structure is considered a major component of the foot strike pattern because less material allows for greater proprioceptive input. Therefore, the influence of different styles of running shoes on the momentary angle of touchdown is likely influenced by amount of cushion and not by the mass of the shoes.

Different foot strike patterns and different running shoes can affect the muscle activation of the lower extremities during running (Nigg and Wakeling, 2001). The results of the current study show that in the pre-activation period, the activation of the TA muscle is affected by the mass of the shoe. When the subjects wore 175 g and 415 g shoes for running, the activation of the TA increased relative to that for the 255 g condition; therefore, light and heavy running shoes used for jogging over a long period of time may cause problems related to TA muscle fatigue. In this study, which used running shoes of the same structure but with different mass, heavier shoes led to higher muscle activation of the semitendinosus, GAS lat and SO muscles during the push-off phase. As a result, runners who habitually adopt RFS may excessively use these highly activated muscles during running in heavy shoes, resulting in muscle fatigue in the lower extremities, which may lead to a higher risk of injury (Shih et al., 2013). However, muscle activation does not infer a direct relationship with the risk of injury (Yong et al., 2014). The current study found light and heavy shoes may cause fatigue in the pre-activation phase. GAS lat and SO muscles may fatigue during the push-off phase. As SI changes, it causes increased muscle activation that may lead to TA, GAS lat and SO muscle fatigue. Therefore, wearing light shoes or heavy shoes may both lead to high levels of muscle fatigue. In FFS runners, the TA demonstrated lower activation with 175 g shoes during the pre-activation phase than with the other shoes of different mass, while TA muscle activation increased as the mass of the shoe increased.

Limitations

The runners were obviously not blinded to the mass of the shoe since they could see the mass attached; this could introduce performance bias. The mass of the shoes could be increased only by using a larger lead weight.

Conclusion

In this study, the shoe design remained consistent while the shoe mass was varied. Heavier shoes increased the peak VGRF force during running, which might increase the risk of lower extremity injuries. The SI with heavier shoes changed to that of the MFS pattern. The MFS pattern may allow for improved attenuation of impact forces throughout the kinematic chain and may be a more comfortable strike pattern for running than the RFS pattern. In this study a greater percentage of runners demonstrated a MFS SI when shoe mass increased, and the percentage of runners using a RFS SI decreased. FFS or MFS landing reduces the ground impact force. To reduce discomfort, it is possible that runners shift to the MFS pattern to mitigate these impact forces.

When runners run in either light or heavy shoes, they increase the activation of their TA, which may cause increased muscle fatigue. When wearing heavier shoes, muscle fatigue may increase more rapidly in the semitendinosus, GAS lat and SO muscles during the push-off phase. When running in light shoes, muscle fatigue may increase more rapidly in the GAS during the braking phase. This study suggests that individuals have unique running strategies, but the mass of the shoes may change the strike pattern and coordination of the running strategy. Both light and heavy shoes may induce muscle fatigue and subsequent muscle damage. Therefore, runners are advised to consider running shoes based on their own muscle strength and to adapt to changes in shoe mass before long periods of exercise to reduce the occurrence of injuries and discomfort.

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Key points

- As shoe mass increases, the peak VGRF also increases.
- The lightest shoes might increase tibialis anterior muscle activity during the pre-activation phase.
- The heaviest shoes could increase semitendinosus, gastrocnemius lateralis and soleus muscle activity during the push-off phase.
- The heaviest shoes could increase gastrocnemius lateralis and soleus muscle activity during the stance phase.

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