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Research article

The Influence of a Bout of Exertion on Novice Barefoot Running Dynamics

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Abstract
Barefoot, forefoot strike (FFS) running has recently risen in popularity. Relative to shod, rear-foot strike (RFS) running, employing a FFS is associated with heightened triceps surae muscle activation and ankle mechanical demand. Novice to this pattern, it is plausible that habitually shod RFS runners exhibit fatigue to the triceps surae when acutely transitioning to barefoot running, thereby limiting their ability to attenuate impact. Therefore, the purpose was to determine how habitually shod RFS runners respond to an exertion bout of barefoot running, operationally defined as a barefoot run 20% of mean daily running distance. Twenty-one RFS runners performed novice barefoot running, before and after exertion. Ankle peak torque, triceps surae EMG median frequency, foot-strike patterns, joint energy absorption, and loading rates were evaluated. Of the 21 runners, 6 maintained a RFS, 10 adopted a mid-foot strike (MFS), and 5 adopted a FFS during novice barefoot running. In-response to exertion, MFS and FFS runners demonstrated reductions in peak torque, median frequency, and ankle energy absorption, and an increase in loading rate. RFS runners demonstrated reductions in peak torque and loading rate. These results indicate that a short bout of running may elicit fatigue to novice barefoot runners, limiting their ability to attenuate impact.

Key words: Fatigue, Footwear, Foot-Strike, Loading Rate, EMG, Torque.

Introduction

Barefoot running has recently risen in popularity, largely due to its association with a forefoot-strike (FFS) contact pattern (Lieberman, 2012). Moving from a rear-foot strike (RFS) – the predominant pattern among shoe runners – to a mid-foot strike (MFS) or FFS, results in a shift in mechanical demand from the knee to the ankle, and presumably, a reduced likelihood of knee joint injury (Bonacci et al., 2013; 2014; Williams et al., 2012). The increase in work by the ankle allows for a slower lowering of the body over the compliant ankle-foot complex and a reduction in the vertical loading rate (Lieberman et al., 2010; Shih et al., 2013; Squadrone and Gallozzi, 2009). Control about the ankle during a mid-foot strike MFS or FFS is likely attributed to passive tension of the Achilles/Tibialis Posterior tendons and/or muscle action of the triceps surae (gastrocnemii and soleus) (Ahn et al., 2014).

A recent investigation of 1065 runners indicates novice barefoot runners are more likely to maintain a RFS rather than adopt a MFS/FFS (Nunns et al., 2013). Yet, similar to their FFS counterparts, novice barefoot RFS runners contact the ground in a more plantar-flexed foot posture relative to traditional shod running (Williams et al., 2012). This presumably increases reliance upon the triceps surae to help mitigate ground reaction forces, albeit to a lesser extent than when employing a MFS/FFS. This claim is supported by recent findings of Williams et al. (2012) whom instructed novice barefoot runners to RFS and reported an increase in ankle mechanical demand relative to when they ran shod.(Williams et al., 2012)

High-level exertion elicits muscle fatigue, which is characterized by a failure to generate the required force/power output for a given task (Fitts, 1994; Phinyoomark et al., 2012). As measured during maximal isometric contractions, in response to fatigue, muscles exhibit a reduction in peak tetanic tension and prolonged fiber contraction and relaxation times (Bigland-Ritchie and Woods, 1984). This leads to lower rates at which muscle fibers shorten and develop tension (Fitts, 1994) and is correlated with reduced motor unit firing frequency (as measured by an EMG median frequency analysis) (Stulen and DeLuca, 1981). Consequently, the ability for lower-extremity muscles to attenuate forces diminishes following fatigue (Milgrom et al., 2007). This possibly explains reports of altered ground reaction force profiles (Zadpoor and Nikooyan, 2012) and reduced shock attenuation (Mercer et al., 2003) among fatigued runners.

The aforementioned findings indicate to an increased potential for injury among fatigued runners, and a heightened demand to the triceps surae during barefoot running, regardless of preferred foot-strike pattern. The findings of Morio et al. (2012) – whom reported an increase in lower limb stiffness among novice barefoot runners following an exhaustive stretch shortening cycle exercise to the plantar-flexors – further indicate to a heightened potential for injury in response to a controlled fatiguing task of the triceps surae (Morio et al., 2012). Similarly, Paquette et al. (2016) suggested an increase potential for repetitive load injury among non-RFS runners due to their findings of reduced foot contact angle variability following 40 minutes of continuous running (Paquette et al., 2016).

Habitually shod RFS runners who employ a MFS or FFS may over-exert their triceps surae during novice barefoot running, inducing localized muscle fatigue. Theoretically, this would result in a reduced capability to slowly lower the heel (e.g. under eccentric control), and thereby attenuate impact, possibly leading to injury and/or a reversion in running kinematics. This is deemed deleterious as it exposes the calcaneus to high frequency collisions without a shoe sole to attenuate the impact (Lieberman, 2010; 2012). Despite this conjecture, no study to date has examined the potential for triceps surae

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fatigue among novice barefoot runners, the influence this has on their running dynamics, and how this may differ across foot-strike patterns.

A previous investigation conducted by the authors demonstrated variability in lower-limb dynamics among habitually shod RFS runners who perform an acute (i.e. within day) transition to barefoot running (Hashish et al., 2015). Expanding upon these findings, the purpose of the present investigation was to determine how these runners respond to an exertion bout of novice barefoot running performed at their self-selected running speed. We hypothesized that novice barefoot runners would exhibit fatigue to the triceps surae in response to exertion, resulting in altered lower-extremity movement patterns and a reduced ability to attenuate impact.

Methods

Participants

An *a priori* power analysis was conducted using data from pilot work for this study. Using the variable with the highest standard deviation (loading rate), it was revealed that 5 subjects were required to adequately power this study (effect size = 1.38, α =0.05, β=0.20). To account for variability in preferred foot-strike patterns, 21 (9 male, 12 female) recreational runners, who ran between 15-40 $km \cdot wk^{-1}$, participated. Their mean age, height, weight and weekly running distance were 26.6±4.4 years, 1.70 ± 0.13 m, 65.3 ± 11.9 kg, and 22.6 ± 6.5 $km \cdot wk^{-1}$, respectively.

A center of pressure analysis was conducted to determine the shod foot-strike index for the respective participants. To qualify, each runner was required to be a habitually shod RFS runner; (Cavanagh and Lafortune, 1980) between the age of 19-40 years; free of injury for the past 6 months; and have no previous experience in barefoot running, or recreational barefoot activities that require running (e.g. gymnastics, beach volleyball).

Ground reaction forces (and therefore, calculated lower-extremity dynamics) during running are influenced by shoe hardness and cushioning properties (Bonacci et al., 2013; Clarke et al., 1983). Accordingly, repeated use of softer, minimalist footwear may result in accommodative running patterns (Lieberman, 2010; 2012). Thus, participants were also excluded if they had any experience with minimalist shoe running, operationally defined as a shoe with a heel-to-toe drop less than 8mm, and/or marketed by the respective manufacturer as a “barefoot shoe” or “minimalist shoe.”

This study was approved by the Institutional Review Board for the University of Southern California (USC) Health Sciences Campus, and informed consent was obtained from all participants.

Instrumentation

All testing was conducted at the USC Musculoskeletal Biomechanics Research Laboratory. Three-dimensional motion analysis data were collected using an 11-camera Qualisys system (Gothenburg, Sweden) at 250 Hz. Ground reaction forces were collected at 3000 Hz using a ground-embedded force platform (AMTI, Newton, MA). Muscle activation was collected at 3000 Hz using the Noraxon Desktop DTS (Noraxon USA, Inc., Scottsdale, AZ). The EMG signal was telemetered to a receiver that contained a differential amplifier with an input impedance of > 100 M ohm and a Common Mode Rejection Ration of > 100 dB. An amplifier gain of 500 was used, and the signal to noise ration was < 1 μV RMS of the baseline. Isometric ankle torque was collected on a computer-based dynamometer (Norm Humac Cybex, CSMi Inc., Stoughton, MA). This signal was sampled at 3000Hz on a 16-bit external analog-to-digital board. The ground reaction force, EMG, and dynamometer signals were all synced to, and analyzed in real-time with the Qualisys system.

Protocol

Participants were instrumented with bipolar EMG electrodes on the lateral gastrocnemius, medial gastrocnemius, and soleus of their dominant limb, operationally defined as the leg with which they kick a ball. Electrodes were placed according to SENIAM standards. Integrity of the EMG signal was confirmed through visual analysis of the digital signal. Subsequently, participants performed dynamometer-based maximal voluntary isometric plantarflexion contractions. Predicated upon the length-tension relationship of muscle, peak isometric force production can be achieved when a muscle is in neither in an excessively lengthened or shortened position (Brughelli and Cronin, 2007). Thus, in order to place the triceps surae in a mechanically advantageous position, testing and evaluation of the EMG signal was conducted with the ankle in 90^o (Brughelli and Cronin, 2007; Jones et al., 1997). To limit the influence of proximal muscle groups (e.g. quadriceps) on force ankle torque production, the knee was placed in 90^o of flexion. A goniometer was used to ensure proper positioning.

Participants were familiarized with the dynamometer and provided adequate practice trials to achieve reliable measurements; all trials were performed barefoot. The barefoot was tightly secured to the footplate with Velcro straps to avoid movement artifacts. Each subject performed three, three-second maximum effort trials with online visual feedback on the computer screen and vocal cueing, with the instructions being “push as fast and as hard as possible.” Trials with an initial countermovement, defined as a reduction in baseline torque just before the plantar flexor push, were disallowed. The system was preloaded at each ankle position with 10-14 Newton-meters torque, in order to eliminate mechanical delay in the testing set up. A similar method was previously been described for the knee joint (Aagaard et al., 2002).

Following dynamometry testing, skin-mounted markers and tracking clusters were affixed to the pelvis, thigh, shank, and foot segments bilaterally (Figure 1). The foot was tracked with a dorsal plate according to previously established methodology (Hashish et al., 2014). The participants were then instructed to “warm-up” for 3-minutes with a series of barefoot jogging and jumping jacks. The warm-up period helped prepare the participants for physical activity, and to become acclimated with the testing procedure and confines of the laboratory. Each subject then completed 6 successful over-ground barefoot running trials in the laboratory. Timing gates were used to determine running velocity. In order to better understand the natural response in running mechanics (and thereby, help make the findings more clinically relevant), all trials were conducted at the participant's self-selected speed; this was defined as the (self-reported) speed at which they would perform a maximum effort 1.6km run. All successful running trial velocities were within a range of 5% of one another. A successful trial was operationally defined as a running trial in which the stance phase of the dominant limb was entirely on the force plate and running speed was within the prescribed range. The laboratory path was approximately 30 feet in length. The force plate was located approximately midway down this path.

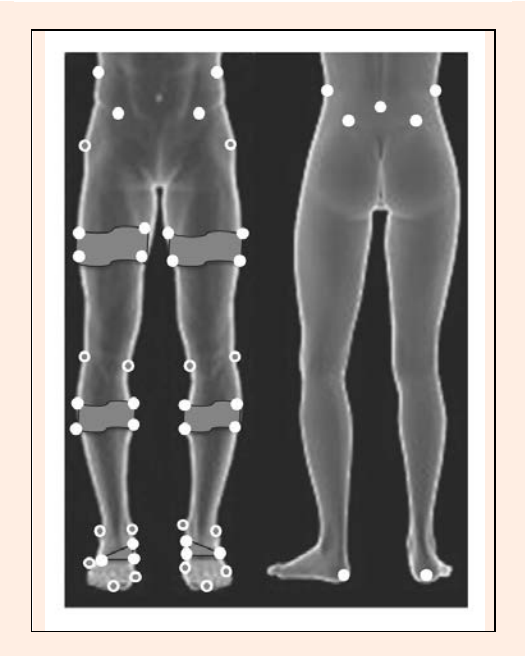

Figure 1. Rendering of lower-extremity marker placement. Static (gray-filled) markers were removed after the standing calibration. Dynamic (white-filled) markers were kept on during the running trials. Static markers were placed on the L5/S1 joint space, and the bilateral iliac crests, anterior and posterior superior iliac spines, greater trochanters, lateral and medial knee joint spaces, lateral and medial malleoli, distal second (foot) phalanges, base of the 1st and 5th metatarsals, and calcanei.

Immediately following the running trials, participants performed an exertion bout of running on an established, outdoor concrete surface, running course. The exertion bout was operationally defined as a barefoot run at 20% of the distance of the respective participant's mean daily shod run. The exertion bout was conducted at the participant's self-selected speed (Table 1).

Following this bout of running, the participants completed another 6 successful trials of laboratory barefoot running at the pre-bout matched average running speed. The participants then repeated the dynamometer testing. The time between the respective running trials and dynamometry testing, as well as the exertion run, was less than 5 minutes. Tracking clusters and EMG electrodes were kept on the participants during the duration of the testing (Figure 1).

Table 1. Running volumes (km), velocity ($m \cdot s^{-1}$) and durations (minutes), delineated by novice barefoot foot-strike pattern. Data are means (±SD).

	RFS (n =6)	MFS (n=10)	FFS (n=5)
Weekly run volume	23.9 (3.7)	24.6 (4.8)	22.9 (5.5)
Exertion run velocity	2.7 (0.4)	2.7 (0.7)	2.7 (0.3)
Exertion run volume	1.3 (0.2)	1.3 (0.3)	1.3 (0.2)
Exertion run duration	8.1 (1.0)	8.1 (2.7)	8.3 (1.0)

RFS= rear-foot strike ; MFS = mid-foot strike; FFS = forefoot-strike.

Data analysis

Three-dimensional marker coordinates during the laboratory running trials were reconstructed using Qualisys Track Manager Software (Qualisys, Gothenburg, Sweden). Hip joint centers were calculated according to the method proposed by Bell et al. (1989). The remaining lower-extremity joint centers were calculated as the three-dimensional midpoint between adjacent segments. Visual 3D (C-motion, Rockville, MD) software was used to process the raw coordinate data and compute segmental kinematics and kinetics for the dominant lower-extremity. Trajectory data were filtered with a fourth-order zero lag Butterworth 12 Hz low-pass filter (Ford et al., 2007). The local coordinate systems of the pelvis, thigh, shank, and foot were derived from the standing calibration trial. Joint kinematics were calculated using the joint coordinate system approach with a six-degree of freedom model (Grood and Suntay, 1983). Running trials were eliminated during post-processing if there was significant marker dropout that degraded the model; there were at least 4 successful trials used from the pre- and post-exertion running bouts for each participant. Internal net joint moments of these successful trials were calculated using inverse dynamics equations and were normalized to body mass. Initial contact was defined as the first instance when the vertical ground reaction force exceeded 20 Newtons (Hashish et al., 2014). The loading phase of running was operationally defined as initial contact to 13% of stance (Willy et al., 2008). The absorption phase was operationally defined as initial contact to peak knee flexion (Heiderscheit et al., 2011). Processing of the EMG and dynamometer recordings were conducted using a custom program written in MATLAB (The Mathworks, Natick, MA). All outcome variables were averaged over the successful trials.

Outcome variables

Contact Patterns: The length of the foot was determined from the standing calibration trial during both the shod and barefoot data collections, and was tracked during movement using a dorsal triad (Hashish et al., 2014). Foot-strike patterns were categorized into a RFS, MFS, or FFS, according to the strike index method (Cavanagh and Lafortune, 1980). As indicated by center of pressure analysis, initial contact made with the rear one-third of the foot was categorized as a RFS, initial contact made with the middle third was categorized as a MFS, and initial contact made with the anterior third was categorized as a FFS. Sagittal ankle initial contact angles were also determined from motion analysis.

Ankle peak torque: The dynamometer signals were low passed filtered at 12Hz with a fourth-order, zero-lag

Butterworth filter (Aagaard et al., 2002). Thereafter, the voltage signal was converted to torque (measured in Newton-meters). The highest peak torque among the three repetitions was ascertained and was used for analysis.

Median frequency: The EMG recordings from the dynamometry trials were assessed. The signals were initially band pass filtered at 30 and 450 Hz. Subsequently, the peak amplitude for each respective muscle was ascertained, and a one-second interval around this peak (0.5 seconds before and after) was extracted. This extracted signal was then subjected to a Fast Fourier Transformation, such that that $f(x) = \int_{-\infty}^{\infty} \hat{f}(\xi)e^{2\pi\xi ix}\, d\xi$, where x represents time and ξ represents frequency in hertz (Hz). We reported the median frequency for the soleus and gastrocnemii. The gastrocnemii was a measure of the average median frequency of the lateral gastrocnemius and medial gastrocnemius.

Energy absorption: Internal net joint moments were calculated using inverse dynamics equations and were normalized to body mass. Net joint powers were calculated as the product of the angular velocity and moment for each joint (Heiderscheit et al., 2011). Energy absorption was determined by integrating the negative net joint power for the ankle and knee, respectively, during the absorption phase.

Loading rate: The loading rate was determined by taking the peak, positive derivative of the vertical ground reaction force during the loading phase (De Wit et al., 2000).

Statistical analysis

Runners were grouped according to their novice (i.e. pre-exertion) barefoot foot-strike pattern (i.e. RFS, MFS, or FFS), as determined from the average location of their center of pressure across trials. Shapiro-Wilk test was used to confirm that the data were normally distributed and studentized residuals were used to confirm the absence of outliers. A 3x2 mixed model ANOVA (strike-pattern [3] x exertion [2]) was used to analyze each outcome variable. In the event of a significant F ratio, paired sample t-tests were used to evaluate differences between pre- and post-exertion barefoot running ($p \leq 0.05$). When data violated Mauchly's test of sphericity, The Greenhouse-Geisser correction was used. Cohen's d_z effect size (d) and confidence intervals of the difference (CI) are reported for significant pairwise comparisons. All statistical calculations, with the exception of effect sizes, were conducted using PASW Version 18.0 (IBM Corporation; New York, USA). Effect sizes were determined using G*Power Version 3.1.9.2 (University of Dusseldorf; Dusseldorf, Germany), where Cohen's $d_z = M_{diff} / \sqrt{\Sigma(X_{diff} - M_{diff})^2 / N\text{-}1}$.

Results

There was no difference in running velocity across groups, or between conditions (Table 2).

Of the 21 RFS shod runners, 5 adopted a FFS, 10 adopted a MFS, and 6 maintained a RFS during novice barefoot running. Following the exertion protocol, all 5 FFS runners impacted the ground with a MFS. Of the 10 MFS runners, 8 maintained a MFS, 1 adopted a FFS, and 1 reverted to a RFS. Of the 6 RFS runners, 5 maintained a RFS and 1 adopted a FFS.

The main effect of foot-strike pattern on sagittal ankle initial contact angle was significant, $F(2, 36) = 14.631$, $p < 0.001$, and there were significant differences between FFS and RFS runners ($p = 0.001$), as well as MFS and RFS runners ($p < 0.001$). The main effect of exertion was also found to be significant, $F(1, 18) = 14.038$, $p = 0.001$, as FFS and MFS runners demonstrated significant reductions in plantar flexion following exertion (FFS: $\Delta = -6.5 \pm 3.6^{o}$; $p = 0.017$; $d = 0.72$; CI = -3.78 – 6.45 | MFS: -4.3 ± 5.7^{o}; $p = 0.042$; $d = 0.84$; CI = -3.54 – 7.42).

Peak ankle torque differed among foot-strike patterns, $F(2, 36) = 5.728$, $p = 0.012$, as there were significant differences between FFS and RFS runners ($p = 0.034$), as well as MFS and RFS runners ($p = 0.019$). The main effect of exertion on peak torque was also significant, $F(1, 18) = 118.755$, $p < 0.001$, as each group of novice barefoot runners demonstrated a significant reduction in peak torque following exertion (Table 2).

There was a significant interaction between strike-pattern and exertion for soleus median frequency, $F(2, 18) = 8.553$, $p = 0.002$. Soleus median frequency did not differ across strike-patterns, $F(2, 36) = 1.483$, $p = 0.051$, but did differ in response to exertion, $F(1, 18) = 13.007$, $p = 0.002$. FFS and MFS runners demonstrated significant reductions in soleus median frequency following exertion (FFS: $\Delta = -14.6 \pm 7.6$ units; $p = 0.013$; $d = 5.05$; CI = 11.58 – 19.00 | MFS: -6.7 ± 4.0 units; $p < 0.001$; $d = 0.74$; CI = 0.55 – 22.12) (Figure 2).

There was a significant interaction between strike-pattern and exertion for gastrocnemii median frequency, $F(2, 18) = 8.720$, $p=0.002$ (Figure 2). However, the main effect of strike-pattern was not significant, $F(2, 36) = .213$, $p=0.727$, nor was the main effect of exertion on gastrocnemii median frequency, $F(1, 18) = 1.385$, $p = 0.206$.

Table 2. Running velocities and peak ankle plantar-flexor torques for novice barefoot runners, pre- and post-exertion. Data are means (±SD).

	RFS		MFS		FFS	
	Pre-Exertion	Post-Exertion	Pre-Exertion	Post-Exertion	Pre- Exertion	Post-Exertion
Velocity ($m{\cdot}s^{-1}$)	3.5 (.5)	3.6 (.5)	3.6 (.4)	3.6 (.4)	3.7 (.4)	3.7 (.4)
Peak Torque (Nm)	106 (32)	93 (34)*	89 (16)	78 (13)*	90 (23)	72 (20)*
p-value	.001		<0.001		< 0.001	
Cohen's d Effect Size	4.93		2.53		10.54	
Confidence Interval	10.40 – 15.96		9.53 – 17.17		15.45-19.56	

RFS= rear-foot strike ; MFS = mid-foot strike; FFS = forefoot-strike. There were significant differences between RFS runners and both, MFS and FFS runners, for peak ankle plantar-flexor torque. P-values, effect sizes, and confidence intervals are presented for significant within-group differences.

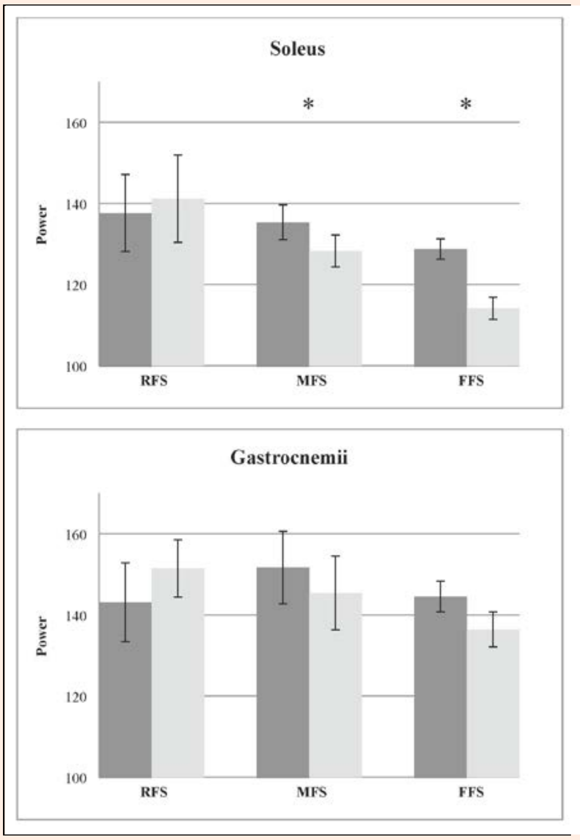

Figure 2. Mean and standard error, soleus (top) and gastrocnemii (bottom) median frequency for novice barefoot runners, pre- (dark gray) and post-exertion (light gray), delineated by foot-strike group. RFS= rear-foot strike ; MFS = mid-foot strike; FFS = forefoot-strike. * indicates a significant ($p < 0.05$) within group difference.

The main effect of foot-strike pattern on ankle energy absorption was significant, $F(2, 36) = 8.690$, $p = 0.002$ as there were significant differences between FFS and RFS runners ($p = 0.005$), as well as MFS and RFS runners ($p = 0.007$). The main effect of exertion on ankle energy absorption was also significant, $F(1, 18) = 14.038$, $p = 0.001$, with both FFS and MFS runners demonstrating significant reductions in ankle energy absorption following exertion (FFS: $\Delta = -1.6 \pm 0.77$ J/kg; p= 0.010; d = 1.97; CI = 0.55 – 2.30 | MFS: Δ= -0.9 ± 1.2 J/kg; p = 0.050; d = 0.69; CI = 0.00 – 1.87). There were no significant differences in knee energy absorption (Figure 3).

There was a significant interaction between strike-pattern x exertion for loading rate, $F(2, 18) = 8.112$, $p = 0.003$. Loading rate did not differ across foot-strike groups, $F(2, 36) = 3.328$, $p = 0.61$, but did differ in response to exertion, $F(1, 18) = 4.476$, $p = 0.046$. FFS and MFS runners demonstrated significant increases in loading rate following exertion (FFS: Δ= 19.0 ± 14.9 $BW \cdot s^{-1}$; p = 0.047; d = 1.38; CI = -25.10 – -1.27 | MFS: Δ= 10.3 ± 13.7 $BW \cdot s^{-1}$; p = 0.040; d = 0.78; CI = -29.21 – -1.34), whereas RFS runners demonstrated a significant reduction in loading rate (Δ= -10.6 ± 8.8 $BW \cdot s^{-1}$; p = 0.032; d = 0.62; CI = -7.19 – 25.09) (Figure 4).

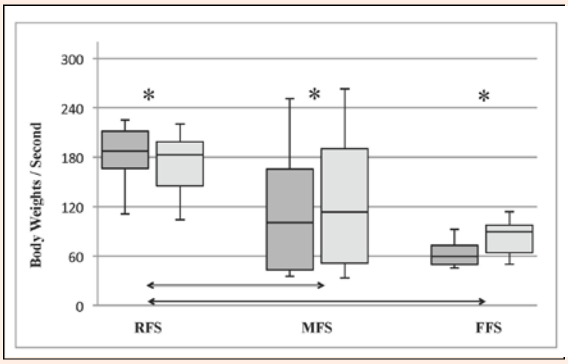

Figure 4. Loading rates for novice barefoot runners, pre- (dark gray) and post-exertion (light gray), delineated by foot-strike group. RFS= rear-foot strike ; MFS = mid-foot strike; FFS = forefoot-strike. Boxes, median, first and third quartiles; whiskers, minima and maxima. (↔) Indicates a significant between group difference. (*) Indicates a significant within group difference.

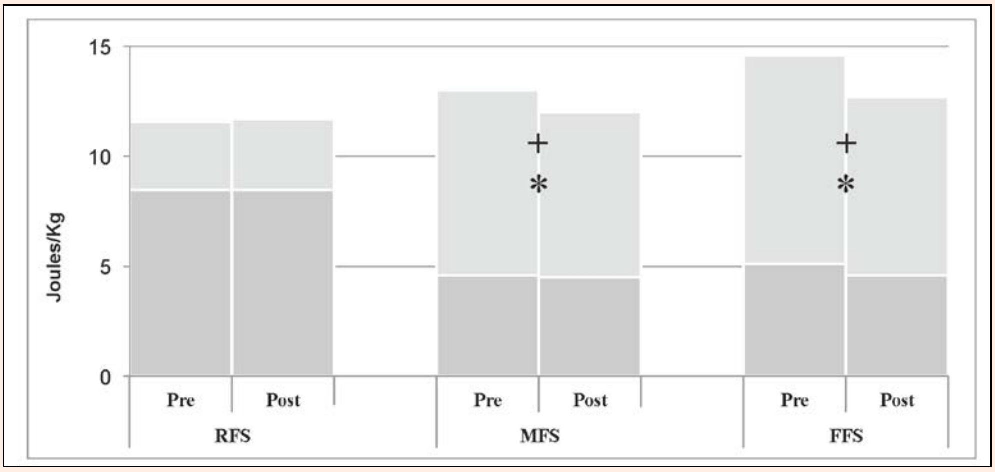

Figure 3. Mean knee (dark gray) and ankle (light gray) energy absorption for novice barefoot runners, pre- and post-exertion, delineated by foot-strike group. RFS= rear-foot strike ; MFS = mid-foot strike; FFS = forefoot-strike. (*) Indicates a significant within group difference. (+) Indicates a significant difference between RFS runners.

Discussion

We sought to assess the influence of a bout of exertion on novice barefoot running patterns. The bout's magnitude was predicated upon a previous investigation which examined the transition from shod to barefoot running, and instructed participants to incrementally increase barefoot running volume (as a percentage of total) by up to 20% per week and no more than 1.6 km during the first week (Lieberman et al., 2010). Based upon this prescription, the 20% volume of running was chosen for the exertion bout and no runner exceeded 2.1 kilometers for the exertion run. In accordance with previous investigations (Hamill et al., 2011; Lieberman et al., 2010; Nunns et al., 2013; Willy and Davis, 2014) novice barefoot runners demonstrated variability in foot-strike patterns. Likely as a result of the variable loading patterns, there were significant differences in how these runners responded to the exertion protocol.

Forefoot strikers

In agreement with our hypothesis, the short bout of running was sufficient to elicit a fatigue response in novice barefoot FFS runners. As a "gold standard" indicator of muscle fatigue, (Phinyoomark et al., 2012) the significant reduction in soleus median frequency (Figure 2) indicates fatigue of the soleus among these runners. This is further corroborated by the significant reduction in ankle peak torque following exertion, and the fact that all 5 FFS runners reverted to mid-foot striking following the exertion protocol.

Following exertion, FFS runners demonstrated a significant reduction in ankle energy absorption (Figure 3). This can be attributed to the posterior shift in the center of pressure at foot contact. A more dorsiflexed foot likely offloads the (fatigued) soleus and increases demand to the gastrocnemii (Cresswell et al., 1995; Hebert-Losier et al., 2012).

The reversion from forefoot, to mid-foot (and particularly, rear-foot) striking, limits the capability of the triceps surae to eccentrically control ankle motion, and presumably, to absorb demand. It appears that this reduced mechanical advantage limits the ability to attenuate lower-extremity forces as indicated by the significant reductions in ankle energy absorption and loading rate among these runners (Figures 3 and 4). Retrospective examinations that have associated accumulated high loading rate events to tibial stress fractures (Milner et al., 2006; Zadpoor and Nikooyan, 2011) support the claim that this change is maladaptive.

Mid-foot strikers

Similar to their FFS counterparts, the exertion bout induced fatigue to novice barefoot MFS runners. This is evidenced by the significant reductions in soleus median frequency and ankle peak torque. Nonetheless, these individuals predominantly continued to MFS following exertion.

Upon initial contact during MFS running, there is a posterior and medial migration of the center of pressure as the ankle dorsiflexes (Cavanagh and Lafortune, 1980). Simultaneously, the center of mass moves forward, resulting in a vertical ground reaction force vector that remains anterior to the ankle throughout the absorption phase. This results in high mechanical demand to the ankle's posterior muscles and tendons. Unaccustomed to this loading pattern, it is plausible that novice MFS runners were attempting to limit the demand to the fatigued soleus and guard against ankle negative work following the exertion bout. The now reduced capability of the posterior ankle's muscles and tendons to absorb demand likely explains the significant increase in loading rate among these runners following exertion.

Rear-foot strikers

Novice barefoot RFS runners predominantly continued to RFS following exertion, and expectedly, demonstrated no changes in joint energy absorption. However, following exertion, RFS runners did demonstrate a significant reduction ($p = 0.032$; $d = 0.62$; CI = -7.19 – 25.09) in loading rate (Fig. 4). These findings are in contrast with those of Willy and Davis (2014), who reported an increase in loading rate among novice minimalist shoe RFS runners following 10-minutes of running (Willy and Davis, 2014). The discrepancy in the findings may be attributed to the fact that the runners in our study were entirely barefoot.

Unable to be reliant on a shoe sole, it is plausible that the painful collision with the ground obliges the barefoot runner to adopt a softer landing pattern. The seemingly beneficial changes gleaned from exertion aside, calculated loading rates were more than double that of rear-foot striking with a minimal (85.4 ± 24.6 body weights per second), (Willy and Davis, 2014) or traditional shoe (69.7 ± 28.7 body weights per second)(Lieberman et al., 2010). It is important to note that although RFS running has recently drawn the ire of researchers and clinicians alike, a recent investigation conducted by Valenzuela et al. (2015) reports higher ground reaction forces among shod RFS runners who were instructed to utilize a FFS when shod (Valenzuela et al., 2015). The aggregate findings indicate that independently running barefoot, or shifting from a RFS to a MFS/FFS, may be insufficient to lower ground reaction forces. Accordingly, we encourage further research to investigate the contributing patterns to high ground reaction forces, as well as the influence of such patterns on injury risk.

Limitations

There are important methodological considerations associated with this study. Although our overall sample size ($n = 21$) was larger than previous investigations, the group sizes were relatively small, particularly the FFS group. Whilst we could have instructed participants to utilize each respective strike-pattern, the purpose of this investigation was to examine the "natural," response to a bout of barefoot running. Furthermore, without a control group, we cannot definitively conclude whether the stated changes were attributed to the exertion protocol.

Conclusion

The present investigation indicates variability in foot-strike patterns and stance phase movement dynamics for habitually shod runners after performing an exertion bout of novice barefoot running. Those who initially adopt a MFS or FFS are largely unable to sustain this pattern attributed to fatigue of the soleus; this in turns limits their ability to attenuate impact. Accordingly, if transitioning to barefoot running is desired, instruction promoting forefoot strikes is suggested. Initial negative work exercises are also recommended to prepare the ankle and foot for the shift in mechanical demand. This may help in preventing fatigue of the calf and foot, and the associated maladaptive contact patterns.

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Key points

- In response to exertion, novice barefoot runners demonstrate fatigue to their soleus.
- In response to exertion, novice barefoot runners demonstrate a reduction in ankle energy absorption
- In response to exertion, novice barefoot runners demonstrate an increase in loading rate

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