The influence of foot-strike technique on the neuromechanical function of the foot

Dr Luke A Kelly¹, Dr Dominic J Farris¹, A/Professor Glen A Lichtwark¹ & Professor Andrew G Cresswell¹

¹ - School of Human Movement and Nutrition Sciences, The University of Queensland, Australia

Corresponding Author:
Dr Luke Kelly
School of Human Movement and Nutrition Sciences
The University of Queensland
Blair Drive
St Lucia
QLD, Australia
Abstract

Purpose: The aim of this study was to investigate the influence of foot-strike technique on longitudinal arch mechanics and intrinsic foot muscle function during running.

Methods: 13 healthy participants ran barefoot on a force-instrumented treadmill at 2.8 ms\(^{-1}\) with a forefoot (FFS) and rear-foot (RFS, habitual) running technique, while kinetic, kinematic and electromyographic (EMG) data from the intrinsic foot muscles were collected simultaneously. The longitudinal arch was modeled as a single “mid-foot” joint representing motion of the rear-foot (calcaneus) relative to the forefoot (metatarsals). An inverse dynamic analysis was performed to estimate joint moments generated about the mid-foot, as well as mechanical work and power.

Results: The mid-foot was more plantar flexed (higher arch) at foot contact when running with a forefoot running technique (RFS 0.2 ± 1.8° v FFS 6.9 ± 3.0°, ES = 2.7), however there was no difference in peak mid-foot dorsiflexion in stance (RFS -11.6 ± 3.0° v FFS -11.4 ± 3.4°, ES = 0.63). When running with a forefoot technique, participants generated greater moments about the mid-foot (27% increase, ES = 1.1) and performed more negative work (240% increase, ES = 2.2) and positive work (42% increase, ES = 1.1) about the mid-foot. Stance phase muscle activation was greater for Flexor Digitorum Brevis (20% increase, ES = 0.56) and Abductor Hallucis (17% increase, ES = 0.63) when running with a forefoot technique.

Conclusion: Forefoot running increases loading about the longitudinal arch and also increases the mechanical work performed by the intrinsic foot muscles. These findings have substantial implications in terms of injury prevention and management for runners who transition from a rear-foot to a forefoot running technique.

Keywords

Intrinsic foot muscles, longitudinal arch, forefoot running, plantar fascia, foot biomechanics
Introduction

The orientation of a runner’s foot at the moment of foot-ground impact is often employed as a method to classify running technique (2, 7, 29, 33). Foot-strike patterns are generally classified in three groups; (1) runners who land in a heel-first manner (rear-foot strike), (2) runners who land on the forefoot (forefoot strike), and (3) runners who land with a simultaneous heel and forefoot contact (mid-foot strike) (3, 7). Although the majority (>75%) of distance runners adopt a rear-foot running technique (14, 25), it has been suggested that a forefoot running technique is mechanically advantageous, as it affords the runner greater opportunity to recycle the energy associated with foot-ground impact via elastic stretch and recoil of the tendons and ligaments of the ankle and foot (28, 29, 33). Consequently, the popularity of running “re-training” programs that promote a forefoot landing pattern to reduce injury risk and improve running performance have surged (3, 10, 15).

The longitudinal arch (LA) of the human foot behaves in a spring-like manner when running, temporarily storing and then subsequently returning a considerable portion of the mechanical energy required for each stride (22, 36, 39). Compression (lowering and lengthening) of the LA during the first half of stance phase allows mechanical energy to be temporarily stored within the stretched ligaments, muscles and tendons that span this structure (20, 22, 39). This energy is subsequently returned in late stance, as the resultant ground reaction force (GRF) declines and the stretched elastic structures shorten to allow the LA to recoil (rise and shorten) (20, 22, 33).

Recently studies have explored the hypothesis that a forefoot running technique enhances foot-spring function by allowing a greater proportion of mechanical energy to be recycled during
each stride (30, 33, 39). Consistent with the hypothesis of enhanced foot-spring function, these studies have all reported an increase in LA compliance with a forefoot technique, compared to a rear-foot technique, when running at matched velocities. However, the increased compliance appears to be due to a higher LA at foot contact, rather than an increase in peak LA compression in mid-stance (30, 33, 39). A key finding in each of these studies was that peak LA compression and model derived plantar aponeurosis strains were similar, regardless of foot-strike technique (30, 33, 39). Given that peak strain primarily determines the magnitude of stored elastic energy, this finding suggests no energetic benefits to a forefoot running technique. The studies by McDonald et al. (2016) and Wager & Challis (2016) provide valuable insight into the biomechanics of the LA when running with rear-foot and forefoot running techniques. However, these studies have a methodological limitation in their assumption that the foot is a passive structure with no active muscular control of its biomechanical function.

Forefoot running is characterized by the centre of pressure (COP) being located further anterior in the foot and higher peak GRF (3, 6, 7, 31). Thus in a similar manner to the ankle joint (35), an increase in the length of the GRF moment arm, combined with the higher peak GRF, will likely increase loading of the LA when running with a forefoot technique. Given that the intrinsic foot muscles have similar anatomical pathways to the plantar aponeurosis (24, 27, 38) and have the capacity to alter the stiffness of the LA (18, 20) it is highly likely that these muscles also contribute to the observed alterations in LA mechanics between foot-strike techniques. An increase in activation of the intrinsic foot muscles when running with a forefoot technique could potentially reduce excessive LA compression and reduce plantar aponeurosis strain, providing an explanation for the lack of difference in plantar aponeurosis strain observed in previous studies (30, 39). Furthermore, these muscles are also known to display preparatory activation prior to foot contact (20, 21), thus it is conceivable that increased activation of these muscles in
late swing may also explain the increased LA height at foot-contact when running with a forefoot technique.

It is currently unknown if the intrinsic foot muscles contribute to the regulation of LA motion when switching from a rear-foot to a forefoot running technique. It is also unknown if loading of the LA changes when converting to forefoot running technique. Therefore, the aim of this study was to test the hypothesis that a forefoot running technique results in higher magnitudes of activation in the intrinsic foot muscles during the stance and swing phases while running, due to the higher mechanical loading occurring about the LA.

**Methods**

**Participants**

Thirteen healthy participants (five females and eight males, age 24 ± 6 years; height: 172 ± 8 cm; mass: 72 ± 12 kg) with no diagnosed history of lower limb injury in the previous six months or known neurological impairment volunteered to participate in the study. All participants were habitually shod recreational runners, who classified themselves as non-forefoot strikers. Written informed consent was obtained from each subject. The study protocol was approved by the institutional human research ethics committee and conducted in accordance with the Declaration of Helsinki.

**Experimental Protocol**

Following a 3-min warm up period and familiarization procedure, participants ran barefoot on a force-instrumented treadmill (AMTI, force-sensing tandem treadmill, Watertown, MA, USA) at 2.8 m.s\(^{-1}\) with their habitual foot-strike (HFS) pattern and a forefoot strike pattern (FFS), in a counter-balanced order. During the familiarization session, foot-strike pattern was confirmed.
by the investigator, using visual analysis. None of the participants displayed a habitual forefoot running technique. Prior to commencement of each experimental condition, participants were given as much time as necessary to familiarize themselves with running on the treadmill with a forefoot strike pattern. Participants were considered familiarized with the forefoot running technique once they were able to perform 20 seconds of consecutive forefoot strikes at the experimental speed and reported to be comfortable running with this technique. Foot-strike pattern during the familiarisation period was confirmed visually by the investigator (LK).

Kinetic, kinematic and electromyographic (EMG) data were collected simultaneously for approximately 15-20 strides (toe-off to ipsilateral toe-off) across a 15 s data collection period for each running condition (HFS and FFS).

Data Acquisition

Kinematic and kinetic measurements

Motion data was captured at 200 Hz using an eight-camera 3D optoelectronic motion capture system (Oqus, Qualysis, Gothenburg, Sweden). GRF and EMG data were synchronously captured with the motion data at 4000 Hz using a 14-bit analogue to digital converter using and the Qualisys Track Management software (Qualisys, Gothenburg, Sweden).

Three-dimensional (3D) motion of the foot and shank was modelled from motion capture data during each running trial. In order to quantify motions of the foot segment and the LA (21, 26) retro-reflective markers (9.0 mm diameter) were secured on the skin of the right foot overlying the medial and lateral malleoli, posterior calcaneus, navicular tuberosity and head of the first and fifth metatarsals. Additional markers were applied to the medial and lateral femoral condyles and a rigid cluster of four markers was placed on the antero-lateral aspect of the shank. During a standing calibration trial, markers located on the segment endpoints were used to
generate a two-segment model of the shank and foot. A virtual marker (med-knee) corresponding to the position of the medial marker was also generated in order to reduce the influence of soft tissue artefact on skin-mounted markers located at the knee. Following the calibration trial, the medial and lateral knee markers were removed and the motion of the shank and med-knee virtual marker was tracked using the rigid marker cluster located on the shank. Markers were adhered with double sided adhesive and further secured with cohesive bandage, allowing secure positioning for both running conditions. Markers were not removed between experimental trials.

Electromyography

Identification of the abductor hallucis (AH) and flexor digitorum brevis (FDB) muscles in the right foot of each subject was conducted using real-time B-mode ultrasound imaging (10 MHz linear array, Ultrasonix RP, USA). Subsequently, bi-polar fine-wire electrodes (0.051 mm stainless steel, Teflon coated wires, Chalgren, USA) with a detection length of 4 mm and inter-electrode distance of 4 mm were inserted using delivery needles (0.5 mm x 50 mm) into the muscle tissue of AH and FDB under ultrasound guidance, in accordance with previously described methods (19). Sterile techniques were used for the insertion of all wires. Surface EMG data was additionally collected from medial gastrocnemius (MG) and soleus (SOL) from the right leg of all participants using Ag-AgCl electrodes with a diameter of 10 mm and an inter-electrode distance of 20 mm (Tyco Healthcare Group, Neustadt, Germany). A surface reference electrode, of the same type and size, was placed over the right fibula head. Prior to electrode placement, the areas of the leg corresponding to the electrode placement sites were shaved, lightly abraded and cleaned with isopropyl alcohol. All EMG signals were amplified 1000 times and hardware filtered with a bandwidth of 30 to 1000 Hz (MA300, Motion Labs, LA, USA). In order to minimize movement artefacts, the fine-wire electrodes, surface
electrodes, connectors, cabling and pre-amplifiers were secured with cohesive bandage around the foot and shank.

Prior to data collection, each participant was asked to perform foot manoeuvres known to activate each of the muscles separately. Activation of the AH muscle was assessed by asking the participant to abduct and flex the great toe about the metatarso-phalangeal (MTP) joint, while avoiding flexion of the interphalangeal (IP) joint. The FDB activation was assessed by asking the participant to flex toes 2-4 about the MTP joints, while minimising flexion of the distal IP joints (Kelly, Kuitunen, Racinais, & Cresswell, 2012; Kelly, Racinais, & Cresswell, 2013). When predicted EMG patterns could be detected, it was concluded that the fine-wire electrodes were in the correct location. If not, the electrodes were withdrawn by approximately 1 mm until appropriate activation patterns could be detected and possible crosstalk excluded.

In order to confirm signal integrity and quality between trials, the EMG signals were re-tested following each trial using the same foot manoeuvres applied above. A Velcro strap was secured around the participant’s waist, which enabled the EMG amplifier box to be secured to the subject without interfering with their gait. A lightweight optical cable connected the amplifier box to the analogue to digital converter that was positioned well away from the subject and treadmill.

Data analysis

Marker trajectories and GRF data were exported to Matlab (Matlab, The Mathworks Inc., USA) while EMG data files were exported to Visual3D (C-motion Inc., Germantown, MD, USA) for post processing and analysis.

Ground reaction forces
GRF data was filtered using a 35 Hz recursive, low-pass second order Butterworth filter. A vertical GRF threshold was set to define each toe-off as occurring when vertical GRF fell below 50 N, while foot contact was defined as occurring when vertical force rose above 50 N. The magnitude of the peak vertical and antero-posterior (A-P) components of the GRF were calculated and normalized to body weight (BW) for each participant. Peak propulsive force was defined as the peak positive value of the A-P component of the GRF.

Stride temporal characteristics

Swing phase was defined as the period from right toe-off to right foot contact, while stance phase was defined as occurring between right foot contact and right toe-off. One stride cycle was considered as right toe-off to the subsequent right toe-off based on GRF data. Stride duration and ground contact times were calculated for both conditions, for all participants.

Joint kinematics

Prior to calculation of joint kinematics, marker trajectories were low pass filtered using a recursive second order Butterworth filter at 20 Hz. Initially, an assumed rigid segment was created for the foot (21, 26). All marker data were transformed to the foot segment co-ordinate system, allowing all sagittal plane angles to be represented relative to the orientation of the z-axis of the foot. Thus, all sagittal plane rotations represent flexion-extension of their respective joints. Ankle angle was defined as the planar angle created by the bisection of a vector projecting from the medial malleolus to the med-knee virtual marker, with another vector projecting from the head of the first metatarsal to the medial malleolus, projected onto the sagittal plane of the foot. Ankle angle was offset to the ankle angle during quiet bipedal stance, calculated during the standing calibration trial. Foot-strike angle was defined as the sagittal plane ankle angle at foot contact. Minimum ankle angle was defined as the minimum angle
during stance phase, representing peak ankle dorsiflexion. Ankle angle at toe-off was also calculated.

In order to gain a deeper understanding of the mechanical behaviour of the foot, we also modelled the foot as a two-segment structure with rear-foot and mid-foot joints. The rear-foot joint angle was defined as the internal angle created by the bisection of a vector projecting from the med-knee virtual marker to the medial malleolus with another vector projecting from the navicular marker to the medial malleolus marker, projected onto the sagittal plane of the foot (Figure 1). Positive rear-foot joint angles indicate plantar-flexion of the rear-foot relative to the shank. Thus the rear-foot angle is representative of the ankle joint in our multi-segment foot model.

The LA is a complex structure comprised of a series of articulations made by a number of different bones (16, 17). Given the difficulties in quantifying the rotations about a series of small joints, located in such close proximity, the LA is often modelled as a single functional ‘mid-foot’ joint, representing the generalised angular motion of this structure (5, 11, 20, 21, 26, 39). Therefore the biomechanical function of the LA during running was represented by a ‘mid-foot’ joint, defined as the internal angle created by the bisection of a vector projecting from the medial malleolus marker to the navicular marker (rear-foot segment) and another vector projecting from the head of the first metatarsal to the navicular marker, projected onto the sagittal plane of the foot (26) (forefoot segment, Figure 1). A decrease in mid-foot angle signifies dorsiflexion, or LA compression (upwards rotation of the distal end of the forefoot segment) relative to the rear-foot. The rear-foot and mid-foot angles were offset to the corresponding angles calculated during quiet bipedal stance, so that 0° represent the angle in
quiet bipedal stance. Angle at foot contact, minimum stance angle (peak dorsiflexion) and angle at toe-off were calculated separately for the rear-foot and mid-foot joints, for all participants.

Joint Kinetics

An inverse dynamic analysis was performed using custom written software (Matlab, The Mathworks Inc., USA), allowing estimation of net joint moments about the ankle, rear-foot and mid-foot during stance phase. Unfiltered marker trajectories, ground reaction force and centre of pressure data were expressed relative to the foot co-ordinate system. Subsequently, proximal segment forces were resolved from the ground up using a link segment model (40). The mid-foot segment was defined as the vector projecting from the navicular marker to the marker located on the head of the first metatarsal, projected onto the sagittal plane of the foot. The rear-foot segment was defined as the vector projecting from the medial malleolus marker to the navicular marker, projected onto the sagittal plane of the foot. A traditional rigid foot segment was also defined in order to estimate ankle joint moments, enabling comparison of multi-segment foot kinetics with traditional calculations based on the assumption of a rigid foot segment. Segment masses and inertial properties were incorporated from previous literature (9, 11), with the centre of mass location for each segment being defined as 50% of the distance between the segment end points (11). Each joint was modelled as a pin joint with rotation occurring about the joint created by the bisection of the two vectors (Figure 1). Each joint axis was oriented parallel to the z-axis of the foot co-ordinate system. Following estimation of joint moments, joint powers were calculated for the stance phase by multiplying the net joint moment by the angular velocity of each corresponding joint (40). During early stance in a rear-foot running technique, GRF is applied to the rear-foot and forefoot segments simultaneously when the foot is flat on the ground. In order to overcome the difficulty in determining the relative percentage of GRF that is acting on the forefoot and rear-foot segments during early stance,
joint moments for the rear-foot and mid-foot were only calculated during stance phase after the COP had progressed distal to the axis of the mid-foot joint, minimising any potential errors associated with segment force assignments (37). Subsequently, joint moments and joint powers for the ankle, rear-foot and mid-foot were filtered using a 20 Hz recursive, second order low-pass Butterworth filter (Edwards, Troy, & Derrick, 2011). Net negative and positive work performed by each joint during stance phase was calculated by applying a trapezoidal integration of joint power curves across the stance phase. Mid-foot stiffness was quantified by dividing the change in mid-foot joint moment by the change in mid-foot angle during stance phase (Farley & Morgenroth, 1999).

Muscle activation

All EMG signals were high-pass filtered using a recursive second order Butterworth filter at 35 Hz to remove any unwanted low-frequency movement artefact. Subsequently, EMG signals were visually inspected in order to identify any remaining artefact, which was defined as an abnormal spike in the signal, typically associated with foot contact. Any remaining artefacts resulted in the EMG data for that particular stride being excluded from further analysis. Following DC-offset removal, root mean square (RMS) signal amplitude was calculated across all EMG data using a moving window of 50 ms to generate an EMG envelope. The EMG envelope for each muscle was then normalized to the maximum RMS amplitude found across all conditions for the respective muscle. The normalized peak EMG RMS envelope amplitude was calculated during stance phase, additionally the normalized mean EMG RMS was calculated during both the stance and swing phases, allowing comparisons in magnitude of both stance and swing phase muscle activation between HFS and FFS conditions.
For each individual, the motion, force and EMG data were time normalized and ensemble averaged across a minimum of 8 stride cycles to form individual variable means for each condition. Paired Student’s t-tests were used to describe the influence of foot-strike pattern on stride temporal characteristics, peak ground reaction forces, joint kinematics and kinetics, as well as muscle activation. Statistical differences were established at $P \leq 0.05$. Effect sizes (ES) are presented as standardized mean differences (Cohen’s D) with $\leq 0.70$ representing a small effect, $>0.70$ and $<1.2$ representing a moderate effect, and $\geq 1.2$ representing a large effect (Hopkins, 2017). Results are presented as mean ± standard deviation (SD) unless otherwise stated.

**Results**

**Stride temporal characteristics**

Participants ran with similar stride durations (ES = 0.07, $P = 0.54$) and ground contact times (ES = 0, $P = 0.77$, Table 1), regardless of the orientation of their ankles at ground contact.

**Ground reaction forces**

When running with a habitual foot strike pattern, vertical GRF-time profiles displayed a characteristic initial early impact transient, followed by a second, much later, larger peak (Figure 2). The initial impact peak transient seen for habitual foot strike running was absent when the participants ran with a forefoot technique (Figure 2), however the magnitude of the peak vertical GRF was significantly higher with the forefoot technique (ES = 0.81, $P \leq 0.05$, Table 1). Propulsive and breaking force was similar between conditions (Figure 2).

**Joint kinematics**
When running with a habitual foot strike pattern, participants generally made initial ground contact with the heel or mid-foot. One participant landed with the ankle in slight plantar flexion (5.4°), while the remaining participants landed with ankle the ankle in slight dorsiflexion (range -8.5° to -2.0°, Figure 3). None of the participants landed with a forefoot strike as their habitual foot-strike pattern. Participants landed with their ankles in a significantly more plantar flexed orientation when instructed to run with a forefoot strike (range 6.4° to 20.6°, ES = 3.5, P ≤ 0.05, Table 2). Peak ankle dorsiflexion occurred in mid-stance for both conditions and was significantly greater when running with a habitual foot-strike (ES = 0.88, P ≤ 0.05, Table 2). At toe-off the ankle was in a more plantar flexed orientation in the forefoot running condition (ES = 1.1, P ≤ 0.05, Table 2).

The rear-foot rotated in a similar manner to the ankle, with dorsiflexion occurring from forefoot contact through to mid-stance and plantar flexion occurring from mid-stance to toe-off. The rear-foot was in a more plantar flexed orientation at foot contact when running with a forefoot strike (ES = 1.1, P ≤ 0.05, Table 2 and Figure 3). Peak rear-foot dorsiflexion occurred in mid-stance and was reduced in magnitude when running with a forefoot strike pattern (ES = 3.5, P ≤ 0.05, Table 2). The rear-foot was more plantar flexed at toe-off with the forefoot running technique (ES = 1.1, P ≤ 0.05, Table 2).

For both running conditions, the mid-foot dorsiflexed (LA compression) from early stance through to mid-stance (Figure 3) and subsequently plantar flexed through to toe-off. At foot contact, the mid-foot joint was more plantar flexed when running with a forefoot strike (ES = 2.7, P ≤ 0.05, Table 2), indicating a higher LA at foot contact. However, there was no difference in peak mid-foot dorsiflexion between running conditions (ES =0.06, P = 0.63,
The mid-foot was in a position of greater plantar flexion at toe-off when participants ran with a forefoot strike (ES = 0.9, P ≤ 0.05, Table 2).

**Joint kinetics**

Peak ankle plantar flexion moment was greater when running with a forefoot strike pattern (ES = 1.3, P ≤ 0.05, Table 3 and Figure 3). Peak negative ankle power (ES = 2.4, P ≤ 0.05, Table 2) and positive ankle power (ES = 0.74, P ≤ 0.05, Table 2) were also greater when running with a forefoot strike technique. A greater amount of negative work (ES = 2.1, P ≤ 0.05, Table 2) and positive work (ES = 1.0, P ≤ 0.05, respectively) was performed at the ankle when running with a forefoot strike.

For both running conditions, peak rear-foot plantar flexion moment occurred in mid-stance (Figure 3). Peak rear-foot plantar flexion moment was greater when running with a forefoot technique (ES= 1.3, P ≤ 0.05, Table 3 and Figure 3). Peak negative power was greater when running with a forefoot technique (ES = 1.6, P ≤ 0.05, Table 3), however there was no difference in peak positive power between running conditions (ES = 0.3, P = 0.13, Table 3). A greater amount of negative work (ES = 1.8, P ≤ 0.05, Table 3) and positive work (ES = 0.6, P ≤ 0.05, Table 3) was performed at the rear-foot in the forefoot running condition.

In a similar manner to the ankle and rear-foot, peak mid-foot plantar flexion moment occurred in mid-stance for both running conditions (Figure 3). Peak mid-foot plantar flexion moment was greater when running with a forefoot technique (ES = 1.1, P ≤ 0.05, Table 3 and Figure 3). Peak negative mid-foot power (ES = 2.7, P ≤ 0.05, Table 3) and positive mid-foot power (ES = 0.77, P ≤ 0.05, Table 3) was also greater when running with a forefoot strike. A greater amount of negative work (ES = 2.2, P ≤ 0.05) and positive work (ES = 1.1, P ≤ 0.05, Table 3)
was also performed at the mid-foot when running with a forefoot technique, suggesting that a
greater magnitude of mechanical energy is absorbed, returned and/or generated about the mid-
foot. Despite similar peak deformation and an increased plantar flexion moment, the mid-foot
was less stiff (more compliant) when running with a forefoot technique (ES = 0.65, P ≤ 0.05,
Table 3).

Muscle activation

The AH muscle displayed a large burst of activity that commenced during late swing-phase,
and continued through to toe-off, with peak activation occurring in mid-stance (Figure 4).
When running with a forefoot strike, mean AH activity was greater in swing (ES = 0.67, P ≤
0.05, Table 4) and stance (ES = 0.63, P ≤ 0.05, Table 4) phase. Peak stance phase muscle
activation was also greater when running with a forefoot strike (ES = 0.71, P ≤ 0.05, Table 4).

The FDB activation pattern was characterised by a large burst of activity commencing at foot
contact, with a mid-stance peak and de-activation shortly after toe-off (Figure 4). As this
muscle is primarily active in stance phase, mean swing phase activation was similar between
conditions (ES = 0.01, P = 0.95, Table 4). Mean FDB stance phase activation substantially
higher when running with a forefoot strike (ES = 0.56, P ≤ 0.05, Table 4), however there was
no significant difference in peak stance phase activation for this muscle (ES = 0.49, P = 0.19,
Table 4).

For both conditions, MG displayed a large burst of activation commencing in late swing,
continuing through stance, with de-activation occurring during late stance-phase (Figure 4).
Mean swing phase activation was substantially greater when running with a forefoot strike
while mean (ES = 0.57, P = 0.09, Table 4) and peak (ES = 0.06, P = 0.84, Table 4) stance phase activity was similar for both running conditions.

The SOL activation pattern was similar for both running conditions, with a single burst of activity that commenced in terminal swing phase, peaked during mid-stance and ceased prior to toe-off (Figure 4). Swing phase activation was comparable for both running techniques (ES = 0.2, P = 0.39), as was mean (ES = 0.50, P = 0.19) and peak (ES = 0.69, P = 0.21) stance phase activity (Table 4).

**Discussion**

This study provides novel evidence for active alterations to the mechanical function of the foot when acutely switching between non-forefoot and forefoot running techniques. When running at matched velocities, forefoot running resulted in higher mid-foot joint moments, more positive and more negative work being performed about the mid-foot. Increases in joint moments for forefoot striking occurred concurrently with increased intrinsic foot muscle activation and no increase in peak mid-foot dorsiflexion. Thus, it seems that the intrinsic foot muscles may be activating to stiffen the foot during mid-stance when running with a forefoot technique.

**LA kinematics**

Peak LA deformation was not influenced by foot-strike technique, despite the increased joint moments about the mid-foot when running on the forefoot. This finding is analogous to a number of previous studies that also reported no difference in peak LA deformation when running with rear-foot or forefoot running techniques (30, 33, 39). Interestingly, the previous studies by Wager & Challis (2016) and McDonald et al (2016) have made the assumption in their modelling that the foot is a passive structure with no muscular control of LA stiffness. The findings of our current
study highlight that the foot is not passive and the plantar aponeurosis is not the sole structure responsible for maintaining LA stiffness during running. If the LA compression was indeed controlled purely by passive elastic structures (e.g. plantar aponeurosis), the increased torques observed when running with a forefoot technique would produce an increase in LA deformation (12, 22). This was not the case in the current study, or in the previously mentioned studies (30, 33, 39), indicating that active muscular control must be contributing to control of LA mechanics when switching between foot-strike techniques.

Recently published studies performed within our laboratory have highlighted the importance of the plantar intrinsic foot muscles in actively controlling the LA during quiet stance and locomotion (19-21). When activated, these muscles have the capacity to provide buttressing support for the LA, resisting excessive deformation that may occur at loads exceeding body weight (18). Furthermore, it appears that the central nervous system (CNS) can vary the relative contribution of these muscles in controlling the LA, depending on the magnitude of forces encountered when running (20). Based on our knowledge of the function of the intrinsic foot muscles (Kelly, Cresswell, Racinais, Whiteley, & Lichtwark, 2014a; Kelly, Lichtwark, & Cresswell, 2014b), it becomes apparent that the observed increase in AH and FDB activation during stance may be responsible for the lack of difference in peak LA deformation between running techniques, despite obvious differences in joint moments. An increase in force output from the intrinsic foot muscles when running with a forefoot technique may serve to reinforce the LA in mid-stance, preventing excessive LA deformation and any associated increase in plantar aponeurosis strain. Recently McDonald et al. (2016) suggested that the risk of injury to the plantar aponeurosis (i.e. plantar fasciopathy) might not be influenced by foot-strike technique, due to the similarity in peak plantar aponeurosis strains between techniques. The current study reveals that there is a substantial increase in loading and energy absorbed about the mid-foot when running with a forefoot technique.
and that this increase is likely associated with an increase in force output from the plantar intrinsic foot muscles. Therefore, without specific training, habitual rear-foot runners may not have the strength or endurance in the intrinsic foot muscles to sustain a forefoot running technique for prolonged periods. This may subsequently influence the loading on the plantar aponeurosis. Further investigations may now be warranted to determine the time required sufficiently strengthen the intrinsic foot muscles, in order to safely transition to a forefoot running technique.

Within the current study we observed a reduction LA stiffness when participants ran with a converted forefoot technique, despite an increase in mid-foot joint moment and no change in peak mid-foot deformation. This reduction in stiffness appears to be due to an increase in mid-foot plantar-flexion (LA height) at foot contact, allowing the mid-foot to move through a greater total range of motion in the forefoot running condition. We observed an increase AH activation in late swing, preceding an increase in LA height at foot contact when participants ran with a forefoot technique. Thus it appears that late swing activation of AH may increase the height of the LA prior to foot contact. An increase in total excursion of the LA and the associated reduction in LA stiffness may enhance the capacity for energy storage and return within this structure, without the potentially harmful effect of increased plantar fascia strain. It must however be noted that due to the lack of difference in peak LA deformation between running techniques, any increase in mechanical work performed about the LA is likely to be performed by active muscle rather than passive elastic tissues. Therefore any increase in energy absorption and return may come with increased metabolic cost.

Joint Moments

Forefoot running resulted in higher plantar flexion moments being generated about the ankle and mid-foot. When running with a forefoot technique, the resultant GRF is generally positioned within
the anterior portion of the foot, as the forefoot makes initial contact with the ground. Conversely, with a rear-foot technique the COP is initially located on the lateral aspect of the heel and progresses forward as the body progresses over the foot (6, 7, 32). If the COP is located further anterior when running with a forefoot technique, it will have a greater moment arm about the ankle and mid-foot, which, alongside an increase in peak vertical GRF, likely explains the increase in plantar flexion moments that are generated about these joints. While the finding of an increase in plantar flexion moment about the mid-foot is quite novel, a similar increase in ankle joint moment has been reported in a number of previous studies examining differences in ankle joint kinetics when switching between rear-foot and forefoot running techniques (23, 35).

Mechanical work and power

The energetic benefits of the spring-like function of the LA during running have been well documented, with this mechanism providing between 8 – 17% of the mechanical energy required for each stride via the elastic stretch and recoil of the plantar aponeurosis (22, 36). Previous studies exploring the influence of foot-strike technique on elastic energy storage in the LA (30, 39) have reported that a forefoot running technique does not promote increased utilization of the elastic energy, due to the lack of difference in peak plantar aponeurosis strain between running techniques. Within our current experiment we observed an increase in mechanical power absorption and generation and an associated increase in negative and positive work about the LA, in the absence of any additional increase in LA deformation. The plantar intrinsic foot muscles are known to actively lengthen and shorten during the stance phase of running and therefore have the potential to absorb and generate mechanical power about the LA (20). The increase in intrinsic foot muscle activation when running with a forefoot technique indicates that the contribution of these muscles to mechanical work about the LA increases when running with a forefoot technique. The intrinsic foot muscles have relatively long tendons and short muscle fibres (24) and thus may be well suited
to elastic energy storage and return. The additional negative work done at the mid-foot joint when running with a forefoot technique may allow greater utilization of the foot spring mechanism, allowing additional energy to be stored in the tendons of the intrinsic foot muscles, without the potentially detrimental effects of excessive strain on the plantar aponeurosis. In this sense we might consider the intrinsic foot muscles as reserve actuators, able to act in parallel with the plantar aponeurosis when a certain level of force generation is required. However, it must also be noted that the increase in work performed about the LA when running with a forefoot technique may not translate directly into metabolic energy savings, due to the added energetic cost of muscle activation required to perform this work. The increased cost of activation in the intrinsic foot muscles and also the posterior leg muscles when running with a forefoot technique may in part explain the lack of difference in energetic cost between habitual rear-foot and forefoot runners (4, 13) as the increased utilization of spring-like running mechanics may be offset by the additional energetic cost of muscle activation.

Runners with an habitual forefoot technique generally display shorter ground contact times, shorter stride length and an increased cadence, compared to runners who are habitual rear-foot strikers (Gruber, Umberger, Braun, & Hamill, 2013; Lieberman et al., 2010; Stearne, Alderson, Green, Donnelly, & Rubenson, 2014). In the current experiment, our group of habitual rear-foot runners maintained relatively constant stride temporal characteristics when acutely changing to a forefoot technique. This finding is in line with a number of previous studies that have shown a lack of immediate change in stride temporal characteristics, when participants shifted from an habitual rear-foot technique to an imposed forefoot technique (Almeida, Davis, & Lopes, 2015). Thus, it is apparent that changes in stride temporal characteristics do not occur automatically when changing foot-strike technique. Specific training to encourage a shorter stride length
and increased cadence, may also need to be included as part of a habituation process in a transition to forefoot running.

Methodological considerations

The LA is a multi-articular structure (17). We have modelled this structure as single functional mid-foot joint. This technique allows important general insights into the function of the LA, however, it may also be an over-simplification of the complex sequence of rotations and translations that occur about a series of small joints within this overall structure. Furthermore, we have only modelled the function of this structure in the sagittal plane. It is likely that motion of the LA also occurs in the frontal and transverse planes (Kelly et al., 2014a; Leardini et al., 2007).

The use of an inverse dynamic solution allowed us to provide unique insight to the mechanical function of the LA when running with rear-foot and forefoot running techniques. A limitation of this approach is that it is difficult to quantify the relative percentage of GRF that is acting on the forefoot segment at any given time in stance, as some force may be simultaneously applied to the rear-foot. However, as the COP rapidly progresses anteriorly in stance phase and the forefoot bears a substantial portion of the GRF force for the majority of stance duration (6, 7), any errors associated with this assumption are likely to be small and localized to the initial period of stance in the habitual running condition. In order to mitigate this limitation, we have calculated rear-foot and mid-foot moments and powers after the COP has progressed anterior to the axis of the mid-foot joint. The COP passed the axis of the mid-foot at approximately 25% stance phase in the habitual running condition and after this time we are confident that the vast majority of contact forces are applied to the forefoot. When running with a forefoot running technique the COP was anterior to the axis of the mid-foot joint for the entire duration of stance phase, therefore the GRF is applied to the forefoot segment from the instant of foot contact. If small errors do exist based on
our GRF assignment assumption, these are limited to the rear-foot running condition and will lead to an over-estimation in LA joint moments, therefore the magnitude of differences in joint kinetics between conditions may actually be slightly larger than those reported. Providing further support to the validity of our inverse dynamic approach, the joint moments at the rear-foot are comparable to the ankle joint moments using a traditional rigid foot segment, both within our current study and also to the values presented in previous studies reporting ankle joint moments during running at similar speeds (34).

Participants within this experiment were shod rear-foot runners who were instructed to run with their habitual foot-strike technique and a forefoot technique. It is possible that different observations may have been made if habitual forefoot runners were included in this experiment. However based on the findings of previous studies (30, 35) it is unlikely that LA kinematics would differ between habitual rear-foot and forefoot runners when instructed to run with forefoot and rear-foot techniques. Furthermore, given that activation of the intrinsic foot muscles is dependent on loading requirement (18), it is unlikely that the activation patterns of these muscles would substantially differ either.

Conclusion

Foot-strike technique substantially influences the mechanical function of the LA during running. A forefoot running technique leads to an increase in loading about the mid-foot and an increase in positive and negative mechanical work performed by the intrinsic foot muscles. These findings suggest that adequate strength and endurance training for the intrinsic foot muscles should be included as part of training programs for runners seeking to transition from a rear-foot to a forefoot running technique.
Acknowledgments

The results of the study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation. Results of the present study do not constitute endorsement by ACSM. The authors would like to acknowledge the contribution of Mr Zachary Goodchild to data collection and processing.

Conflict of interest

The authors declare no conflict of interest.

Funding

Funding for this study was provided via an industry research grant from Asics Oceania. Grant identification number 2014000885.

References


Figure 1. Depiction of sagittal plane angular rotations about the Ankle (top), Rear-foot (middle) and Mid-foot (bottom) joints. Angles in the text are relative to the respective joint angle in quiet stance, with 0° representing the joint angle in quiet stance. Negative angles denote dorsiflexion (upwards rotation of the distal segment).

Figure 2. Group mean ± standard deviation (shaded area) for vertical ground reaction force (GRF, top) and anterior-posterior GRF (bottom). Data is recorded from each participant running with a habitual (solid) and forefoot running technique (dashed) at 2.8 ms⁻¹. Data are shown for stance only; from right foot contact (FC) to right toe off (TO). All data are normalised to body weight (BW). * denotes a significant difference in peak vertical GRF between the running conditions.
Figure 3. Group mean ± standard deviation (shaded area) for joint angle (top), normalised joint moment (middle) and normalised joint power (bottom) calculated during stance phase for the mid-foot (left), rear-foot (middle) and ankle (right). Data is recorded from each participant running with a habitual (solid) and forefoot running technique (dashed) at 2.8 ms\(^{-1}\). Data are shown for stance only; from right foot contact (FC) to right toe off (TO). Vertical dashed lines denote the time point that the centre of pressure progressed anterior to the axis of rotation of the mid-foot joint in the habitual foot-strike condition. Joint work is calculated by trapezoidal integration of the joint power data. \(\delta\) denotes a significant difference between running conditions in joint angle (at FC, peak or TO) for each joint. \(\gamma\) denotes significant difference in peak joint moment between the running conditions. \(\beta\) denotes a significant difference in peak joint power (negative or positive) and \(\alpha\) denotes a significant difference in negative or positive joint work between running conditions.

Figure 4. Group mean ensembles ± standard deviation (shaded area) for electromyography (EMG) normalised root mean square signal amplitude for medial gastrocnemius (MG), soleus (SOL), flexor digitorum brevis (FDB) and abductor hallucis (AH). Group mean ensembles are defined from toe off (TO) to ipsilateral toe off for the right foot. Data recorded during running at 2.8 ms\(^{-1}\). For each muscle EMG data is normalised to the maximal amplitude recorded for all trials. The habitual foot-strike condition is the solid lines and the forefoot condition the dashed lines. Vertical lines denote onset of stance phase at foot contact (FC). * denotes significant difference between running conditions.
Figure 1. Depiction of the Lower limb marker set employed for collection of kinematic data. Knee markers are removed following a static calibration trial, with the cluster of four markers on a rigid plastic shell used to track the motion of the shank. The posterior calcaneus marker cannot be viewed in this image.
Figure 2. Depiction of angular rotations about the longitudinal arch (LA) and rear-foot (RF) joints. Positive angles denote plantar flexion. Increase in LA angle indicates an increase in height of the LA.
Figure 3. Group mean ± standard deviation (shaded area) for vertical force (top) and antero-posterior force (bottom). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at 2.8 ms⁻¹ and presented from foot contact to toe off from the right foot. All data is normalised to body mass.
(bottom) calculated during stance phase for the longitudinal arch (left), rear-foot (middle) and ankle (right). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at 2.8 ms\(^{-1}\). All data is normalised to body mass. Vertical dashed lines denote initial forefoot loading in preferred foot-strike condition.
Figure 5. Group mean ensembles ± standard deviation (shaded area) for electromyography (EMG) normalised root mean square signal amplitude for gastrocnemius medialis (MG), soleus (SOL), flexor digitorum brevis (FDB) and abductor hallucis (AH). Group mean ensembles are defined from toe off (TO) to ipsilateral toe off for the right foot. Data recorded during running at 2.8 ms⁻¹. For each muscle EMG data is normalised to the maximal amplitude recorded for all trials. The preferred foot-strike condition is the solid lines and the forefoot condition the dashed lines. Vertical lines denote onset of stance phase.