The pursuit of improved running performance: Can changes in cushioning and somatosensory feedback influence running economy and injury risk?

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**Purpose:** There is currently no consensus regarding the effect that barefoot (BFT) running has on running economy (RE). Stride length and shoe mass are confounding variables, with a BFT stride length being shorter than a shod (SH) stride length. Comparison of SH, minimalist shod (MS) and BFT allows controlled variation of cushioning and somatosensory feedback to determine the effect that either and/or both have on RE and running mechanics. **Methods:** Fifteen female habitually shod, recreational runners visited the laboratory twice. Familiarisation to BFT and SH treadmill running occurred during visit one, in addition to determining SH stride length and BFT stride length. During visit two participants ran BFT, SH and MS with BFT stride length and MS with SH stride length at 10 km·h⁻¹ for six minutes with ten minute rest periods between each condition. Lower limb kinematics, EMG, impact acceleration and \( \dot{\text{VO}}_2 \) were recorded during the final two minutes of each run. **Results:** BFT RE was significantly better than SH and MS with BFT stride length. SH RE was significantly worse than MS with SH stride length, but similar to MS with a BFT stride length. Low vertical oscillation, peak eversion and peak dorsiflexion, less plantarflexion at toe-off, in addition to an earlier occurrence of heel off, higher impact accelerations and greater tibialis anterior activity were observed during the most economical condition. **Conclusions:** Heightened somatosensory feedback and lack of cushioning (BFT) offered an advantage to economy over less somatosensory feedback (MS) and cushioning (SH). Whilst the low vertical oscillation and low plantarflexion at toe-off appear to contribute to the improved RE, other changes to running mechanics whilst BFT could potentially influence injury risk. **Keywords:** Barefoot running, minimalist footwear, running mechanics, muscle activity, oxygen consumption
1. Introduction

Potential benefits of barefoot (BFT) running have long been of interest to both runners and scientists. Research in the late ‘80s first suggested that running BFT could have possible benefits in terms of reducing the risk of injury (Robbins et al. 1989, Robbins & Hanna 1987). Recently, attention had been given to addressing the question of whether BFT or minimalist shod (MS) running is more economical than shod running (SH) (Franz et al. 2012, Hanson et al. 2011, Squadrone & Gallozzi 2009). The anecdotal notion that BFT running may improve a runner’s performance stems from Abebe Bikila and Zola Budd breaking world records in 1960 and 1985 for the marathon and 5000m respectively, whilst BFT. However, these runners represent a small minority, as the vast majority of runners are, and have been, SH whilst competing and breaking other world records in running.

A recent review concluded that currently there is no consensus in the literature regarding the affect that BFT and/or MS running has on running economy (RE) (Nigg & Enders 2013), a crucial factor in determining performance in long distance running events (Conley & Krahenbuhl 1980). The differing methodologies are perhaps, in part, the reason for contrasting evidence. For example, Divert et al. (2008) demonstrated that the difference in oxygen consumption ($\dot{V}O_2$) when BFT and SH was due to the effect of added mass. Based on this, researchers have looked to adjust for the ‘mass effect’ by attaching equivalent masses to the feet when running BFT or MS (Franz et al. 2012, Perl et al. 2012). Typically, such investigations report MS or SH running to have a better RE (lower $\dot{V}O_2$) than BFT running (Franz et al. 2012). Such rigor aids the control of the confounding variable, increasing internal validity, but actually detracts from the essence of running BFT whereby nothing is on your feet, and thus reduces external validity. Furthermore kinematic adjustments to BFT running, which have been extensively reported (De Wit et al. 2000, Squadrone & Gallozzi 2009), are likely to be affected by the extra mass being carried on the foot (Martin 1985). Mathematically adjusting the absolute $\dot{V}_O_2$ consumed to not just a runner’s mass, but also the difference in footwear mass provides an approach to account for shoe mass without potentially affecting running mechanics.

One of the main gait parameters that changes when running BFT is stride length. Previously, researchers have found the naturally chosen BF stride length to be 3-7% shorter than the naturally chosen SH stride length (Franz et al. 2012, Squadrone & Gallozzi 2009). Based on evidence that the naturally chosen stride length is at or near economically optimal for runners (Cavanagh & Williams 1982), researchers have argued that stride length may play an important role in determining RE during BFT running (Perl et al. 2012). That said, previous investigations have either not controlled for stride length (Franz et al. 2012, Squadrone & Gallozzi 2009) or instructed runners to run with their SH stride length whilst MS (Perl et al. 2012). Perl and colleagues (2012) hypothesized that if they had instructed runners to shorten their stride lengths whilst MS they would have found greater improvements in RE compared to SH. However, given that SH and MS have been reported to have similar stride lengths (Squadrone & Gallozzi 2009), then it could be argued that shorter MS strides may not result in RE improvements.

Another factor influencing any observed differences between BFT and SH running is the level of somatosensory feedback experienced (Robbins & Hanna 1987). SH running offers a cushioning protective
layer between the foot and the ground that reduces somatosensory feedback. BFT running removes this layer, thus heightening the somatosensory feedback experienced (Robbins & Hanna 1987). It must also be noted that previous investigations have often instructed participants to wear socks in their BFT condition (Divert et al. 2008, Franz et al. 2012), which interferes with the level of somatosensory feedback the runner receives. Minimalist footwear typically also lacks a cushioning layer, but does include a thin protective layer. This reduces the potential benefit of increased somatosensory feedback provided when BFT, but also lacks cushioning in a similar manner to BFT. Thus MS, like SH, has reduced somatosensory feedback, but differs in the level of cushioning. Therefore comparison of SH, MS and BFT allows controlled variation of cushioning and somatosensory feedback to determine the effect that either and/or both have on RE and running mechanics.

The pursuit of improved running performance through changes in running mechanics is also likely to affect a runner’s risk of sustaining an injury. Whilst BFT gait adjustments have been reported to reduce injury likelihood (Lieberman et al. 2010, Robbins & Hanna 1987), there is no conclusive evidence to support this notion (Nigg & Enders 2013). In fact, there is some evidence of an increase in peak tibial acceleration when running BFT compared with SH (McNair & Marshall 1994, Sinclair et al. 2013a). On the other hand, the lack of cushioning experienced when BFT, in addition to heightened somatosensory feedback, are believed to promote a forefoot strike pattern (Lieberman et al. 2010). Such a footstrike modality has been associated with reduced impact force (Lieberman et al. 2010) and injury rate (Daoud et al. 2012) in comparison to rearfoot striking. Yet RE is similar between the two strike patterns (Perl et al. 2012). Together these findings suggest changes in running mechanics that potentially reduce injury likelihood are not specifically beneficial or detrimental to performance. Consideration of biomechanical changes for both their performance and injury implications is needed to generate greater understanding of the differences between BFT, MS and SH.

It has also been argued that there is a metabolic cost to cushioning the body during BFT running due to muscles having to actively protect the lower limb upon impact and during ground contact. This is known as the ‘cost of cushioning’ hypothesis (Frederick 1984). Due to the lack of external cushioning present in minimalist footwear, it is likely that this extra metabolic cost will apply to MS running too. Contrary to this expectation, recent evidence has shown that transitioning to running MS can improve RE (Warne & Warrington 2012), and thus performance. Yet it can also increase bone marrow edema (Ridge et al. 2013), which is an early sign of bone stress injury. Researchers must therefore consider the affect that any changes in gait have on both performance and injury mechanisms before being able to justify advocating either BFT, MS or SH running to individuals.

There is a relative scarcity of studies that have assessed muscular activity during BFT/MS running. One notable exception identified that the tibialis anterior (TA) activity was adjusted to suit external conditions, such as changes in cushioning (von Tscharner et al. 2003). These authors reported greater activity in the TA during ground contact, but less activity just prior to touchdown whilst BFT compared to SH. Although Hamill and colleagues (2011) did not record muscular activity, they did calculate a high level of ankle stiffness to be present during BFT running (with a forefoot strike pattern). This suggests a high level of muscle coactivation, which is the simultaneous contraction of two muscles. Such a muscular strategy can be detrimental to the
metabolic cost of running (Moore et al. 2013). To fully investigate the impact of changing cushioning and somatosensory feedback during running muscular activity must also be considered. This will allow the ‘cost of cushioning’ hypothesis, first proposed by Frederick (1984), to be evaluated.

This study aimed to assess the mechanisms behind changes in RE during running at different stride lengths when varying cushioning and somatosensory feedback. This was achieved by comparing SH, MS and BFT running economy, biomechanics (lower limb kinematics and heel impact accelerations) and electromyography (EMG). Since the focus was on mechanisms, specific hypotheses concerning kinematics and muscular activity/coactivity were not constructed. Regarding performance benefits and injury concerns, it was hypothesised that: 1) BFT running would be the most economical condition. 2) Running with a SH stride length during MS running would be more economical than MS with a BFT stride length, and only running with a SH stride length when MS would be more economical than SH. And secondly regarding acceleration; 3) BFT and MS running would have greater impact accelerations than SH.

2. Methods
Fifteen female (mass: 62.0 ± 6.4 kg, height: 1.66 ± 0.1 m, age: 20.5 ± 1.4 yrs, weekly distance: 74.1 ± 16.6 km-wk⁻¹), habitually shod, recreational runners volunteered for the study. Only participants with no prior BFT and MS running were included in the study. Each participant provided informed consent and was free from injury for at least three months prior to testing. Ethical approval was granted from the Ethics committee of the University’s Sport and Health Sciences department.

2.1 Procedure
Two laboratory visits were completed by each participant. During the first visit participants underwent familiarisation to running BFT and SH on a motorised treadmill (PPS 43med; Woodway, Weil am Rhein, Germany). The experimental procedure was undertaken during the second visit. Movement analysis and muscle activity of the right leg were simultaneously recorded, along with oxygen consumption.

2.2 Treadmill familiarisation
Based on results from our laboratory and literature evidence, 20 minutes (2 x 10 minutes) was given for BFT running and 6 minutes for SH (Lavcanska et al. 2005, Moore 2013). At the end of each familiarisation period the participant’s natural stride length was recorded using a Basler camera (100 Hz) positioned approximately 1.5 m in front of the treadmill. To determine each stride length the following equation (1) was used:

\[ \text{SL} = \frac{\text{ST} \times \text{V}}{\text{K}} \]  

SL is the stride length, ST is the time taken for each stride (right foot contact to right foot contact) and V represents the treadmill speed. Six strides were recorded during each familiarisation condition and the average stride length was calculated for both BFT and SH running. The Root Mean Square Error (RMSE) for stride length
using this approach was 0.06 ± 0.02 m (coefficient of variation: 1.3 ± 0.4%) for BFT and 0.05 ± 0.02 m (coefficient of variation: 1.0 ± 0.4%) for SH running.

2.3 Experimental procedure

Participants ran each stride length (BFT and SH) during each footwear condition (BFT, MS and SH), performed in a randomised order to reduce fatigue and learning effects. Each run was performed at 10 km·h⁻¹ for six minutes with 10 minute rest periods between each condition. Only data from the BFT with BFT stride length, MS with BFT stride length, MS with SH stride length and SH with SH stride length are presented here and will be referred to as BFT, MS with BFT stride length, MS with SH stride length and SH from here on in. The rest period was used to change over footwear and check the attachment of kinematic markers. A metronome was used to control stride length, with participants instructed to strike the treadmill in time with each beat.

The treadmill used for the experimental procedure was the same as that used for the familiarisation. Heart rate was recorded during the final two minutes of each exercise bout via a wireless chest strap telemetry system (Polar Electro T31; Kempele, Finland). Respiratory gas exchange was measured every 10 s throughout each run using an automated gas analysis system (Cortex Metalyzer II; Cortex Biophysik, Leipzig, Germany).

Three-dimensional kinematic data were collected using an eight camera optical system (Vicon Peak, 120 Hz, automatic optoelectronic system; Peak Performance Technologies, Inc., Englewood, CO) positioned around the treadmill. Synchronisation of the kinematic and EMG data occurred via a manual trigger pressed during the final two minutes of each run. Spherical reflective markers were affixed to the right lower limb of each participant on the following anatomical landmarks to denote the thigh, shank and foot using a modified Soutas-Little et al. model (1987): the proximal greater trochanter (hip), the medial and lateral condyles (knee), midline of the posterior shank, the musculotendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon, the mid tibia below the belly of the tibialis anterior, the lateral malleolus (ankle), the superior and inferior calcaneus and the proximal head of the third metatarsal. Additionally an accelerometer (Trigno Wireless EMG, Delsys, Boston, MA, USA, 148 Hz) was attached to the heel, in between the calcaneus markers, and used to determine heel impact acceleration.

Surface EMG (Trigno Wireless EMG, Delsys, Boston, MA, USA; parallel bar configuration, contact material 99.9% Ag, interelectrode spacing 10 mm, electrode size 37 x 26 x 15 mm) was used to analyse the activity of six lower limb muscles on the right leg: rectus femoris (RF); vastus lateralis (VL); biceps femoris (BF); gastrocnemius medialis (GM); gastrocnemius lateralis (GL); and TA. The electrodes were placed longitudinally with respect to the muscle fibre direction following standardised criteria recommended by SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles project) (Hermens et al. 2000). The skin surface area was prepared using an abrasive gel and then wiped clean with an alcohol swab. The electrodes were affixed with double-sided tape to the lower limb, with those on the shank covered with elastiacted tubular bandage and those on the thigh with self-adhesive elastic bandage to minimize their movement. The kinematic markers were then affixed on top of the bandages.
2.4 Data analysis

RE was defined as the average $\dot{V}O_2$ during the final two minutes of each run. Carbon dioxide expiration ($\dot{V}CO_2$), respiratory exchange ratio (RER) and minute ventilation ($\dot{V}E$) were also determined over the same time period. To account for the mass of the shoe absolute $\dot{V}O_2$, $\dot{V}CO_2$ and $\dot{V}E$ values were normalized to each individual’s mass, plus the shoe mass difference. Subsequently two sets of normalization occurred, the first to compare BFT and MS to SH and the second to compare BFT to MS. Given that the condition with the greatest added mass in each set is assumed to lead to the highest $\dot{V}O_2$ whilst running, and hence a worse RE, the mathematical adjustments should impose a similar effect upon RE by decreasing the denominator (i.e. mass). Therefore to account for the trainer mass (SH condition) the following adjustments were made: 1) BFT adjusted to body mass minus the trainer mass; 2) MS adjusted to body mass minus the difference in mass between the trainer and minimalist footwear and; 3) SH adjusted to body mass. To account for the minimalist footwear mass (MS condition) two adjustments were made: 1) BFT adjusted to body mass minus the minimalist footwear mass and; 2) MS adjusted to body mass. This method has not previously been utilised by researchers, however it provides more realistic BFT and MS running conditions than adding weights to the foot. This latter approach increases the leg moment of inertia and has been shown to interfere with running mechanics (Martin 1985).

Stance was defined as initial foot touchdown (TD) to toe-off (TO) and was determined using the vertical heel acceleration data. When comparing the accuracy of using acceleration data to identify stance to force plate data Sinclair and colleagues (2013b) reported an error of 5.2 ms. Six consecutive gait cycles were collected for each condition during the final two minutes of data collection for that condition. Participants were not aware of when data collection was taking place. Kinematic data were filtered using a fifth-order quintic spline within the Peak Motus system. Dynamic angles were normalized to standing trials recorded after each run to provide anatomically meaningful angles. As the focus of the study was changes to foot-surface interaction via varying degrees of cushioning and somatosensory feedback, only angles recorded during ground contact were used. The hip, knee, ankle and rearfoot angles at TD, peak (excluding hip) and TO were analysed. Additionally, to identify foot strike modality the foot angle at TD was calculated. The instant of heel off was determined from the kinematic data. This would provide information about the time spent loading the forefoot, and thus the metatarsals, during propulsion. The final kinematic variable measured was vertical oscillation, defined by the maximum vertical displacement of the hip marker during one gait cycle.

Data from the GM and BF electrodes could not be used for two participants due to electrode movement. Within the Delsys hardware the EMG signal (sampled at 2000 Hz) was amplified and bandpass filtered (20-500 Hz). Offline analysis was performed using a customized Matlab (Math Works Inc., Cambridge, MA, USA) script. EMG data was base-line adjusted and underwent full-wave rectification. To determine muscle on-off times the rectified data was submitted to a nonlinear Teager-Kaiser energy operator (TKEO) (Hortobagyi et al. 2009, Li et al. 2007). Individual muscle on-off thresholds were determined by trialing a range of thresholds between 3 and 25% of the peak muscle activity during the six steps. The TKEO data had to rise and stay above each threshold for at least 50 ms. These muscle on-off times were then compared to manually derived thresholds, and specific cut-offs were chosen for each muscle.
The rectified muscle activity data for each muscle was low-pass filtered with a recursive filter and cut-off frequency of 6 Hz. The filtered EMG data was then integrated (iEMG) during preactivity (100 ms prior to TD) and stance. The iEMG was then normalized to the iEMG of each muscle across five gait cycles during the SH. Normalisation for the coactivation calculation was similar to the above, but average EMG amplitude rather than iEMG was used. The normalised EMG data were then entered into the coactivation calculation, previously used in gait studies (Franz & Kram 2012, Peterson & Martin 2010). A total of three agonist-antagonist pairs were analysed, two from the thigh (RFBF and VLBF) and one from the shank (GLTA), using equation (2).

\[
\text{\text{Normalized EMG}} = 2 \left( \frac{\int_{\text{EMG}_{1} - \text{EMG}_{2}}}{\text{EMG}_{1} + \text{EMG}_{2}} \right) \times 100
\]

EMG\textsubscript{1} and EMG\textsubscript{2} denote the two sets of EMG data used to calculate muscular coactivation and min refers to the minimum of the two sets of EMG.

2.5 Statistical analysis
Intra-class correlations (ICC) demonstrated strong within-subject reliability over the six strides (ICC range: .904 - .987). Therefore, each individual’s EMG measures and kinematics were expressed as the mean of the six strides recorded for each condition. Two one-way, repeated measures ANOVA were performed on RE. The first consisted of each condition with their RE values adjusted to SH mass. The second consisted of MS mass adjusted RE values, which included the BFT and both MS conditions. Post-hoc analysis was conducted using paired T-tests. For those conditions that exhibited significant differences in RE, the EMG measures and kinematics were compared using paired T-tests or non-parametric Wilcoxon signed rank tests, with statistical significance set at p \leq 0.05. This allowed for direct comparison of running mechanics only in those conditions where there were differences in RE. Additionally, effect sizes (ES) were calculated for all RE comparisons, with the following criteria: small (0.2 – 0.49), medium (0.5 – 0.79) and large (\geq 0.8) (Cohen 1992).

3. Results
The average BFT stride length was significantly shorter by 2.58% than the SH stride length (2.02 ± 0.10 vs. 2.08 ± 0.09 m), indicating a significantly higher stride frequency (165 ± 8 vs. 160 ± 7 steps-min\textsuperscript{-1}, BFT and SH respectively). BFT running had a significantly lower \(\dot{V}\text{O}_2\) than the SH condition (ES = 0.52, p = 0.002) and MS with BFT stride length (ES = 0.33, p = 0.014), but not MS with SH stride length (ES = 0.24, p = 0.062). SH running had a similar \(\dot{V}\text{O}_2\) to MS with BFT stride length running (ES = 0.14, p = 0.168), but greater \(\dot{V}\text{O}_2\) than MS with SH stride length running (ES = 0.26, p = 0.010). There was no difference between the two MS conditions (Table 1). SH running had a significantly higher \(\dot{V}\text{E}\) than BFT (4.8%), MS with BFT stride length (3.7%) and MS with SH stride length (2.7%). There was no change in HR across conditions.

There was no change in iEMG preactivation of any of the lower limb muscles. During stance iEMG of two muscles, BF and TA, exhibited significant differences across conditions (Table 2). Furthermore GLTA
Coactivation was ~ 12% lower during MS with BFT stride length than BFT. There were several kinematic variables that were found to change across conditions, with the lowest vertical oscillation, plantarflexion at TO and lowest peak dorsiflexion (Figure 1) and eversion recorded during BFT running (Table 2). Impact acceleration was highest during BFT running (7.0 ± 0.8 g), with impact acceleration remaining consistent across the other conditions (MS with BFT stride length: 5.6 ± 1.0 g, MS with SH stride length: 5.7 ± 0.9 g and SH: 5.5 ± 0.6 g).

Measurement of initial foot angle provided data on the footstrike modality employed for each condition. Based on the classification of Altman and Davis (forefoot striking: foot angle < -1.6°, rearfoot striking: foot angle > 8°, and midfoot striking: -1.6° < foot angle < 8°) BFT running resulted in the greatest number of forefoot strikers (7) and the least amount of rearfoot strikers (3). The MS conditions demonstrated consistent distributions of strike patterns [MS with SH stride length: 4, 5, 6 (forefoot, midfoot, rearfoot respectively) and MS with BFT stride length: 4, 6, 5]. The greatest number of rearfoot strikers were seen in the SH condition (10) together with the least amount of midfoot and forefoot strikers (2 and 3 respectively). The latest occurrence of heel off was observed during SH running, with BFT and MS running demonstrating similar timings that were on average up to 22 ms (~12.7%) earlier than SH.

4. Discussion

This study aimed to assess the mechanisms behind changes in RE during running at different stride lengths when varying cushioning and somatosensory feedback. Supporting our first hypothesis, the most economical condition was BFT. The MS running was only more economical than SH when running with a SH stride length, which partially supports our second hypothesis. Yet there was no significant difference in RE between the two MS conditions, thus contradicting our second hypothesis. This suggests therefore, that when footwear and speed are kept constant stride length deviations by approximately 3% do not affect RE.

Perl and colleagues (2012) found a similar metabolic advantage of MS over SH running when instructing runners to maintain a SH stride length, but suggested a greater advantage may have been recorded with higher stride frequencies (shorter stride lengths). Contrastingly our data suggest that if shorter stride lengths (i.e. BFT stride lengths) were taken then Perl et al. (2012) may not have found any economical advantage. On the other hand, several studies fail to find a significant difference in RE between BFT, MS and SH conditions (Burkett et al. 1985, Franz et al. 2012, Squadrone & Gallozzi 2009). Suggested reasons for these disparities include: a lack of familiarisation with BFT running (Burkett et al. 1985); added shoe mass affecting running mechanics (Franz et al. 2012); or failure to account for differences in footwear mass (or lack of) when calculating relative \( \dot{V}O_2 \) (Squadrone & Gallozzi 2009).

4.1 Mechanisms behind changes in RE

A significant reduction in vertical oscillation was observed in the BFT condition compared to the SH condition (6.8%). Previous studies have advocated a low vertical oscillation for economical running (Anderson 1996, Saunders et al. 2004), predominantly based on reported trends for the most economical runners to have the lowest vertical oscillation and gait manipulation studies showing that increasing a runners vertical oscillation is detrimental to their RE (Tseh et al. 2008, Williams & Cavanagh 1987). Furthermore, a lower vertical oscillation
has been associated with a reduced mechanical cost of centre of mass displacement (Slawinski & Billat 2004), which is likely to be due to performing less work against gravity. Therefore when runners were BFT, minimizing their vertical motion enabled them to perform less work against gravity, which could have contributed to the lower metabolic cost (Grabowski et al. 2005). Additionally, reducing vertical oscillation by decreasing the vertical displacement has been argued as more beneficial for RE improvements than trying to reduce the absolute height of the centre of mass (Halvorsen et al. 2012). The participants in the current study achieved a decrease in vertical displacement for BFT through a reduced rotation of the lower leg during stance, and thus a lower peak ankle dorsiflexion, whilst maintaining consistent peak knee flexion compared to the MS and SH conditions. This suggests a higher centre of mass during stance for BFT running and a stiffer, less compliant leg. Greater leg stiffness has previously been associated with a lower metabolic cost of running, and appears to be a characteristic of BFT running compared with SH running (Divert et al. 2005, Heise & Martin 1998).

The observed reduction in plantarflexion at TO when BFT compared to SH and MS is suggested as another economical running strategy adopted by runners in this study. This implies that runners were trying to ‘push-up’ to a lesser degree. Previous research has shown that less plantarflexion at TO is a factor in economical running (Moore et al. 2012), possibly due to a greater proportion of propulsive force being directed forward, a feature of BFT running (Paquette et al. 2013). This highlights the importance of the propulsive phase of stance during economical running, whilst previous research has tended to focus on the initial contact and impact phase (Gruber et al. 2013b, Perl et al. 2012).

The findings support suggestions that the initial foot angle (or footstrike) does not directly affect RE (Perl et al. 2012), as MS with SH stride length had a similar foot angle at TD to SH running. Furthermore both BFT and MS with BFT stride length have no external heel lift and produce similar foot angles at TD, implying that foot angle is not just a consequence of having different heel heights (Gruber et al. 2013a). Rather the observed change in foot angle for BFT to one that contributes more to plantarflexion results from shorter stride lengths. By adopting a shorter stride length it is easier to position the foot closer to the line of the centre of gravity and initially strike the ground more anteriorly on the foot. Interestingly, the shorter stride lengths appear more influential than heightened somatosensory feedback at altering foot angle, as BFT and MS with BFT stride length do not differ significantly. Recently, it has been reported that surface characteristics contribute to foot strike patterns with rearfoot strike patterns being maintained on soft surfaces (Gruber et al. 2013a). Together with our results, this suggests that foot inclination is a combination of both stride length and surface characteristics. With cushioned shoes, a foot angle contributing more to dorsiflexion is demonstrated, but with the removal of cushioning, the same foot angle is only true with long stride lengths, demonstrating it is these two factors combined that contribute to strike pattern.

The high TA iEMG exhibited during MS with SH stride length and BFT running, consistent with previous research (von Tscharner et al. 2003), does not appear to be directly detrimental to the metabolic cost of running. BFT running was the most economical condition and only when running with a SH stride length was MS running more economical than SH running. Although this disagrees with the metabolic cost of cushioning hypothesis (Frederick 1984), it is likely, especially during the BFT condition, that less cushioning is needed due
to the reduced vertical oscillation. Therefore whilst the increase in TA iEMG may not affect the metabolic cost, it could be increasing the mechanical cost which has been shown to be higher when BFT than when SH (Divert et al. 2008). Further research is needed to examine the mechanical and metabolic cost of BFT running in combination with muscular activity, to understand whether there are direct changes in efficiency and economy based on muscular activation strategies.

When removing the cushioning layer (SH to BFT/MS) there were no changes to muscular preactivity in any of the muscles. It has been argued that to adjust to impact changes, such as the removal of cushioning, runners use an anticipatory strategy termed ‘muscle tuning’ to gear the leg for impact (Boyer & Nigg 2004, 2007). Even though no statistical difference in preactivity was found, there were changes in initial TD kinematics at the knee and foot. This suggests that different leg geometries can be achieved with similar muscular activations and that both kinematic and EMG data are needed to understand how runners gear the leg for impact. It must be noted that there was no change in TD ankle angle, which is contrary to previous reports (De Wit et al. 2000, Squadrone & Gallozzi 2009). This is possibly a result of adequate barefoot treadmill familiarisation being given to runners prior to experimental testing in the current study, since TD ankle angle was found to change over this familiarisation period (Moore 2013).

Our final hypothesis was also only partially supported, as only BFT running exhibited greater impact accelerations whilst MS running demonstrated similar values to SH running. The high impact accelerations observed during BFT running may have been a result of a potential reduction in effective mass. The knee was more flexed and foot angled more towards the ground (toes lower than heel) during BFT running. Both these factors, in isolation, have been shown to reduce the effective mass (Derrick 2004, Lieberman et al. 2010). Together, along with the removal of the footwear mass, these adjustments imply that the BFT condition had the lowest effective mass. If effective mass denominates in the relationship between impact force and acceleration, as explained by Derrick (2004), then you would expect impact acceleration to increase and impact force to potentially decrease, as observed previously (Lieberman et al. 2010, Squadrone & Gallozzi 2009). High impact accelerations have been shown to lead to increased energy absorption at the hip, knee and ankle (Derrick et al. 1998). Greater energy absorbed via eccentric contractions can lead to greater energy released during concentric contractions, enhancing the efficiency of the stretch-shortening cycle. This implies that when increasing somatosensory feedback (MS to BFT) and removing external cushioning (SH to BFT) performance improvements (enhanced RE) are prioritised over reducing high impact acceleration and possible injury risk.

Comparing the MS with SH stride length condition to SH supports this notion that performance improvements are prioritised over injury risk. By removing the external cushioning layer of the SH condition, the knee became more extended at TD. Similar changes in leg geometry have been observed with increases in surface hardness (Hardin et al. 2004). These authors argued that this was an attempt to minimise metabolic cost at the expense of increased impact shock. However this was not the case for the MS conditions as impact acceleration was similar to SH. This most likely resulted from a similar impact force present even with differing external cushioning (Clarke et al. 1983, Nigg et al. 1987) and a similar effective mass, with a lower footwear mass countering the effects of increased knee extension.
Influence on injury risk

The two conditions where RE was improved compared to SH also had the highest TA iEMG. Alterations to TA activity, which is responsible for regulating impact loading (Christina et al. 2001), could reduce its capabilities to protect the bone (Mizrahi et al. 2000), and be the cause of the shin discomfort reported by runners transitioning to MS running in a recent study (Giandolini et al. 2013). The increase in TA activity is consistent with previous research (von Tscharner et al. 2003), and together with the GLTA coactivation, may help stabilize and increase ankle stiffness. Such a coactivation strategy has been documented in women with a history of falls, who look to increase their stability during walking (Marques et al. 2013). Furthermore, this may be a response to, or the cause of, the small amount of pronation (low peak eversion and dorsiflexion) during BFT running. Previous findings have reported similar differences in eversion (Paquette et al. 2013, Stacoff et al. 1991), which suggest the subtalar joint is more stable when BFT, but may be a concern for mechanical shock absorption. It could be argued that rather than a compliant ankle, with relatively high amounts of pronation absorbing the impact, the muscles in the lower limb act as shock absorbers. MS with BFT stride length displays low coactivation and TA activity, so optimal ankle stiffness and stability may not have been achieved which may explain why there was greater BF activity in the MS with BFT stride length condition to aid lower limb stability. This may be to the detriment of RE as evidence shows the BF to be positively related with \( \dot{V}O_2 \), meaning greater levels of activity during stance are related to higher metabolic costs (Kyrolainen et al. 2001).

The implications on bone stress and muscle strain of the greater TA activity reported in the economical conditions should not be overlooked. Greater muscle activity of the TA would lead to greater compressive stress on the anterior side of the tibia and greater tensile stress on the posterior side (assuming all else is equal/unchanged). Such muscle imbalances are potentially harmful as bone is weaker in tension than in compression (Reilly & Burstein 1975). Thus by trying to maintain joint stiffness to aid performance, the removal of a cushioning layer without shorter strides (MS with SH stride lengths) could lead to muscular imbalances across the tibia and greater tensile stress. Similarly, the removal of cushioning with shorter strides and heightened somatosensory feedback (BFT) could produce muscle imbalances and tensile stress.

Another factor that warrants consideration is the timing of heel off. By lifting the heel off earlier in the contact phase, as observed during BFT and MS conditions, longer time is spent loading the metatarsals. This, in addition to the forefoot being more angled towards the ground (BFT and MS with BFT stride length) suggesting a forefoot strike pattern, is a potential concern for injuries such as metatarsal stress fractures (Giuliani et al. 2011, Ridge et al. 2013). The forefoot would be loaded twice during each stance period (at initial footstrike and in midstance) and for a significantly longer time period. This suggests that runners who have previously experienced forefoot injuries should be cautious about running BFT/MS, regardless of the potential performance improvements.

The acute alterations to running mechanics due to changes in footwear that affect muscular activity and metatarsal and impact loads identified in this study highlight the need for an appropriate transition period to MS or BFT running. Merely gradually increasing running volume does not appear to adequately transition an individual to MS running due to increases in bone edema (Ridge et al. 2013). This, together with the current
findings, suggests that a more diverse transition programme is needed, which perhaps encompasses strengthening exercises targeting the smaller muscles around the ankle as these appear crucial for both performance (increasing reaction time) and the protection (reducing joint forces) of the ankle joint (Nigg 2005).

4.3 Limitations

It must be acknowledged that by mathematically adjusting for shoe mass we are technically adding mass to the centre of mass, rather than to the lower limb. Evidence suggests that adding mass to the foot increases $\dot{V}O_2$ to a greater degree than adding mass to the centre of mass (Myers & Steudel 1985). Yet, when the raw RE values are compared the percentage change from SH to MS, SH to BFT and MS to BFT all are greater than would be expected by the mass effect, which is generally considered to be 1% $\dot{V}O_2$ increase per 100g added to the foot (Frederick 1984). For example, for SH to BFT, the average shoe mass was 223g, therefore only a 2.23% change in RE would be expected. However, our results show a 5.83% change in raw RE. Additionally, the difference in SH to MS mass was 85g, yet the change in raw RE was 2.85%. Thus the results identified in this study cannot purely down to the removal of shoe mass and it is possible that self-optimisation through modifications in running mechanics may explain the extra improvements in RE which are unexplained by the mass effect. Furthermore, the mathematical adjustment does not directly account for the effect of the change in leg moment of inertia during both BFT and MS running. Previous results show that the effect on the energetic cost of walking is slightly higher when changing leg mass than leg moment of inertia (4 and 3.4% increase respectively) (Royer & Martin 2005). Therefore it is possible that the potential effect of decreasing leg moment of inertia on RE was encompassed in the calculation to adjust for the decrease in mass. Yet on the other hand, the adjusted RE could have slightly under/overestimated the overall effect of running BFT or MS on the metabolic cost of running. However, given the current level of understanding, the method utilised in the current study was thought to be the most suitable approach for accounting for different shoe mass.

When switching between conditions, the markers affixed to the foot whilst barefoot had to be re-affixed to the upper and heel of both the minimalist and traditional shoe. A review has argued that using shoe kinematics to infer foot motion is inappropriate (Arnold & Bishop 2013). However it was deemed the best approach to take, as affixing the markers to the foot in the shod conditions requires interfering with the structural integrity of the shoe (Arnold & Bishop 2013), which could possibly affect running mechanics. Additionally, as markers were re-affixed after changes in footwear a new set of standing angles were recorded for each condition, meaning dynamic angles could be normalized to the specific marker set-up.

Generalizing the results to overground running should be done with caution, as it is possible that the level of cushioning offered by the treadmill was optimal for BFT running but in fact when combined with the shoe cushioning was too much for the SH condition. Recent evidence has shown that increasing treadmill cushioning using 10mm foam improves the metabolic power of BFT running, yet using 20mm foam does not improve metabolic power (Tung et al. 2013). This suggests that there is an optimal level of cushioning which can be beneficial for running. However, if the results were merely due to the treadmill cushioning then both MS conditions would have an improved RE compared to SH, which was not the case.
5. Conclusions

BFT running offers a small economical advantage over MS with BFT stride length and SH running. Therefore heightening somatosensory feedback and removing an external cushioning layer (SH/MS to BFT) may produce slight performance benefits. To a lesser degree the same is true when just removing cushioning (SH to MS), but only when a SH stride length is maintained. However, when footwear is kept constant deviations in stride length by 3% do not appear to influence RE. Several changes in running mechanics were observed which may have contributed to the improved RE, with BFT running producing the lowest vertical oscillation and plantarflexion at TO. Notwithstanding this evidence, there were also alterations in muscular activity, foot inclination, impact accelerations and heel off occurrence suggesting that BFT running can potentially influence injury risk.

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References


Table 1. Physiological results for each condition.

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT</th>
<th>MS with BFT stride length</th>
<th>MS with SH stride length</th>
<th>SH</th>
</tr>
</thead>
<tbody>
<tr>
<td>RE adjusted to SH mass (ml·kg⁻¹·min⁻¹)</td>
<td>35.77 ± 3.70ᴬ</td>
<td>37.16 ± 4.19</td>
<td>36.73 ± 3.67ᴬ</td>
<td>37.71 ± 3.74ᴮ,ᴰ</td>
</tr>
<tr>
<td>(\dot{V}_\text{CO}_2) adjusted to SH mass (ml·kg⁻¹·min⁻¹)</td>
<td>32.61 ± 3.94ᴬ</td>
<td>33.62 ± 4.59ᴬ</td>
<td>34.03 ± 4.28</td>
<td>34.68 ± 4.21ᴮ,ᴰ</td>
</tr>
<tr>
<td>RER adjusted to SH mass</td>
<td>0.91 ± 0.04</td>
<td>0.90 ± 0.03</td>
<td>0.92 ± 0.04</td>
<td>0.92 ± 0.04</td>
</tr>
<tr>
<td>(\dot{V}_E) adjusted to SH mass (ml·kg⁻¹·min⁻¹)</td>
<td>1049.654 ±</td>
<td>1062.16 ±</td>
<td>1077.87 ±</td>
<td>1102.28 ±</td>
</tr>
<tr>
<td>RE adjusted to MS mass (ml·kg⁻¹·min⁻¹)</td>
<td>35.67 ± 3.69</td>
<td>37.06 ± 4.18ᴮ</td>
<td>36.63 ± 3.65</td>
<td>NA</td>
</tr>
<tr>
<td>(\dot{V}_\text{CO}_2) adjusted to MS mass (ml·kg⁻¹·min⁻¹)</td>
<td>32.52 ± 3.92</td>
<td>33.64 ± 4.60</td>
<td>34.19 ± 4.30</td>
<td>NA</td>
</tr>
<tr>
<td>RER adjusted to MS mass</td>
<td>0.91 ± 0.04</td>
<td>0.91 ± 0.03</td>
<td>0.93 ± 0.04</td>
<td>NA</td>
</tr>
<tr>
<td>(\dot{V}_E) adjusted to MS mass (ml·kg⁻¹·min⁻¹)</td>
<td>1046.65 ±</td>
<td>1061.60 ±</td>
<td>1073.04 ±</td>
<td>NA</td>
</tr>
<tr>
<td>HR (beats·min⁻¹)</td>
<td>166 ± 15</td>
<td>167 ± 13</td>
<td>167 ± 16</td>
<td>168 ± 15</td>
</tr>
</tbody>
</table>

ᴬ denotes significantly different to SH. ᴮ denotes significantly different to BFT. ᶜ denotes significantly different to MS with BFT stride length. ᴰ denotes significantly different to MS with SH stride length.
Table 2. Kinematic and EMG variables that were significantly different between conditions.

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT</th>
<th>MS with BFT stride length</th>
<th>MS with SH stride length</th>
<th>SH</th>
</tr>
</thead>
<tbody>
<tr>
<td>TD foot angle (°)</td>
<td>-3.79 ± 16.12(^{A})</td>
<td>-1.1 ± 17.75</td>
<td>4.41 ± 13.89</td>
<td>4.97 ± 17.61(^{B})</td>
</tr>
<tr>
<td>Peak dorsiflexion (°)</td>
<td>-9.61 ± .014(^{A,C})</td>
<td>-16.68 ± 5.50(^{B})</td>
<td>-16.94 ± 6.00</td>
<td>-20.28 ± 5.80(^{B})</td>
</tr>
<tr>
<td>Plantarflexion at TO (°)</td>
<td>11.31 ± 7.17(^{A,C})</td>
<td>15.29 ± 9.07(^{B})</td>
<td>17.21 ± 6.46</td>
<td>18.59 ± 8.49(^{B})</td>
</tr>
<tr>
<td>TD knee flexion (°)</td>
<td>12.35 ± 4.14(^{X})</td>
<td>9.90 ± 3.46(^{X,R,D})</td>
<td>8.26 ± 2.96(^{A,C})</td>
<td>11.35 ± 2.65(^{D})</td>
</tr>
<tr>
<td>Peak eversion (°)</td>
<td>-3.13 ± 2.82(^{A,C})</td>
<td>-5.18 ± 3.43(^{B})</td>
<td>-6.04 ± 4.00</td>
<td>-5.40 ± 3.39(^{B})</td>
</tr>
<tr>
<td>Heel off (ms)</td>
<td>151 ± 15(^{A})</td>
<td>155 ± 14(^{A})</td>
<td>160 ± 12(^{A})</td>
<td>173 ± 18(^{B,C,D})</td>
</tr>
<tr>
<td>Vertical oscillation (cm)</td>
<td>8.96 ± 2.06(^{A})</td>
<td>9.26 ± 2.08</td>
<td>9.33 ± 2.29</td>
<td>9.61 ± 1.95(^{B})</td>
</tr>
<tr>
<td>TA activity (a.u)</td>
<td>72.4 ± 50.5(^{A})</td>
<td>58.9 ± 32.9</td>
<td>72.7 ± 50.5(^{A})</td>
<td>56.2 ± 32.0(^{B,D})</td>
</tr>
<tr>
<td>BF activity (a.u)</td>
<td>29 ± 23.1(^{C})</td>
<td>38.6 ± 29.6(^{B})</td>
<td>34.9 ± 39.1</td>
<td>28.1 ± 17</td>
</tr>
<tr>
<td>GLTA (%)</td>
<td>61.63 ± 13.89(^{C})</td>
<td>49.35 ± 18.23(^{B})</td>
<td>58.22 ± 17.42</td>
<td>55.96 ± 13.15</td>
</tr>
</tbody>
</table>

\(^{A}\) denotes significantly different to SH. \(^{B}\) denotes significantly different to BFT. \(^{C}\) denotes significantly different to MS with BFT stride length. \(^{D}\) denotes significantly different to MS with SH stride length. \(^{X}\) denotes nearing significantly (p=.055). a.u. represents arbitrary units.
Figure 1. Representative sample ankle angle data for each condition. Positive values indicate plantarflexion and negative values indicate dorsiflexion.