©Journal of Sports Science and Medicine (2023) **22**, 512-525
http://www.jssm.org DOI: https://doi.org/10.52082/jssm.2023.512

Research article

An Acute Transition from Rearfoot to Forefoot Strike does not Induce Major Changes in Plantarflexor Muscles Activation for Habitual Rearfoot Strike Runners

Bálint Kovács [1,2]✉, Leonidas Petridis [3], János Négyesi [4], Örs Sebestyén [2], Ye Jingyi [1] Jingfeng Zhang [5], Yaodong Gu [1] and József Tihanyi [2]

[1] Ningbo University, Faculty of Sport Science, Ningbo, China; [2] Department of Kinesiology, Hungarian University of Sport Sciences, Budapest, Hungary; [3] Research Centre for Sport Physiology, Hungarian University of Sport Sciences, Hungary; [4] Division of Biomedical Engineering for Health and Welfare, Tohoku University Graduate School of Biomedical Engineering, Sendai, Japan; [5] University of Chinese Academy of Sciences, Department of Radiology, HwaMei Hospital, Ningbo, China

Abstract

Footstrike pattern has received increased attention within the running community because there is a common belief that forefoot strike running (FFS) is more advantageous (i.e., improve performance and reduce running injuries) than rearfoot strike running (RFS) in distance running. Literature reports suggest greater knee joint flexion magnitude and initial knee angle during stance in FFS compared with RFS running We examined the EMG activation of the triceps surae muscles during an acute transition from RFS to FFS strike. We tested the hypothesis that due to larger knee flexion in FFS the gastrocnemius muscles possibly decrease their EMG activity because muscle fascicles operate under unfavorable conditions. Fourteen competitive healthy middle- and long-distance runners who were habitual RFS runners ran on a treadmill at three speeds: 12, 14, and 16 km·h^{-1}. Each running speed was performed with both FFS and RFS patterns. Lower limb kinematics in the sagittal plane and normalized electromyography (EMG) activity of medial gastrocnemius proximal, middle and distal regions, lateral gastrocnemius and soleus muscles were compared between footstrike patterns and running speeds across the stride cycle. Contrary to our expectations, the knee joint range of motion was similar in FFS and RFS running. However, the sagittal plane ankle joint motion was greater ($p < 0.01$) while running with FFS, resulting in a significantly greater muscle-tendon unit lengthening ($p < 0.01$) in FFS compared with RFS running. In addition, medial and lateral gastrocnemius showed higher EMG activity in FFS compared with RFS running in the late swing and early stance but only for a small percentage of the stride cycle. However, strike patterns and running speed failed to induce region-specific activation differences within the medial gastrocnemius muscle. Overall, well-trained RFS runners are able to change to FFS running by altering only the ankle joint kinematics without remarkably changing the EMG activity pattern.

Key words: Running biomechanics, forefoot strike pattern, rearfoot strike pattern, EMG, joint kinematics, muscle mechanics.

Introduction

Footstrike pattern during running has received increased attention within the running community, mainly because it has been related to injury risk and running performance (Ahn et al., 2014; Hamill and Gruber, 2017; Hasegawa et al., 2007; Lin et al., 2021; Perl et al., 2012; Stearne et al., 2014; Yong et al., 2020; 2014). Three main types of footstrike patterns have been recognized according to the initial contact of the foot with the ground: rearfoot strike (RFS), forefoot strike (FFS), and midfoot strike which is often joined with FFS and referred to as non-rearfoot strike. Elite runners use FFS on long track events e.g., 10000m (Hanley et al., 2021) and on road races (de Almeida et al., 2015; Hanley et al., 2021; 2019; Hasegawa et al., 2007) yet the majority of sub-elite long-distance runners mostly use RFS. Typically, runners use their preferred strike pattern consistently and are commonly characterized as either habitual FFS or RFS runners (Hanley et al., 2021; 2019).

The plantarflexor muscle-tendon unit (MTU) spans the ankle joint hence its function is associated with footstrike pattern in running. FFS running is often characterized by a greater ankle dorsiflexion range of motion which likely induces increased lengthening of the triceps surae muscle-tendon unit (Nunns et al., 2013; Yong et al., 2020, 2014) and greater knee flexion during the stance phase (Kuhman et al., 2016; Lieberman et al., 2010; Nigg, 1988; Nunns et al., 2013; Valenzuela et al., 2014; Yong et al., 2020). However, different footstrike patterns seem to be associated with muscle-specific changes in activation and mechanics (Ahn et al., 2014; Yong et al., 2020; 2014). For example, Yong et al.(2014) found lower soleus tendon energy storage and fiber work but higher gastrocnemius muscle activation and fiber work in FFS than in RFS running suggesting an increase of the stored strain energy during FFS running due to the larger tendon excursion. In addition, FFS runners activate their plantarflexor muscles earlier and for a longer period compared with RFS runners (Ahn et al., 2014), which may, in turn, enhance the amount of stored elastic strain energy in the Achilles tendon (Perl et al., 2012). Nonetheless, others found no differences in peak Achilles tendon force and loading between FFS and RFS running however, during FFS running peak tendon force occurs earlier and the tendon is subjected to a higher load (Almonroeder et al., 2013). There is a lack of direct evidence that the Achilles tendon actually stores and reuses more elastic strain energy in FFS running (Ahn et al., 2014; Almonroeder et al., 2013; Kelly et al., 2018; Perl et al., 2012; Swinnen et al., 2018; Wearing et al., 2019). Addi-

Received: 23 February 2023 / Accepted: 14 August 2023 / Published (online): 01 September 2023

tionally, running economy has been reported either similar for both FFS and RFS runners (Gruber et al., 2013) which may increase the energy consumption of the muscle's contractile elements, or superior in RFS running at a slow to moderate speed (Ogueta-Alday et al., 2014). It is reasonable to think that in FFS running a greater magnitude of force production is needed to lengthen the tendon, which may increase the energy consumption of the muscle's contractile elements. This increased muscle's energy consumption might be similar in magnitude to the assumed additional strain energy return which consequently may not lower the net energy cost of running (Fletcher and MacIntosh, 2017).

Besides ankle kinematics, muscle activation is also affected by the degree of knee flexion during the stance phase. Gastrocnemius muscles originate above the knee therefore knee flexion will most likely influence the operational length of this muscle. Whereas the soleus origin is below the knee thus the knee joint displacement does not influence it. In FFS running, during the early stance phase there may be a greater knee flexion compared with RFS (Ahn et al., 2014; De Wit et al., 2000) and this may result in an unfavorable contractile condition for gastrocnemius muscles to operate, which can consequently decrease the efficiency of muscle contraction (Yong et al., 2020). In this case, medial (MG) and lateral gastrocnemius (LG) fascicles are likely to operate at the ascending region of the force-length relationship since there is a higher knee flexion in FFS running, resulting in a shorter operating fascicle length (Yong et al., 2020). To counteract this effect gastrocnemii muscles may increase muscle tension via greater activation which can be an alternative explanation for their higher EMG activity during FFS. Another possibility to compensate for the assumed decreased gastrocnemii muscle tension is if the soleus muscle increases its contribution (i.e., muscle tension) to the torque generation by increasing the elastic strain energy in the tendon (Kovács et al., 2020; Lai et al., 2018).

Acute changes from RFS to FFS pattern and its effects on the plantarflexor mechanics are not well understood. In the present study, we examined muscle activity of the plantarflexor muscles in relation to the strike pattern and running speed. Previous studies applied running protocols only with one predefined speed (Ahn et al., 2014; Yong et al., 2020; 2014), and obtaining muscle EMG activity at one location to compare FFS and RFS running characteristics. Based on literature reports footstrike pattern seems to be speed dependent suggesting that when running at higher speeds, runners are more likely to use FFS (Forrester and Townend, 2015; Pizzuto et al., 2016). In addition, faster running speeds are expected to increase the EMG activity of the triceps surae muscles in both forefoot or rearfoot strike patterns. However, limited data exist, on how acute changes in striking pattern and running speed alter joint kinematics and muscle activity. It is unclear if changing to FFS from RFS has any advantage in acute transition (for example during repeated intermittent running with relatively short distances), therefore it is important to understand better the changes occurring in the lower leg induced by a different footstrike pattern. The purpose of this study was to examine the kinematic changes of lower limb joints and muscle activation of the triceps surae muscles after an acute transition from RFS to FFS run at three different running speeds. MG is a biarticular muscle and it acts as a knee flexor beside plantarflexion. It has been shown that the proximal and distal regions of the MG muscle selectively recruit motor units during different tasks probably corresponding to their functional roles (Hodson-Tole et al., 2013; Watanabe et al., 2021) thus we assume that during FFS running a greater knee flexion range of motion can influence MG function. More specifically we assume that the proximal region, which is closer to the knee joint, is more affected by the knee rotation during the stance phase than the other regions of the MG. Accordingly, we expect lower EMG activity in the proximal region of the MG during FFS running. Additionally, a greater lengthening of the muscle tendon unit will likely increase EMG activity of the distal region of the MG. We hypothesize that SOL activity will follow changes in the EMG activity at the proximal region of MG, i.e., for lower activation of the MG proximal region SOL activity will increase, or for increased activation at the MG proximal region SOL activity will remain unchanged to compensate the unfavorable contractile condition.

Methods

Participants

Fourteen competitive middle- and long-distance self-reportedly habitual RFS runners (mean ± SD age 26.4 ± 5.9 years, body height 1.77 ± 0.08 m, and body mass 63.4 ± 6.2 kg, 10 male and 4 female) participated in this study. All runners had participated in both national and international level competitions and had an average training volume of 80 - 200 km per week. All participants were healthy and regularly exercised with no musculoskeletal injury or pain in the lower extremities at the time of the measurements. All participants gave written informed consent to take part in the study, which was performed in accordance with the Declaration of Helsinki and was approved by the ethics committee of the Hungarian University of Sports Science (TE-KEB/No22/2019).

Experimental protocol

First, participants ran 5 min at a self-selected speed on a treadmill. Then, similarly to previous protocols (Ahn et al., 2014; Kelly et al., 2018; Wessbecher and Ahn, 2019; Yong et al., 2014) they performed 30-s running trials at three different running speeds (12 $km \cdot h^{-1}$, 14 $km \cdot h^{-1}$, 16 $km \cdot h^{-1}$) using FFS and RFS techniques with five min of rest between trials. The participants wore their own shoes during all conditions. During the running trials, footstrike pattern was confirmed visually by the principal investigator (B K) and was analyzed afterwards via video from the lateral view of the foot as previously described (Valenzuela et al., 2014). RFS was defined as when the heel is the first region to contact the ground and FFS was defined as when the ball of the foot contacts the ground prior to the heel. For each trial, kinematic data and EMG activity of the dominant (right) lower leg muscles were simultaneously recorded during running. Ten consecutive strides across a 10 s data collection period at the end of each running trial for each running condition (RFS and FFS) were averaged and then included in data analysis.

Data collection and analysis

Joint kinematics were estimated from 19 skin-attached retro-reflective markers placed on the right lower limbs following the Helen-Hayes model. Marker trajectories were recorded in three dimensions sampling at 100 Hz with a motion capture system (OptiTrack Flex 13, NaturalPoint, OR, USA). To estimate segment lengths, joint centers, and joint coordinates and to optimize the model, a static standing pose was recorded before recording marker displacements during running. The marker trajectories were smoothed with a 10-Hz low-pass Butterworth filter (Hegyi et al., 2019). The angular displacement of the knee and ankle joints was calculated for each stride using inverse kinematics in Opensim (SimTK v. 4.0.1.) (Delp et al., 2007; Seth et al., 2018). The neutral ankle joint is 0° and it was considered as 90° between the foot and shank. Neutral knee was considered 0° when the angle between the shank and thigh was 180°. A custom Matlab script was used to separate motion into gait cycles and to determine foot contact duration as described elsewhere (Pálya and Kiss, 2018). Briefly, at the beginning of the gait cycle the distance between the hip and heel is the greatest in the sagittal plane (initial foot contact). Thus, we located the maximal distance between the marker set on the iliac crests and the marker set on the lateral ankle in the horizontal axis. The maximum distance in the opposite direction was also located (toe off). Locating these maximum values helped us to determine the gait cycles and divide them into subphases. Initial contact was then defined as the maximal distance when the heel is in front of the hip, whereas toe off as the maximal distance when the heel is behind the hip (Pálya and Kiss, 2018). Sagittal plane ankle and knee angles were exported and used to calculate MG, LG and SOL MTU lengths relative to shank length in Matlab (MathWorks, Natick, MA, USA). MTU lengths were computed with the following equation:

$$L_{MTU} = C0 + C2\alpha + C4\beta$$

where L_{MTU} is the normalized MTU length, C0, C2 and C4 are the regression coefficients (see in Hawkins and Hull 1990), and α and β are the ankle and knee angles in degree. Joint angular data were time-normalized to the strides cycle (1 - 101 points), then these strides were averaged for each running speed and footstrike pattern.

After skin preparation, surface EMG electrodes (Blue Sensor M-00-S/25, Ambu, Denmark, 20 mm interelectrode distance, 10 mm diameter) were placed on the calf muscles, and EMG activity was recorded using TeleMyo telemetric hardware system (Noraxon U.S. Inc., Scottsdale, Az, USA) at a sampling frequency of 1 kHz. MG was divided into three parts of equal size between the origin and insertion point of the muscle (Figure. 1.). Then we placed the electrodes over the middle of each region (proximal, middle, and distal). In the case of the LG and SOL electrode placement was followed by the SENIAM recommendations (Hermens et al., 2000). Ultrasonography was used to ensure that electrodes were aligned parallel with the fascicle orientation and to minimize cross-talk between the observed and the neighboring muscles. A reference electrode was placed on the ipsilateral patella. EMG cables were taped over the skin to minimize movement artifacts. A fourth-order zero-lag Butterworth filter (band-pass 20 - 450 Hz) was used to remove movement artifacts and signal noise from the raw EMG signals. To account for the time difference between the stride durations at different running speeds, the root mean square (RMS) window duration was adjusted to the duration of the stride, allowing a comparison of time-normalized curves across different running speeds (Mark Burden et al., 2014). For the longest stride during 12 km·h^{-1}, a 50 ms RMS window was applied, and then the RMS window was reduced proportionately according to the duration of the stride at a higher running speed. The EMG signal amplitudes were normalized to the mean of the peaks over 10 consecutive strides at 16 km·h^{-1} for each muscle.

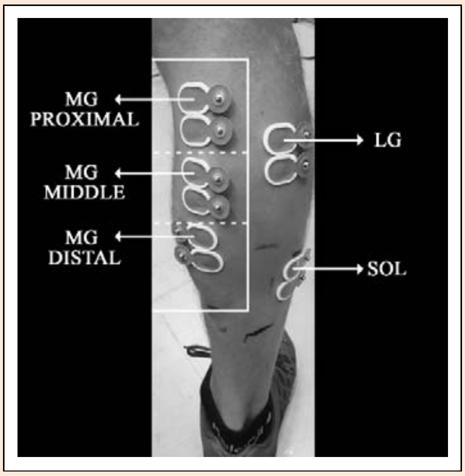


Figure 1. Installation of EMG electrodes on the right shank. Bipolar surface EMG electrodes were placed over the right medial gastrocnemius (MG) proximal, middle and distal region, lateral gastrocnemius (LG), and soleus (SOL) muscles.

Statistical analysis

To confirm the normal distribution of the data we used Statistical Parametric Mapping (SPM) normality test. SPM two-tailed paired t-tests were used to compare time-normalized kinematic and MTU length variables between FFS and RFS running in Matlab using the open-source spm1d code (SPM 30, v.0.4.8, www.spm1d.org). First, the scalar output (SPM{t}) was calculated, then the critical threshold (t*), at which only α% of smooth random curves are expected to be traversed. If the t-test statistic trajectory crossed the critical threshold, the difference was statistically significant. In order to retain a type 1 family-wise error rate considering multiple pairwise comparisons (adjustment was made for the three speeds) we used the Bonferroni correction setting α level at 0.017. To compare the EMG curves of plantarflexor muscles and MG regional activity, one-way repeated measures ANOVA SPM test was performed across the time-normalized running strides. First SPM{F} test statistics were calculated to test regional EMG interaction with speed. In case of interaction at any timepoint across the stride cycle, locations of the differences were tested using paired-sample t-tests with Bonferroni correction. Family-wise type I error rate was set at 0.05. SPM technical details are described elsewhere (Pataky et al., 2016a; 2016b).

Results

As it was expected, there was a significant difference in ankle joint angular displacement (i.e., greater dorsiflexion and plantarflexion in FFS) between FFS and RFS running at all speeds (0 - 9% $p < 0.001$, 21% - 41% $p < 0.001$, 73% - 100% $p < 0.001$ at 12 km·h^{-1}, 0 - 7% $p = 0.008$, 23% - 36% $p = 0.001$, 83% - 100% $p = 0.001$ for 14 km·h^{-1}, 0 - 10% $p = 0.001$, 23% - 41% $p < 0.001$, 81% - 100% $p < 0.001$ for 16 km·h^{-1}). On the other hand, the knee joint angle at ground contact and the knee range of motion during the entire stride cycle were almost identical during FFS and RFS running at each running speed (Figure 2). In accordance with the ankle joint rotation, MTU length changes were significantly greater for each MTU during FFS compared with RFS running at early and late stance and at the late swing as well (Figure. 2). The differences were significant in early stance and in the push-off phase and also from mid-swing to foot contact at each MTU (Figure 2.).

The comparison between FFS and RFS running showed different EMG activity in the investigated muscles (Figure 3). The MG proximal region showed higher EMG activity in FFS running at the early stance and late swing phase mainly at 14 km·h^{-1} (0 - 3%, $p = 0.002$, 84 - 100%, $p < 0.001$) and 16 km·h^{-1} (0 - 4%, $p < 0.001$, 96 - 100%, $p < 0.001$) running speeds and greater EMG activity during mid-swing in FFS running at 12 km·h^{-1} (80 - 86%, $p = 0.001$). The middle region of MG showed a similar pattern to the proximal region at 14 km·h^{-1} (0 - 3%, $p = 0.003$, 87 - 100%, $p < 0.001$) and 16 km·h^{-1} (0 - 2%, $p = 0.003$, 98 - 100%, $p = 0.005$) running speeds however, at 12 km·h^{-1} RFS had higher EMG activity in the middle regions during midstance phase (21 - 26%, $p = 0.005$). In the distal region, during early stance and late swing FFS showed higher EMG activity at all speeds (12 km·h^{-1}: 0 - 3%, $p = 0.004$, 98 - 100%, $p = 0.008$, 14 km·h^{-1}: 0 - 2%, $p = 0.003$, 93 - 100%, $p < 0.001$, 16 km·h^{-1}: 0 - 1%, $p = 0.007$, 64 - 67 $p = 0.003$, 95 - 100%, $p = 0.004$). Similarly in LG, FFS running resulted in higher EMG activity at the end of the stride (12 km·h^{-1}: 90 - 100%, $p < 0.001$, 14 km·h^{-1}: 93 - 100%, $p < 0.001$) and during the late push-off and late swing at 16 km·h^{-1} (38 - 43%, $p = 0.005$, 96%, $p = 0.002$). There was no difference in the EMG activity of SOL when running at all running speeds (Figure 3.).

The SPM {*F*} test on MG regional EMG activity (see in the Supplementary Figures 1, 2, 3, 4, 5, 6, and 7) showed no significant effects between regions at either measured running speed or footstrike pattern (Figure 4).

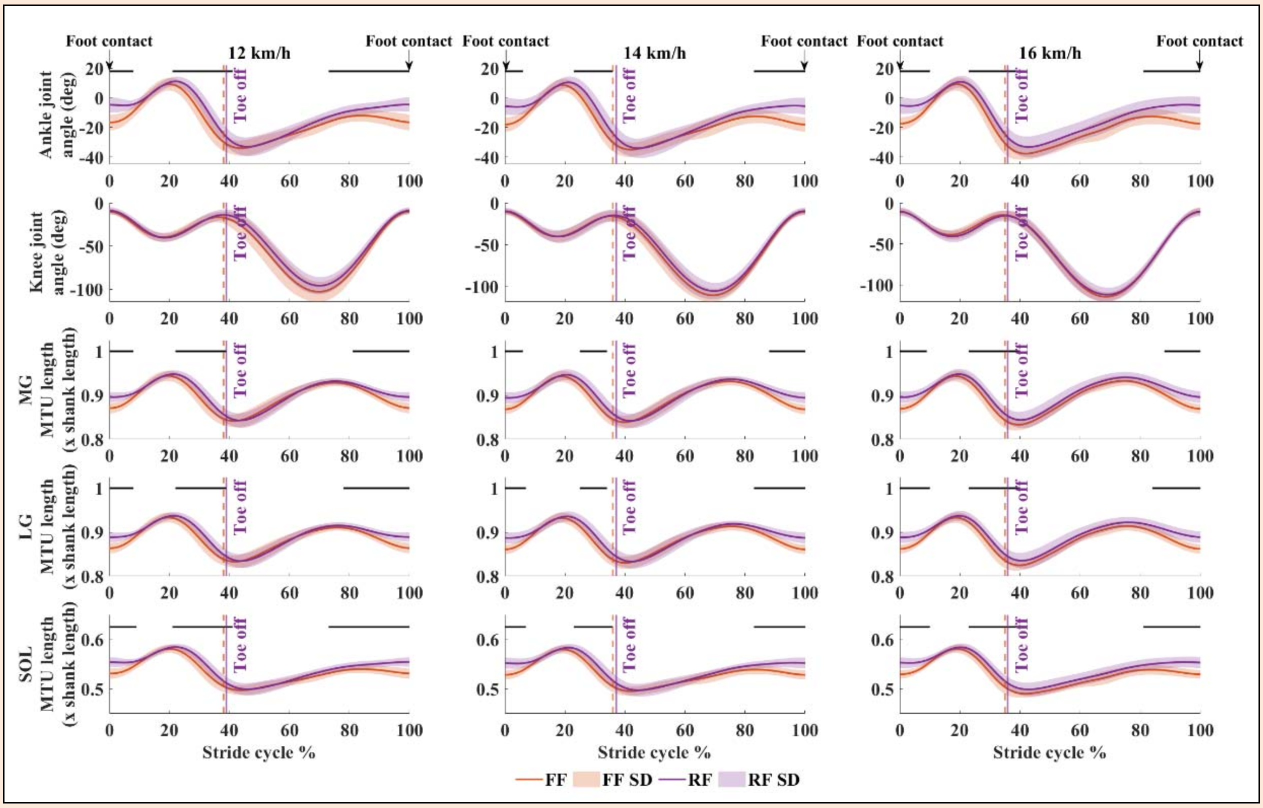


Figure 2. Ankle and knee joints displacement and plantarflexor MTU length changes. Group means and SD of the ankle and knee joint displacements and MTU length changes of medial (MG) and lateral gastrocnemius (LG) and soleus (SOL) during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The solid purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The black solid lines show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a black line.

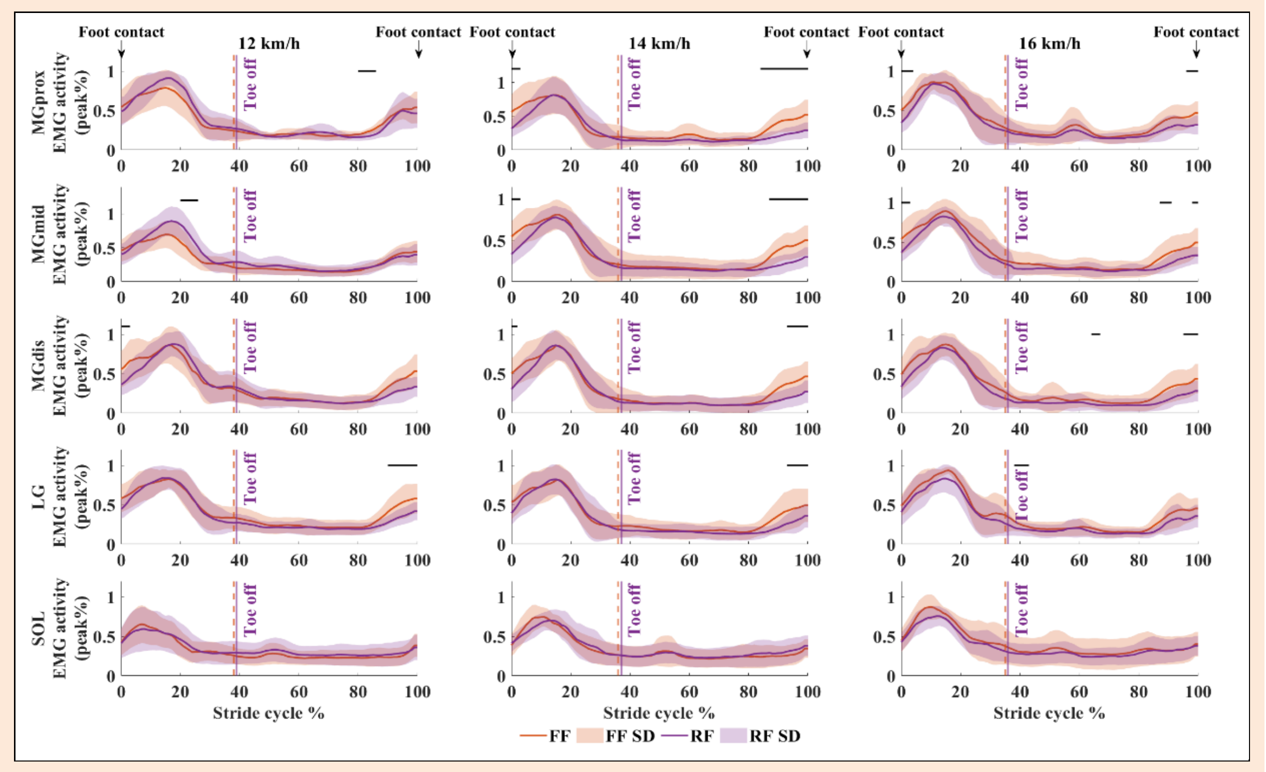


Figure 3. Plantarflexor EMG activity during forefoot and rearfoot strike running. Comparisons of group means and SD of the normalized electrical activity (all EMG channels averaged per muscle and normalized to maximum activity) at 12 km·h^{-1}, 14 km·h^{-1}, and 16 km·h^{-1} running speeds between forefoot (FFS) and rearfoot strike (RFS) running. Vertical dashed lines define the sub-phases of the running stride (0-100%) as stance phase (from foot strike at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The solid purple and orange vertical lines show the end of stance phase during FFS and RFS running, respectively. The black solid lines show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and was marked with a black line.

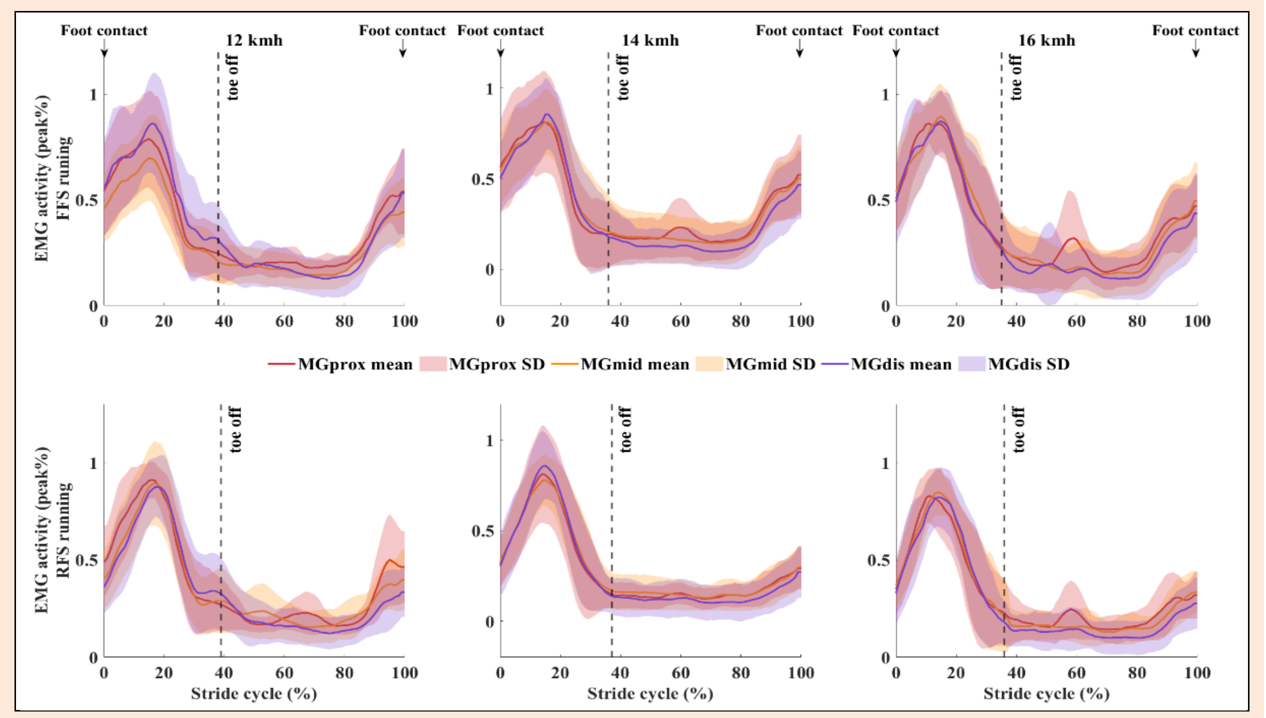


Figure 4. Regional EMG activity of medial gastrocnemius. Regional EMG activity (normalized to maximum EMG activity) of the medial gastrocnemius at different running speeds during forefoot and rearfoot strike running. Colored lines represent group means and SD across the stride cycle at different speeds for each region, upper panels during forefoot strike while lower panels during rearfoot strike running. The dashed black vertical line shows the end of the stance phase at the corresponding running speed. Statistically significant differences are marked with black horizontal lines above the EMG curves. Dashed line: differences between the proximal and middle region, solid line: differences between proximal to distal region, dotted line: differences between middle and distal region. MGprox - medial gastrocnemius proximal region, MGmid – medial gastrocnemius middle region, MGdis – medial gastrocnemius distal region, FFS – forefoot strike running, RFS – rearfoot strike running.

Discussion

The present study aimed to examine the kinematic changes of lower limb joints and muscle activation of the triceps surae muscles after an acute transition from RFS to FFS run at three different running speeds. Based on the preponderance of studies showing greater knee flexion during FFS (Kuhman et al., 2016; Lieberman et al., 2010; Nigg, 1988; Nunns et al., 2013; Valenzuela et al., 2014; Yong et al., 2020) and region-specific activation pattern of the MG (Watanabe et al., 2021), we hypothesized (1) a lower EMG activity at the proximal region of the MG during FFS running, attributed to the greater knee flexion and (2) a higher EMG activity at the distal portion. Contrary to our expectations, kinematic changes were not observed at the knee but only at the ankle in well-trained, habitually RFS runners after an acute transition to FFS at different running speeds. MG and LG muscles showed higher EMG activity in FFS than in RFS before and after initial ground contact for a short percentage of the stride cycle while SOL showed no difference in EMG activity between footstrike patterns at any running speeds.

Our results correspond with previous investigations that switching from RFS to FFS running increased sagittal plane ankle joint motion. On the other hand, and contrary to our assumption, there was no difference in knee joint rotation between FFS and RFS running, regardless of the running speed. Previous results indicated the possibility of greater knee flexion angle during FFS running (Ahn et al., 2014; De Wit et al., 2000), which would likely be suboptimal for the gastrocnemius muscles due to shorter fascicle operating lengths than their optimal lengths (Kovács et al., 2020). However, in our study sagittal knee joint kinematics was similar in both FFS and RFS running. This discrepancy between our result and the results of some previous studies can possibly be explained by long-term neuromuscular adaptations at the hip and knee joints as a result of their long training background with RFS running (Landreneau et al., 2014; Lin et al., 2021). Thus, the acute transition from RFS to FFS applied in this study did not result in changes in knee joint kinematics. Despite increased knee flexion angle in FFS as compared with RFS in some studies, plantarflexor MTUs lengthen more during FFS as compared with RFS running due to the greater ankle dorsiflexion (Sano et al., 2015; Takeshita et al., 2021; Yong et al., 2020). These studies compared habitual RFS runners to habitual FFS runners. The majority of increased MTU lengthening seems to occur rather in the Achilles tendon than in the muscle fascicles (Bohm et al., 2019; Lai et al., 2018; 2015; Sano et al., 2015; Takeshita et al., 2021). In the current study, where habitually RFS runners were instructed to acutely switch to FFS running, we also detected greater ankle joint dorsiflexion range of motion and also a greater MTU lengthening in all investigated MTUs when running with FFS compared with RFS running. The increased MTU lengthening induces greater mechanical load and stress on the Achilles tendon in FFS running (Almonroeder et al., 2013; Kernozek et al., 2018; Kulmala et al., 2013; Lyght et al., 2016), which is likely associated with higher muscle force production to lengthen the tendon. The need for higher force production in FFS running may be linked to higher muscle activation. This may be the reason why previous studies found higher EMG activation in MG when switching from RFS to FFS running (Ahn et al., 2014; Landreneau et al., 2014; Yong et al., 2020; 2014). Our results showed a higher MG and LG activity during early stance and late swing phase at most speeds, but only for a small percentage of the stride cycle in FFS running. We found that MTU lengthening was greater in FFS than in RFS running, but since the increase in muscle tension (i.e., EMG activity) was evident for a shorter period than that of MTU lengthening (at late swing and early stance phase), we can assume that the additional lengthening of the MTU occurs evenly in the muscle and tendon tissues, or even dominantly in the muscles during stance. At the same time, RFS runners have been shown to have different ground reaction force-time characteristics markedly different (i.e., first impact peak) from that of FFS running (Lieberman et al., 2010). Upon impact, the force development rate rapidly increases creating a muscle tension (higher EMG activity) that would stretch the tendon in FFS running due to the dorsiflexion occurring after ground contact (i.e., MTU lengthening), while in RFS running after initial ground contact, a slight plantarflexion occurs shortening slightly the MTU. In addition, before landing the ankle joint is already at a more dorsiflexed position in RFS in order to prepare for ground contact which magnifies the difference in MTU length at the beginning of foot contact. Without data on the magnitude of muscle and tendon elongation, we can only speculate how the total MTU lengthening occurred. In a recent study, Yong et al. (2020) suggested that higher MG activity during FFS running is associated with greater muscle and tendon forces allowing for greater tendon lengthening velocities. However, in our study, MG EMG activity was higher in FFS compared with RFS for a shorter period during the early stance than in the study of Yong et al. (2020) likely indicating moderate differences in muscle and tendon force between the two footstrike techniques.

Kelly et al. (2018) reported that SOL activity was similar across the entire stride cycle between the two strike patterns which we also observed in this study. The increased MTU lengthening in SOL during FFS running did not result in elevated EMG activity most probably because SOL fascicles may operate at similar lengths during different footstrike patterns and the additional length changes may occur in the Achilles tendon when running with FFS (Ahn et al., 2014; Wiesinger et al., 2017; Yong et al., 2020). Our findings indicate that acute change to FFS running did not increase the demand on the SOL as no changes in EMG activity were found. The observed muscle activity patterns in this study are consistent with previously reported differences in EMG activity between habitual RFS and FFS runners (Yong et al., 2014) and habitual RFS runners (Ahn et al., 2014; Yong et al., 2020). Based on the similar findings by Yong et al. (2020), we can assume that the negative work performed by the MG might be higher in FFS compared to RFS. In addition, the positive work of SOL might decrease possibly counteracting the increased MG work and this seems to be the case at each examined speed in this study. However, this assumption needs to be examined experimentally, since our study lacks the necessary data to support it. We also investigated the region-specific EMG activity of MG during FFS and RFS running initially assuming that changes in knee joint rotation will affect mainly the proximal region. Contrary to this expectation, we found no difference in muscle activity between MG regions, regardless of strike pattern. This might be explained by the lack of kinematic changes in knee joint rotation that would have affected the level of gastrocnemius muscle contraction considering that gastrocnemius muscles are biarticular muscles and their activation is closely related to the knee joint rotation and ankle rotation as well (Watanabe et al., 2021). Recently it was suggested that biarticular muscles show greater regional differences than monoarticular muscles (Watanabe et al., 2021). Therefore, we could not support the presence of regional differences within the MG because we failed to detect significant region-specific differences within MG.

Differences in joint kinematics and kinetics should be also considered in the interpretation of these results. For example, sagittal plane kinematics may differ as knee and ankle flexion range of motion can be smaller in treadmill running compared to overground running, but these differences seem to be consistent regardless of footstrike pattern (van Hooren et al., 2020). More importantly, muscle activity may be different between treadmill and overground running, which may have affected our EMG results. Irrespective of the footstrike pattern, joint kinetics have been reported to be lower on treadmill, due to the lack of whole-body acceleration. During overground running larger forces are needed to accelerate the body, which results in different force generation mechanism and accordingly muscle activity strategy (van Caekenberghe et al., 2013a; 2013b). Therefore, it may be possible that EMG activity during FFS and RFS strike may show a different pattern in overground running.

Also, changes in moment arm length may possibly explain our results. It has been documented that the Achilles tendon moment arm length depends on the joint angle (Fath et al., 2010; Maganaris et al., 1998; Wade et al., 2019). For instance, compared with the anatomically neutral position (Deforth et al., 2019), the moment arm lengthens in a plantarflexed position, while it shortens in a more dorsiflexed position. Thus, when running with FFS, the plantar-flexed ankle can increase the length of the Achilles tendon moment arm. In turn, this may lower the muscle force production while maintaining the same joint torque around the ankle, which may possibly explain the lack of an increase in EMG activity during the late stance phase in FFS running. Based on our results, a short-duration change in footstrike pattern from RFS to FFS does not induce substantial changes in the muscle activity level of the plantarflexor muscles with respect to stance duration. For example, changing from RFS to FFS for short-duration interval runs is not recommended at moderate running speeds (12-16 km·h^{-1}) since we did not find clear evidence that it would be beneficial compared to RFS. Also, the changes we found in MTU lengthening during FFS running may impose a higher load on the tendon increasing the risk of a repetitive stress injury (Pizzuto et al., 2016). Footstrike pattern seems to be influenced by running speeds (Pizzuto et al., 2016) as higher running speeds are often performed with FFS especially above 18 - 21.6 km·h^{-1} (Forrester and Townend, 2015; Pizzuto et al., 2016) thus it is possible that changing to FFS at this level might have benefits. It must be noted that the experimental protocol most likely did not induce fatigue, thus the results can be different in a case of a prolonged run or in the case of a high repetition interval run.

This study has some methodological limitations. We did not measure or calculate the length changes of the muscle-tendon unit separately, which could provide more valuable information about muscle-tendon interaction and possibly could support our assumption. It should also be stressed that the MTU length calculations were possibly affected by the model we used and the cutoff frequency we applied to the joint angular data. The MTU calculation method was developed based on male participants (Hawkins and Hull, 1990), who were, however similar in body height with both the male and female participants in our study. Although MTU behavior was not the main focus of this study these limitations must be considered when interpreting the results of this study. The applied footfall detection method we used in this study is applied for RFS running method and it may have some inaccuracy for FFS running which should be noted. Considering the kinematic data of the knee range of motion it is more likely that retracting the foot earlier in FFS may not have happened most of the time, thus the distance between the ankle and the iliac crest was similar for both strides. It is important to note, that this is true only for this study also considering the kinematic data of the knee range of motion. In addition, the retraction of the foot can also occur via earlier hip extension which we did not obtained in this study. In order to calculate joint kinetics with inverse dynamics in OpenSim ground reaction force data is required which was not available during the measurements. It must be highlighted that joint kinetic data would help us to understand better the findings of this study. It must be considered when interpreting the data that the small sample size raises some concerns about generalizing the conclusion. A detailed visualization of the magnitude of the differences may help to understand and interpret better our results therefore, we provided the results of the spm statistics in the supplementary materials. In addition, the participants ran for 30 s with each strike technique and each running speed. Although it is possible, that such a short duration of running may be inadequate to induce significant kinematic and kinetic changes, particularly in well-trained runners, previous studies also applied a short 30 s or not more than 2 min duration trials (Ahn et al., 2014; Kelly et al., 2018; Lin et al., 2021; Stearne et al., 2014; Wessbecher and Ahn, 2019; Yong et al., 2020, 2018). Nevertheless, future research should reconsider performing longer running intervals. Running speed is also an important aspect and must be taken into account when interpreting the results. The examined three running speeds (12 km·h^{-1}, 14 km·h^{-1}, and 16 km·h^{-1}) for this treadmill protocol were similar to previous studies (Ahn et al., 2014; Kelly et al., 2018; Yong et al., 2020; 2014). However, considering that the racing speed of the participants involved in this study is higher, our conditions represented a considerably lower load compared with a race-like situation. The applied running speeds in this study are commonly used speeds during training sessions,

thus our results reflect more on the training situation rather than on a race-like
situation.

Conclusion

In conclusion, our data suggest that kinematic changes are not evident in the knee but are evident in the ankle joint of well-trained habitual RFS runners after an acute transition to FFS at various running speeds. Although the MG and LG appeared to be more active in FFS at moderate speeds (12 - 14 km·h^{-1}), strike patterns and running speed failed to induce region-specific activation differences within the MG muscle. Therefore, we can conclude that well-trained RFS runners are able to change to FFS running by altering mainly the ankle but not the knee joint kinematics to adequately mimic the mechanical loading condition with elevated MG and LG EMG activity around ground contact.

Acknowledgements

The authors wish to thank all participants for volunteering in this study. The authors have no conflicts of interest to declare. The experiments complied with the current laws of the country in which they were performed. The datasets generated during and/or analyzed during the current study are not publicly available, but are available from the corresponding author who was an organizer of the study.

References

Ahn, A.N., Brayton, C., Bhatia, T. and Martin, P. (2014) Muscle activity and kinematics of forefoot and rearfoot strike runners. *Journal of Sport and Health Science* **3**, 102-112. https://doi.org/10.1016/j.jshs.2014.03.007

Almonroeder, T., Willson, J.D. and Kernozek, T.W. (2013) The effect of foot strike pattern on achilles tendon load during running. *Annals of Biomedical Engineering* **41,**1758–1766. https://doi.org/10.1007/s10439-013-0819-1

Bohm, S., Mersmann, F., Santuz, A. and Arampatzis, A. (2019) The force-length-velocity potential of the human soleus muscle is related to the energetic cost of running. *Proceedings of the Royal Society B: Biological Sciences* **286,** 20192560. https://doi.org/10.1098/rspb.2019.2560

Brughelli, M., Cronin, J. and Chaouachi, A. (2011) Effects of running velocity on running kinetics and kinematics. *Journal of Strength and Conditioning Research* **25,** 933-9. https://doi.org/10.1519/JSC.0b013e3181c64308

de Almeida, M.O., Saragiotto, B.T., Yamato, T.P. and Lopes, A.D. (2015) Is the rearfoot pattern the most frequently foot strike pattern among recreational shod distance runners? *Physical Therapy in Sport* **16**, 29–33. https://doi.org/10.1016/j.ptsp.2014.02.005

De Wit, B., De Clercq, D. and Aerts, P. (2000) Biomechanical analysis of the stance phase during barefoot and shod running. *Journal of Biomechanics* **33**, 269–278. https://doi.org/10.1016/S0021-9290(99)00192-X

Deforth, M., Zwicky, L., Horn, T. and Hintermann, B. (2019) The effect of foot type on the Achilles tendon moment arm and biomechanics. *Foot* **38**, 91–94. https://doi.org/10.1016/j.foot.2018.10.003

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E. and Thelen, D.G. (2007) OpenSim: Open-source software to create and analyze dynamic simulations of movement. *IEEE Transactions on Biomedical Engineering* **54**, 1940–1950. https://doi.org/10.1109/TBME.2007.901024

Fath, F., Blazevich, A.J., Waugh, C.M., Miller, S.C. and Korff, T. (2010) Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans. *Journal of Applied Physiology* **109**, 1644–1652. https://doi.org/10.1152/japplphysiol.00656.2010

Fletcher, J.R. and MacIntosh, B.R. (2017) Running economy from a muscle energetics perspective. *Frontiers in Physiology* **8**, 433. https://doi.org/10.3389/fphys.2017.00433

Forrester, S.E. and Townend, J. (2015) The effect of running velocity on footstrike angle - A curve-clustering approach. *Gait and Posture* **41**, 26–32. https://doi.org/10.1016/j.gaitpost.2014.08.004

García-Pinillos, F., Jerez-Mayorga, D., Latorre-Román, P., Ramirez-Campillo, R., Sanz-López, F. and Roche-Seruendo, L.E. (2020) How do Amateur Endurance Runners Alter Spatiotemporal Parameters and Step Variability as Running Velocity Increases? A Sex Comparison. *Journal of Human Kinetics* **72**, 39-49. https://doi.org/10.2478/hukin-2019-0098

Gruber, A.H., Umberger, B.R., Braun, B. and Hamill, J. (2013) Economy and rate of carbohydrate oxidation during running with rearfoot and forefoot strike patterns. *Journal of Applied Physiology* **115**, 194–201. https://doi.org/10.1152/japplphysiol.01437.2012

Hamill, J. and Gruber, A.H. (2017) Is changing footstrike pattern beneficial to runners? *Journal of Sport and Health Science* **6**, 146–153. https://doi.org/10.1016/j.jshs.2017.02.004

Hanley, B., Bissas, A., Merlino, S. and Gruber, A.H. (2019) Most marathon runners at the 2017 IAAF World Championships were rearfoot strikers, and most did not change footstrike pattern. *Journal of Biomechanics* **92**, 54–60. https://doi.org/10.1016/j.jbiomech.2019.05.024

Hanley, B., Tucker, C.B., Bissas, A., Merlino, S. and Gruber, A.H. (2021) Footstrike patterns and race performance in the 2017 IAAF World Championship men's 10,000 m final. *Sports Biomechanics* **10**, 1–10. https://doi.org/10.1080/14763141.2020.1856916

Hasegawa, H., Yamauchi, T. and Kraemer, W.J. (2007) Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *Journal of Strength and Conditioning Research* **21**, 888–893. https://doi.org/10.1519/R-22096.1.

Hawkins, D. and Hull, M.L. (1990) A method for determining lower extremity muscle-tendon lengths during flexion/extension movements. *Journal of Biomechanics* **23**, 487–494. https://doi.org/10.1016/0021-9290(90)90304-L

Hegyi, A., Gonçalves, B.A.M., Finni, T. and Cronin, N.J. (2019) Individual Region- and Muscle-specific Hamstring Activity at Different Running Speeds. *Medicine & Science in Sports & Exercise* **51**, 2274–2285. https://doi.org/10.1249/MSS.0000000000002060

Hermens, H.J., Freriks, B., Disselhorst-Klug, C. and Rau, G. (2000) Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology* **10**, 361–374. https://doi.org/10.1016/S1050-6411(00)00027-4

Hodson-Tole, E.F., Loram, I.D. and Vieira, T. (2013) Myoelectric activity along human gastrocnemius medialis: Different spatial distributions of postural and electrically elicited surface potentials. *Journal of Electromyography and Kinesiology* **23,** 43-50. https://doi.org/10.1016/j.jelekin.2012.08.003.

Kelly, L.A., Farris, D.J., Lichtwark, G.A. and Cresswell, A.G. (2018) The Influence of Foot-Strike Technique on the Neuromechanical Function of the Foot. *Medicine & Science in Sports & Exercise* **50**, 98–108. https://doi.org/10.1249/MSS.0000000000001420

Kernozek, T.W., Knaus, A., Rademaker, T. and Almonroeder, T.G. (2018) The effects of habitual foot strike patterns on Achilles tendon loading in female runners. *Gait and Posture* **66**, 283–287. https://doi.org/10.1016/j.gaitpost.2018.09.016

Kovács, B., Kóbor, I., Gyimes, Zs., Sebestyén, Ö. and Tihanyi, J. (2020) Lower leg muscle–tendon unit characteristics are related to marathon running performance. *Scientific Reports* **10**, 17870. https://doi.org/10.1038/s41598-020-73742-5

Kuhman, D., Melcher, D. and Paquette, M.R. (2016) Ankle and knee kinetics between strike patterns at common training speeds in competitive male runners. *European Journal of Sport Science* **16**, 433–440. https://doi.org/10.1080/17461391.2015.1086818.

Kulmala, J.P., Avela, J., Pasanen, K. and Parkkari, J. (2013) Forefoot strikers exhibit lower running-induced knee loading than rearfoot strikers. *Medicine & Science in Sports & Exercise* **45**, 2306–2313. https://doi.org/10.1249/mss.0b013e31829efcf7

Lai, A., Lichtwark, G.A., Schache, A.G., Lin, Y.-C., Brown, N.A.T. and Pandy, M.G. (2015) In vivo behavior of the human soleus muscle with increasing walking and running speeds. *Journal of Applied Physiology* **118**, 1266–1275. https://doi.org/10.1152/japplphysiol.00128.2015

Lai, A., Lichtwark, G.A., Schache, A.G. and Pandy, M.G. (2018) Differences in in vivo muscle fascicle and tendinous tissue behavior between the ankle plantarflexors during running. *Scandinavian Journal of Medicine & Science in Sports* **28**, 1828–1836. https://doi.org/10.1111/sms.13089

Landreneau, L.L., Watts, K., Heitzman, J.E. and Childers, W.L. (2014) Lower limb muscle activity during forefoot and rearfoot strike running techniques. *International Journal of Sports Physical Therapy* **9**, 888–97. PMID: 25540704

Lieberman, D.E., Venkadesan, M., Werbel, W.A., Daoud, A.I., Dandrea, S., Davis, I.S., Mangeni, R.O. and Pitsiladis, Y. (2010) Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature* **463**, 531–535. https://doi.org/10.1038/nature08723

Lin, J.Z., Chiu, W.Y., Tai, W.H., Hong, Y.X. and Chen, C.Y. (2021) Ankle muscle activations during different foot-strike patterns in running. *Sensors* **21**, 3422. https://doi.org/10.3390/s21103422

Lyght, M., Nockerts, M., Kernozek, T.W. and Ragan, R. (2016) Effects of foot strike and step frequency on Achilles tendon stress during running. *Journal of Applied Biomechanics* **32**, 365–372. https://doi.org/10.1123/jab.2015-0183

Maganaris, C.N., Baltzopoulos, V. and Sargeant, A.J. (1998) Changes in Achilles tendon moment arm from rest to maximum isometric plantarflexion: In vivo observations in man. *Journal of Physiology* **510**, 977–985. https://doi.org/10.1111/j.1469-7793.1998.977bj.x

Burden M., Lewis SE. and Willcox, E. (2014) The effect of manipulating root mean square window length and overlap on reliability, inter-individual variability, statistical significance and clinical relevance of electromyograms. *Manual Therapy* **19**, 595–601. https://doi.org/10.1016/j.math.2014.06.003

Nigg BM. and Bashlen.M. (1988) Influence of Heel Flare and Midsole Construction on Pronation Supination and Impact Forces for Heel-Toe Running. *Journal of Applied Biomechanics* **4**, 205–219. https://doi.org/10.1123/ijsb.4.3.205

Nunns, M., House, C., Fallowfield, J., Allsopp, A. and Dixon, S. (2013) Biomechanical characteristics of barefoot footstrike modalities. *Journal of Biomechanics* **46**, 2603–2610. https://doi.org/10.1016/j.jbiomech.2013.08.009

Ogueta-Alday, A., Rodríguez-Marroyo, J.A. and García-López, J. (2014) Rearfoot striking runners are more economical than midfoot strikers. *Medicine & Science in Sports & Exercise* **46**, 580–585. https://doi.org/10.1249/MSS.0000000000000139

Pálya Zs. and Kiss R.M. (2018) Az emelkedő teljesítményű futás hatása a mozgásparaméterekre. *Biomechanica Hungarica* **11**, 37–45. (In Hungarian: English abstract). https://doi.org/10.17489/2018/1/03

Pataky, T.C., Robinson, M.A. and Vanrenterghem, J. (2016a) Region-of-interest analyses of onedimensional biomechanical trajectories: Bridging 0D and 1D theory, augmenting statistical power. *PeerJ* **11**, 27833816. https://doi.org/10.7717/peerj.2652

Pataky, T.C., Vanrenterghem, J. and Robinson, M.A. (2016b) The probability of false positives in zero-dimensional analyses of one-dimensional kinematic, force and EMG trajectories. *Journal of Biomechanics* **49**, 1468–1476. https://doi.org/10.1016/j.jbiomech.2016.03.032

Perl, D.P., Daoud, A.I. and Lieberman, D.E. (2012) Effects of footwear and strike type on running economy. Medicine & Science in Sports & Exercise **44**, 1335–1343. https://doi.org/10.1249/MSS.0b013e318247989e

Pizzuto, F., Rago, V., Bailey, R., Tafuri, D. and Raiola, G. (2016) The importance of foot-strike patterns in running: A literature review. *Sport Science* **9**, 87-96.

Sano, K., Nicol, C., Akiyama, M., Kunimasa, Y., Oda, T., Ito, A., Locatelli, E., Komi, P. V. and Ishikawa, M. (2015) Can measures of muscle–tendon interaction improve our understanding of the superiority of Kenyan endurance runners? *European Journal of Applied Physiology* **115**, 849–859. https://doi.org/10.1007/s00421-014-3067-7

Seth, A., Hicks, J.L., Uchida, T.K., Habib, A., Dembia, C.L., Dunne, J.J., Ong, C.F., DeMers, M.S., Rajagopal, A., Millard, M., Hamner, S.R., Arnold, E.M., Yong, J.R., Lakshmikanth, S.K., Sherman, M.A., Ku, J.P. and Delp, S.L. (2018) OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. *PLOS Computational Biology* **14**, 1006223. https://doi.org/10.1371/journal.pcbi.1006223

Stearne, S.M., Alderson, J.A., Green, B.A., Donnelly, C.J. and Rubenson, J. (2014) Joint kinetics in rearfoot versus forefoot running: Implications of switching technique. *Medicine & Science in Sports & Exercise* **46**, 1578–1587. https://doi.org/10.1249/MSS.0000000000000254

Swinnen, W., Hoogkamer, W., de Groote, F. and Vanwanseele, B. (2019) Habitual foot strike pattern does not affect simulated triceps surae muscle metabolic energy consumption during running. *Journal of Experimental Biology* **22**, jeb212449. https://doi.org/10.1242/jeb.212449

Swinnen, W., Hoogkamer, W., Delabastita, T., Aeles, J., de Groote, F. and Vanwanseele, B. (2018) Effect of habitual foot-strike pattern on the gastrocnemius medialis muscle-tendon interaction and muscle force production during running. *Journal of Applied Physiology* **126**, 708–716. https://doi.org/10.1152/japplphysiol.00768.2018

Takeshita, T., Noro, H., Hata, K., Yoshida, T., Fukunaga, T. and Yanagiya, T. (2021) Muscle-tendon behavior and kinetics in gastrocnemius medialis during forefoot and rearfoot strike running. *Journal of Applied Biomechanics* **37**, 240–247. https://doi.org/10.1123/jab.2020-0229

Valenzuela, K.A., Lynn, S.K., Mikelson, L.R., Noffal, G.J. and Judelson, D.A. (2014) Effect of acute alterations in foot strike patterns during running on sagittal plane lower limb kinematics and kinetics. *Journal of Sports Science and Medicine* **14**, 225–232. PMID: 25729311

van Caekenberghe, I., Segers, V., Aerts, P., Willems, P. and de Clercq, D. (2013a) Joint kinematics and kinetics of overground accelerated running versus running on an accelerated treadmill. *Journal of the Royal Society Interface* **10**. 1-11. https://doi.org/10.1098/rsif.2013.0222

van Caekenberghe, I., Segers, V., Willems, P., Gosseye, T., Aerts, P. and de Clercq, D. (2013b) Mechanics of overground accelerated running vs. running on an accelerated treadmill. *Gait and Posture* **38,** 125-131. https://doi.org/10.1016/j.gaitpost.2012.10.022

van Hooren, B., Fuller, J.T., Buckley, J.D., Miller, J.R., Sewell, K., Rao, G., Barton, C., Bishop, C. and Willy, R.W. (2020) Is Motorized Treadmill Running Biomechanically Comparable to Overground Running? A Systematic Review and Meta-Analysis of Cross-Over Studies. *Sports Medicine* **50**, 785–813. https://doi.org/10.1007/s40279-019-01237-z

Wade, F.E., Lewis, G.S. and Piazza, S.J. (2019) Estimates of Achilles tendon moment arm differ when axis of ankle rotation is derived from ankle motion. *Journal of Biomechanics* **90**, 71–77. https://doi.org/10.1016/j.jbiomech.2019.04.032

Watanabe, K., Vieira, T.M., Gallina, A., Kouzaki, M. and Moritani, T. (2021) Novel Insights into Biarticular Muscle Actions Gained from High-Density Electromyogram. *Exercise and Sport Sciences Reviews* **49**, 179–187. https://doi.org/10.1249/JES.0000000000000254

Wearing, S.C., Davis, I.S., Brauner, T., Hooper, S.L. and Horstmann, T. (2019) Do habitual foot-strike patterns in running influence functional Achilles tendon properties during gait? *Journal of Sports Sciences* **37**, 2735–2743. https://doi.org/10.1080/02640414.2019.1663656

Wessbecher, L.S. and Ahn, A.N. (2019) Lower leg morphology in runners: forefoot strikers have longer heels but not bigger muscles than rearfoot strikers. *Journal of Experimental Biology* **222**, 1–4. https://doi.org/10.1242/jeb.187815

Wiesinger, H.P., Rieder, F., Kösters, A., Müller, E. and Seynnes, O.R. (2017) Sport-specific capacity to use elastic energy in the patellar and Achilles tendons of elite athletes. *Frontiers in Physiology* **8**, 132. https://doi.org/10.3389/fphys.2017.00132

Yong, J.R., Dembia, C.L., Silder, A., Jackson, R.W., Fredericson, M. and Delp, S.L. (2020) Foot strike pattern during running alters muscle-tendon dynamics of the gastrocnemius and the soleus. *Scientific Reports* **10**, 5872. https://doi.org/10.1038/s41598-020-62464-3

Yong, J.R., Silder, A. and Delp, S.L. (2014) Differences in muscle activity between natural forefoot and rearfoot strikers during running. *Journal of Biomechanics* **47**, 3593–3597. https://doi.org/10.1016/j.jbiomech.2014.10.015

Yong, J.R., Silder, A., Montgomery, K.L., Fredericson, M. and Delp, S.L. (2018) Acute changes in foot strike pattern and cadence affect running parameters associated with tibial stress fractures. *Journal of Biomechanics* **76**, 1–7. https://doi.org/10.1016/j.jbiomech.2018.05.017

Key points

- Acute transition to forefoot strike running style from rearfoot running style can be performed changing mainly the ankle joint range of motion, but not definitely the knee joint range of motion.
- Increased MTU lengthening during FFS running is not paired with elevated EMG activity indicating similar muscle tension as in RFS running, hence the additional lengthening probably occurs in the tendon and muscle evenly.
- Long-term adaptations to habitual rear-foot strike may explain the absence of changes in knee joint kinematics and EMG activation after an acute transition from rearfoot to forefoot strike.
- Both footstrike patterns (FFS and RFS), or the examined running speeds failed to induce region-specific EMG activation differences within the MG muscle.

AUTHOR BIOGRAPHY

Bálint KOVÁCS
Employment
Post-doctoral fellow at Ningbo University, Ningbo, China and Research assistant at Hungarian University of Sports Science, Budapest Hungary
Degree
PhD
Research interests
Biomechanics, Effect of lower leg structure on motor performance, Determinants of running economy
E-mail: k.balin828@gmail.com

Leondias PETRIDIS
Employment
Associate Professor at Hungarian University of Sports Science, Budapest Hungary
Degree
PhD
Research interests
Exercise testing and monitoring, training adaptation in young athletes
E-mail: petridis.leonidas@tf.hu

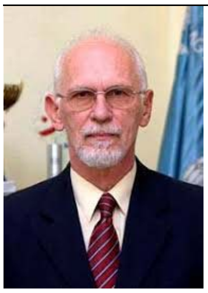
János NÉGYESI
Employment
Research assistant professor at Hungarian University of Sports Science, Fit4Race Kft., Budapest, Hungary, Department of Medicine and Science in Sports and Exercise, Tohoku University Graduate School of Medicine, Sendai, Japan Budapest Hungary
Degree
PhD
Research interests
The effects of sensory inputs on motor control and learning
E-mail: negyesi@tohoku.ac.jp

Örs SEBESTYÉN
Employment
PhD student at Hungarian University of Sports Science, Budapest Hungary
Degree
Msc in Human Kinesiology
Research interests
Biomechanics, Impact of muscle and tendon structure on muscle performance
E-mail: sebestyénors10@gmail.com

Ye JINGYI
Employment
Msc student at Ningbo University, Ningbo China
Degree
Msc in Physical Education
Research interests
Biomechanics, Changes in lower limb muscle parameters during exercise, The impact of footwear changes on the lower limbs of adolescents
E-mail: yejingyi1999@gmail.com

Jingfeng ZHANG
Employment
Radiology Department of HwaMei Hospital, University of Chinese Academy of Sciences
Degree
PhD
Research interests
Radiology, MRI imaging
E-mail: jingfengzhang73@zju.edu.cn

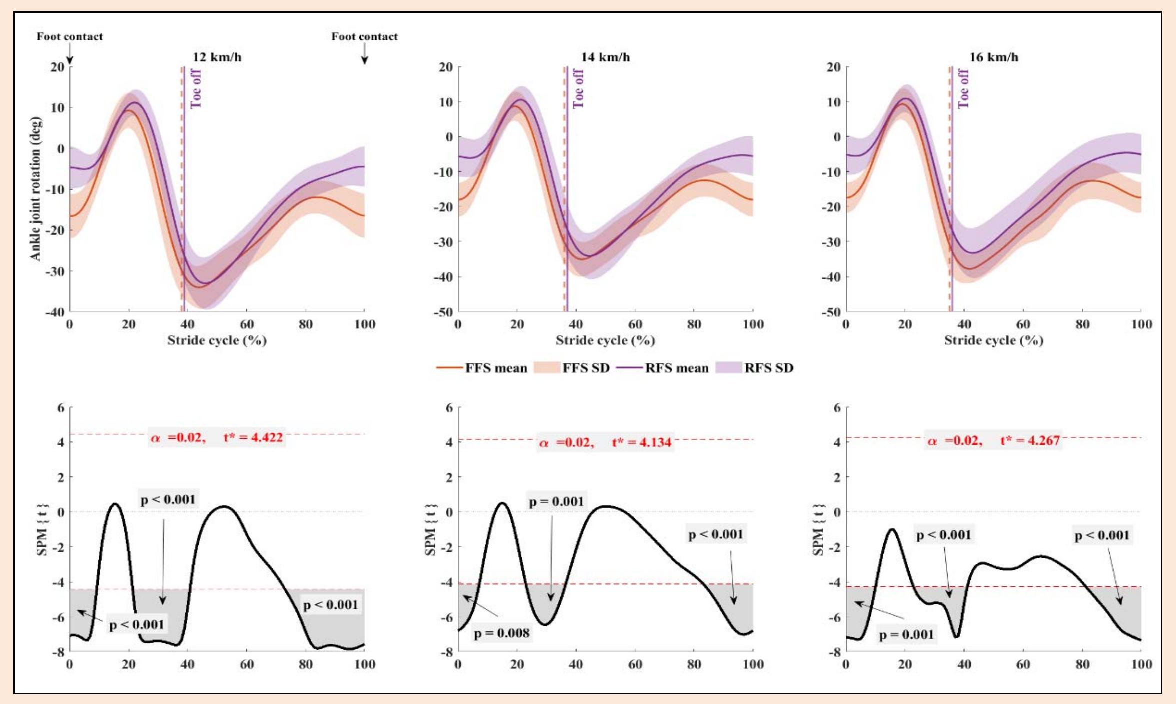
Gu YAODONG
Employment
Professor at Ningbo University, Ningbo China
Degree
PhD
Research interests
Biomedical Engineering; Sports Biomechanics; Foot biomechanics
E-mail: guyaodong@hotmail.com



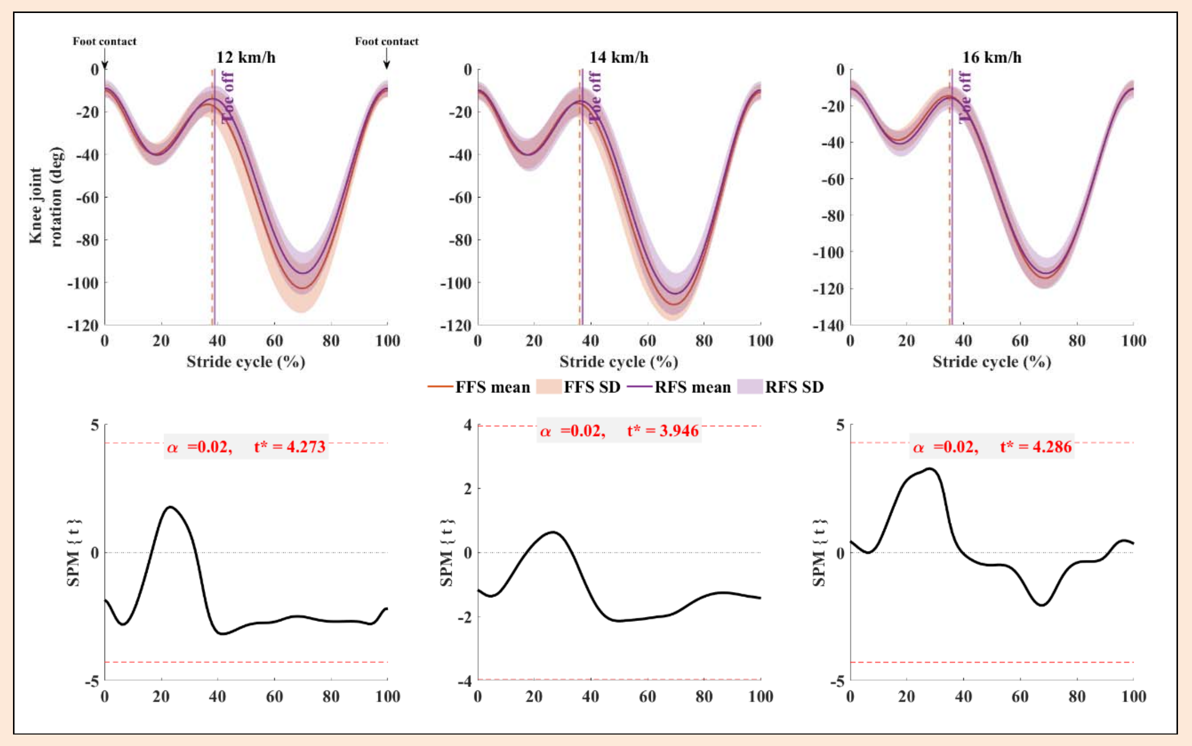
József TIHANYI
Employment
Professor, Hungarian University of Sports Science, Budapest Hungary
Degree
PhD, DSc
Research interests
Functional biomechanics. Biomechanics of the musculoskeletal system, the effect of lower leg structure on muscle function during locomotion, Eccentric exercise induced muscle damage, muscle fiber adaptation
E-mail: tihanyi.jozsef@tf.hu

✉ **Bálint Kovács, Gu Yaodong**
Faculty of Sports Science, Ningbo University, Ningbo China.

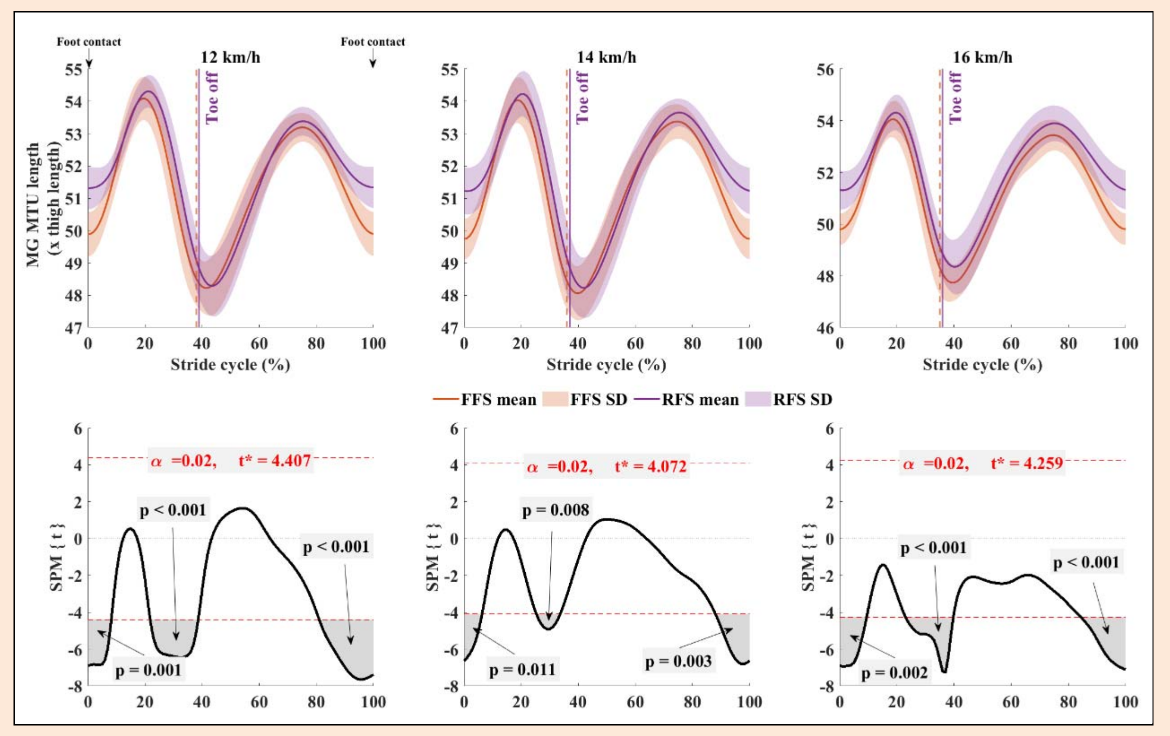
Supplementary Figures (SPM t-test results)



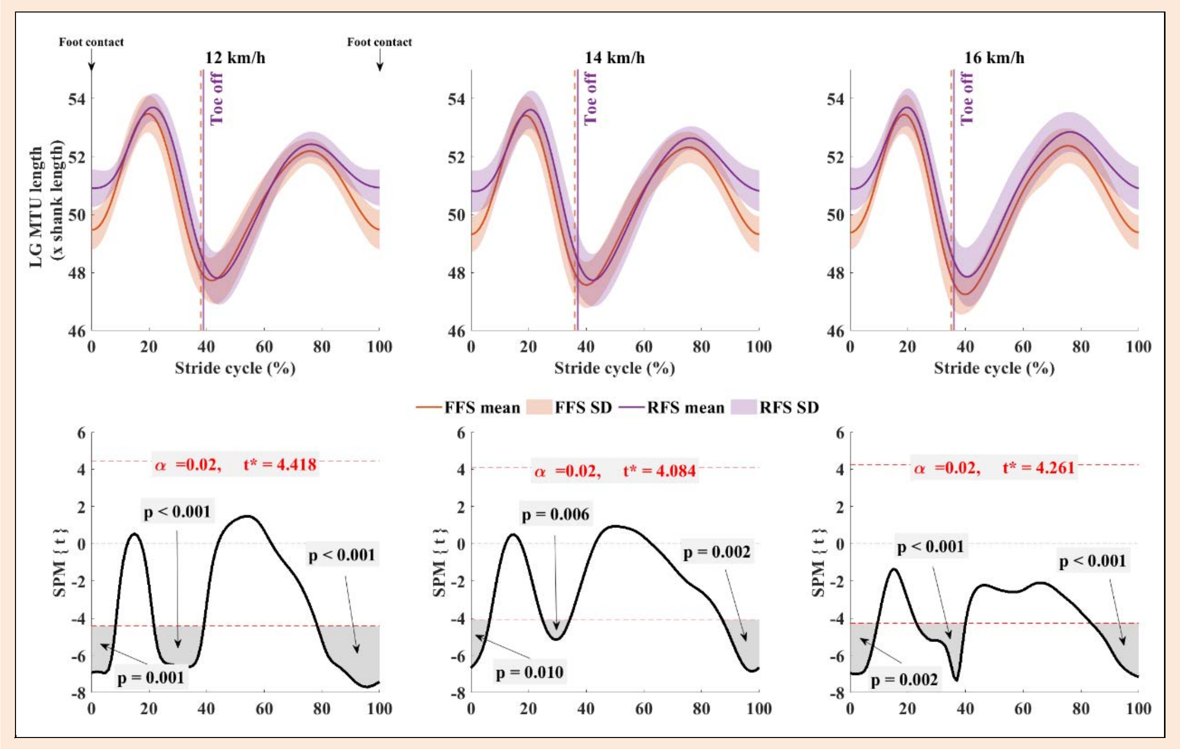
Supplementary Figure 1. Group means and SD of ankle joint displacements during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The dashed purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The panels below the joint displacement panels show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a grey shaded area. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



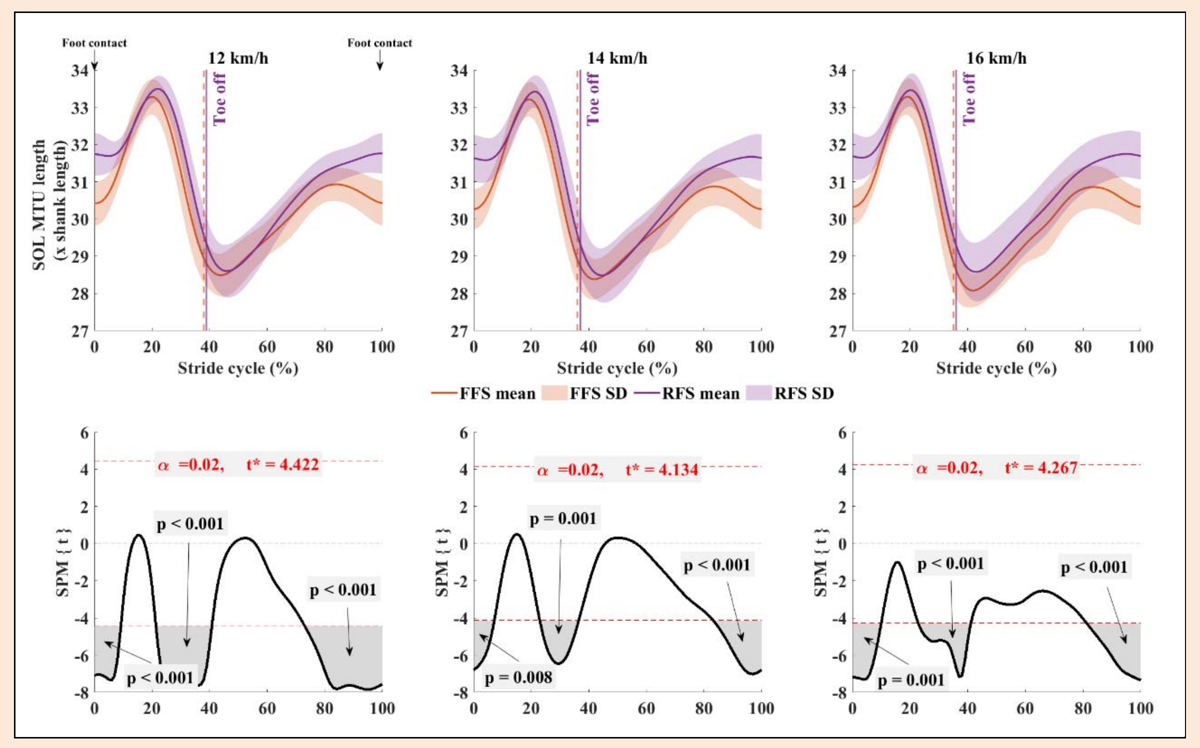
Supplementary Figure 2. Group means and SD of knee joint displacements during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The dashed purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The panels below the joint displacement panels show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a grey shaded area. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



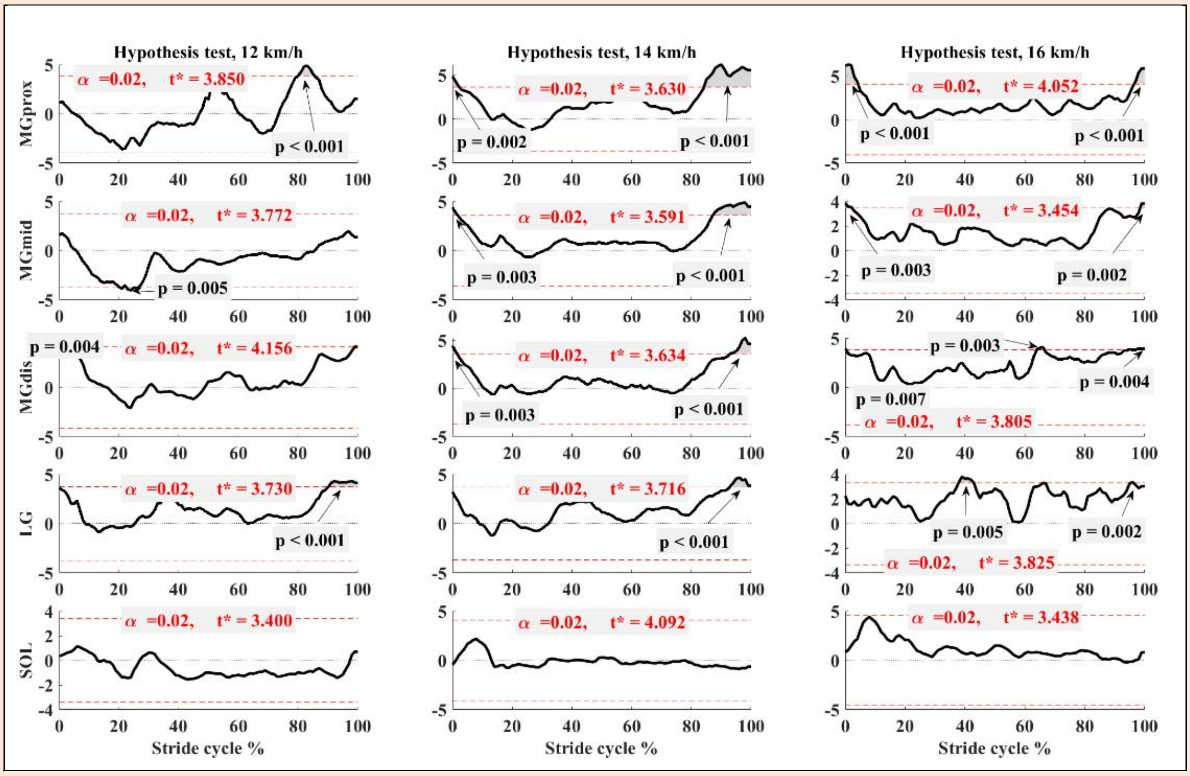
Supplementary Figure 3. Group means and SD of the MTU length changes of medial gastrocnemius (MG) during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The dashed purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The panels below the MTU length changes panels show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a grey shaded area. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



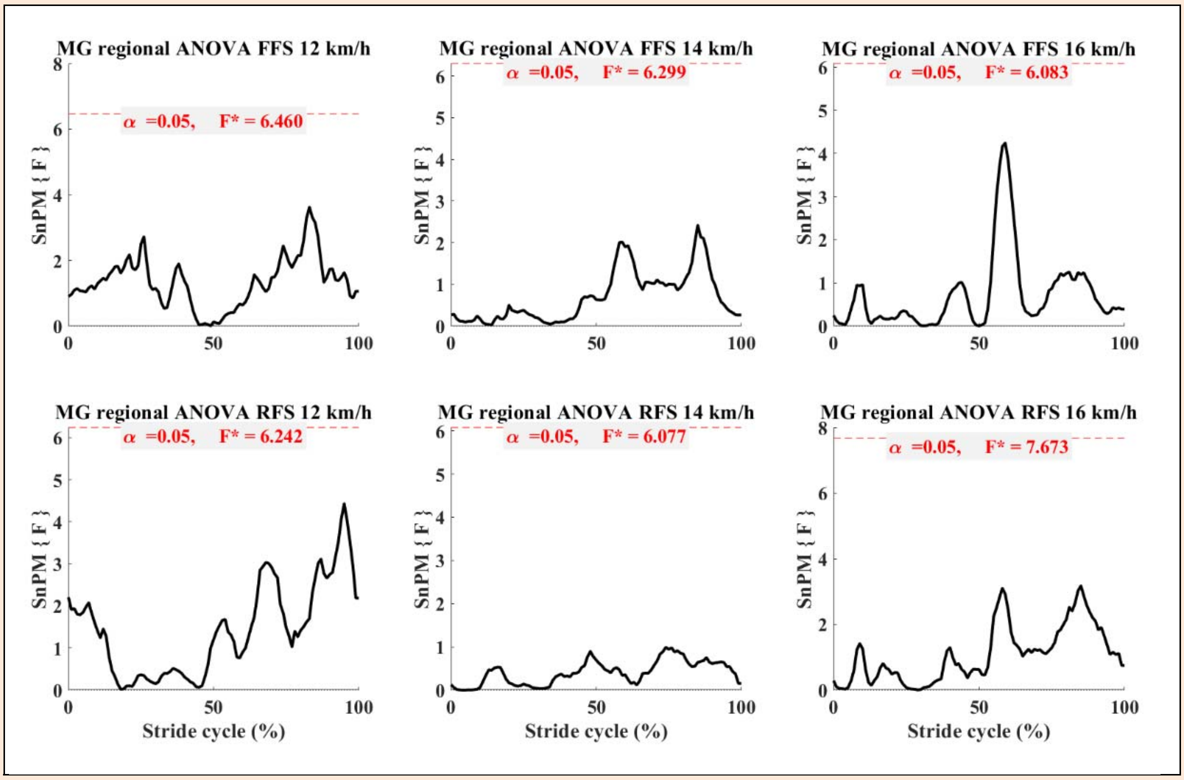
Supplementary Figure 4. Group means and SD of the MTU length changes of lateral gastrocnemius (LG) during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The dashed purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The panels below the MTU length changes panels show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a grey shaded area. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



Supplementary Figure 5. Group means and SD of the MTU length changes of soleus (SOL) during forefoot (FFS) and rearfoot running (RFS) at three different running speeds. Vertical dashed lines define the subphases of the running stride (0-100%) as stance phase (from initial foot contact at 0% to maximum knee angle in stance at toe-off) and swing phase (from toe-off to initial foot contact at 100%). The dashed purple vertical line shows the end of stance phase during FFS running and the orange dashed line during RFS running. The panels below the MTU length changes panels show the results of the SPM paired t-test between FFS and RFS. When the calculated SPM{t}value crossed the critical threshold at any point or region, the difference was statistically significant and marked with a grey shaded area. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



Supplementary Figure 6. Comparisons of group means and SD of the normalized electrical activity (all EMG channels averaged per muscle and normalized to maximum activity) at 12 km·h-1, 14 km·h-1, and 16 km·h-1 running speeds between forefoot (FFS) and rearfoot strike (RFS) running. Statistical Parametric Maps (SPM) show the comparisons between the foot strikes for each muscle. SPM{f} trajectories reflect the magnitude of the test statistics. When the calculated SPM{t}value crossed the critical threshold at any point or region it indicates statistically significant differences at group level. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.



Supplementary Figure 7. Comparison within the medial gastrocnemius proximal, middle and distal region. Statistical Parametric Maps (SPM) show the comparisons between the foot strike and speed conditions for each muscle. SPM{F} trajectories reflect the magnitude of the test statistics. No thresholds are crossed by the SPM{F}, which indicating no statistically significant regional differences at any time point of the stride at group level. Abbreviations: FFS – forefoot strike running, RFS – rearfoot strike running.