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ORIGINAL RESEARCH

LOWER LIMB MUSCLE ACTIVITY DURING FOREFOOT AND REARFOOT STRIKE RUNNING TECHNIQUES

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ABSTRACT

Purpose/Background: Distance running offers a method to improve fitness but also has a risk of lower limb overuse injuries. Foot strike technique has been suggested as a method to alter loading of the lower limb and possibly minimize injury risk. However, there is a dearth of information regarding neuromuscular response to variations in running techniques. The purpose of this investigation was to compare the EMG activity that occurs during FFS running and RFS running, focusing on the biceps femoris, semitendenosis, rectus femoris, vastus medialis oblique, tibialis anterior (TA), medial head of gastrocnemeus (MGas), lateral head of gastrocnemius (LGas), and soleus.

Methods: 14 healthy adults (6 male, 8 female; age, 24.2 $\pm$ 0.8 years, height 170.1 $\pm$ 7.8 cm; mass 69.8 $\pm$10.9 kg; Body Mass Index 24.1 $\pm$ 3.0 kg·m2) participated in the study. All participants performed a RFS and FFS running trial at 8.85 kph. A 3D motion capture system was used to collect kinematic data and electromyography was used to define muscle activity. Two-tailed paired t-tests were used to examine differences in outcomes between RFS and FFS conditions.

Results: The ankle was significantly more plantarflexed during FFS running ($p = .0001$) but there were no significant differences in knee and hip angles ($p = .618$ & .200, respectively). There was significantly less activity in tibialis anterior (TA) ($p < .0001$) and greater activity in the MGas ($p = .020$) during FFS running. The LGas and soleus did not change activity ($p = .437$ & .490, respectively).

Conclusions: FFS running demonstrated lower muscular activity in the TA and increased activation in the MGas.

Clinical Relevance: FFS and RFS running have the potential to off-load injury prone tissues by changing between techniques. However, future studies will be necessary to establish more direct mechanistic connections between running technique and injury.

Key Words: Electromyography (EMG), kinematics, neuromuscular adaptation

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INTRODUCTION

Running is a simple, cost efficient solution for many who wish to lose weight, decrease stress, or improve their cardiovascular fitness. The popularity of endurance running has increased over the years and as the popularity increases, so do running related injuries.[1] In 2002, below the knee leg injuries accounted for approximately 37% of all running injuries.[2] The most common area for injuries to occur is in the knee, followed by the lower leg and foot, respectively.[3] Recorded data regarding acute and chronic injuries in elite runners suggest that 56.6% of these athletes have sustained an achilles tendon overuse injury, 46.4% anterior knee pain, 35.7% shin splints, and 12.7% have had plantar fasciitis.[4] A 2011 survey identified plantar fasciitis (15%), shin splints (15%), and achilles tendonitis (11%) among the top injuries experienced by runners.[5] These injuries can result from running technique, overuse, and/or training errors, leading to irritation of the lower leg.[3,4,6,7]

Ground reaction forces (GRF) occurring during running may exceed three times that of walking and this may be related to the prevalence of running injuries.[6,8,9,10] These large forces require the lower limb to compensate in order to absorb a significant amount of energy during the impact of each step. [6,8,9,10] Compensation occurs through increased flexion of the hip, knee, and ankle joints in order to disperse the forces.[6] The attenuation of forces occurs due to both the overall shortening of the lower extremity combined with the eccentric loading of the muscles surrounding the hip, knee, and ankle joints.[6] In the foot, the plantar fascia, which runs from the calcaneus to the phalanges, helps provide support for the foot architecture during weight bearing. The calcaneus, midtarsal joint, and metatarsals form an arch or triangle shape on the plantar surface of the foot with the plantar fascia acting as the base of the triangle.[11,12] This structure helps absorb GRFs as well as provides the elasticity needed for propulsion in gait. [11,12]

Foot strike patterns are defined as 1) the rear foot strike (RFS) pattern, in which the heel of the foot lands on the ground surface first 2) the forefoot strike (FFS) pattern, in which the ball of the foot lands on the ground surface first.[8] Lieberman and colleagues[8] explored both foot strike patterns (RFS and FFS) and how each pattern assisted in managing GRFs. They discovered that FFS may decrease the probability of injuries.[8] There has been a debate, both in peer reviewed[1,8,12,13,14,15,16] and popular literature,[17] on the benefits of RFS versus FFS techniques. Compared to RFS running, the proposed benefits of FFS running include decreased energy absorption at the knee, better running economy, and faster running pace. [15,16,18]

A thorough search of the literature revealed that the majority of research on RFS and FFS running has focused primarily on GRFs and kinematics.[8,12,16,19] The type of foot strike pattern utilized alters the mechanics of the running task via changing the GRFs and lower limb kinematics.[8,17,19] Therefore, the magnitude of muscle activity should be expected to change in order to manage the differences in impact absorption between RFS and FFS running. EMG measurements of the magnitude of muscular activity may be a key component to understanding potential injury mechanisms between these two techniques. Shih et al focused on the changes that occurred in GRFs during RFS and FFS running.[16] However, they did not include results regarding the magnitude of muscle activity in the rectus femoris, tibialis anterior, biceps femoris, and the gastrocnemius and concluded that an increase in gastrocnemius activity occurred during FFS running based on their EMG data collected with surface electrodes.[16] In spite of their results, a thorough analysis of muscle activity was difficult because the authors did not address any other muscles crossing the ankle joint, (ie soleus), and did not specify which head of the gastrocnemius was utilized during analysis.[16] Therefore, currently a dearth of knowledge exists regarding muscle responses to RFS and FFS running and the relation of muscle activity to possible injury mechanisms between these two techniques.

The purpose of this investigation was to compare the EMG activity that occurred during FFS running and RFS running, focusing on the biceps femoris (BF), semitendenosis (ST), rectus femoris (RF), vastus medialis oblique (VMO), tibialis anterior (TA), medial head of gastrocnemeus (MGas), lateral head of gastrocnemius (LGas), and soleus (Sol). A greater understanding of the muscle activity in RFS and FFS running could assist clinicians in determining the causes for overuse injuries and developing adequate therapeutic programs. The hypothesis tested was that FFS running would have reduced muscle activity in the TA and increased muscle activity in the triceps surae due to the reduction of ankle dorsiflexion required during FFS running.

METHODS

Fourteen participants (n = 14) that consistently engaged in cardiovascular activity were selected by a sample of convenience from the university student population. After nine participants completed the experimental protocol, a power analysis was performed using the central limit theorem to determine the number of total participants necessary to compare EMG data at a statistical power of 0.95. The results influenced recruitment of additional participants. Participants were included if they participated in running activities at least twice a week and did not have medically diagnosed musculoskeletal, neuromuscular, and cardiovascular dysfunctions within the last five years that could have influenced adaptation to the studied running techniques. These included but were not limited to achilles tendonitis, plantar fasciitis, lower limb bone fracture, muscle paralysis, knee meniscus or ligamentous injury, and diabetes. The inclusion criteria were established to minimize cardiovascular risk and were: age (males 19-44; females 19-54), BMI less than or equal to 30 $kg \cdot m^{-2}$, no smoking within the last six months, blood pressure below 140/90 mm Hg, and females who were not pregnant.[20] The participants preferred foot strike pattern was recorded. All testing procedures were approved by the University of Alabama Institutional Review Board. Participants gave informed consent after having all procedures explained to them and prior to testing.

Participants' height and weight (Double beam scale 3p7044, Detecto®, Webb City, MO), along with blood pressure (BP) and heart rate (HR) (Omron digital BP monitor, BP789, Omron Healthcare, Inc, Lake Forest, IL) were recorded. The participants' maximum heart rate was calculated based on age (HRmax = 206.9 – (0.67 * age)[21] and BMI was calculated based upon height and weight (BMI = (Wt (lbs) / Ht^2 (inches)) * 703).[20] Subjects used their own pair of athletic shoes because the shoes each subject commonly wears during running would be more comfortable to the individual and this should minimize EMG variability within each participant.[22] The foot strike pattern not normally preferred by the participant were then explained and demonstrated to each participant. For example, if the participant preferred RFS, then FFS strike pattern was explained and demonstrated and vice versa. The participants were given several minutes to acclimate to the treadmill and experiment with the different foot strike patterns before the warm up began. Initial foot strike pattern was randomly assigned.

Lower extremity muscle activity of the eight selected muscles on the right limb was recorded using electromyography (EMG) at 1500 Hz (Noraxon TeleMyo DTS, Noraxon USA, Inc, Scottsdale, AZ, USA). The skin preparation included shaving of and the use of alcohol wipes on selected areas for placement of the disposable, self-adherent bipolar electrodes with fixed 2.0 cm spacing (Noraxon USA, Scottsdale, AZ, USA). Electrode placements were based upon Surface EMG for Noninvasive Assessment of Muscles (SENIAM)[23] standards on the muscle bellies of the BF, ST, RF, VMO, TA, MGas, LGas, and Sol on the right lower extremity. The electrodes were secured with cotton surgical tape and a compression sleeve.

Kinematic data was captured using an 8-camera, infrared motion capture system (Vicon Motion Capture System, Vicon Motion Systems Ltd, Oxford, UK) at 100 Hz. Sixteen reflective markers (Vicon Plug-In-Gait, Vicon Inc., Oxford, UK) were placed bilaterally, on the shoe over the base of the 2nd metatarsal phalangeal joint (MTP), on the shoe over the calcaneus, and over the lateral malleoli, shank, lateral epicondyle of femur, and thigh, and on the pelvis at the anterior superior iliac spine and posterior superior iliac spine using double sided tape (Figure 1). [24]

A standard treadmill was utilized for testing after the participants were oriented to the usage and safety of the device. The treadmill belt speed was set at 8.85 kph (5.5mph), which represents 115% of the average walk-to-run transition speed demonstrated in prior research.[25] A belt speed of 8.85 kph ensured a speed high enough to elicit running yet low enough to ensure submaximal exercise. Each participant was given a two-minute warm up period at a self-selected pace**.** After one minute of steady state running, 20 seconds of data were recorded. This protocol ensured stable EMG and kinematic data well after the 1-20 cycles necessary for the neuromuscular system to adapt to drastically altered task mechanics[26,27,28] and was further supported by pilot data specific to this experiment.

Figure 1. *Placement of reflective markers. A leg warmer was used on the right leg to help stabilize the EMG electrodes.*

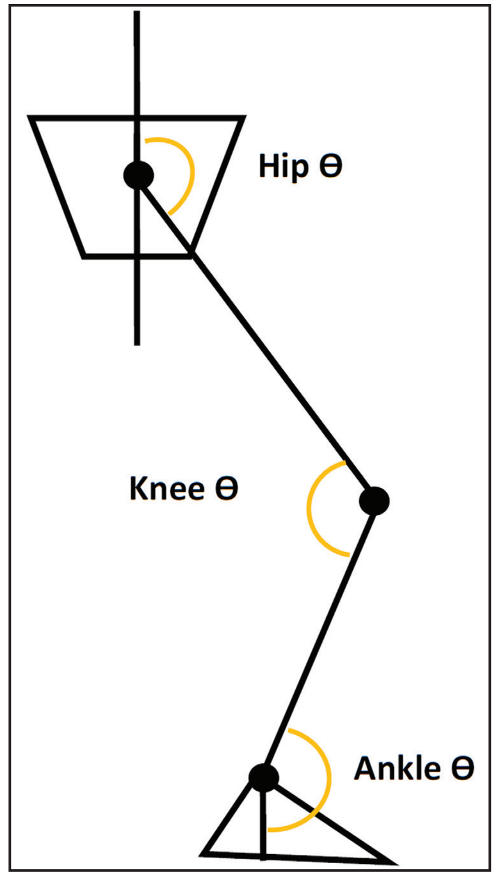

Figure 2. *Graphic representation of joint angles measured in the study.*

Kinematic and EMG recordings occurred simultaneously. To conclude the testing session, the participant performed a two-minute cool down session at their self-selected walking pace. BP and HR were taken immediately post cool down and five minutes later after the electrodes and markers were removed.

Marker coordinate data were smoothed using a woltering quintic spline.[29] Joint centers and joint angles (Figure 2) were calculated based on marker coordinate data.[24] Positive values represent ankle joint dorsiflexion, knee flexion, and hip flexion. The data were exported and combined with the raw EMG data in Matlab 2013a for final processing. Surface EMG data were digitally bandpass filtered (Butterworth digital, fourth order, zero lag, 20 – 500Hz) to attenuate high frequency and low frequency (motion artifact) noise. The data were then rectified and low pass filtered (Butterworth digital, fourth order, zero lag, 10 Hz cutoff).

Twenty consecutive strides were used for analysis. Foot strike on the treadmill was defined as the forward most position of the right MTP marker and was used to define the beginning and end of each stride. Each kinematic and EMG variable was then time normalized to 100 data points (each data point represents 1% of the stride cycle) and the twenty consecutive strides were averaged together for each participant. EMG signal amplitude of each muscle was normalized to the greatest value during the RFS condition from the average of twenty strides. This enables a comparison of relative EMG intensity between RFS and FFS conditions and is consisted with EMG normalization procedures for dynamic tasks.[30, 31]

Two-tailed paired t-tests were used to test the differences in kinematic and EMG outcomes between RFS and FFS conditions. Differences were considered significant at the 0.05 level of probability for all data analysis ($p \leq .05$).

RESULTS

Power analysis revealed the minimum number of participants necessary to achieve power of 0.95

between EMG variables was 13. Fourteen participants (10 identified as habitual FFS runners and four identified as habitual RFS runners) were tested in both RFS and FFS patterns. The sample included six males and eight females; age, 24.2 ± 0.8 years; height, 170.1 ± 7.8 cm; body mass, 69.8 ± 10.9 kg; body mass index (BMI), 24.1 ± 3.0 kg·m^{-2}, who participated in cardiovascular training 3.8 ± 1.4 times per week. Nine participants were African American and five were Caucasian.

Kinematic data revealed a significant difference between ankle angles in the sagittal plane between conditions**.** There were no significant differences between hip and knee angles. (Figure 3) Mean ankle angle during each stride was greater during RFS running (10.9° ± 4.3°) compared to FFS running cycle (-0.43° ± 4.8°). There was no significant difference between hip, knee, and ankle inversion/eversion angles between conditions (Table 1). The ankle angle at foot strike was greater (more dorsiflexed) (p = .0001) during RFS running (17.3° ± 5.2°) compared to FFS running (-6.3° ± 7.2°) (Figure 4).

There were significant differences in normalized EMG of the TA (*p* = .0001) and MGas (*p* = .020) (Figure 5). There was greater normalized activity of the TA during RFS running (0.51 ± 0.07) compared to FFS running (0.27 ± 0.07) (Figure 6), in contrast to greater normalized activity of the MGas during FFS running (0.33 ± 0.07) compared to RFS running (0.27 ± 0.06) (Figure 7). There were no significant differences in the normalized EMG for LGas, Sol,

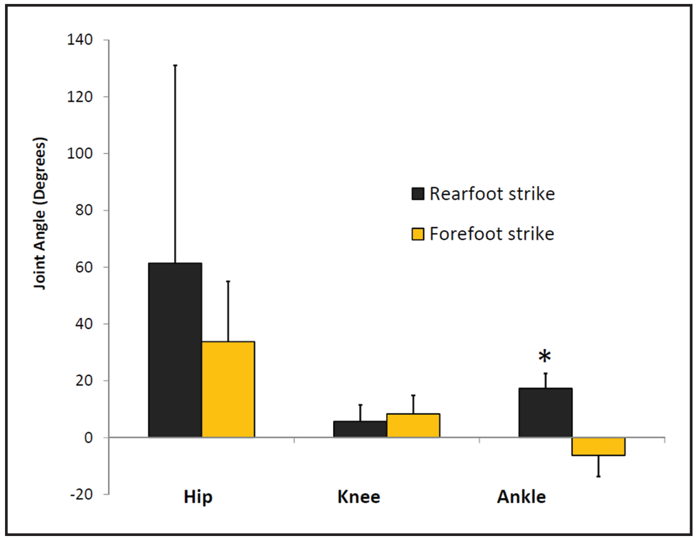

Figure 3. *Comparison of mean (one standard deviation) joint angles during rearfoot strike and forefoot strike running.*

VMO, RF, ST, or BF between RFS and FFS running conditions.

DISCUSSION

The most significant finding of this study was a statistically significant decrease in TA muscle activity and a statistically significant increase in MGas muscle activity with significant differences in SOL or LGas muscle activity in FFS compared to RFS running (Figure 5). Sagittal plane ankle angles were significantly different between RFS and FFS conditions while more proximal joints (knee and hip) did not demonstrate any differences. The participants' ankle alignment at foot contact was dorsiflexed during RFS

Table 1. *Mean (one standard deviation) for the hip, knee and ankle joint angles during rearfoot (RFS) and forefoot (FFS) strike running. * Indicates a significant difference (P < .05) between joint angles in RFS and FFS running.*

	Hip Angle			Knee Angle		
	HS	FFS	P value	HS	FFS	P value
Mean	44.9 ± 66.4	20.0 ± 17.1	0.20	34.9 ± 8.4	36.6 ± 10.4	0.62
Min	18.7 ± 64.0	-3.1 ± 14.3	0.24	5.2 ± 5.7	7.3 ± 5.8	0.36
Max	67.0 ± 73.3	40.0 ± 18.3	0.21	76.2 ± 13.9	80.2 ± 19.3	0.50
ROM	48.3 ± 11.1	43.0 ± 6.9	0.18	10.9 ± 11.2	72.9 ± 14.6	0.68
	Ankle Angle			**Ankle E/I Angle**		
	HS	FFS	P value	HS	FFS	P value
Mean	10.9 ± 4.3	-0.4 ± 4.8	0.0001*	2.9 ± 3.9	2.4 ± 3.4	0.76
Min	- 11.3 ± 6.6	-20.6 ± 5.5	0.005*	-2.1 ± 4.6	-2.6 ± 3.4	0.81
Max	29.4 ± 4.5	23.6 ± 5.3	0.005*	7.3 ± 4.0	7.4 ± 3.9	0.99
ROM	40.6 ± 8.2	44.2 ± 5.5	0.34	9.5 ± 2.9	9.9 ± 2.1	0.69

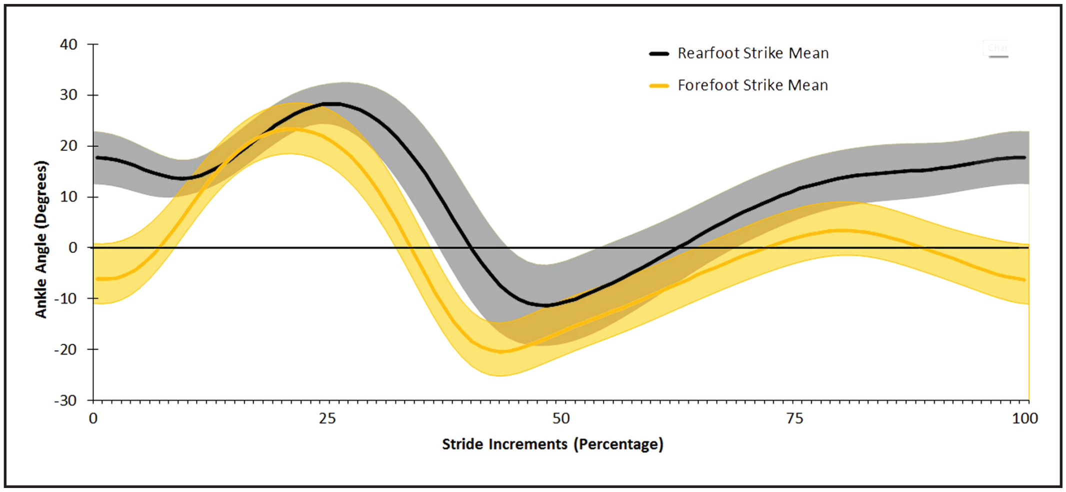

Figure 4. *Comparison of mean ankle angle during rearfoot strike and forefoot strike running normalized to 100 data points.*

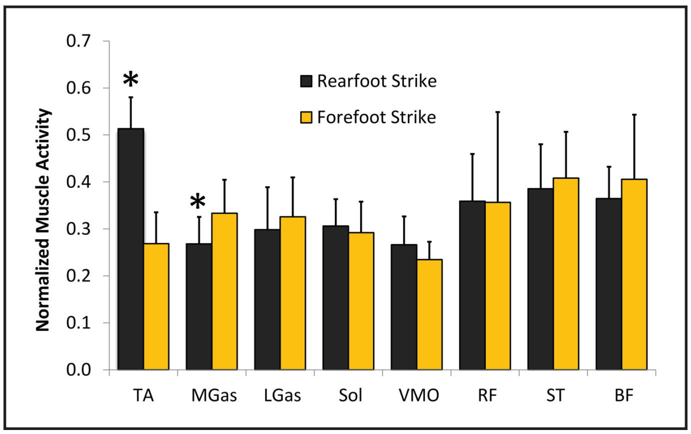

Figure 5. *Comparison of mean (one standard deviation) muscle activity during rearfoot strike (RFS) and forefoot (FFS) strike running for the, tibialis anterior (TA), medial gastrocnemius (MGas), lateral gastrocnemeus (LGas), soleus (Sol), vastus medialis oblique (VMO), rectus femoris (RF), semitendonosis (ST), and biceps femoris (BF).*

and plantarflexed during FFS indicated these participants did indeed adopt the correct foot strike for each condition. The lack of changes at more proximal joints demonstrated that this group of runners was able to limit compensation to striking patterns to the ankle joint. Previous researchers have demonstrated differences in either knee angle[32] or differences in both knee and hip angles[16] between the foot strike patterns. These differences could be a result of these researchers using habitual RFS runners. This study used predominantly habitual FFS runners, who may have made long-term neuromuscular adaptations at the hip and knee. Therefore, the short term transition between RFS and FFS used in the current study resulted only in changes observable at the ankle.

As stated, with FFS running, there was a significantly less TA activity with a corresponding increase in MGas activity but no differences in LGas, SOL, or more the

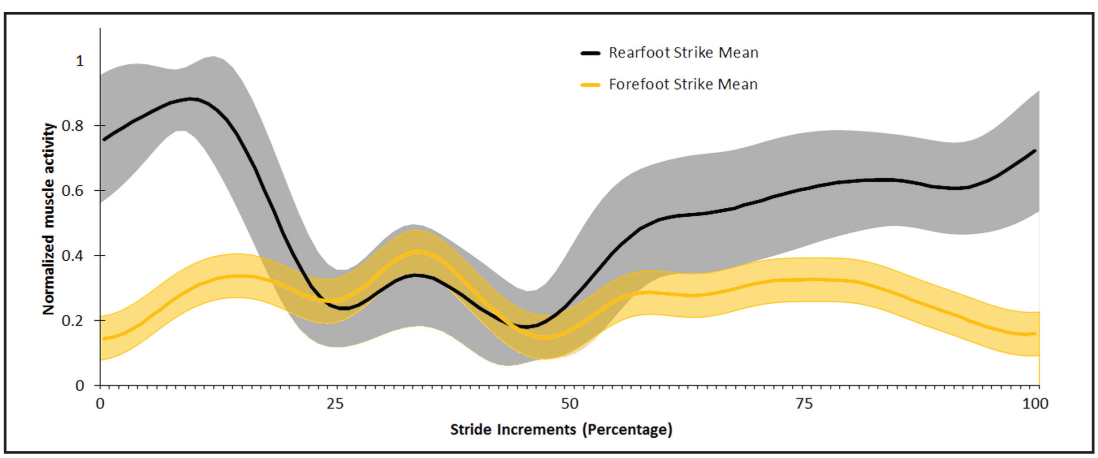

Figure 6. *Comparison of mean muscle activity of the tibialis anterior during rearfoot strike and forefoot strike running normalized to 100 data points.*

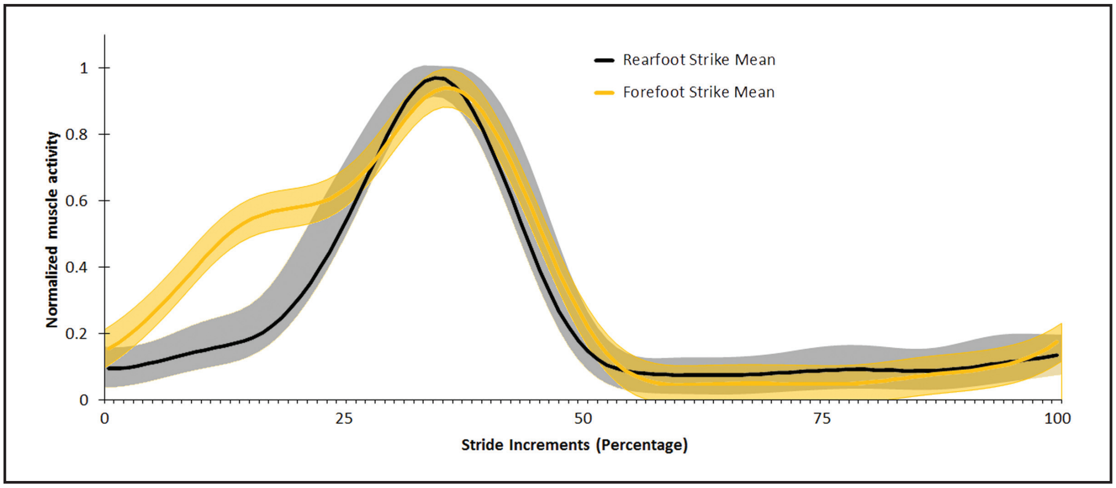

Figure 7. *Comparison of mean muscle activity of the medial head of the gastrocnemeus during rearfoot strike and forefoot strike running normalized to 100 data points.*

proximal muscles (VMO, RF, BF, and ST). The lack of significant differences in knee and hip kinematics could explain the lack of change in muscle activity in the muscles that control the knee and hip joints. Interestingly, the greater activation in MGas during the FFS condition did not correspond with significant differences in ankle eversion/inversion angles despite this muscle's contribution to frontal plane mechanics.[33] This could be related to the different biomechanics utilized to perform RFS and FFS running which has been established in prior literature.[8,16,19]

RFS and FFS running both involve energy absorption from foot strike through midsupport running phases although this is accomplished differently.[12,13] Foot strike during RFS involves general laxity of the plantar fascia and surrounding structures of the foot/ankle complex to accomplish energy absorption and transfer forces to proximal bones (such as the tibia), and is aided by eccentric contraction of the TA for controlled plantar flexion.[6,8,12] Later in the RFS running cycle, the tibia externally rotates, the MGas is active in conjunction with metatarsophalangeal joint extension in order to engage the windlass mechanism, tensioning the plantar fascia and stiffening the ankle/foot complex to enable propulsion.[8,11,32,33,34,35] In contrast, during FFS running, energy absorption is accomplished through the plantar fascia and eccentric contraction of lower limb joint extensors.[12,13] This technique requires a stiff ankle/foot complex at foot strike[8,12] and therefore, requires preservation of the longitudinal arches of the foot and maintained tension in the plantar fascia. One method to accomplish this would be preventing forefoot pronation and tibia internal rotation. There was a significantly greater activity of the MGas from foot strike through midsupport without a difference in inversion/eversion kinematics between RFS and FFS. The increased activity in this muscle during this phase could serve to prevent forefoot pronation, thereby maintaining ankle/foot complex stiffness and tension in the plantar fascia. Additionally, energy absorption is suggested to occur in via the achilles tendon, plantar fascia, and the elastic elements present in the triceps surae. This energy is then utilized later during propulsion through the stretch shortening contraction mechanism, potentially increasing the mechanical efficiency of the muscle/tendon unit during runnng.[12,36,37] The TA is not necessary for FFS mechanics and this explains the lower level of activity in this muscle in the FFS versus the RFS condition as seen in Figure 6. Interestingly, only the MGas required additional activation during FFS running, which was in contrast to the stated hypothesis that the entire triceps surae would demonstrate greater activity in the FFS condition. This indicates the mechanical energetic requirements to perform FFS task mechanics may be accomplished by an increase in only one portion of the muscle, the MGas, and that this muscle may have the additional benefit of stabilizing coronal plane kinematics. The lack of difference in coronal plane kinematics and greater MGas activity may indicate that an increase in mechanical demand exists in this plane and provides another example of the neuromuscular system leveraging musculoskeletal redundancy[38,39] in order to optimize control across multiple joints or axes.[40]

Clinical evaluation and treatment of running injuries should take into account the joint kinematics, muscle activity, GRFs, and plantar pressures acting on the lower extremity and foot. Changes made in foot strike pattern affect these factors therefore potentially affecting the probability of injuries. Prior researchers' have suggested that runners who habitually use a RFS pattern are at least twice as likely to experience a repetitive stress injury as those using a FFS pattern.[1] The injuries caused by RFS running noted in their study included higher rates of hip pain, knee pain, lower back pain, tibial stress injuries, plantar fasciitis, and stress fractures of the lower limb bones (excluding the metatarsals).[1] In contrast, higher rates of achilles tendinopathies, foot pain, and metatarsal stress fractures were noted during FFS running.[1] The injury mechanisms are related to task mechanics and how energy absorption is achieved by and distributed across anatomical structures. The forces during FFS at foot strike are absorbed through the small midfoot bones and muscles in comparison to the energy absorption in the calcaneus, talus, and tibia as seen in foot strike of RFS.[12] The presumed increase in stretch and tension placed on the plantar fascia during RFS running increases the risk of plantar fasciitis while the presumed increase stretch and tension on the Achilles tendon during FFS running increases the risk for Achilles tendinopathies. However, the submaximal tension on the Achilles tendon in FFS and the maximal tension on the plantar fascia at midsupport during RFS suggest a possible explanation for increased rate sof injury during RFS. Also, the prolonged pronation during RFS when compared to FFS can result in repetitive stress to the plantar fascia.[11,34]

Potential limitations of this study may include the participants' use of their own footwear. Participant use of comfortable footwear may have introduced shoe stiffness variability across the participants. Footwear stiffness does influence limb kinematics to a minor degree[41] and kinetics[42] in order to regulate total limb stiffness.[41,42] The lower limb responds to surface/terrain stiffness (these stiffness differences being larger than what can be attributed to shoes) through small changes in limb kinetics and muscle activity in order to regulate total limb stiffness[43,44] whereas shoe discomfort significantly influences muscle activity.[22] Taken together, the effect of footwear stiffness variability (induced by the participants using their own shoes) would have affected inter-participant variability in the current data but this effect could have been greater if the shoe was standardized across participants, potentially inducing greater changes due to discomfort. Inter-participant variability (through footwear or other factors) was minimized by performing a power analysis, ad hoc, on the variables with the greatest inter-participant variability (EMG data) to inform recruitment numbers.

Another possible study limitation could be related to time spent at each running technique before data collection. Muscle activity (measured through EMG) must adapt rapidly to changes in task mechanics in order to demonstrate stable locomotor performance.[45,46] This has been verified in human running through demonstrable stabilization of limb stiffness in as little as one to two running strides when presented with a change in ground stiffness.[26] In a study of human hopping (used as a simplified model of running) inter-cycle EMG variability of muscles in the lower limb between cycles became stabilized in the first couple of hops, allowing for data collection after 10 seconds.[27] Inter-cycle EMG variability is also stabilized in 10-20 cycles after being introduced to drastically altered intra-pedal stroke mechanical loading.[28] Pilot work for the present study confirmed stable intra-participant EMG recordings after 20 strides when foot strike changed to the non-habitual technique. Data collection for this study occurred after ~50 complete strides in order to minimize intra-subject EMG variability associated with adaptation to foot strike technique.

CLINICAL IMPLICATIONS

Shin splints are another common injury noted with runners. The term *shin splint* is non-specific term used to label to leg pain occurring between the tibial tuberosity and the ankle.[47,48] Shin splints can be a result of overstress to the bone, compartment syndromes, biomechanical variations,, or muscle fatigue.[47,48,49,50,51,52,53] A previous study looked at fatigue related loading imbalances on the shank and determined that during RFS running there is evidence of fatigue in the TA after 20 minutes of sustained running.[53] Results from the current study displayed a significantly greater activity of TA activity during RFS as compared to FFS running. Although not specifically addressed in this study, it might be that the high activity of the TA during RFS running could lead to muscle fatigue causing shin pain. FFS running has not been proven as a guaranteed technique to prevent injury but does appear to be a technique to change stress distribution between muscles and tendons anterior and posterior to the ankle joint and alter the loading of bones structures in the lower limb.[8,16,19] Altering a runner's technique is not the only solution to an injury but understanding the muscle activity that occurs during RFS and FFS running may assist the clinician in deciding whether to alter running foot strike techniques in order to address injury.

CONCLUSION

The findings of this study demonstrate that RFS running results in a significantly greater TA activity while FFS running results in greater activity of the MGas. The activation of the MGas may help maintain the arch of the foot and tension in the Achilles tendon producing an efficient stretch shortening cycle in order to propel the body forward. The greater levels in MGas activity during FFS running may be advantageous because it has the potential to affect the bodies ability to control both sagittal plane and coronal plane motions simultaneously. Future studies could investigate the effects of different types of footwear on RFS and FFS mechanics, the transition from RFS to FFS, the injuries involved with FFS running, muscle/tendon and plantar fascia dynamics with FFS running as well as the effects of fatigue.

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