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The Effect of Foot Strike Pattern on Achilles Tendon Load During Running

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Abstract—In this study we compared Achilles tendon loading parameters during barefoot running among females with different foot strike patterns using open-source computer muscle modeling software to provide dynamic simulations of running. Muscle forces of the gastrocnemius and soleus were estimated from experimental data collected in a motion capture laboratory during barefoot running for 11 runners utilizing a rearfoot strike (RFS) and 8 runners utilizing a non-RFS (NRFS) pattern. Our results show that peak Achilles tendon force occurred earlier in stance phase ($p = 0.007$), which contributed to a 15% increase in average Achilles tendon loading rate among participants adopting a NRFS pattern ($p = 0.06$). Stance time, step length, and the estimated number of steps per mile were similar between groups. However, runners with a NRFS pattern experienced 11% greater Achilles tendon impulse each step ($p = 0.05$) and nearly significantly greater Achilles tendon impulse per mile run ($p = 0.06$). This difference equates to an additional 47.7 body weights for each mile run with a NRFS pattern. Runners considering a NRFS pattern may want to account for these novel stressors and adapt training programs accordingly.

Keywords—Muscle force, Gastrocnemius, Soleus, Barefoot.

INTRODUCTION

It has been estimated that more than 35 million Americans use running as a mode of exercise.[1] Unfortunately, up to 74% of runners are expected to experience a musculoskeletal injury that results in pain or limited training each year.[7] The Achilles tendon, which functions to transmit forces from the medial and lateral gastrocnemius and soleus muscles to the calcaneous,[26] may be involved in 10% of running-related injuries.[37] Injury to the Achilles tendon may result from repetitive submaximal loading, creating microscopic tears to collagen fibers which experience high tensile forces during running.[18,23] Collagenous material such as tendon may also be prone to soft tissue injury when loads are repeatedly applied at a high rate.[6,27,41] Although tendon collagen synthesis is initiated by mechanical loads, inadequate rest between running sessions may result in collagen degradation which outpaces synthesis, leaving the tendon both weakened and vulnerable to injury. This may be especially true for female runners as estrogen has been observed to have an inhibitory effect on collagen and matrix synthesis in response to loading.[24,43]

Different foot strike patterns during running may affect Achilles tendon force, rate of loading, and the potential for Achilles tendon injury. In the United States, over 70% of elite runners[15] and over 85% of recreational runners[20] traditionally demonstrate a rearfoot strike (RFS) pattern where the heel makes first contact with the ground. Running shoes with a cushioned heel may promote the RFS strike pattern observed in these runners.[21,31–33] However, a RFS pattern has recently been observed as the predominant strike pattern during running at endurance speeds among habitually unshod people,[16] which suggests that foot strike pattern during running may be independent of footwear.

There is widespread interest in barefoot running among recreational and competitive athletes. One recent survey revealed that 76% of runners are interested in barefoot running and 22% of runners have already tried barefoot running.[34] This interest is based largely on the premise that barefoot running will result in a transition from a RFS pattern to a non-RFS (NRFS) pattern where the midfoot or forefoot make first

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contact with the ground during running. Mechanical consequences from the shift from a RFS to NRFS pattern are thought to minimize some risk factors associated with running-related injuries such as tibial stress injuries, plantar fasciitis, tibial stress fractures, and patellofemoral pain.[7]

To date, there is limited evidence that runners utilizing a NRFS pattern are at decreased risk for running-related injury. A recent cross sectional study suggested that individuals running barefoot or in minimalistic footwear who reported using a NRFS pattern were significantly less likely to report a running-related injury.[13] Additionally, runners utilizing a RFS pattern have been found to be over 2.5 times as likely to report a repetitive stress injury than runners with a NRFS pattern.[7] However, in this same study females who reported a NRFS pattern were four times as likely to complain of Achilles tendinopathy as females with a RFS pattern. This suggests that a NRFS pattern may reduce injury risk factors for some tissues but may increases Achilles tendon load and the overall propensity for tendinopathy.

Achilles tendon force during running has been reported in relatively few studies. In most of those studies, Achilles tendon force during running has been estimated by dividing net ankle joint moments estimated using the inverse dynamics approach by the Achilles tendon moment arm.[12,17,29,35] To our knowledge, only one previous comparison of Achilles tendon force has been made between runners with different foot strike patterns using these methods. Perl *et al.*[29] reported that running barefoot with a NRFS pattern increases Achilles tendon impulse by 24% for each step compared to running with a RFS pattern. However, a principle limitation of using the net joint moments calculated using inverse dynamics to estimate Achilles force is that this method does not account for cocontraction of the tibialis anterior observed during the loading phase while running with a RFS pattern.[40] Thus, the magnitude of Achilles tendon force may have been underestimated during the loading phase of running with a RFS pattern with this type of approach.

Open-source computer muscle modeling software has been developed to provide dynamic simulations of movements that allow the researcher to estimate muscle forces during running from experimental data commonly collected in a motion capture laboratory.[8] This approach can be used to better estimate the force generated by the individual plantarflexor muscles and thus the force transmitted to the Achilles tendon during running. The purpose of this study was to compare the Achilles tendon force during running with a RFS and NRFS pattern among people beginning a transition to barefoot running. We hypothesized that peak Achilles tendon force and loading rate will be lower among people who run barefoot with a RFS pattern than those who run with a NRFS pattern. We also hypothesized that Achilles tendon impulse per step and total impulse per mile will be lower among barefoot runners with a RFS pattern.

MATERIALS AND METHODS

The study procedures were approved by an Institutional Review Board and all participants provided informed consent prior to their participation. As part of a larger study on accommodation effects of running barefoot or in minimalistic footwear, we recruited 19 healthy female runners from three area universities, 18–35 years old, who were running at least 10 miles per week (Table 1). Participants who reported any cardiovascular pathology or surgery to either lower extremity in the last 12 months were excluded. Those who reported lower extremity symptoms during running that interfered with their desired training schedule over the last 2 years were also excluded. Each participant reported that they traditionally ran for exercise outdoors using cushioned heel footwear but that they were attracted to the study due to their interest in training barefoot or in minimalistic footwear.

We asked participants to run barefoot using their preferred foot strike pattern, which was determined during a practice session where participants ran barefoot over several lengths of a 125 foot concrete runway covered with linoleum tiles. During this practice session, participants were informed that many people choose to land on their forefoot while running barefoot but that they should experiment with their own strike pattern and choose the running style that was most comfortable for them. Participants were allowed to practice running barefoot as long as they wished prior to further testing. Practice time varied among the participants but no participant practiced more than 10 min prior to data collection.

Following the barefoot running practice, we prepared participants for 3D motion analysis measurements. Reflective markers were used to track three dimensional motions of the trunk, pelvis, bilateral femurs, shanks, and feet of the participants as they ran,

TABLE 1. Participant demographics from our investigation of runners.

	Mean (*SD*)
Height (m)	1.66 (0.06)
Mass (kg)	60.7 (5.7)
Age (years)	21.2 (1.6)
Weekly mileage (mi/week)	15.2 (9.5)
Running experience (years)	7.4 (3.3)

with each segment modeled as a rigid body. Anatomical markers used to establish the segmental-coordinate systems were placed over the bilateral acromion, iliac crests, greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, and the first and fifth metatarsal heads. Tracking markers, which remained in place for all of the running trials, were positioned as a cluster of four markers on the trunk, three markers for the pelvis on each anterior superior iliac spine and at the L5–S1 interspace, a cluster of four markers on the lateral thigh, a cluster of four markers on the posterior shank, and three markers on the posterior calcaneus. Following a standing calibration trial all anatomical markers were removed for the running performance trials.

All running mechanics were recorded as participants ran along a 20-m runway covered with low pile carpet tiles with a fiberglass and thermoplastic composite backing. Running velocity was between 3.52 and 3.89 m/s as indicated by the forward velocity of the L5–S1 marker in the lab coordinate system at the time of contact with the force platform. Following at least five practice trials along the laboratory runway, five trials were collected for further analysis. During each trial, marker data were collected at 120 Hz using eight Eagle digital cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) positioned around the runway. Ground reaction forces measured with a force platform flush with the surface of the runway were recorded at 1200 Hz (Model 4080, Bertec Corporation, Columbus, OH). Marker trajectories and analog signals from the force platform were digitally filtered at 15 Hz using a low pass, fourth order Butterworth recursive filter.[4]

We used OpenSim (open source software developed by Simtk.org) to estimate lower extremity muscle forces during running among runners who preferred to run barefoot with a RFS pattern and those who did not.[8] An eight segment, 19 degree-of-freedom (DOF) model was developed based on the gait2392 model using OpenSim 2.3.2 software (Simtk.org). The head and trunk were modeled as a single segment with three rotational DOF relative to the pelvis. The pelvis segment had six DOF and was able to rotate and translate in all three dimensions with respect to the ground. The hip joint was modeled as a ball in socket joint with three rotational DOF. The knee joint was modeled as a single DOF hinge joint where any tibio-femoral translations and non-sagittal rotations were constrained as a function of knee flexion. The ankle joint was modeled as a single DOF, sagittal plane hinge joint with both the subtalar and metatarsal-phalangeal joints being locked in a neutral position allowing for no motion. The inertial characteristics of the segments used in the model were based on subjects' total body mass and segment lengths.[38]

Ninety-two separate musculotendon actuators represented the muscle–tendon units in the model. Eighty-six actuators represented the muscle–tendon units of the lower limbs and six represented the trunk. The physiological properties of the muscle–tendon actuators were based on a Hill-type model where the force–length–velocity relationships for the muscle–tendon unit was based on Zajac.[44] These muscle properties were then scaled for each individual based on peak isometric muscle force, optimal muscle fiber length, pennation angle, and tendon slack length based on Delp *et al*.[9] Other muscle parameters such as muscle insertion points and wrapping points were determined by Delp *et al*.[9]

We used inverse kinematics calculated by Visual 3D software (C-Motion Inc, Rockville, MD) and ground reaction forces as inputs into OpenSim's residual reduction algorithm (RRA). This algorithm computes the joint moments needed to follow the experimental motion as closely as possible by altering the subject-specific model's torso center of mass to correct excessive left–right or anterior–posterior movements. The pelvis in our model was represented as a six DOF joint relative to the ground and therefore, it had its own torque actuator, called the residual actuator. Three residual actuators represented the translational DOF between the pelvis and ground, and are referred to as residual forces (F_x, F_y, F_z). The other three residual actuators represent rotational DOF and are called residual torques (M_x, M_y, M_z). Since the model does not have arms, dynamic inconsistencies are created leading the measured forces and moments to deviate from Newton's Second Law ($\Sigma F = \text{ma}$). Therefore, the models six residuals essentially add a new variable to Newton's equation ($\Sigma F + F_{\text{Residual}} = \text{ma}$). After the RRA algorithm was performed on the experimental data, the root-mean-squared (RMS) error between experimental and RRA calculated pelvic rotations and translations, right and left hip flexion, hip adduction, hip rotation, knee angle, ankle angle, lumbar extension, lumbar bending, and lumbar rotation were calculated to ensure that all RMS errors fell below 1° (or 10 mm for the pelvic translations).

After the RRA algorithm, we used the computed muscle control (CMC) algorithm to estimate a set of muscle excitation patterns by moving the model to match the desired kinematics.[39] This algorithm uses a static optimization technique to estimate muscle excitations at each time step of the model to match the net joint moments. Due to the infinite number of muscle excitation patterns that could generate the desired kinematics, the model was optimized to minimize the weighted sum of squared muscle activations.[38] The model was moved forward a single time step and the results were compared to the original kinematics. The CMC processing was repeated until the tolerance of the optimizer convergence was met for our model segment angle and angular velocity output with the desired kinematics from the RRA processing.[3,39] A optimizer convergence criterion of 0.1° was used.

Following CMC processing we calculated peak Achilles tendon force, average Achilles loading rate, and Achilles impulse during the stance phase from the CMC time series data of each running trial using custom software (LabView 8.6, National Instruments, Austin, TX). The stance phase was defined as the period of time when the vertical ground reaction force was greater than 25 Newtons. Achilles tendon force was calculated by summing the estimated actuator forces representing the medial gastrocnemius, lateral gastrocnemius, and soleus muscles. This value was then normalized to the body weight (BW) of each participant. Peak instantaneous and average Achilles loading rate were also calculated as the peak change in tendon force between sequential samples of data and change in tendon force from initial contact to peak force divided by the time to peak force, respectively. To facilitate comparison of our simulation-based estimates to previous studies using net joint moments to estimate Achilles tendon force during running, we also calculated peak Achilles tendon force using methods described by previous authors. Specifically, we divided net ankle sagittal plane moment calculated using inverse dynamics during the running trials by an estimated Achilles tendon moment arm of 0.05 m.[19,35]

Changing foot strike pattern during running may also affect a runner's step length, which would affect the number of steps used to complete the distance of a mile. This may also influence the total Achilles tendon impulse per mile during running. Thus, we also estimated the total Achilles tendon impulse per mile run by multiplying the Achilles tendon impulse estimated during a single stance phase by the number of steps required to run a mile. Step length was used to calculate number of steps required to run a mile and was determined by taking the difference in location of the distal heel marker during each running trial between the right and left legs at initial contact during the laboratory performance trials.

We compared dependent variables of interest between groups of participants who preferred a RFS pattern while running barefoot and those who did not (NRFS) using nonparametric Mann–Whitney U tests with α set to 0.05 (SPSS version 20, SPSS Inc., Chicago, IL). Effect sizes (difference in group means relative to the pooled standard deviation) were also calculated to quantify the magnitude of the observed effects between foot strike patterns of the runners. We classified participants by their preferred foot strike pattern during the running trials based on the orientation of the three calcaneal markers used to track the foot segment relative to horizontal in the laboratory coordinate system at initial contact with the force plate. Positive foot angles greater than 5° were considered RFS running trials. Negative foot angles less than −5° were considered NRFS running trials. Reference of the foot segment to the laboratory coordinate system has been found to be an accurate method of classifying foot strike pattern during running.[2] However, for trials with foot angles within 5° horizontal at initial contact, we also determined the location of the center of pressure relative to the origin of the foot segment (posterior calcaneus) at the time of initial contact with the force platform (foot strike index).[5] If the strike index indicated the center of pressure was posterior to a point half the distance from the midpoint of the 1st and 5th metatarsal heads to the origin of the posterior calcaneus, the trial was considered a RFS pattern. Every trial was evaluated. No data were excluded based on foot strike pattern.

RESULTS

Following the acclimation to barefoot running during the practice trials, 11 of the 19 participants (58%) preferred a RFS pattern while running barefoot. Ensemble average muscle forces for those muscles inserting into the Achilles tendon for each set of runners that exhibited a RFS and NRFS pattern are displayed in Figs. 1–4. The medial gastrocnemius and soleus muscles produced the greatest force magnitude during stance for each foot strike pattern (roughly 2.0

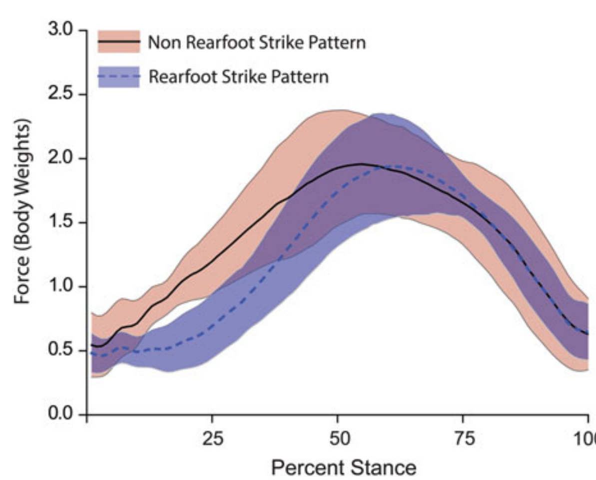

FIGURE 1. Time normalized medial gastrocnemius force during the stance phase of running with a rearfoot (RFS) or non-rearfoot strike (NRFS) pattern. Shaded regions represent one standard deviation from the mean.

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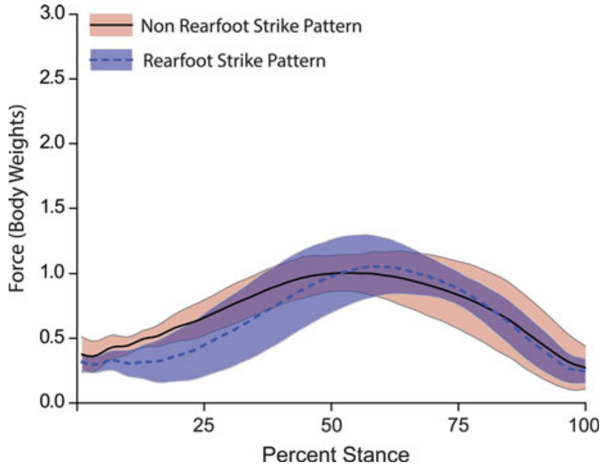

FIGURE 2. Time normalized lateral gastrocnemius force during the stance phase of running among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS). Shaded regions represent one standard deviation from the mean.

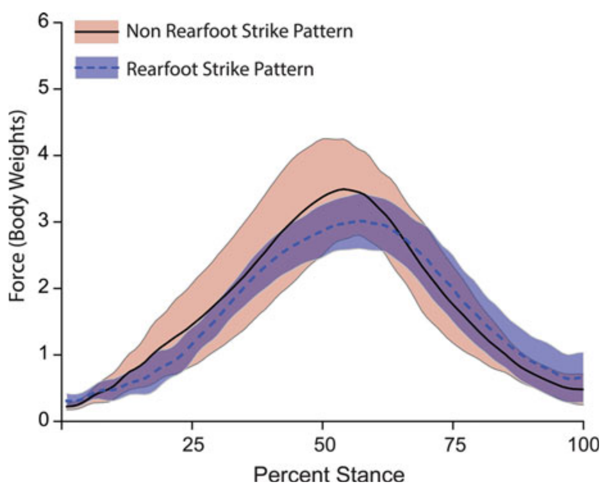

FIGURE 3. Time normalized soleus muscle force during the stance phase of running among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS). Shaded regions represent one standard deviation from the mean.

and 3.0 BW, respectively). The estimated tibialis anterior muscle force ranged from 0.5 to 1.0 BW and was larger during the first half of stance among participants who preferred a RFS pattern (Fig. 5). The peak Achilles tendon force occurred earlier in the stance phase ($p = 0.007$, $ES = 1.61$) but it was not greater among participants running with a NRFS pattern ($p = 0.31$, $ES = 0.69$) (Table 2). Runners with a NRFS pattern generally experienced a higher Achilles tendon average loading rate. However the difference between groups (15%) was not statistically significant ($p = 0.06$, $ES = 0.93$).

Temporal–spatial differences were not observed between participants who chose to run barefoot with a

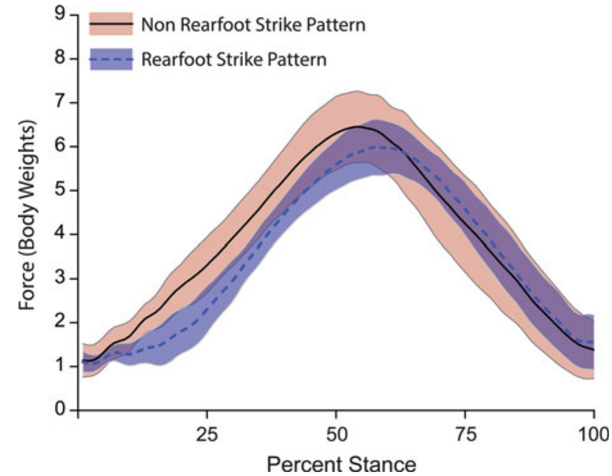

FIGURE 4. Time normalized total Achilles force during the stance phase of running among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS). Shaded regions represent one standard deviation from the mean.

RFS or NRFS pattern (Table 2). Stance time, step length, and the estimated number of steps per mile were very similar between groups. However, runners who preferred to NRFS experienced an 11% greater Achilles impulse each step ($p = 0.05$, $ES = 1.07$) and a greater Achilles tendon impulse per mile run that approached statistical significance ($p = 0.06$, $ES = 0.95$). This finding suggests a runner will experience an additional Achilles tendon impulse of 47.7 BWs for each mile run with a NRFS pattern (Table 3).

DISCUSSION

The purpose of this study was to compare the estimated Achilles tendon force among people running with a RFS and NRFS pattern at the beginning of their transition to barefoot running. Our hypothesis that peak Achilles force and loading rate would be lower among runners using a RFS pattern was not statistically supported. However, there were large effect sizes and percentage differences observed between RFS and NRFS groups. The magnitude of these effects suggests that compared to runners with a NRFS pattern, large, clinically relevant decreases in peak force and average loading rate may exist among runners with a RFS pattern. We also hypothesized that Achilles tendon impulse per step and total impulse per mile would be lower among barefoot runners with a RFS pattern. Achilles tendon impulse per step and impulse per mile were both substantially lower among runners with a RFS pattern. However, only decreased Achilles tendon impulse per step was statistically different between groups.

The large but statistically insignificant differences in Achilles tendon average loading rate and Achilles tendon impulse per mile suggest this study was statistically underpowered to identify potentially meaningful effects. A *post hoc* sample size calculation using $\alpha = 0.05$, $\beta = 0.2$, and the average within-group variability of these variables suggests that 14 runners per group would be necessary for the group means reported here to achieve statistical significance. Unfortunately, it was not feasible to recruit these participants. Therefore, we recommend that future studies with adequate sample size be conducted to cross-validate these findings.

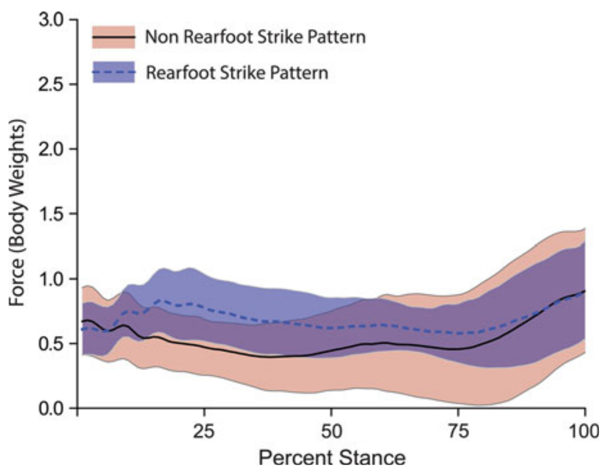

FIGURE 5. Time normalized total tibialis anterior force during the stance phase of running among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS). Shaded regions represent one standard deviation from the mean.

The novelty of the methods utilized in our study merits a comparison to previously published findings. In this study and others, Achilles force in running has been depicted as a parabolic curve, beginning at initial contact and peaking shortly after midstance.[11,12,19,29,35] Peak Achilles tendon force estimated in this study was between 6 and 7 BWs while running at 3.7 m/s ($\pm$5%), regardless of foot strike pattern. Based on custom and proprietary musculoskeletal modeling software (SIMM, Musculo-Graphics Inc., Santa Rosa, CA), Edwards reported average peak Achilles force values between 7 and 8 BWs while running at 4.4 m/s for four runners with a NRFS pattern and six runners with a RFS pattern.[11] However, those data were considered together and comparisons between runners with different strike patterns were not performed. *In vivo* peak Achilles tendon force trends measured with a buckle transducer in a single male subject running barefoot with a RFS (5.2 BW at 3.9 m/s) and NRFS (5.4 BW at 3.8 m/s) pattern were also similar to our data.[19] Peak Achilles tendon force estimated using net ankle joint moment and estimated Achilles tendon moment arms has been reported as 7.2 BW at 4 m/s, 6.3 BW at 5.1 m/s, and 7.4 BW at 5.3 m/s for three runners with a RFS pattern.[35] Variability in running speed between studies likely contributes to the differences reported in previously reported Achilles tendon force estimates during running.

A potential advantage of the methods used in this study to estimate Achilles tendon force is that ankle dorsiflexion muscle activity is not assumed to be zero during the stance phase of running. This assumption is inherent in the use of net ankle joint moments derived using inverse dynamics to calculate Achilles tendon

TABLE 2. Average (*SD*) Achilles tendon force measurements during stance phase of running among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS).

	Peak force (BW)	Time to peak force (% stance)	Average loading rate (BW/s)	Peak instantaneous loading rate (BW/s)
NRFS ($n = 8$)	6.61 (0.84)	52.24 (3.31)	45.61 (7.16)	135.5 (25.3)
RFS ($n = 11$)	6.11 (0.63)	56.61 (2.14)	39.44 (6.15)	127.4 (21.5)
p value	0.31	0.007	0.06	0.54
Effect size	0.69	1.61	0.93	0.34
% Diff	8.3	7.7	15.7	6.3

TABLE 3. Average (*SD*) temporal–spatial characteristics and Achilles tendon load per mile run barefoot among participants who preferred a rearfoot strike (RFS) pattern and those who did not (NRFS).

	Running speed (m/s)	Stance time (s)	Step length (m)	Steps per mile	Achilles load per step (BW $\times$ s)	Achilles load per mile (BW $\times$ s)
NRFS ($n = 8$)	3.68 (0.07)	0.23 (0.02)	1.302 (0.05)	616 (23.5)	0.85 (0.09)	522.1 (54.3)
RFS ($n = 11$)	3.67 (0.07)	0.23 (0.01)	1.294 (0.07)	621 (35.7)	0.76 (0.07)	474.4 (46.5)
p value	0.65	0.39	0.90	0.90	0.05	0.06
Effect size	0.22	0.26	0.14	0.18	1.07	0.95
% Diff	0.4	1.6	0.7	0.9	11.02	10.1

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force during running. One consequence of this assumption is the potential to underestimate Achilles tendon muscle force. Peak Achilles tendon force was determined to be 5.9 ($SD = 0.7$) BW for runners with a NRFS pattern and 5.0 ($SD = 0.5$) BW for runners with a RFS pattern. Therefore, our simulation approach results in greater Achilles tendon force estimates than those calculated using inverse dynamics by 10% for NRFS and 18% for RFS patterns. The greater disparity in Achilles tendon force estimates between these two methods for runners with a RFS pattern may be at least partially explained by greater tibialis anterior muscle activity during stance phase while running with a RFS pattern compared with a NRFS pattern.[36]

Our findings may be relevant to the pathogenesis of Achilles tendinopathy, a common injury among runners. The etiology of Achilles tendinopathy is believed to be associated with repeated mechanical loading of the tendon. Repetitive tendon loads such as those experienced during distance running initiates collagen and extracellular matrix synthesis and tissue degradation. With sufficient rest between bouts of running (36–72 h), a net positive balance in collagen and extracellular matrix synthesis exists that may contribute to hypertrophic changes observed in the Achilles tendon among habitual runners.[22] However, with inadequate rest between running sessions, tissue degradation outpaces synthesis and the tendon is believed to be both weakened and vulnerable to injury.[22] Our results suggest that habitual runners who convert to a NRFS pattern will experience an additional 10% Achilles tendon impulse each mile run, or an additional 47.7 BW/mile. This change in Achilles loading during running may hasten collagen remodeling, and may require additional rest between exercise sessions to avoid a net loss of collagen and extracellular matrix that may predispose the tendon to injury. As such, runners making a transition to a NRFS pattern may benefit from additional rest between bouts of running to allow for tissue remodeling. This result may also underscore the importance of a gradual transition to a NRFS pattern for individuals interested in that barefoot running.

Our investigation was on female recreational runners. It may be particularly important that female runners gradually convert to a NRFS pattern or to allow adequate time for recovery between bouts of running. Females have been hypothesized to have an attenuated adaptive tendon response to repetitive loading.[42] This attenuated response may be associated greater estrogen levels among females. Estrogen has been observed to have an inhibitory effect on collagen and matrix synthesis in response to loading.[24,43] Indeed, female barefoot runners with a NRFS pattern have been reported to experience Achilles tendinopathy at a prevalence four times greater than females who prefer to run with a RFS pattern.[7] This appears consistent with the hypothesized attenuated adaptive tendon response to loading and our finding of greater Achilles tendon impulse among women with a NRFS pattern compared with a RFS pattern.

Advocates for conversion to running with a NRFS pattern attribute the benefits of the technique largely to a reduction in the large and rapid vertical ground reaction force that typically occurs when the heel first makes contact with the ground (impact transient) during a RFS pattern.[5,21,28] The magnitude of this impact transient and vertical loading rate has been found to predict some running-related injuries such as tibial stress fractures and plantar fasciitis in female runners.[25,30] Running with a NRFS pattern may decrease vertical ground reaction force loading rate.[10] However, our results suggest that this reduction may be at the expense of greater Achilles loading rate and impulse per step. As such, this technique may not be suitable for all runners with running-related injuries, particularly in individuals with a history of Achilles tendon injury.

Greater Achilles tendon impulse per step among runners with a NRFS pattern is consistent with findings of Perl *et al.*[29] who analyzed 13 male and 2 female experienced barefoot runners and reported a 24% greater Achilles tendon impulse per step when running with a RFS pattern in minimalistic footwear compared to those using a RFS pattern. In this previous study, Achilles tendon force was estimated by dividing the net ankle plantarflexion moment obtained from the inverse dynamics approach by the Achilles tendon moment arm. As such, this method does not account for antagonistic muscle force produced by the anterior tibialis. In our results and in previous studies, the tibialis anterior is active during the loading phase of running with a RFS pattern as the foot plantarflexes after initial contact.[40] Among female runners with a RFS pattern in the present study, the tibialis anterior muscle force was generally double the force measured among runners using a NRFS strike pattern during the first half of stance. This greater anterior tibialis force among runners with a RFS may lead to underestimation of the Achilles tendon force over time in this previous study. This may account for the smaller effect of strike pattern on Achilles tendon impulse per step observed in this study as compared to the data provided by Perl *et al.*[29]

In this study we identified a shorter time to peak Achilles tendon force ($p = 0.007$, $ES = 1.61$) and greater Achilles tendon impulse per step at the level of statistical significance ($p = 0.05$, $ES = 1.07$) among females running barefoot with a NRFS strike pattern compared with females using a RFS pattern. Further, although not statistically significant, runners who prefer a NRFS pattern while running barefoot tended to experience a greater Achilles tendon loading rate ($p = 0.06$, $ES = 0.93$) and greater Achilles tendon impulse per mile run ($p = 0.06$, $ES = 0.95$). These results should be viewed in the context of several limitations. For example, participants in this study were habitual runners with no previous experience running barefoot. It has been suggested that barefoot running mechanics change with time and experience.[10] As such, our results may only generalize well to the growing number of individuals beginning the transition to barefoot running. Additionally, kinematic data and ground reaction forces were collected as participants ran on a runway covered with low pile commercial carpet tiles. The compliance of the running substrate may affect foot strike pattern preference and lower extremity mechanics.[14,16] The running surface in this study may be more or less compliant than the typical training surface preferred among our sample. Large effect sizes were observed between groups of runners with a NRFS and RFS pattern during barefoot running. However, several of these comparisons failed to reach statistical significance, suggesting a larger sample size is necessary for subsequent studies on this topic. Finally, computer modeling is an estimate of muscle forces during dynamic movement performance. Accuracies of these simulations of movement depend largely on the underlying mathematical models of the neuromuscular and skeletal system used and their assumptions as well as approximations. For example, in this study the ankle is modeled as a hinge joint which likely affects the validity of these results. Given these limitations, future studies to cross-validate these results or test if these findings apply to other populations or settings appear justified.

In conclusion, foot strike pattern is a potentially modifiable aspect of running that may increase the risk of Achilles tendon injury. Within the context of the limitations of this study, our interpretation of these results is that barefoot running with a NRFS pattern may not be advisable for all runners. This may be particularly true for runners with Achilles tendon pain or a history of Achilles tendon injury.

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