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Acute changes in foot strike pattern and cadence affect running parameters associated with tibial stress fractures

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ABSTRACT

Tibial stress fractures are a common and debilitating injury that occur in distance runners. Runners may be able to decrease tibial stress fracture risk by adopting a running pattern that reduces biomechanical parameters associated with a history of tibial stress fracture. The purpose of this study was to test the hypothesis that converting to a forefoot striking pattern or increasing cadence without focusing on changing foot strike type would reduce injury risk parameters in recreational runners. Running kinematics, ground reaction forces and tibial accelerations were recorded from seventeen healthy, habitual rearfoot striking runners while running in their natural running pattern and after two acute retraining conditions: (1) converting to forefoot striking without focusing on cadence and (2) increasing cadence without focusing on foot strike. We found that converting to forefoot striking decreased two risk factors for tibial stress fracture: average and peak loading rates. Increasing cadence decreased one risk factor: peak hip adduction angle. Our results demonstrate that acute adaptation to forefoot striking reduces different injury risk parameters than acute adaptation to increased cadence and suggest that both modifications may reduce the risk of tibial stress fractures.

1. Introduction

Long distance running is plagued with high injury rates; in one study, 79% of long distance recreational runners experienced an injury that required training adjustments (Lun et al., 2004). Tibial stress fractures, one of the most common running injuries (Taunton et al., 2002), are the most common stress fracture in athletes (Matheson et al., 1987) and require an average recovery time of 12 weeks (Beck et al., 2008). Biomechanical parameters associated with a history of tibial stress fracture suggest possible mechanisms of injury. Higher average and peak loading rates and peak tibial accelerations may contribute to stress fracture development (Milner et al., 2006b). Runners with a history of tibial stress fractures tend to have higher peak hip adduction angles and peak absolute free moments than runners with no prior fracture (Pohl et al., 2008); these parameters are thought to increase fracture risk due to poor alignment during running (Milner et al., 2005) and excessive torques (Milner et al., 2006a), respectively. Although we cannot directly determine changes to injury risk, it is possible to investigate the effects of changing running pattern on parameters associated with tibial stress fractures.

A small retrospective study showed that habitual forefoot striking runners had a lower incidence of tibial stress fractures than rearfoot striking runners (Daoud et al., 2012), suggesting that converting to forefoot striking may reduce tibial stress fracture risk. Converting from rearfoot striking to forefoot striking is associated with lower average and peak loading rates (Boyer et al., 2014; Shih et al., 2013), but can increase tibial accelerations near the time of foot contact (Laughton et al., 2003). Forefoot striking is typically taught to habitual rearfoot strikers through verbal explanation (Giandolini et al., 2013; Laughton et al., 2003; Olin and Gutierrez, 2013) or video (Boyer and Derrick, 2015). Adjusting to forefoot striking is associated with changes in kinematics – e.g., higher cadence (Arendse et al., 2004), a more plantarflexed ankle (Lieberman et al., 2010) and flexed knee (Laughton et al., 2003) at foot contact – and changes in muscle activity (Landreneau et al., 2014; Shih et al., 2013; Yong et al., 2014). It is unclear which kinematic and kinetic differences associated with forefoot striking contribute to the potential decrease in injury rate, and whether injury risk parameters can be reduced with a different adaptation, such as increasing cadence.

Increasing cadence may reduce the risk of tibial stress fractures, but, unlike forefoot striking, does not require major adjustments to

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running style. Increasing cadence while rearfoot striking was shown to lower peak tibial accelerations during overground running (Derrick et al., 1998) and lower peak hip adduction angle during treadmill running (Heiderscheit et al., 2011; Willy et al., 2016a). Previous studies increased subjects' cadence during overground running by using tape marks on the floor to shorten stride length (Boyer and Derrick, 2015; Edwards et al., 2009), but this method does not ensure the ability to recreate the motion without visual feedback. Cadence has been successfully increased using a metronome during treadmill running (Giandolini et al., 2013; Hamill et al., 1995; Heiderscheit et al., 2011; Hobara et al., 2012; Snyder et al., 2012), overground running (Lyght et al., 2016), and during in-field gait retraining (Willy et al., 2016a, 2016b). It remains unknown, however, if changes to parameters associated with tibial stress fracture, which were previously observed during treadmill running, translate to overground running without visual feedback.

The effectiveness of reducing injury risk parameters by increasing cadence alone compared to that of forefoot striking during overground running is unclear. Studies that compared the effects of foot strike type and increased cadence on tibial stress fracture risk parameters were either restricted to treadmill running (Giandolini et al., 2013) or investigated a limited set of parameters (Boyer and Derrick, 2015; Giandolini et al., 2013). Previous work comparing these acute adaptations independently assessed the effects of foot strike or cadence by having subjects run using a forefoot striking pattern without adjusting cadence and changing cadence while maintaining a rearfoot striking pattern (Boyer and Derrick, 2015; Giandolini et al., 2013; Lyght et al., 2016). In our study, we aimed to capture how runners naturally adjusted their running pattern when being asked to only focus on changing foot strike or increasing cadence. Our goal was to study how natural adaptations to changing foot strike or increasing cadence affects tibial stress fracture risk during overground running in a single population by analyzing the following set of injury risk parameters: peak tibial acceleration, peak absolute free moment, peak hip adduction angle, and average and peak loading rates.

To achieve this goal, we first demonstrated that training runners with visual feedback of foot pressure allows them to perform a forefoot striking pattern, and providing runners with auditory cueing allows them to increase cadence while running overground. We then tested the hypotheses that in comparison to rearfoot striking, (1) average and peak loading rate decrease during forefoot striking, (2) peak tibial acceleration and free moment decrease during increased cadence, and (3) peak hip adduction angle decreases during both forefoot striking and increased cadence.

2. Methods

2.1. Subjects

Seventeen healthy, habitual rearfoot striking recreational runners (11 females, 6 males; age: 32.1 ± 9.8 years; height: 168 ± 11 cm; mass: 64.9 ± 12.5 kg), participated in this study. We ensured subjects landed on their heels using a high speed camera (Casio Exilim Pro Ex-F1, Casio Computer Co., Ltd., Shibuya-ku, Tokyo, Japan) during overground running ($\geq$120 frames/s). All subjects were experienced long distance runners who reported running a minimum of 10 km/week (33.5 ± 17.5 km/week). Each subject gave informed consent prior to participation according to a protocol approved by the Stanford University Institutional Review Board.

2.2. Experimental protocol

Subjects were equipped with neutral running shoes (Saucony Ride 7, Saucony, Lexington, MA, USA) to eliminate footwear effects on injury risk parameters (Giandolini et al., 2013) and ensure neither foot strike pattern was promoted (Squadrone et al., 2015). Pressure sensitive insoles (Pedar, Novel Electronics Inc., St. Paul, MN, USA) were inserted into the shoes. Subjects were given a minimum of 5 min to warm up on a treadmill (Pro XL, Woodway USA Inc., Waukesha, WI, USA), and acclimate to the running shoes at their self-selected pace (3.00 ± 0.34 m/s) and preferred cadence. During the treadmill warmup, rearfoot strike type was confirmed using high speed video and by ensuring pressure within the heel during foot strike using pressure insoles, and preferred cadence was calculated using the number of strides in one minute.

We next placed 43 reflective markers and a tri-axial accelerometer (Lumo Bodytech Inc., Palo Alto, CA, USA) on each subject. We used a full-body marker set adapted from Hamner and Delp (2013) that included 27 markers placed on bony landmarks and 16 tracking markers on the lower limbs, and placed the accelerometer on the distal tibia of each subject's dominant limb. Marker positions were tracked at 200 Hz using a motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA), tibial accelerations were collected at 400 Hz with an acceleration range of ±16 g using a modified iOS application, and ground reaction forces were collected at 2000 Hz using in-ground force plates (Bertec Corporation, Columbus, OH, USA). Data were collected from a static standing pose and bilateral hip circumduction trials, which included five stiff-knee full range-of-motion leg circles for each leg (Piazza et al., 2001). Baseline running data were collected with subjects running continuously overground at their self-selected pace, preferred cadence, and with a rearfoot strike pattern along a 16.5 m runway, with force plates located 9.8 m from the beginning of the runway, for approximately 10 min. Subjects' running speed was verified to be within five percent of their self-selected pace throughout the protocol using timing gates (Fusion Sport, New Zealand).

Overground data were collected after subjects converted to a forefoot striking pattern and after increasing cadence, in a randomized order (Fig. 1a). We did not restrict the subjects' cadence during the forefoot striking trials or their foot strike during the increased cadence trials. Subjects were given 5–7 min to adapt to each new running pattern while running on a treadmill. While practicing a forefoot striking pattern, subjects were given visual feedback from the pressure insoles and told to keep the pressure on the front half of their feet (Fig. 1b); this feedback was not provided during overground running. Subjects practiced increasing their cadence by 10% compared to their original treadmill cadence using auditory cueing from a metronome; this auditory cueing continued during overground running. During data collection, runners were given feedback about their ability to maintain their preferred speed, but not their ability to convert strike pattern or increase cadence. Runners' compliance to these running adaptations was confirmed during post hoc analysis. After training, subjects ran overground for a maximum of 10 min using each new running pattern. Between experimental conditions, subjects were given 2 min to run on the treadmill at their preferred cadence, calculated to be within 1–2 strides per minute of their original cadence, and strike type. All runners successfully returned to baseline before practicing the other running pattern.

2.3. Kinematic and kinetic analysis

Joint kinematics were estimated using a 29 degree-of-freedom musculoskeletal model (Rajagopal et al., 2016). The model included six degrees of freedom describing the position and orientation of the pelvis in space, ball-and-socket joints representing the hips, one degree-of-freedom custom joints representing the knees, and revolute joints representing the ankles. Additionally, ball-and-socket joints captured lumbar and shoulder rotations, and revolute joints captured elbow flexion and forearm pronation. For

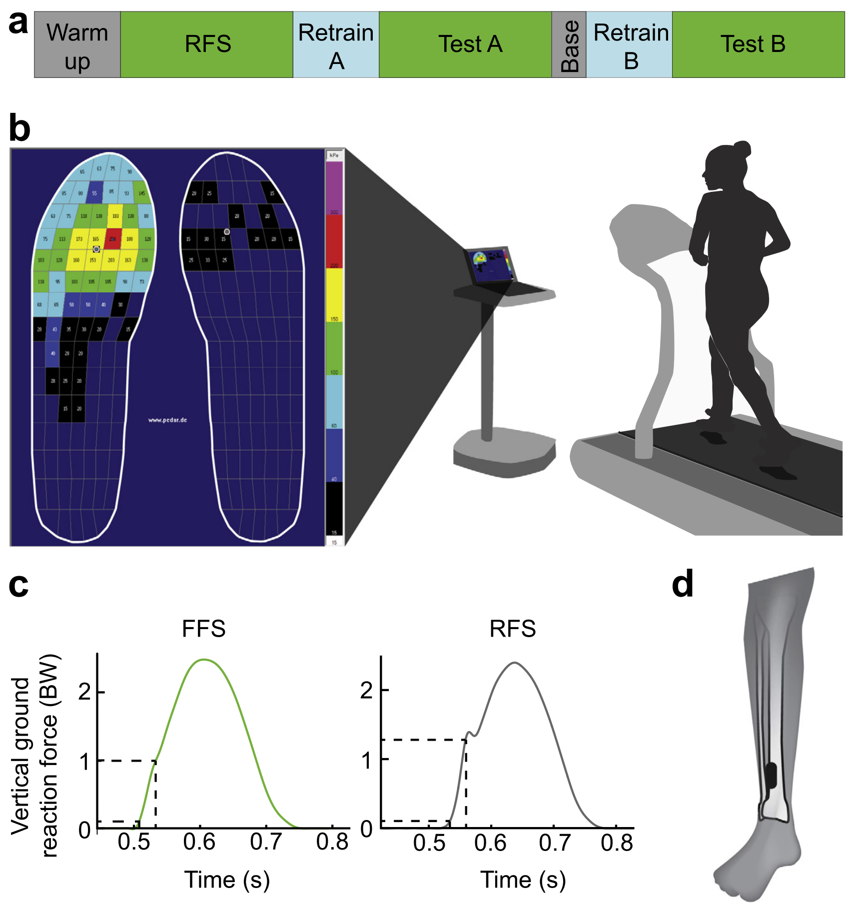

Fig. 1. Habitual rearfoot striking (RFS) runners were taught new running patterns, forefoot striking (FFS) and increasing cadence (CAD), using short-term training interventions while data were collected on parameters associated with tibial stress fractures. (a) Subjects were tested (green) during three different running patterns beginning with a RFS pattern, and then were retrained to FFS and CAD (blue) in a random order. During warm-up and between retraining sessions, subjects were asked to run with their habitual RFS pattern (gray). (b) Pressure sensitive insoles were used to train runners to switch to a FFS pattern and visually confirm if the runner was RFS or FFS during data collection. (c) Average loading rate was calculated from vertical ground reaction forces by evaluating the difference in force between a threshold of 50 N and 25 ms after that threshold was reached (dotted lines). (d) Tibial accelerations were collected using an accelerometer (Lumo Body Tech) that was placed on the distal anteromedial shank of the dominant leg. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

each subject, we scaled the model using markers placed on anatomical landmarks, taken from the static trial, and virtual hip joint centers, estimated from hip circumduction trials (Piazza et al., 2004). Kinematics were then calculated using an inverse kinematics algorithm that minimized differences between experimentally measured marker positions and virtual model markers (Delp et al., 2007).

Marker positions were used to determine foot strike pattern and calculate cadence and speed. The vertical displacement between markers placed on the heel and toe of the shoe were used to examine foot orientation at contact relative to static standing and differentiate between rearfoot and forefoot striking (Yong et al., 2014). Foot orientation greater than 0.035 m at foot strike was classified as rearfoot striking, and less than 0.02 m was classified as forefoot striking. Overground cadence was calculated after data collection by estimating the timing between successive foot strikes; foot strike was detected using the maximum vertical displacement between markers placed at the heel and sacrum (Zeni et al., 2008). Consistency in speed was ensured using the timing gates, and post hoc confirmed using the sacral marker.

Ground reaction forces and moments were filtered at 50 Hz using a 4th order low-pass filter. Average loading rate was calculated using the change in vertical ground reaction force for the first 25 ms after reaching a threshold of 50 N (Fig. 1c). Peak loading rate was defined as the peak slope found during the first 25 ms after reaching a threshold of 50 N. This method, chosen over methods that rely on percentage of the gait cycle or after reaching a force threshold, allowed for calculation of loading rates without relying on the presence of an impact peak, and yielded loading rates that were highly correlated with loading rates calculated using an impact peak dependent method ($R^2 = 0.87$, Milner et al., 2006b). The free moment was calculated from experimental ground reaction forces (e.g., Holden and Cavanagh, 1991), and the peak absolute value of the free moment during the stance phase was identified.

All kinetic and kinematic variables were calculated for each subject based on three force plate strikes from their dominant limb, and kinetic variables were normalized by body weight (BW).

2.4. Tibial accelerations

Tibial accelerations were collected from the distal anteromedial part of the dominant leg with one axis of the accelerometer aligned along the length of the tibia (Fig. 1d). Accelerations were filtered at 50 Hz using a 4th order low-pass filter. We averaged 12 acceleration peaks from the stance phase of steady-state running.

2.5. Statistics

Repeated measures ANOVA was used to compare differences in average and peak loading rates, peak tibial acceleration, peak absolute free moment and peak hip adduction angle during the three running conditions: rearfoot striking at preferred cadence, forefoot striking at free cadence and increased cadence with free strike type. If differences were detected, paired t-tests were used with Bonferroni corrections. This method was also used to compare cadence, foot orientation at initial contact, speed, and hip, knee and ankle flexion angles at initial contact during the three different conditions. All analyses were conducted using SPSS (SPSS, IBM, Armonk, NY, USA), and significance for all analyses, before any corrections, was set at $p < 0.05$. All results are reported as mean ± one standard deviation.

3. Results

Individuals were able to perform forefoot striking and increased cadence running patterns while running overground after practicing with feedback on a treadmill. We confirmed runners converted foot strike pattern while running overground, and observed significant differences in foot orientation at foot contact between rearfoot striking and forefoot striking ($p < 0.001$) (Fig. 2a). We also found 4.0 ± 3.3% greater cadence in forefoot striking compared to overground rearfoot striking trials ($p < 0.001$). During the increased cadence trials, runners increased their cadence by 11.4 ± 4.4% compared to overground rearfoot striking trials (Fig. 2b, $p < 0.001$). Foot orientation also changed during increased cadence trials compared to rearfoot striking trials ($p = 0.004$), and strike type changed in two subjects (Fig. 2c). Although runners increased their cadence during the forefoot striking trials and adjusted their foot orientation towards forefoot striking during the increased cadence trials, these changes were not to the same degree as during the alternate conditions (Fig. 2c).

We found differences in injury risk parameters between rearfoot striking, forefoot striking and increased cadence trials. Average and peak loading rates decreased from 48.8 ± 8.4 and 77.7 ± 19.0 BW/s during rearfoot striking trials to 28.8 ± 6.7 and 41.6 ± 9.3 BW/s during forefoot striking trials ($p < 0.001$) (Fig. 3a). Increased cadence trials had average and peak instantaneous loading rates of 48.0 ± 9.3 and 77.7 ± 20.0 BW/s, which were not significantly different from rearfoot striking trials ($p = 1.000$) (Fig. 3a). After Bonferroni corrections, peak hip adduction angle during forefoot striking trials (13.4 ± 3.6°) was not different compared to rearfoot striking trials (14.4 ± 4.2°, $p = 0.062$), but a significant decrease in peak hip adduction angle was observed during increased cadence trials (12.5 ± 3.6°) compared to rearfoot striking trials ($p = 0.001$) (Fig. 3b).

We did not find differences in peak tibial accelerations or the peak absolute free moments among running patterns (Fig. 3). During rearfoot striking trials, peak tibial acceleration was 5.4 ± 1.3 g, compared to 5.2 ± 1.4 g during forefoot striking trials and 5.5 ± 1.5 g during increased cadence trials. Peak absolute free moment was 9.3E−3 ± 3.3E−3 Nm/BW during rearfoot striking trials, 10.6E−3 ± 2.9E−3 Nm/BW during forefoot striking trials and 10.0E−3 ± 2.8E−3 Nm/BW during increased cadence trials.

We also examined how subjects' kinematics differed at initial contact among these three running patterns (Fig. 4). Subjects had a more flexed knee ($p < 0.001$) and a more plantarflexed ankle ($p < 0.001$) at initial contact during forefoot striking compared to rearfoot striking trials, but no significant differences in hip flexion angle ($p = 1.000$). During increased cadence trials, subjects had a more plantarflexed ankle ($p < 0.001$) at initial contact, and no significant differences in knee ($p = 0.106$) or hip flexion ($p = 0.124$) compared to rearfoot striking.

4. Discussion

This study's findings suggest that parameters associated with tibial stress fractures can be reduced by altering running pattern; forefoot striking with free cadence reduces loading rates, and increasing cadence with free foot strike decreases peak hip adduction angle. In our study, we independently evaluated both running pattern modifications in the same population, which allowed us to compare changes to injury risk parameters and improve understanding of differences between the two modifications. We additionally showed that providing visual feedback on a treadmill successfully prepared runners to use a forefoot striking pattern during overground running, and that recreational runners were able to increase cadence without changing speed during overground running while receiving auditory cueing. These methods are valuable tools for retraining foot strike pattern or cadence in future studies.

Compared to rearfoot striking, average and peak loading rates decreased during forefoot striking, but were unchanged when increasing cadence. These trends are consistent with a study that focused on acute adaptations, including transitioning to midfoot striking and increasing cadence, during treadmill running (Giandolini et al., 2013), and studies that compared rearfoot strikers transitioning to forefoot striking (Boyer et al., 2014; Shih et al., 2013), but were inconsistent with one study that found a decrease in loading rate with increased cadence (Willy et al., 2016a). In our study, average and peak loading rates decreased by 38.1% and

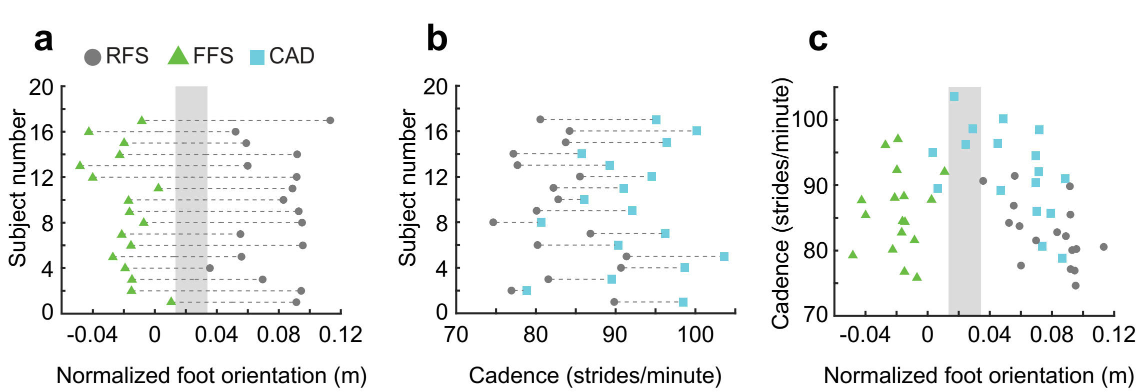

Fig. 2. Cadence and foot strike changed between RFS (circle), FFS (triangle) and CAD (square). (a) Normalized foot orientation, calculated from markers placed on the subjects' shoes, during trials for RFS and FFS. (b) Cadence during trials for RFS and CAD. (c) Average cadence and normalized foot orientation for all subjects during trials for RFS, FFS and CAD. Gray areas in (a) and (c) represent transition between FFS and RFS.

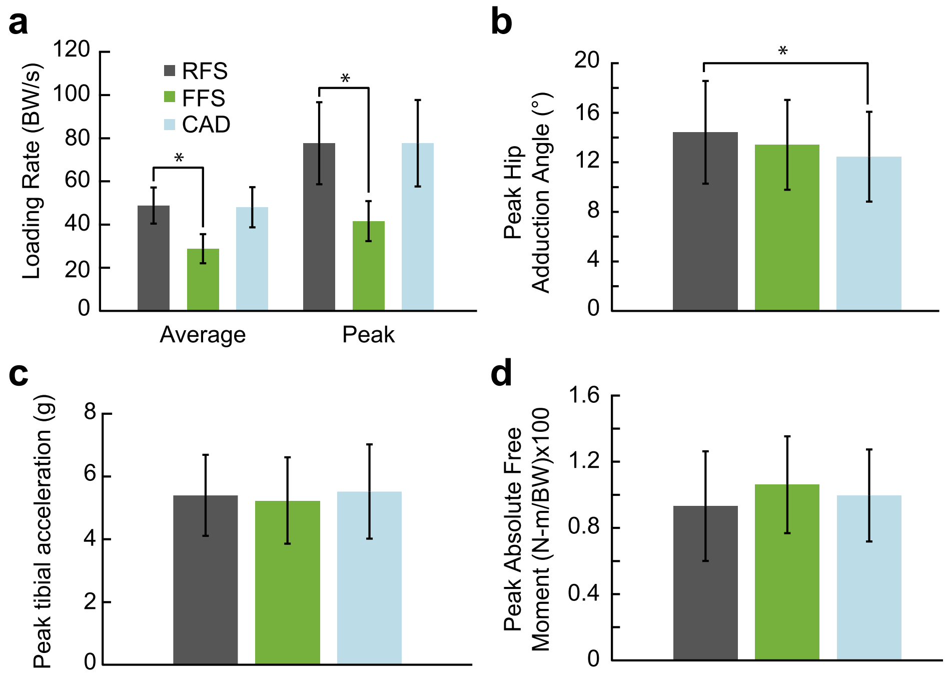

Fig. 3. Some parameters associated with tibial stress fractures differ between RFS (gray), FFS (green) and CAD (blue). Ensemble averaged ± one standard deviation for (a) average and peak loading rates, (b) peak hip adduction angle, (c) peak tibial acceleration, and (d) peak absolute free moment. * indicates a significant difference ($p < 0.05$) after Bonferroni corrections. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

39.3% respectively, compared to the 29% reduction in loading rate reported by Boyer et al. (2014) and the 56.9% reduction reported by Giandolini et al. (2013). These reductions in loading rates are greater than the 13.9% and 16.0% reduction in peak and average loading rates between runners with a history of tibial stress fractures and healthy controls observed by Milner et al. (2006b). Unlike studies whose subjects ran on treadmills (Heiderscheit et al., 2011; Lenhart et al., 2013), we did not observe increases in knee flexion angle at foot contact during increased cadence trials. Due to greater energy absorption in the knee (Pollard et al., 2010), increases in knee flexion angle during forefoot striking trials, but not during increased cadence trials, may contribute to differences in loading rates. We did not find significant differences in average peak tibial accelerations among the three different running patterns, which is inconsistent with previous studies that reported increases in tibial accelerations during forefoot striking (Laughton et al., 2003), and decreases in tibial accelerations at higher cadences (Derrick et al., 1998). These inconsistencies may be due to study design differences. While our study provided visual feedback during retraining and confirmed conversion to forefoot striking, Laughton et al. (2003) verbally instructed runners, but did not provide feedback nor report success of forefoot striking conversion. To increase cadence, Derrick et al. (1998) altered stride length using floor markers whereas we directly altered step rate, which may affect kinematic changes. Finally, in contrast to our study, Crowell and Davis (2011) focused on runners with high baseline tibial accelerations. Although we found no significant differences in peak tibial accelerations, forefoot striking may be better for reducing loading rates.

Compared to rearfoot striking, small but consistent reductions in peak hip adduction angle were observed when running with increased cadence; 14 of the 17 subjects reduced peak hip adduction angle. The 1.9° average reduction in peak hip adduction angle is consistent with previous studies (Boyer and Derrick, 2015; Heiderscheit et al., 2011; Willy et al., 2016a) which also found 1.1°–2.9° decreases, with Willy et al. (2016a) finding a change of 2.5° after one month. Since peak hip adduction angle was reported to be 4° higher in runners with a history of tibial stress fractures (Pohl et al., 2008), running with an increased cadence may be beneficial because this modification reduces hip adduction angle without increasing peak absolute free moment.

A subset of the risk parameters reported here may be useful in assessing risk for other injuries, such as the correlation between high peak hip adduction angle and iliotibial band syndrome (Ferber et al., 2010) and patellofemoral pain (Willson and Davis, 2008). More work is needed to identify how changes to running pattern affect risk for these injuries.

Several limitations of our study should be considered. First, some of the subjects were unable to reliably achieve a 10% increase from their preferred treadmill cadence while running overground during the increased cadence trials. On average, however, cadence increased by 11.4% from overground preferred cadence, and, using the sacral marker for estimating speed, we found no difference in speeds between the rearfoot striking and increased cadence trials. We did not limit our subjects to those with a naturally low cadence, potentially biasing our results to favor greater changes when converting to forefoot striking pattern. However, when retesting our hypotheses with the subset ($n = 8$) of runners with a low cadence (Luedke et al., 2016), we found the same changes in injury risk parameters.

Second, because our study investigated short-term retraining effects, we cannot claim the changes in injury risk parameters are representative of changes after long-term adaptation. While our study subjects confirmed comfort with the new running patterns after practicing for 5–7 min, it has not been reported how much time is needed to adapt to new running patterns. However, runners without experience on a treadmill require 6–10 min to acclimate to treadmill running (Lavcanska et al., 2005).

Finally, our study is unable to identify the independent effects of altering foot strike or increasing cadence. Since we calculated a very weak negative correlation ($r = -0.187$) between cadence and normalized foot orientation ($p = 0.010$), we addressed this

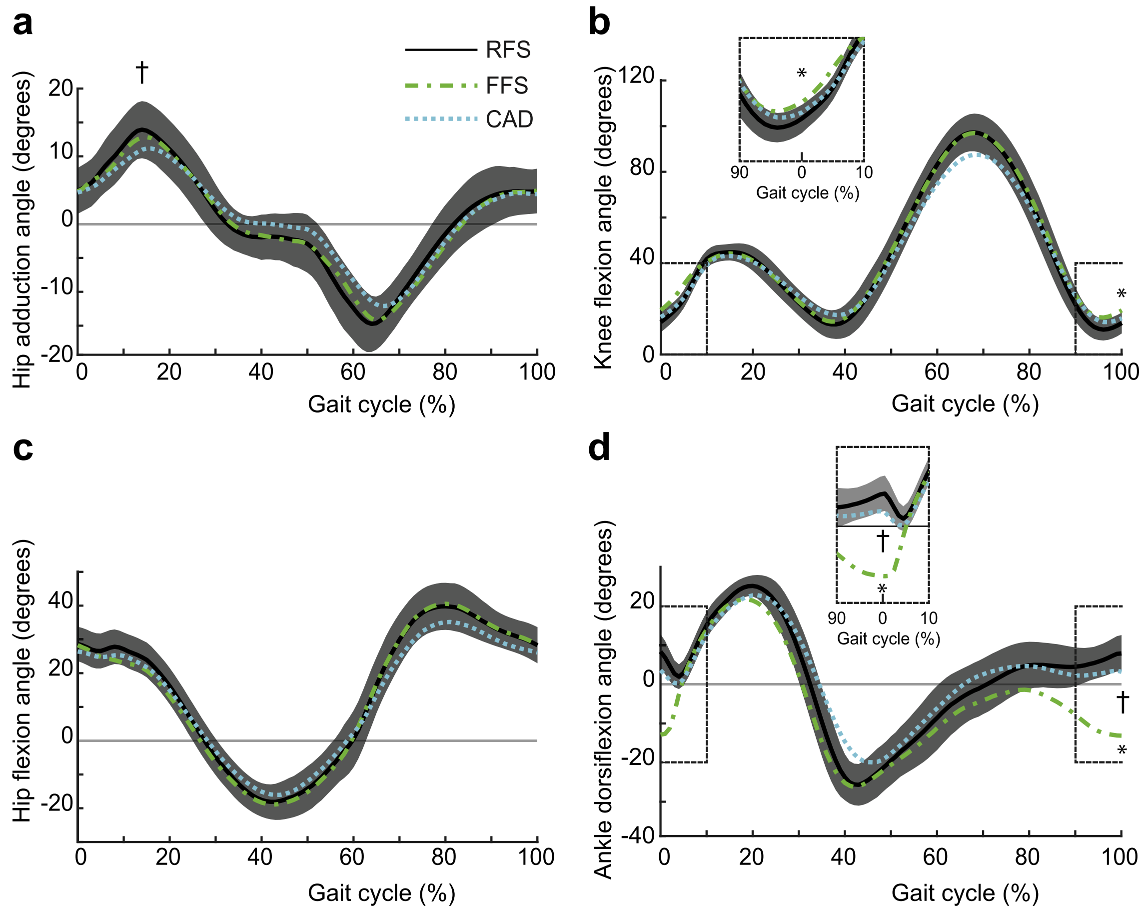

Fig. 4. Kinematics differ between running patterns. Ensemble averaged kinematics including (a) hip adduction, (b) knee flexion, (c) hip flexion and (d) ankle dorsiflexion angles for RFS (black solid with ± one standard deviation in gray), FFS (green dashed) and CAD (blue dotted). Peak hip adduction angle was reduced during CAD ($p = 0.001$), but not during FFS ($p = 0.062$). During FFS, knee flexion angle increased ($p < 0.001$) and ankle plantarflexion angle increased ($p < 0.001$) at initial contact compared to RFS. During CAD, ankle plantarflexion angle also increased ($p < 0.001$) compared to RFS. * indicates a significant difference ($p < 0.05$) between RFS and FFS after Bonferroni corrections. † indicates a significant difference ($p < 0.05$) between RFS and CAD after Bonferroni corrections. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

limitation by creating descriptive linear regression models for our injury risk parameters using all subject trials in all running conditions with cadence and normalized foot orientation as independent variables and dummy variables to account for repeated measures. The results of these regression models were consistent with our reported results. Only normalized foot orientation was a significant predictor for peak and average loading rates with a lower normalized foot orientation corresponding to lower loading rates. Only cadence was a significant predictor for peak hip adduction angle with increasing cadence corresponding to lower angles. Neither normalized foot orientation nor cadence were significant predictors for peak absolute free moment. Although we used post hoc analyses to identify independent effects, we were able to analyze natural changes in gait pattern that occur when runners were asked to focus on changing only one aspect of their gait pattern. With this study design, we observed how cadence increased when runners transitioned to forefoot striking and how foot strike changed when runners increased cadence. Our processed data, including cadence, normalized foot orientation, and injury risk parameters for all subject trials, are available to download at simtk.org/projects/rfs-ffs-cad-tsf.

We trained runners to change their strike pattern and cadence using tools that may be available in running clinics. We identified that rearfoot striking, which is most commonly found in long distance recreational runners (Hasegawa and Yamauchi, 2007), may not be the best running pattern to minimize parameters associated with a history of tibial stress fracture. Our results indicate that forefoot striking reduces average and peak loading rates, and that increasing cadence can reduce peak hip adduction angle, which may reduce risk of tibial stress fractures in addition to other injuries. While we cannot determine the effects of a long-term adaptation to these different running patterns or identify an ideal running pattern for all recreational runners, this study provides new insight into techniques for modifying running patterns and how different gait modifications may be beneficial for parameters associated with injury.

Conflict of interest statement

None of the authors had any financial or personal conflict of interest with regard to this study.

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References

Arendse, R.E., Noakes, T.D., Azevedo, L.B., Romanov, N., Schwellnus, M.P., Fletcher, G., 2004. Reduced eccentric loading of the knee with the pose running method. Med. Sci. Sports Exerc. 36, 272–277.

Beck, B.R., Matheson, G.O., Bergman, G., Norling, T., Fredericson, M., Hoffman, A.R., Marcus, R., 2008. Do capacitively coupled electric fields accelerate tibial stress fracture healing?: A randomized controlled trial. Am. J. Sports Med. 36, 545–553.

Boyer, E.R., Derrick, T.R., 2015. Select injury-related variables are affected by stride length and foot strike style during running. Am. J. Sports Med. 43, 2310–2317.

Boyer, E.R., Rooney, B.D., Derrick, T.R., 2014. Rearfoot and midfoot or forefoot impacts in habitually shod runners. Med. Sci. Sports Exerc. 46, 1384–1391.

Crowell, H.P., Davis, I.S., 2011. Gait retraining to reduce lower extremity loading in runners. Clin. Biomech. (Bristol, Avon) 26, 78–83.

Daoud, A.I., Geissler, G.J., Wang, F., Saretsky, J., Daoud, Y.A., Lieberman, D.E., 2012. Foot strike and injury rates in endurance runners: a retrospective study. Med. Sci. Sports Exerc. 44, 1325–1334.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 54, 1940–1950.

Derrick, T.R., Hamill, J., Caldwell, G.E., 1998. Energy absorption of impacts during running at various stride lengths. Med. Sci. Sports Exerc. 30, 128–135.

Edwards, W.B., Taylor, D., Rudolphi, T.J., Gillette, J.C., Derrick, T.R., 2009. Effects of stride length and running mileage on a probabilistic stress fracture model. Med. Sci. Sport. Exerc. 41, 2177–2184.

Ferber, R., Noehren, B., Hamill, J., Davis, I.S., 2010. Competitive female runners with a history of iliotibial band syndrome demonstrate atypical hip and knee kinematics. J. Orthop. Sports Phys. Ther. 40, 52–58.

Giandolini, M., Arnal, P.J., Millet, G.Y., Peyrot, N., Samozino, P., Dubois, B., Morin, J.B., 2013. Impact reduction during running: efficiency of simple acute interventions in recreational runners. Eur. J. Appl. Physiol. 113, 599–609.

Hamill, J., Derrick, T.R., Holt, K.G., 1995. Shock attenuation and stride frequency during running. Hum. Mov. Sci. 14, 45–60.

Hamner, S.R., Delp, S.L., 2013. Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. J. Biomech. 46, 780–787.

Hasegawa, H., Yamauchi, T., 2007. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. J. Strength Cond. Res. 21, 888–893.

Heiderscheit, B.C., Chumanov, E.S., Michalski, M.P., Wille, C.M., Ryan, M.B., 2011. Effects of step rate manipulation on joint mechanics during running. Med. Sci. Sports Exerc. 43, 296–302.

Hobara, H., Sato, T., Sakaguchi, M., Nakazawa, K., 2012. Step frequency and lower extremity loading during running. Int. J. Sports Med. 33, 310–313.

Holden, J.P., Cavanagh, P.R., 1991. The free moment of ground reaction in distance running and its changes with pronation. J. Biomech. 24, 887–897.

Landreneau, L.L., Watts, K., Heitzman, J.E., Childers, W.L., 2014. Lower limb muscle activity during forefoot and rearfoot strike running techniques. Int. J. Sports Phys. Ther. 9, 888–897.

Laughton, C., Davis, I., Hamill, J., 2003. Effect of strike pattern and orthotic intervention on tibial shock during running. J. Appl. Biomech. 19, 153–168.

Lavcanska, V., Taylor, N.F., Schache, A.G., 2005. Familiarization to treadmill running in young unimpaired adults. Hum. Mov. Sci. 24, 544–557.

Lenhart, R.L., Thelen, D.G., Wille, C.M., Chumanov, E.S., Heiderscheit, B.C., 2013. Increasing running step rate reduces patellofemoral joint forces. Med. Sci. Sports Exerc., 557–564

Lieberman, D.E., Venkadesan, M., Werbel, W.A., Daoud, A.I., D'Andrea, S., Davis, I.S., Mang'Eni, R.O., Pitsiladis, Y., 2010. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature 463, 531–535.

Luedke, L.E., Heiderscheit, B.C., Williams, D.S.B., Rauh, M.J., 2016. Influence of step rate on shin injury and anterior knee pain in high school runners. Med. Sci. Sports Exerc. 48, 1244–1250.

Lun, V., Meeuwisse, W.H., Stergiou, P., Stefanyshyn, D., 2004. Relation between running injury and static lower limb alignment in recreational runners. Br. J. Sports Med. 38, 576–580.

Lyght, M., Nockerts, M., Kernozek, T.W., Ragan, R., 2016. Effects of foot strike and step frequency on Achilles tendon stress during running. J. Appl. Biomech. 32, 365–372.

Matheson, G.O., Clement, D.B., McKenzie, D.C., Taunton, J.E., Lloyd-Smith, D.R., MacIntyre, J.G., 1987. Stress fractures in athletes. A study of 320 cases. Am. J. Sports Med. 15, 46–58.

Milner, C.E., Davis, I.S., Hamill, J., 2006a. Free moment as a predictor of tibial stress fracture in distance runners. J. Biomech. 39, 2819–2825.

Milner, C.E., Davis, I.S., Hamill, J., 2005. Is dynamic hip and knee malalignment associated with tibial stress fracture in female distance runners? Med. Sci. Sport. Exerc. 37, s346.

Milner, C.E., Ferber, R., Pollard, C.D., Hamill, J., Davis, I.S., 2006b. Biomechanical factors associated with tibial stress fracture in female runners. Med. Sci. Sports Exerc. 38, 323–328.

Olin, E.D., Gutierrez, G.M., 2013. EMG and tibial shock upon the first attempt at barefoot running. Hum. Mov. Sci. 32, 343–352.

Piazza, S.J., Erdemir, A., Okita, N., Cavanagh, P.R., 2004. Assessment of the functional method of hip joint center location subject to reduced range of hip motion. J. Biomech. 37, 349–356.

Piazza, S.J., Okita, N., Cavanagh, P.R., 2001. Accuracy of the functional method of hip joint center location: Effects of limited motion and varied implementation. J. Biomech. 34, 967–973.

Pohl, M.B., Mullineaux, D.R., Milner, C.E., Hamill, J., Davis, I.S., 2008. Biomechanical predictors of retrospective tibial stress fractures in runners. J. Biomech. 41, 1160–1165.

Pollard, C.D., Sigward, S.M., Powers, C.M., 2010. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. Clin. Biomech. 25, 142–146.

Rajagopal, A., Dembia, C., DeMers, M., Delp, D., Hicks, J., Delp, S., 2016. Full body musculoskeletal model for muscle-driven simulation of human gait. IEEE Trans. Biomed. Eng. 63, 2068–2079.

Shih, Y., Lin, K.L., Shiang, T.Y., 2013. Is the foot striking pattern more important than barefoot or shod conditions in running? Gait Posture 38, 490–494.

Snyder, K.L., Snaterse, M., Donelan, J.M., 2012. Running perturbations reveal general strategies for step frequency selection. J. Appl. Physiol. 112, 1239–1247.

Squadrone, R., Rodano, R., Hamill, J., Preatoni, E., 2015. Acute effect of different minimalist shoes on foot strike pattern and kinematics in rearfoot strikers during running. J. Sports Sci. 33, 1196–1204.

Taunton, J., Ryan, M., Clement, D., McKenzie, D., Lloyd-Smith, D., Zumbo, B., 2002. A retrospective case-control analysis of 2002 running injuries. Br. J. Sports Med., 95–101

Willson, J.D., Davis, I.S., 2008. Lower extremity mechanics of females with and without patellofemoral pain across activities with progressively greater task demands. Clin. Biomech. 23, 203–211.

Willy, R.W., Buchenic, L., Rogacki, K., Ackerman, J., Schmidt, A., Willson, J.D., 2016a. In-field gait retraining and mobile monitoring to address running biomechanics associated with tibial stress fracture. Scand. J. Med. Sci. Sport. 26, 197–205.

Willy, R.W., Meardon, S.A., Schmidt, A., Blaylock, N.R., Hadding, S.A., Willson, J.D., 2016b. Changes in tibiofemoral contact forces during running in response to in-field gait retraining. J. Sports Sci. 34, 1602–1611.

Yong, J.R., Silder, A., Delp, S.L., 2014. Differences in muscle activity between natural forefoot and rearfoot strikers during running. J. Biomech. 47, 3593–3597.

Zeni, J.A., Richards, J.G., Higginson, J.S., 2008. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait Posture 27, 710–714.