Forefoot Strikers Exhibit Lower Running-Induced Knee Loading than Rearfoot Strikers

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ABSTRACT

KULMALA, J.-P., J. AVELA, K. PASANEN, and J. PARKKARI. Forefoot Strikers Exhibit Lower Running-Induced Knee Loading than Rearfoot Strikers. Med. Sci. Sports Exerc., Vol. 45, No. 12, pp. 2306-2313, 2013. Purpose: Knee pain and Achilles tendinopathies are the most common complaints among runners. The differences in the running mechanics may play an important role in the pathogenesis of lower limb overuse injuries. However, the effect of a runner's foot strike pattern on the ankle and especially on the knee loading is poorly understood. The purpose of this study was to examine whether runners using a forefoot strike pattern exhibit a different lower limb loading profile than runners who use rearfoot strike pattern. Methods: Nineteen female athletes with a natural forefoot strike (FFS) pattern and pair-matched women with rearfoot strike (RFS) pattern (n = 19) underwent 3-D running analysis at 4 m·s⁻¹. Joint angles and moments, patellofemoral contact force and stresses, and Achilles tendon forces were analyzed and compared between groups. **Results**: FFS demonstrated lower patellofemoral contact force and stress compared with heel strikers $(4.3 \pm 1.2 \text{ vs } 5.1 \pm 1.1 \text{ body weight},$ P = 0.029, and 11.1 ± 2.9 vs 13.0 ± 2.8 MPa, P = 0.04). In addition, knee frontal plane moment was lower in the FFS compared with heel strikers (1.49 \pm 0.51 vs 1.97 \pm 0.66 N·m·kg⁻¹, P =0.015). At the ankle level, FFS showed higher plantarflexor moment (3.12 \pm 0.40 vs 2.54 ± 0.37 N·m·kg⁻¹; P = 0.001) and Achilles tendon force (6.3 ± 0.8 vs 5.1 ± 1.3 body weight; P = 0.002) compared with RFS. Conclusions: To our knowledge, this is the first study that shows differences in patellofemoral loading and knee frontal plane moment between FFS and RFS. FFS exhibit both lower patellofemoral stress and knee frontal plane moment than RFS, which may reduce the risk of running-related knee injuries. On the other hand, parallel increase in ankle plantarflexor and Achilles tendon loading may increase risk for ankle and foot injuries. Key Words: RUNNING, PATELLOFEMORAL STRESS, ACHILLES TENDON FORCE, KNEE MO-MENT, FOREFOOT STRIKE, REARFOOT STRIKE

I t is expected that 37%–56% of all runners have runningrelated overuse injuries each year (37). The knee joint and Achilles tendon are the most commonly injured sites, both covering about one-fifth of all running-related injuries (37). Despite the great effort used for developing better running shoes during the past decades, the injury rate has remained at the same level (27).

Pathomechanics of the overuse injuries involves the cumulative effects of repeated overloading of the musculoskeletal structures. Compared with walking, relatively high ground reaction forces (GRF) are directed to the lower limb during the stance phase of running (23). Findings of previous studies suggest that high vertical GRF impact peak

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0195-9131/13/4512-2306/0 MEDICINE & SCIENCE IN SPORTS & EXERCISE® Copyright © 2013 by the American College of Sports Medicine DOI: 10.1249/MSS.0b013e31829efcf7 and/or loading rate may increase the risk of running-related injuries (19,40).

Foot strike pattern across the normal population exhibits a large intersubject variability during running. Runners can be categorized into rearfoot strikers (RFS), midfoot strikers (MFS), or forefoot strikers (FFS) on the basis of the landing strategy at the instant of initial ground contact. Approximately 75% of long-distance shod runners use RFS strategy, whereas the rest of the people run using either MFS or FFS strategy (10). Biomechanical comparison of the different foot strike patterns has shown that MFS and FFS demonstrate lower vertical GRF impact peak and reduced vertical GRF loading rate (5). Therefore, it has been proposed that FFS could potentially decrease running-related injuries (15,16,38). Furthermore, a recent study of Daoud et al. (6) suggests lower injury incidence in the knee and hip joints for the runners with FFS strategy compared with runners who use RFS pattern.

However, not all studies have found the association between injuries and the impact characteristics of the vertical GRF (4,23). This suggests that other biomechanical factors may also play a role in the development of running-related injuries. Moments of forces determined by an inverse dynamics approach have been shown to give a relevant approximation for internal joint loading (29,41). Previous studies have also shown a link between high gait-induced knee frontal plane moment and the presence and progression of knee disorders like osteoarthritis (3,20). Furthermore, the results of previous studies suggest that high running-induced frontal plane moment may be associated with the development of the patellofemoral pain in runners (32,33,39).

A few studies examining the effects of foot strike pattern on lower extremity mechanics have focused mainly on the sagittal plane joint moments. Findings of these studies have shown that running with FFS increases ankle contribution by causing higher plantarflexor moment (38) and Achilles tendon strain (25) compared with RFS. Arendse et al. (2) reported that FFS demonstrated lower eccentric quadriceps work during the braking phase of the running gait compared with RFS. Hence, it has been speculated that increased ankle contribution during FFS could potentially decrease efforts of the quadriceps muscle, resulting in lower knee loading; however, only limited evidence exists. Furthermore, no studies have quantified the effects of different foot strike pattern on the knee frontal plane loading.

Therefore, the purpose of this study was to examine whether runners with FFS pattern exhibit different lower limb, especially knee, loading than those who use RFS pattern. It was hypothesized that FFS would demonstrate lower internal knee extensor moment and reduced patellofemoral contact force (PFCF) and patellofemoral stress (PFS) compared with RFS. Second, because the differences in running mechanics between FFS and RFS have been previously reported to take place in the sagittal plane, we also hypothesized that no differences exist in the peak internal knee abduction moment between FFS and RFS.

METHODS

Participants. In total, 286 team sport athletes (227 women and 66 men) underwent 3-D running analysis. Foot strike patterns were first visually identified and later confirmed with motion analysis by calculating foot strike angle (FSA) according to Altman and Davis (1). The criterion for FFS was FSA $< 8^{\circ}$ (1). Those who showed changes in the foot strike pattern (FSA >8°) between running trails were not accepted for the FFS group. Finally, 19 female and 4 male athletes demonstrated acceptable FFS pattern (average FSA, $-8.6^{\circ} \pm 7.8^{\circ}$). Because of gender-related differences in running mechanics (9), only women with FFS were selected for further analysis (n = 19; age, 18.6 ± 5.0 yr; height, 1.69 ± 0.05 m; weight, 63.2 ± 9.2 kg). Pair-matched group of RFS (FSA >8°) included women with similar weight and height (n = 19; age, 17.5 ± 3.6 vr; height, 1.69 ± 0.05 m; weight, 62.8 ± 8.6 kg; FSA, $24.3^{\circ} \pm 5.6^{\circ}$). At the beginning of the study, questionnaire on players' background information was used to confirm that there was no previous history of any musculoskeletal problems, such as a recent injury or surgery, that could have an effect on the running pattern of the subject. We included players if they were injury- and

symptom-free at the onset of the study. The study was approved by the ethics committee of the Pirkanmaa Hospital District, Tampere, Finland (ETL code R10169), and was performed in accordance with the Declaration of Helsinki.

Hip strength and anthropometric parameters were measured to exclude their effects on between-group running mechanics. Isometric strength of the hip abductor muscles was tested with a hand-held dynamometer (Baseline; Fabrication Enterprises, Elmsford, NY) on the basis of the previously published method (14,34). Subjects were in a supine position, with the legs extended and the ankles dorsiflexed on the testing table. Two straps were used to stabilize the pelvis and trunk against the testing table. The examiner positioned the dynamometer approximately 2 cm proximal from the lateral malleolus, after which, the subject performed one submaximal practice trial and two test trials (at least 10-s rest period between trials) against the dynamometer fixed by the examiner. A 3-s isometric maximum voluntary contraction was conducted, and the average of two successfully completed test trials was selected for the analvsis. Two research physical therapists carried out the tests. The method has been shown to have intratester and intertester reliabilities ranging from 0.73 to 0.97 (14,34).

Navicular drop was used to assess foot structure (21). Navicular drop was defined as the difference (mm) between navicular height in barefoot standing with the subtalar joint in a neutral position and in a relaxed stance. Navicular tuberosities were palpated and marked with a pen. To determine navicular height in the subtalar joint in a neutral position, the examiner palpated the medial and lateral prominence of the talus with the thumb and forefinger during pronation and supination of the foot. Neutral position of the subtalar joint was determined when the talar prominences were congruent medially and laterally. From this position, the distance between the ground and the navicular mark was measured. The subject was then instructed to perform walking on a place and stop in relaxed stance, then the distance between navicular mark and ground was measured again. One research physical therapist carried out the tests. This method has been found to have intratester and intertester reliabilities ranging from 0.73 to 0.96 (21,31).

Tibiofemoral angle was determined from a static standing position using motion analysis measurement. Knee joint mechanical axis in the frontal plane was defined as an angle between ankle, knee, and hip joint centers calculated by Plug-in Gait model (Vicon Nexus v1.7; Oxford Metrics, Oxford, UK). This procedure has been shown to estimate mechanical axis alignment similar to full-limb weightbearing radiographs ($R^2 = 0.54$) (22). Varus alignment was when angle was >0° and valgus when angle was <0°.

Running analysis. Anthropometric measurements (height, weight, leg length, and knee and ankle diameters) and bilateral placement of 34 retroreflective markers (on the shoe over the second metatarsal head and over the posterior calcaneus, lateral malleolus, lateral shank, lateral knee, lateral thigh, anterior superior iliac spine, posterior superior iliac spine, clavicula,

sternum, seventh cervical vertebra, 10th thoracic vertebra, shoulder, elbow, two wrist markers, finger, and four head markers) were carried out according to Plug-in Gait full body model (Vicon, Oxford, UK). The subjects performed shod running along a 15-m track at $4.0 \text{ m} \text{ s}^{-1}$. Two photocells were used to control the velocity between trials (±0.2 m·s⁻¹). An eight-camera system (Vicon T40, Vicon) and a force platform (AMTI BP6001200; AMTI, Watertown, MA) were used to record marker positions and GRF data synchronously at 300 and 1500 Hz, respectively.

Marker trajectories and GRF data were low-pass filtered using a fourth-order Butterworth filter with cutoff frequencies of 12 and 50 Hz, respectively. Five successful ground contacts of the left leg were selected for the analysis. GRF data were exported into the Signal software (v.4.1; Cambridge Electronic Design, Cambridge, UK) to determine contact time on the basis of 20-N GRF thresholds. Vertical GRF impact peak determination was performed first for the RFS. Mean time from ground contact to impact peak was 0.0282 s, and this time point was used to determine the magnitude of the vertical impact force for FFS (16). Average vertical loading rate was calculated as the total change in force divided by the total change in time between 20% and 80% of the period between ground contact and vertical impact peak (19).

Kinematic and kinetic analyses as well as a calculation of the position of the center of mass (COM) were performed using the Plug-in Gait model (Vicon Nexus v1.7, Oxford Metrics). Foot contact and toe-off events were used to calculate cadence, step length, and width. COM–heel distance in the anterior–posterior direction was determined during initial ground contact as the difference between heel marker and COM. Joint angles and internal joint moments (N·m·kg⁻¹) during the stance phase of running were determined across five successful force plate contacts of the left leg.

Patellofemoral joint contact force during running was then estimated as a function of knee flexion angle (x) and knee extensor moment (M_k) according to the biomechanical model described by Ho et al. (13). First, an effective moment arm of the quadriceps muscle (L_q) was calculated as a function of knee flexion angle using nonlinear equation, which is based on the cadaver data reported by van Eijden et al. (36):

$$L_{\rm q} = 8.0{\rm E}^{-5}x^3 - 0.013x^2 + 0.28x + 0.046$$
 [1]

Second, quadriceps force (F_q) was calculated as follows:

$$F_{\rm q} = M_{\rm k}/L_{\rm q}$$
 [2]

Finally, PFCF was calculated as the product of the quadriceps force (F_q) and a constant (k):

$$PFCF = F_q k$$
[3]

The constant k was estimated for knee joint angle position (x) using the following nonlinear equation on the basis of the curve fitting to the data of van Eijden et al. (35):

$$k(x) = (4.62E^{-1} + 1.47E^{-3}x^2 - 3.84E^{-5}x^2)/(1 - 1.62E^{-2}x + 1.55E^{-4}x^2 - 6.98E^{-7}x^3)$$
[4]

PFS was then calculated as the PFCF divided by the patellofemoral contact area. Contact area was estimated according to Ho et al. (13) by fitting a second-order polynomial curve to the data of Powers et al. (26) (83 mm² at 0° , 140 mm² at 15° , 227 mm² at 30° , 236 mm² at 45° , 235 mm² at 60° , and 211 mm² at 75° of knee flexion).

$$PFS = PFCF/contact area$$
 [5]

Achilles tendon force (ATF) was determined by dividing the plantarflexion moment (calculated by inverse dynamics) by the estimated Achilles tendon lever arm (L_a) as described by Self and Paine (30):

$$ATF = M_a/L_a$$
 [6]

$$L_{\rm a} = -0.5910 + 0.08297a - 0.0002606a^2$$
[7]

where a = ankle angle.

Statistical analysis. Kinematic and kinetic data were time normalized (0%-100%) and averaged across five trials to get individual mean curves for the parameters of interest. Running-induced knee loading was evaluated by defining the magnitude of the peak knee extensor and abduction moment, peak PFCF, and peak PFS. Ankle loading was evaluated by defining the magnitude of the peak ankle plantarflexor moment and the peak ATF. In addition, selected variables including subject, spatiotemporal, and secondary kinematic and kinetic parameters were chosen for comparison on the basis of the previous studies to ensure that the general running mechanics of the FFS and RFS in the current study is in line with those reported earlier (5,16,25,38). Univariate differences between RFS and FFS groups were compared with two-tailed independent *t*-tests, and 95% confidence intervals were calculated and reported for all comparisons (SPSS 18.0; SPSS, Chicago, IL). P values less than 0.05 were considered significant.

RESULTS

FFS and RFS groups were not different as regards age, weight, and height (Table 1). In addition, there were no differences present in the hip abductor strength, navicular drop, or tibiofemoral angle (Table 1).

During running, FFS demonstrated lower peak hip adduction (mean difference, -3.9°) during stance phase than RFS runners (P = 0.010, Table 2). There was also a tendency to lower abduction moment in the hip of the FFS runners (P = 0.095). At the knee level, RFS showed significantly greater peak flexion angle (mean difference, 4.0°) during the stance phase of running (P = 0.003; Fig. 1A, Table 2). In addition, FFS exhibited 16% lower PFCF (P = 0.029) and 15% lower PFS (P = 0.041) compared with RFS (Fig. 1D and E, Table 2) and a trend toward lower peak knee extensor moment (P = 0.088; Fig. 1B, Table 2). In the frontal plane, peak knee abduction moment was 24% lower in the FFS group compared with RFS (P = 0.015; Fig. 1C, Table 2).

TABLE '	1. Mean	(SD)	subject	parameters	for	the	RFS and FFS.
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Subject Parameters	RFS (<i>n</i> = 19)	FFS (<i>n</i> = 19)	Mean Difference (95% CI)	P Value
Age (yr)	17.5 (3.6)	18.6 (5.0)	1.1 (-1.8 to 4.0)	0.439
Weight (kg)	62.8 (8.6)	63.2 (9.2)	0.4 (-5.6 to 6.2)	0.915
Height (m)	1.69 (0.05)	1.69 (0.05)	0.00 (-0.04 to 0.03)	0.749
Hip abductor strength (kg)	10.1 (2.2)	10.8 (2.8)	0.7 (-0.9 to 2.4)	0.371
Navicular drop (mm)	6.7 (3.9)	4.7 (3.1)	-2.0 (-4.3 to 0.4)	0.101
Tibiofemoral angle (°)	-0.4 (3.5)	-0.2 (4.5)	0.2 (-2.5 to 2.8)	0.921

CI, confidence interval.

At the ankle level, FFS runners had less dorsiflexion (mean difference, 22.6°) at the initial ground contact (P = 0.001; Fig. 2A, Table 2), and they demonstrated 19% higher peak plantarflexor moment (P = 0.001) and 19% higher ATF (P = 0.002) during the stance phase of running (Fig. 2B and C, Table 2). Vertical GRF showed 26% lower impact peak (P = 0.001) and 47% lower average loading rate (P = 0.001)in the FFS compared with RFS (Fig. 2D, Table 2). Spatiotemporal comparison showed significantly shorter contact time for the FFS than RFS (P = 0.001, Table 2). In addition, FFS runners demonstrated shorter COM-heel distance (mean difference, 0.057 m) at initial ground contact (P = 0.001). No differences were present in the cadence, step length, or width between groups.

DISCUSSION

The primary aim of the current study was to examine whether FFS runners exhibit lower knee loading than runners with RFS pattern. It was hypothesized that FFS runners would demonstrate lower knee extensor moment, PFCF, and PFS. This hypothesis was partially supported in that FFS showed lower knee PFCF and PFS but not significantly lower knee extensor moment than RFS. For the second hypothesis, it was suggested that there would not be a difference in the frontal plane knee moment between groups. However, our results do not support this hypothesis, suggesting that FFS demonstrates significantly lower peak internal knee abduction moment compared with RFS.

To the best of our knowledge, the present study was the first one to show that runners with FFS exhibit lower PFCF and PFS than those who run with RFS. This finding may be important from a clinical point of view because one of the most widely accepted theories regarding the etiology of the patellofemoral pain suggests that the symptoms are the result of excessive patellofemoral joint stress (12). Because patellofemoral pain is the most common disorder among runners (37), FFS may be a potential way to decrease the knee injury risk via reduction of the patellofemoral joint loading.

Interestingly, our results also suggest that FFS demonstrate lower peak knee abduction moment in the frontal

Parameters	RFS (<i>n</i> = 19)	FFS (<i>n</i> = 19)	Mean Difference (95% CI)	P value	Effect Size
Нір					
Flexion at initial contact (°)	46.0 (5.1)	43.3 (5.1)	-2.7 (-6.0 to 0.7)	0.112	
Peak adduction (°)	17.0 (4.8)	13.1 (3.9)	-3.9 (-6.8 to -1.0)	0.010	0.89
Extensor moment (N·m·kg ⁻¹)	2.11 (1.11)	2.15 (0.66)	0.04 (-0.56 to 0.64)	0.892	
Flexor moment (N·m·kg ⁻¹)	-2.79 (1.62)	-2.98 (0.83)	-0.18 (-1.03 to 0.66)	0.660	
Abduction moment (N·m·kg ⁻¹)	2.36 (0.47)	2.10 (0.46)	-0.26 (-0.56 to 0.05)	0.095	
Knee					
Flexion at initial contact (°)	21.0 (3.6)	22.6 (5.6)	1.6 (-1.5 to 4.7)	0.308	
Flexion max (°)	50.9 (3.1)	46.9 (4.5)	-4.0 (-6.5 to -1.5)	0.003	1.04
Peak adduction (°)	6.3 (6.4)	8.4 (6.5)	2.1 (-2.0 to 8.4)	0.310	
Extensor moment (N·m·kg ⁻¹)	3.54 (0.69)	3.13 (0.77)	-0.42 (-0.90 to 0.06)	0.088	
Abduction moment (N·m·kg ⁻¹)	1.97 (0.66)	1.49 (0.51)	-0.48 (-0.87 to -0.10)	0.015	0.81
PFCF (BW)	5.1 (1.1)	4.3 (1.2)	-0.82 (-1.6 to -0.1)	0.029	0.69
PFS (MPa)	13.0 (2.8)	11.1 (2.9)	-1.9 (-3.8 to -0.08)	0.041	0.67
Ankle					
Dorsiflexion at initial contact (°)	24.8 (4.3)	2.3 (9.2)	-22.6 (-27.3 to -17.9)	0.001	3.13
ATF (BW)	5.1 (1.3)	6.3 (0.8)	1.2 (0.5 to 1.9)	0.002	1.11
Plantarflexion moment (N·m·kg $^{-1}$)	2.54 (0.37)	3.12 (0.40)	0.59 (0.33 to 0.84)	0.001	1.51
GRF					
Vertical impact peak (BW)	1.93 (0.21)	1.23 (0.35)	-0.70 (-0.89 to -0.51)	0.001	2.43
Average vertical loading rate (BW·s ⁻¹)	98.5 (21.2)	51.9 (16.7)	-46.6 (-59.1 to -34.0)	0.001	2.44
Peak vertical force (BW)	2.49 (0.33)	2.69 (0.38)	0.19 (-0.04 to 0.43)	0.105	
Spatiotemporal parameters					
Velocity (m·s ⁻¹)	4.0 (0.2)	4.0 (0.2)	0.0 (-0.1 to 0.1)	0.931	
Cadence (steps⋅min ⁻¹)	175 (17)	186 (20)	11 (-2 to 23)	0.090	
Contact time (s)	0.219 (0.015)	0.196 (0.019)	-0.023 (-0.034 to -0.012)	0.001	1.34
Step length (m)	1.45 (0.09)	1.43 (0.10)	-0.02 (-0.08 to 0.04)	0.451	
Step width (m)	0.05 (0.05)	0.06 (0.05)	0.00 (-0.03 to 0.03)	0.932	
COM-heel distance at initial contact (m)	0.142 (0.030)	0.085 (0.026)	-0.057 (-0.074 to -0.037)	0.001	2.03

BW, body weight: CI, confidence interval.

EFFECTS OF STRIKING PATTERN ON KNEE LOADING



FIGURE 1-Knee kinematics and kinetics.

plane compared with RFS. High knee frontal plane moment is known to correspond with the increased medial compartment knee loading (29,41) and is commonly linked with the presence and progression of degenerative knee disorders such as medial tibiofemoral osteoarthritis (3,20). Furthermore, it has been suggested that high running-induced frontal plane moment and impulse could increase loading of the lateral aspect of the patellofemoral joint and thus contribute to the pathomechanics of the patellofemoral pain (32,33). Lower knee frontal plane moment combined with a shorter contact time in FFS may presumably reduce abductor impulse during a single step. However, although the peak knee load per step demonstrated reduction during FFS, the amount of cumulative knee load (number of steps \times amount of load) may stay nearly the same between FFS and RFS because a shorter contact time and a trend toward higher cadence in the FFS increases the total number of steps needed in the same distance when compared with RFS. There is a lack of knowledge which one is more important with respect to developing knee overuse injury: the amount of repetitive knee load or the magnitude of the peak knee load during running gait.

The factors that are generally linked to altered knee mechanics during running include static knee frontal plane alignment (24), height of the foot arch (38), and hip abductor muscle strength (8). However, these factors do not show significant differences between groups, suggesting that the difference in the magnitude of the knee frontal plane moment is due to distinct running mechanics between FFS and RFS runners.

Previously, higher step rate at the same running speed is reported during FFS when using minimalist footwear (7). The higher step rate is shown to decrease step length and reduce joint energy absorption (negative work) during the first half of running gait (11). However, contrary to expectations, significant differences in the step rate and step length were not observed between FFS and RFS. This may be due to the fact that both FFS and RFS performed running trials with normal footwear in the current study, and that the running velocity in the current study was higher than what was used in the study of Divert et al. (7).

The spatiotemporal comparison showed lower contact time and shorter distance between COM and heel during initial contact for the FFS. In addition, lower hip adduction angle was present in the FFS. It has been known that the magnitude of the knee frontal plane moment is highly related to the length of the GRF vector lever arm at the knee in the frontal plane (28). Because no differences were observed in the frontal plane knee angle, it can be assumed that a closer heel location to the COM and/or lower peak hip adduction in FFS can change the position of the GRF vector in relation to the lower



FIGURE 2—Ankle kinematics and kinetics and vertical GRF.

limb so that the lever arm in the frontal plane decreases, leading to a reduction of the knee frontal plane moment. Future studies are needed to evaluate these theories.

We also evaluated ankle joint kinetics and vertical GRF parameters because these have been shown to differ remarkably between FFS and RFS. The results of the current study are in line with the previous findings (5,16,25,38), suggesting significantly higher ankle plantarflexor moment and ATF but lower vertical GRF impact peak and loading rate when running with FFS pattern. Lower GRF impact loading during FFS has shown to be a result of increased ankle joint energy absorption during the first half of stance (16). On the other hand, greater loading of plantarflexors and Achilles tendon may potentially increase the risk of stress-related injuries of the ankle and foot (Achilles tendinopathy, plantar fasciitis, and metatarsal stress fractures).

Higher ankle contribution may be the main mechanism to explain lower PFCF and PFS in FFS because the role of the knee joint as an energy absorber reduces, resulting in lower knee flexion excursion during FFS. This idea is supported by Arendse et al. (2) who reported reduced eccentric quadriceps work during FFS compared with RFS. Although researchers did not report the magnitude of the knee extensor moment, PFCF, or PFS in their study, they observed lower knee flexion range of motion during FFS, which is in accordance with our findings. Because the quadriceps moment arm decreases as a function of increased knee flexion angle (18,36), greater eccentric quadriceps force is therefore needed to resist knee flexion during the first half of the stance when running with RFS pattern. This explains why greater PFCF and PFS were observed in RFS runners regardless of no significant differences in the peak knee extensor moment in the present study.

Certain limitations with the current study should be considered when interpreting its findings. Participants were relatively young female team athletes, and therefore, caution must be made in generalizing these results to males or even female runners of a different age. Our study may also underestimate the PFCF, PFS, and ATF because these variables were identified from the average of the timenormalized data over the stance phase of five trials and because the biomechanical models use the net joint moments of the knee and ankle joint, respectively, as an input parameter and therefore do not take into account the antagonist forces acting on opposing direction of the joint. As a knee flexor, greater gastrocnemius muscle force, while running with FFS, may thus lead to underestimation of the quadriceps force as well as PFCF and PFS. However, a minimal contribution of the gastrocnemius muscle to knee flexion moment production at knee angles of more than 30° (17), where peak PFCF and PFS typically occur (midstance phase), suggests that using only the net joint moment of the knee for calculation of the PFCF and PFS may have no effect on between-group differences in the current study. Finally, a large set of variables were compared between groups, which raises concern over the familywise type I error rate for this study.

In conclusion, the findings of the current study suggest that runners with FFS pattern demonstrate lower knee PFCF and PFS as well as lower frontal plane moment compared with runners with RFS pattern. This may decrease the risk of

REFERENCES

- Altman AR, Davis IS. A kinematic method for footstrike pattern detection in barefoot and shod runners. *Gait Posture*. 2012;35(2): 298–300.
- Arendse RE, Noakes TD, Azevedo LB, Romanov N, Schwellnus MP, Fletcher G. Reduced eccentric loading of the knee with the pose running method. *Med Sci Sports Exerc.* 2004;36(2):272–7.
- Baliunas AJ, Hurwitz DE, Ryals AB, et al. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis Cartilage*. 2002;10(7):573–9.
- Bredeweg S, Buist I. No relationship between running related injuries and kinetic variables. Br J Sports Med. 2011;45(4):328.
- Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. J Biomech. 1980;13(5):397–406.
- Daoud AI, Geissler GJ, Wang F, Saretsky J, Daoud YA, Lieberman DE. Foot strike and injury rates in endurance runners: a retrospective study. *Med Sci Sports Exerc*. 2012;44(7):1325–34.
- Divert C, Mornieux G, Baur H, Mayer F, Belli A. Mechanical comparison of barefoot and shod running. *Int J Sports Med.* 2005;26(7):593–8.
- 8. Earl JE, Hoch AZ. A proximal strengthening program improves pain, function, and biomechanics in women with patellofemoral pain syndrome. *Am J Sports Med.* 2011;39(1):154–63.
- Ferber R, Davis IM, Williams DS 3rd. Gender differences in lower extremity mechanics during running. *Clin Biomech (Bristol, Avon)*. 2003;18(4):350–7.
- Hasegawa H, Yamauchi T, Kraemer WJ. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *J Strength Cond Res.* 2007;21(3):888–93.
- Heiderscheit BC, Chumanov ES, Michalski MP, Wille CM, Ryan MB. Effects of step rate manipulation on joint mechanics during running. *Med Sci Sports Exerc.* 2011;43(2):296–302.
- Heino Brechter J, Powers CM. Patellofemoral stress during walking in persons with and without patellofemoral pain. *Med Sci Sports Exerc.* 2002;34(10):1582–93.
- Ho KY, Blanchette MG, Powers CM. The influence of heel height on patellofemoral joint kinetics during walking. *Gait Posture*. 2012;36(2):271–5.
- Krause DA, Schlagel SJ, Stember BM, Zoetewey JE, Hollman JH. Influence of lever arm and stabilization on measures of hip abduction and adduction torque obtained by hand-held dynamometry. *Arch Phys Med Rehabil.* 2007;88(1):37–42.
- Lieberman DE. What we can learn about running from barefoot running: an evolutionary medical perspective. *Exerc Sport Sci Rev.* 2012;40(2):63–72.

developing running-related knee injuries. However, increased ankle plantarflexor loading and ATF during FFS may increase the risk of injury of the ankle and foot areas. Prospective studies are needed to determine whether different loading profiles due to distinct striking patterns are associated with specific running-related injuries.

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- Lieberman DE, Venkadesan M, Werbel WA, et al. Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature*. 2010;463(7280):531–5.
- Li L, Landin D, Grodesky J, Myers J. The function of gastrocnemius as a knee flexor at selected knee and ankle angles. *J Electromyogr Kinesiol*. 2002;12(5):385–90.
- Mason JJ, Leszko F, Johnson T, Komistek RD. Patellofemoral joint forces. J Biomech. 2008;41(11):2337–48.
- Milner CE, Ferber R, Pollard CD, Hamill J, Davis IS. Biomechanical factors associated with tibial stress fracture in female runners. *Med Sci Sports Exerc*. 2006;38(2):323–8.
- Miyazaki T, Wada M, Kawahara H, Sato M, Baba H, Shimada S. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. *Ann Rheum Dis.* 2002;61(7):617–22.
- Mueller MJ, Host JV, Norton BJ. Navicular drop as a composite measure of excessive pronation. *J Am Podiatr Med Assoc.* 1993;83(4): 198–202.
- 22. Mundermann A, Dyrby CO, Andriacchi TP. A comparison of measuring mechanical axis alignment using three-dimensional position capture with skin markers and radiographic measurements in patients with bilateral medial compartment knee osteoarthritis. *Knee*. 2008;15(6):480–5.
- Nigg BM. Impact forces in running. Curr Opin Orthop. 1997; 8(6):43–7.
- Park SK, Stefanyshyn DJ. Greater Q angle may not be a risk factor of patellofemoral pain syndrome. *Clin Biomech (Bristol, Avon)*. 2011;26(4):392–6.
- Perl DP, Daoud AI, Lieberman DE. Effects of footwear and strike type on running economy. *Med Sci Sports Exerc.* 2012;44(7): 1335–43.
- Powers CM, Lilley JC, Lee TQ. The effects of axial and multiplane loading of the extensor mechanism on the patellofemoral joint. *Clin Biomech (Bristol, Avon)*. 1998;13(8):616–24.
- Richards CE, Magin PJ, Callister R. Is your prescription of distance running shoes evidence-based? Br J Sports Med. 2009; 43(3):159-62.
- Schache AG, Fregly BJ, Crossley KM, Hinman RS, Pandy MG. The effect of gait modification on the external knee adduction moment is reference frame dependent. *Clin Biomech (Bristol, Avon)*. 2008;23(5):601–8.
- Schipplein OD, Andriacchi TP. Interaction between active and passive knee stabilizers during level walking. *J Orthop Res.* 1991; 9(1):113–9.

- Self BP, Paine D. Ankle biomechanics during four landing techniques. *Med Sci Sports Exerc.* 2001;33(8):1338–44.
- Sell KE, Verity TM, Worrell TW, Pease BJ, Wigglesworth J. Two measurement techniques for assessing subtalar joint position: a reliability study. *J Orthop Sports Phys Ther.* 1994;19(3):162–7.
- Stefanyshyn DJ, Stergiou P, Lun VM, Meeuwisse WH, Worobets JT. Knee angular impulse as a predictor of patellofemoral pain in runners. *Am J Sports Med.* 2006;34(11):1844–51.
- 33. Stefanyshyn D, Stergiou P, Lun V, Meeuwisse W, Nigg B. Knee joint moments and patellofemoral pain syndrome in runners. Part I: a case control study; Part II: a prospective study. In: *Proceedings* of the 4th Symposium of Footwear Biomechanics. 1999 Aug 5–7: Canmore (Canada). University of Calgary; 1999. p. 86–7.
- Thorborg K, Petersen J, Magnusson SP, Holmich P. Clinical assessment of hip strength using a hand-held dynamometer is reliable. *Scand J Med Sci Sports*. 2010;20(3):493–501.
- 35. van Eijden TM, de Boer W, Weijs WA. The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion-extension angle. *J Biomech*. 1985;18(10):803–9.

- van Eijden TM, Kouwenhoven E, Verburg J, Weijs WA. A mathematical model of the patellofemoral joint. *J Biomech*. 1986; 19(3):219–29.
- van Mechelen W. Running injuries. A review of the epidemiological literature. *Sports Med.* 1992;14(5):320–35.
- Williams D, McClay I, Manal K. Lower extremity mechanics in runners with a converted forefoot strike pattern. *J Appl Biomech*. 2000;16:210–8.
- Willy RW, Manal KT, Witvrouw EE, Davis IS. Are mechanics different between male and female runners with patellofemoral pain? *Med Sci Sports Exerc*. 2012;44(11):2165–71.
- 40. Zadpoor AA, Nikooyan AA. The relationship between lowerextremity stress fractures and the ground reaction force: a systematic review. *Clin Biomech (Bristol, Avon)*. 2011; 26(1):23-8.
- Zhao D, Banks SA, Mitchell KH, D'Lima DD, Colwell CW, Fregly BJ. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *J Orthop Res.* 2007;25(6):789–97.