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Analysis of ground reaction forces and muscle activity in individuals with anterior cruciate ligament reconstruction during different running strike patterns

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ABSTRACT

Background: Anterior cruciate ligament reconstruction provides successful clinical outcomes. However, reconstruction cannot restore normative lower limb mechanics during running. While numerous studies have investigated running characteristics in individuals with anterior cruciate ligament reconstruction, no study has been compared foot strike patterns among them.

Research question: If ground reaction forces and lower extremity muscle activities in individuals with anterior cruciate ligament reconstruction and healthy control ones differ during three running strike patterns?

Methods: In this cross-sectional study, fourteen healthy adult males and fourteen adult males with anterior cruciate ligament reconstruction were recruited to participate. Surface electromyography of selected lower limb muscles and ground reaction forces were measured during three-strike patterns: rearfoot strike pattern, midfoot strike pattern, and forefoot strike pattern during barefoot running (~ 3.3 m/s).

Results: The results revealed that the strike patterns influenced the peak lateral ground reaction force ($P < 0.001$) and peak vertical impact ground reaction force ($P = 0.002$) during the stance phase of running for both groups. The strike pattern also influenced the tibialis anterior ($P < 0.001$) and vastus lateralis ($P = 0.035$) activities during the early stance phase for both groups. However, the vastus medialis ($P = 0.030$) presented reduced activity, and the biceps femoris ($P = 0.039$) presented increased activity in the anterior cruciate ligament reconstruction group. Tibialis anterior ($P = 0.021$), gastrocnemius medialis ($P < 0.001$) and vastus medialis ($P < 0.001$) presented lesser activity irrespective of strike patterns in the anterior cruciate ligament reconstruction group.

Significance: Running with a forefoot strike pattern may be associated with lesser rearfoot eversion due to lower peak lateral ground reaction forces than running with a rearfoot strike pattern or midfoot strike pattern. Moreover, the altered muscle activities could contribute to the elevated risk of future joint injury in the anterior cruciate ligament reconstruction population.

1. Introduction

Surgical reconstruction is the most common treatment option for individuals following anterior cruciate ligament injury [1]. The incidence of Anterior cruciate ligament reconstruction (ACLR) in the United States has increased from 86,687 (32.9 per 100,000 person-years) in 1994 to 129,836 (43.5 per 100,000 person-years) in 2006 [2]. The number of ACLR increased in patients younger than 20 years and those who were 40 years or older over 12 years [2]. ACLR provides successful clinical outcomes. However, reconstruction cannot restore normative lower limb mechanics (e.g., lower peak knee flexion angle at the loading phase) during running [3]. Changes in lower extremity mechanics are well known in individuals with ACLR to compensate for knee joint instability [4]. For example, patients with quadriceps strength <80 % of the uninjured side also had reduced knee flexion angles and moments during running between 14 and 21 weeks after ACLR [4]. However, little

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is known about muscle activity or its relationship to knee stability during different running strike patterns in individuals with ACLR.

The running biomechanical analysis is a crucial part of improving running style [5,6], including the determination of foot strike patterns [5,7,8]. Typically, the foot strike pattern is defined as a biomechanical analysis of the way the foot touches the ground [9], and three-foot strike patterns were described: the rearfoot strike (RFS), in which initial contact is made somewhere on the heel or rear one-third of the foot; the midfoot strike (MFS), in which the runner initially contacts the ground throughout the metatarsal heads with the heel subsequently contacting the running surface; and the forefoot strike (FFS), in which initial contact is also on the metatarsal heads, however, the heel never touches the ground [10].

During running, different foot strike patterns may affect Achilles tendon force, knee loading, and the potential for Achilles tendon or knee injury [11]. Mechanical consequences from the shift from an RFS to MFS or FFS pattern are thought to minimize some risk factors associated with running-related injuries such as tibial stress injuries, plantar fasciitis, tibial stress fractures, and knee injuries [12]. Several studies compared the kinetic and kinematic properties and muscle activity of these various foot strike patterns during running [13,14]. However, it is important to explore further how movement behavior can be modified or adapted in individuals with ACLR while running with various foot strike patterns. Although running, jump landing, and cutting kinematics and muscular control during the jump and cutting tasks have been studied extensively [15–18], very little is known about impact forces and the associated muscle activities of various foot strike patterns during running after ACLR. Pamukoff et al. found that the injured leg in individuals with ACLR was maintained in similar impact peak of vertical ground reaction force and greater instantaneous loading rate during the stance phase of running than that healthy control group [17]. Smeets et al. reported that athletes with ACLR show larger hamstring activation and lower vastus medialis activation than controls [18]. A previous study demonstrated that the maximum free moment amplitudes between the ACLR patients and the healthy control group were similar during walking [19]. Also, there are no differences in peak absolute free moments among different running strike patterns in healthy, habitual rearfoot striking recreational runners [20]. While numerous studies have investigated running characteristics in individuals with ACLR [17,18], no study has been compared foot strike patterns among them. The differences in the running mechanics may play an important role in the pathogenesis of lower limb overuse injuries. This study aimed to evaluate ground reaction forces and lower extremity muscle activity in individuals with ACLR during three running strike patterns. Following the relevant literature, we hypothesized that compared with the healthy control group, the individuals with ACLR running with different foot strike patterns would show a similar peak of vertical ground reaction force [17] and free moment [19,20], and increased loading rate [17]. In addition, they would show increased hamstring activity and reduced quadriceps activation [18].

2. Methods

2.1. Participants

This cross-sectional study design recruited 28 males (age 21–25), 14 healthy, and 14 with ACLR. Males and females present different lower limb mechanic profiles while walking or running. For example, males naturally generate lower hip and knee moments in the frontal plane during walking and running [5]. Distinct lower limb muscle recruitment, passive and dynamic knee stiffness, neuromuscular control, femoral notch, and pelvis width (i.e., lower Q angle) are commonly associated with the lower moments generated by males during locomotion [6–9]. Therefore, we used only males as our statistical sample in the present study. The individuals had unilateral ACLR using a hamstring tendon autograft 6–10 months before the study and without other significant ligamentous injuries, confirmed by arthroscopy. Individuals with complete meniscectomy or arthrotomy were excluded from the study, as were those with a concurrent painful condition in the lower limbs. All participants were physically active with at least three years of recreational training experience, such as walking and/or running with three sessions per week, each session lasting 40 min. Also, all participants' natural running strike pattern was RFS. The strike pattern was established through observation and kinetic data [21]. The ACLR individuals have physical activity (recreational training experience) such as walking and/or running, with three sessions per week, each session lasting 40 min after the surgery. The recreational training in ACLR individuals was similar to healthy control ones. We determinate the dominant limb for this test by asking participants who preferred kicking leg. All participants were right-footed. Moreover, the participants in the ACLR group had ACLR in the right foot. All participants provided written consent approved by the local Medical University's Ethics Committee (Code number: IR.ARUMS.REC.1397.201).

2.2. Ground reaction force measurements

Before the experiment, we allowed participants to run freely or warm up exercises for five minutes to become familiar with the experimental environment. Running trials were performed along a 20 m runway, providing practice trials to ensure each participant was comfortable contacting a force platform (Bertec Corporation, Columbus, OH, USA) embedded in the middle of the runway with the right foot, without targeting. Foot strike conditions included RFS, MFS, and FFS. Participants had a period of rest from 30 s to 1 min between each trial, with 5 min provided between conditions to reduce participant fatigue. We also offered to participants instructional videos illustrating the desired foot strike patterns, with participants verbally instructed to match the demonstrated foot strike pattern to the best of their ability. Participants practiced (typically 3 trials) each foot strike condition before data collection. Participants ran through timing gates synchronized via a multi-function timer, separated by 7 m (Bertec Corporation, Columbus, OH, USA). Trials outside of an adequate time window (±5% constant speed of ~ 3.3 m/s) were discarded and repeated, along with tests visibly altered in an attempt to contact the force platform or deviating from the desired foot strike pattern. Participants were considered adapted once the desired foot strike pattern, target running speed, and foot placement were visibly established in each condition. Blocked foot strike conditions were carried out in randomized order for each participant. Participants completed 15 successful running trials (15 trials used during analysis) in 3-foot strike conditions (3 conditions × five trials per condition) at a speed of ~3.3 m/s. Running at this speed has previously been used for determining running-related risk factors of injuries [22].

2.3. Electromyography (EMG) recordings

EMG patterns were recorded during the running trial using bipolar Ag/AgCl surface electrodes with a 10-mm active diameter attached parallel to the muscle fiber direction with an inter-electrode distance of 20-mm. Standardized skin preparation was performed to ensure skin impedance of ≤5000Ω. The EMG data were recorded at 1000 Hz by an EMG system (Data LITE EMG, Biometrics Ltd, England). We collected data from eight major surface muscles of the lower extremity, including tibialis anterior (TA), gastrocnemius medialis (GM), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), semitendinosus (ST), and gluteus medius (GMD) muscles following the SENIAM recommendation [23]. For the TA muscle, the electrode was placed at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus. For GM muscle, the electrode was placed on the most prominent bulge of the muscle. For VM muscle, the electrode was placed at 80 % on the line between the anterior spina iliaca superior and the joint space in front of the anterior border of the medial ligament. For VL muscle, the electrode was placed at 2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella. For RF muscle, the electrode was placed at 50 % on the line from the anterior spina iliaca superior to the superior part of the patella. For BF muscle, the electrode was placed at 50 % on the line between the ischial tuberosity and the lateral epicondyle of the tibia. For ST muscle, the electrode was placed at 50 % on the line between the ischial tuberosity and the medial epicondyle of the tibia. For GMD muscle, the electrode was placed at 50 % on the line from the crista iliaca to the trochanter. We choose these muscles because they provide supportive and propulsive forces during running [24]. We used the medical adhesive tape to fix electrodes and probes on the skin to minimize any motion artifact. EMG signals were sampled at 1 kHz analog-to-digital conversion rate at 16-bit resolution (amplitude range ± 5 V; bandpass filtered 10–500 Hz; input impedance > 10 Ohm; common mode-rejection ratio >110 dB) by a portable Wi-Fi transmission device.

2.4. Data analyze

We calculated the ground reaction force along vertical (z), anterior-posterior (y), and lateral-medial (x) axes, time to peak (TTP), vertical loading rate, and the free moment (FM) [25]. The ground reaction force components in the z-axis were reported for the heel contact (Fz_{HC}) phase. The ground reaction forces in the y-axis also were reported in heel contact (Fy_{HC}) and push-off (Fy_{PO}) phases. The x-axis and ground reaction force during heel contact (Fx_{HC}) and push-off (Fx_{PO}) were reported. All ground reaction force and FM values were normalized by the bodyweight "BW" and "BW × Height," respectively [25]. We considered the beginner of the stance phase to be the frame in which heel contact occurred (10-N vertical force threshold), and the end was the last frame in which the forefoot was on the ground (10-N vertical force threshold). The ground reaction force data were then filtered using a fourth-order low-pass Butterworth filter with a 20 Hz cutoff frequency.

Ground reaction force and EMG data were synchronized using Nexus software (Oxford Metrics, Oxford, UK). We computed the EMG analyses for the first 50 % (early) and the second 50 % (late) stance phase of running. Maximum voluntary isometric contraction (MVIC) was assessed for each recorded muscle to normalize EMG during walking to maximal voluntary activation. Table 1 describes muscle-specific MVIC tests [26].

EMG signals were subsequently processed digitally using a 10-Hz high-pass filter with Butterworth approximation and smoothed using a 100-ms root-mean-square (RMS) window. We computed the peak RMS amplitude (μV) over a 1000-ms window for each muscle.

2.5. Statistical analysis

A priori power analysis software (G*3 Power) revealed that for a statistical power of 0.90 at an effect size of 0.3 and alpha level of 0.05, using an F-test family (ANOVA: Repeated measures, within-between interaction, was required sample size of at least 26 participants [27].

After determining the normal distribution of data using the Shapiro-Wilk test, we carried out the statistical analysis using separate 3 × 2 (foot strike (forefoot, mid-foot, rearfoot strike)) × (group (ACL injured vs. Healthy)) repeated measures ANOVAs. If there was a significant interaction for any dependent variables, simple main effects analysis we used one-way repeated measures ANOVAs for each factor, with pairwise comparisons identifying the location of significant differences. Effect sizes were determined by converting partial eta-squared (η^2_p) to Cohen's d. We ran for all analyses the SPSS (SPSS, IBM, Armonk, NY, USA), and significance, before any corrections, was set at $p \leq 0.05$.

3. Results

The results showed no significant difference in the demographic parameters of individuals between groups (Table 2).

The statistical analyses indicated significant main effects of "strike pattern" for Fx_{HC}, Fy_{HC}, Fy_{Po}, and Fz_{HC} during the stance phase of running ($p < 0.012$, d = 1.31–2.78) (Table 3). The pairwise comparison revealed that Fx_{HC} magnitude was greater in RFS, MFS, and FFS, respectively (i.e., RFS > MFS > FFS; $p < 0.001$) (Table 3). Also, Fy_{HC} was greater during RFS running in comparison with MFS and FFS running pattern ($p < 0.001$) (Table 3). On the other hand, Fy_{PO} during RFS was significantly lesser than that of FFS ($p = 0.011$, d = 1.31) (Table 3). Furthermore, Fz_{HC} during RFS was considerably lesser than that of MFS and FFS ($p = 0.002$, d = 1.63) (Table 3).

In the early stance phase, the statistical analyses indicated significant main effects of the "strike pattern" for TA and VL activity ($p < 0.035$, d = 1.10–2.44). The pairwise comparison revealed significantly greater TA and VL ($p < 0.001$) activity while running with RFS than running with MFS and FFS patterns (Table 4). Also, results demonstrated a significant main effect of "strike pattern" and "group" for VM activity ($p \leq 0.030$, d = 1.15−0.90) (Table 4). The pairwise comparison revealed significantly greater VM ($p = 0.027$) activity while running with RFS and MFS than running with FFS pattern (Table 4). In addition, the pairwise comparison revealed significantly lower VM activity in the ACLR group than that healthy group ($p = 0.039$, d = 0.85).

Also, we found significant main effects of "group" for BF activity ($p = 0.024$, d = 0.88) (Table 4). The pairwise comparison revealed significantly greater BF activity in the ACLR group than that healthy group ($p = 0.039$, d = 0.85).

During the late stance phase, the statistical analyses indicated significant main effects of "group" for TA activity ($p = 0.021$, d = 0.94) (Table 5). We also observed a significant main effect of "strike pattern" and "group" for GM, VL, VM ($p < 0.019$, d = 1.23–1.34). The pairwise comparison showed significantly lesser GM and TA activity among ACLR than healthy counterparts and greater GM activity during RFS and FFS than the MFS pattern (Table 5). Only VL activity was influenced by the group and strike pattern ($P = 0.016$, d = 1.25) (Table 5). The pairwise comparison revealed significantly lesser VL ($p = 0.006$ d = 1.18) activity while running with MFS among ACLR compared to healthy

Table 1
Description of the maximum voluntary isometric contraction (MVIC) tests.

Muscle	Test position
TA	In sitting position with knee flexed to 90° and manual resistance applied against the medial and dorsal aspect of the foot [27].
GM	Seated at the examination table, with the hip flexed at 90° and the knee and ankle in a neutral position. Participants activated their plantar flexors at maximal effort against resistance [27].
BF ST VL VM RF	The sitting position with the knees flexed to 90° using a gait belt around the distal third of the shank during isometric knee extension and knee flexion. We used 90° to normalize quadriceps and hamstring activation to maximal activity during peak knee flexion [27].
GMD	In a side-lying position, with the hip in neutral rotation and slightly extended with minimal resistance applied to the distal lower leg as the hip actively moved into abduction [27].

Legends: TA, tibialis anterior; GM, gastrocnemius medialis; BF, biceps femoris; ST, semitendinosus; VL, vastus lateralis; VM, vastus medialis; RF, rectus femoris; GMD, gluteus medius.

Table 2
Characteristics of participants.

	ACLR (n = 14)	Healthy (n = 14)	Sig
Age (year)	20.8 ± 0.3	21.3 ± 0.4	0.935
Mass (kg)	70.2 ± 4.2	70.1 ± 4.2	0.955
Height (cm)	175.9 ± 9.4	174.8 ± 9.0	0.865
Post-reconstruction duration (months)	7.2 ± 1.1	–	NA

Note: Values are mean ± standard deviation. Abbreviations: ACLR, anterior cruciate ligament reconstruction; NA, not applicable. * Significance level $p < 0.05$.

Table 3
Ground reaction force during the stance phase of running.

Interaction: strike pattern *group Sig. (effect size)	Main effect: group Sig. (effect size)	Main effect: strike pattern Sig. (effect size)	Healthy			ACLR			
			FFS	MFS	RFS	FFS	MFS	RFS	
0.750(0.30)	0.899(0.06)	**$p < 0.001$(2.01)**	13.28 ± 8.93	18.25 ± 5.46	27.68 ± 20.23	11.26 ± 7.02	18.47 ± 8.39	30.67 ± 13.84	Fx_{HC}
0.832(0.24)	0.808(0.09)	0.059(1.00)	−158.93 ± 45.40	−146.20 ± 48.37	−137.68 ± 35.80	−153.22 ± 41.21	−141.33 ± 54.25	−137.72 ± 42.76	Fx_{PO}
0.950(0.12)	0.983(0.00)	**$p < 0.001$(2.51)**	−43.86 ± 18.46	−39.26 ± 14.39	−25.72 ± 15.02	−44.33 ± 15.70	−37.87 ± 19.00	−26.33 ± 14.19	Fy_{HC}
0.922(0.15)	0.786(0.11)	**0.011(1.31)**	43.45 ± 10.73	37.26 ± 11.14	32.99 ± 10.53	45.29 ± 18.12	38.54 ± 8.78	32.60 ± 16.12	Fy_{PO}
0.326(0.61)	0.427(0.31)	**0.002(1.63)**	1821.33 ± 258.31	1788.20 ± 295.78	1598.44 ± 302.50	1881.19 ± 272.62	1811.45 ± 311.90	1737.22 ± 221.91	Fz_{HC}
0.483(0.49)	0.627(0.19)	0.062(0.99)	109.89 ± 19.35	111.16 ± 13.69	118.90 ± 17.49	101.86 ± 21.25	113.13 ± 11.47	118.14 ± 21.63	TTP Fz_{HC}
0.402(0.54)	0.731(0.14)	0.174(0.77)	$0.21*10^{-2}±0.44*10^{-3}$	$0.22*10^{-2}±0.31*10^{-3}$	$0.20*10^{-2}±0.42*10^{-3}$	$0.22*10^{-2}±0.54*10^{-3}$	$0.21*10^{-2}±0.35*10^{-3}$	$0.21*10^{-2}±0.44*10^{-3}$	FM_{max}
0.889(0.19)	0.210(0.50)	0.054(1.02)	$−0.3*10^{-3}±0.19*10^{-3}$	$−0.3*10^{-3}±0.16*10^{-3}$	$0.2*10^{-3}±0.13*10^{-3}$	$−0.2*10^{-3}±0.12*10^{-3}$	$−0.2*10^{-3}±0.11*10^{-3}$	$−0.2*10^{-3}±0.8*10^{-4}$	FM_{min}

Legends: ACLR, anterior cruciate ligament reconstruction; RFS, Rearfoot strike; MFS, Midfoot strike; FFS, Forefoot strike; Fz_{HC}, peak vertical ground reaction force during heel contact; Fy_{HC}, braking force; Fy_{PO}, propulsion force; Fx_{HC}, peak lateral ground reaction force during heel contact; Fx_{PO}, peak medial ground reaction force during the push of phase; TTP, time-to-peak; FM_{max}, maximal free moment; FM_{min}, Minimal free moment; x, medio-lateral direction; y, anterior-posterior direction; z, vertical direction, Sig., Significant.

counterparts. We also observed a significant main effect of "strike pattern" for GMD activity ($p = 0.001$, $d = 1.80$). The pairwise comparison revealed significantly greater GMD ($p = 0.001$) activity while running with RFS than running with FFS pattern (Table 5).

4. Discussion

In the present study, ground reaction forces were influenced by different strike patterns irrespective of the groups. Our results are in line with Pamukoff et al. (2018), who observed that compared with healthy control, the individuals with ACLR maintained a similar impact peak of vertical ground reaction force during the stance phase of running [17]. Previous studies suggest that FFS runners land with a more flexed knee compared to RFS runners. However, sagittal plane kinematics can be replicated by an RFS runner running with an FFS pattern [7,24]. Increased vertical ground reaction force manifests excessive joint contact force, which could lead to knee osteoarthritis development. Increased knee stiffness, a loading strategy frequently seen in ACLR subjects, is associated with excessive joint contact. Increased knee stiffness shifts the load from the knee to the hip, foot, and ankle. The knee stiffening strategy seen in ACLR individuals may reflect the early stages of motor skill acquisition since as the skills level improves, knee stiffening decreases [28]. Our findings demonstrated that Fz_{HC} during RFS was considerably lesser than that of MFS and FFS. Therefore, using RFS while running was suggested for individuals with ACLR due to lower peak vertical ground reaction force. Our results about peak free moment amplitudes for the ACLR and healthy groups are in line with a previous study demonstrating the maximum free moment amplitudes between the ACLR patients and the healthy control group were similar during walking [19]. Also, our results about peak free moment amplitudes while running with different strike patterns are consistent with a previous study that showed there are no differences in peak absolute free moments among different running strike patterns in healthy, habitual rearfoot striking recreational runners [20].

In the present study, we observed greater GM activity during RFS and FFS than the MFS pattern. In contrast, Yong et al. 2014 showed no significant difference between RMS muscle activity during the early stance phase between natural RFS and FFS runners [24]. This discrepancy may have arisen because when running with an FFS pattern, the habitual RFS runners ran with increased stride lengths compared to when they ran with an RFS pattern [24]. Natural FFS runners have similar or shorter stride lengths compared to RFS runners. We noted lesser GM and TA activity among ACLR in comparison with healthy counterparts in the present study. Also, Biererle et al. demonstrated greater TA activity among regular soccer players during barefoot running, consistent with our findings [14]. They also suggested that an increase in TA activity could be reflecting an increased tibial impact, which is a risk factor for shin splints and stress fractures [14]. Alternatively, the increased EMG activity could also be interpreted as diverting stress away from the bone and into the muscle but could result in greater joint stress nonetheless [14].

Furthermore, gastrocnemius can aid knee stability by increasing joint stiffness and restricting anterior tibial shear in weight-bearing [29]. Therefore, the greater GM activity in this study could be postulated to be an effort to facilitate knee stiffness during RFS and FFS. RFS runners have a dorsiflexed ankle during the terminal swing and early stance phase, whereas FFS runners keep their ankles in a more neutral position during the late swing phase and land with a plantarflexed ankle. These differences may be related to the larger ankle plantarflexion moments measured in FFS runners during early stance and greater peak ankle plantarflexion moments and stance phase Achilles' tendon forces [25]. Some other studies didn't report statistically significant differences in the injury rates between the habitual rearfoot and mid/forefoot strikers [30]. In contrast, they reported a difference in the injury site and the type of injury incurred between foot strike pattern groups [31,32]. These may be due to different patterns of muscle activities while running

Table 4
Muscular activity during the early stance phase of running.

Muscle	ACLR			Healthy			Main effect: strike pattern	Main effect: group	Interaction: strike pattern *group
	RFS	MFS	FFS	RFS	MFS	FFS	Sig. (effect size)	Sig. (effect size)	Sig. (effect size)
TA	80.66 ± 16.02	43.89 ± 19.90	47.13 ± 19.54	75.27 ± 16.48	60.46 ± 25.63	54.80 ± 22.80	**p < 0.001(2.44)**	0.261(0.44)	0.097(0.90)
GM	41.14 ± 27.73	29.25 ± 16.94	40.23 ± 29.24	23.54 ± 9.63	23.64 ± 15.18	38.11 ± 17.00	0.051(1.03)	0.148(0.58)	0.217(0.72)
VL	59.41 ± 29.71	47.97 ± 25.36	47.06 ± 22.85	53.82 ± 23.77	57.58 ± 27.39	40.81 ± 19.91	**0.035(1.10)**	0.921(0.00)	0.239(0.69)
VM	39.82 ± 21.89	36.98 ± 22.37	28.45 ± 17.38	53.09 ± 14.29	52.04 ± 26.08	41.34 ± 21.15	**0.027(1.15)**	**0.030(0.90)**	0.974(0.09)
RF	38.43 ± 17.91	27.61 ± 17.04	31.59 ± 18.47	43.15 ± 21.81	36.81 ± 16.80	41.82 ± 19.13	0.108(0.88)	0.166(0.55)	0.743(0.30)
BF	43.14 ± 16.10	39.48 ± 19.50	37.63 ± 15.29	34.32 ± 8.74	34.19 ± 11.81	28.38 ± 12.02	0.266(0.66)	**0.039(0.85)**	0.803(0.26)
ST	39.28 ± 17.93	38.59 ± 16.78	40.72 ± 21.39	39.15 ± 17.19	38.15 ± 20.51	36.91 ± 21.99	0.965(0.11)	0.817(0.09)	0.852(0.23)
GMD	45.09 ± 18.70	48.56 ± 16.94	48.31 ± 19.03	45.26 ± 15.42	62.65 ± 46.37	45.01 ± 14.25	0.253(0.68)	0.555(0.23)	0.534(0.45)

Legends: ACLR, anterior cruciate ligament reconstruction; RFS, Rearfoot strike; MFS, Midfoot strike; FFS, Forefoot strike; TA, tibialis anterior; GM, gastrocnemius medialis; VL, vastus lateralis; VM, vastus medialis; RF, rectus femoris; BF, biceps femoris; ST, semitendinosus; GMD, gluteus medius. Sig., Significant.

Table 5
Muscular activity during the late stance phase of running.

Muscle	ACLR			Healthy			Main effect: strike pattern	Main effect: group	Interaction: strike pattern *group
	RFS	MFS	FFS	RFS	MFS	FFS	Sig. (effect size)	Sig. (effect size)	Sig. (effect size)
TA	62.64 ± 20.66	54.02 ± 20.34	54.98 ± 28.93	77.02 ± 20.50	64.00 ± 20.93	77.94 ± 23.66	0.070(0.97)	**0.021(0**.94)	0.396(0.55)
GM	159.32 ± 49.63	135.88 ± 44.87	155.17 ± 37.04	226.28 ± 52.50	161.77 ± 40.61	213.38 ± 47.26	**0.009(1.34)**	**P < 0.001 (1.71)**	0.314(0.62)
VL	122.44 ± 37.37	**124.85 ± 22.42**	137.22 ± 33.80	112.68 ± 51.52	**185.83 ± 56.23**	144.78 ± 29.73	**0.010(1.33)**	**0.043(0.83)**	**0.016(1.25)**
VM	122.70 ± 39.62	123.74 ± 25.12	89.05 ± 13.04	172.98 ± 48.19	197.08 ± 99.46	154.23 ± 40.43	**0.018(1.23)**	**P < 0.001 (1.71)**	0.640(0.38)
RF	43.69 ± 23.15	30.98 ± 14.40	37.27 ± 24.53	50.84 ± 23.17	52.85 ± 22.39	45.54 ± 19.66	0.476(0.49)	0.036(0.86)	0.339(0.60)
BF	49.65 ± 20.40	49.67 ± 22.58	50.31 ± 24.46	50.60 ± 24.49	53.71 ± 22.34	52.84 ± 20.31	0.927(0.15)	0.672(0.16)	0.957(0.12)
ST	38.68 ± 18.46	41.69 ± 21.04	42.30 ± 20.67	46.07 ± 21.40	44.68 ± 18.71	47.22 ± 23.78	0.849(0.23)	0.356(0.36)	0.917(0.16)
GMD	98.27 ± 33.70	80.00 ± 25.02	69.81 ± 27.32	73.31 ± 21.81	97.05 ± 37.37	64.40 ± 21.84	**0.001(1.80)**	0.520(0.87)	0.052(1.03)

Legends: ACLR, anterior cruciate ligament reconstruction; RFS, Rearfoot strike; MFS, Midfoot strike; FFS, Forefoot strike; TA, tibialis anterior; GM, gastrocnemius medialis; VL, vastus lateralis; VM, vastus medialis; RF, rectus femoris; BF, biceps femoris; ST, semitendinosus; GMD, gluteus medius. Sig., Significant.

with different strike patterns, as observed in the present study.

We found lesser VL activity during running with MFS pattern and lesser VM activity among ACLR than healthy counterparts. In contrast, RF activity was not significantly different among groups and strike patterns during the entire stance phase of running. This finding is inconsistent with Yong et al. and found no significant differences in RMS activity of rectus femoris during the late swing or early stance phases in FFS runners than RFS runners [24]. In line with our results, Smeets et al. reported that athletes with ACLR show lower VM activation than healthy controls [18]. In contrast, Shih et al. reported that RFS runners had greater muscle activity in the RF during the swing phase when running with an FFS running pattern [33]. Thus, the differences between natural FFS runners and the RFS runners running with an FFS pattern may arise due to adaptations in muscle activities made after running habitually with an FFS pattern [24]. Importantly, from a clinical perspective, these reductions in quadriceps muscle activity are potentially hazardous, as less quadriceps activity is thought to lead to reinjury and may contribute to post-traumatic development osteoarthritis [34].

On the other hand, the ACLR group had greater BF activity than healthy counterparts irrespective of strike pattern only in the early stance phase of running. In contrast, ST activity was not significantly different among groups and strike patterns during the entire stance phase of running, consistent with a previous study [24]. The effect of ACLR on quadriceps and hamstring function is well documented and includes quadriceps atrophy and weakness [35]. However, little is known about the effect of ACLR on muscular activity during different strike patterns, making it difficult to draw more conclusions.

Individuals who have ACLR must rely on muscle function to maintain stability. Altered muscle timing and magnitude are proposed compensatory mechanisms that have been demonstrated in this population. Indeed, some evidence exists that the hamstrings act as an ACL agonist in providing anterior knee stability [27]. In addition, the quadriceps, once thought to be purely antagonistic to the ACL, has been shown to co-contract with the hamstrings in larger amplitudes in an attempt to "stiffen'', or, "stabilize'' the ACLR knee [24].

In the present study, greater GMD activity was observed during running with RFS than running with FFS pattern in the late stance phase of running, irrespective of the group. Increased hip adduction excursion and abduction angular impulse after an exhaustive run was reported in previous studies [27]. These changes were thought to play a role in increased patellofemoral forces [36]. The reduced gluteal muscle forces (medius and minimum) during the FFS condition may improve hip adduction and internal rotation by reducing the demand placed on the hip abductor muscles. Thus, these changes in gluteal muscle forces may have clinical merit. Further research is required to elucidate the implications of these findings.

Given the cross-sectional nature of the investigation, the longitudinal effects of the observed kinetic characteristics are unknown. Further considerations include the familiarity of each participant with the instructed foot strike patterns. Although instructional videos were provided and practice trials, impact characteristics may differ with movement pattern accommodation. The relatively small sample size, sex (only males participated in the study), and differences in the natural foot strike patterns in the sampled participants may also limit generalizations.

5. Conclusion

Strike patterns for both groups influenced the ground reaction force and muscle activities, but just muscle activities were different for individuals with ACLR. In the early stance phase of running, the ACLR individuals presented an increased BF activity irrespective of the strike pattern. In contrast, in the late stance phase of running, a reduced VL activity was observed during MFS and reduced GM and TA activities irrespective of strike patterns. These altered muscle activities could contribute to the elevated risk of future joint injury in the ACLR population. Recognizing the differences in muscle activity between FFS, MFS, and RFS runners is an important step toward understanding how foot strike patterns may contribute to ACLR limb.

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Declaration of Competing Interest

The authors report no declarations of interest.

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