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Gait Biomechanics in Individuals Meeting Sufficient Quadriceps Strength Cutoffs After Anterior Cruciate Ligament Reconstruction

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Context: Quadriceps weakness is associated with disability and aberrant gait biomechanics after anterior cruciate ligament reconstruction (ACLR). Strength-sufficiency cutoff scores, which normalize quadriceps strength to the mass of an individual, can predict who will report better function after ACLR. However, whether gait biomechanics differ between individuals who meet a strength-sufficiency cutoff (strong) and those who do not (weak) remains unknown.

Objective: To determine whether vertical ground reaction force, knee-flexion angle, and internal knee-extension moment differ throughout the stance phase of walking between individuals with strong and those with weak quadriceps after ACLR.

Design: Case-control study.

Setting: Laboratory.

Patients or Other Participants: Individuals who underwent unilateral ACLR >12 months before testing were dichotomized into strong (n = 31) and weak (n = 116) groups.

Main Outcome Measures: Maximal isometric quadriceps strength was measured at 90° of knee flexion using an isokinetic dynamometer and normalized to body mass. Individuals who demonstrated maximal isometric quadriceps strength ≥3.0 $N{\cdot}m{\cdot}kg^{-1}$ were considered strong. Three-dimensional gait biomechanics were collected at a self-selected walking speed. Biomechanical data were time normalized to 100% of stance phase. Vertical ground reaction force was normalized to body weight (BW), and knee-extension moment was normalized to BW × height. Pairwise comparison functions were calculated for each outcome to identify between-groups differences for each percentile of stance.

Results: Vertical ground reaction force was greater in the weak group for the first 22% of stance (peak mean difference [MD] = 6.2% BW) and less in the weak group between 36% and 43% of stance (MD = 1.4% BW). Knee-flexion angle was greater (ie, more flexion) in the strong group between 6% and 52% of stance (MD = 2.3°) and smaller (ie, less flexion) between 68% and 79% of stance (MD = 1.0°). Knee-extension moment was greater in the strong group between 7% and 62% of stance (MD = 0.007 BW × height).

Conclusions: Individuals with ACLR who generated knee-extension torque ≥3.0 $N{\cdot}m{\cdot}kg^{-1}$ exhibited different biomechanical gait profiles than those who could not. More strength may allow for better energy attenuation after ACLR.

Key Words: posttraumatic osteoarthritis, knee, rehabilitation

Key Points

- Individuals with quadriceps strength <3.0 $N{\cdot}m{\cdot}kg^{-1}$ (weak group) demonstrated less overall knee flexion between 6% and 52% of stance and less knee extension between 68% and 79% of stance than did individuals with greater quadriceps strength (strong group).
- The weak group displayed less knee-extension moment between 7% and 62% of stance and greater vertical ground reaction force in the first 22% of stance than did the strong group. The biomechanical differences indicated greater use of a stiffened-knee strategy in the weak than the strong group, which may be related to the early development of posttraumatic osteoarthritis.

Persistent quadriceps weakness is a common clinical impairment after anterior cruciate ligament (ACL) injury and ACL reconstruction (ACLR).[1] Individuals with ACLR and weaker quadriceps report more disability,[2,3] demonstrate altered cartilage composition,[4] and exhibit greater tibiofemoral joint space narrowing,[5] which are changes associated with the early development of posttraumatic osteoarthritis (PTOA). Restoring quadriceps strength after ACLR is a critical component of postoperative rehabilitation for improving quality of life and eliminating aberrant movement biomechanics associated with PTOA development[6]; however, the link between restoring strength and normalizing walking gait biomechanics to that of uninjured control individuals after ACLR remains unclear. Developing sufficient quadriceps strength, or strength of the ACLR limb normalized to the mass of the patient (newton meters per kilogram), may be more strongly associated with higher self-reported function after ACLR than quadriceps strength symmetry or strength equal to that of the contralateral limb.[2] In separate cross-sectional studies, Kuenze et al[7] and Pietrosimone et al[2] found that the ability to generate maximal voluntary isometric contractions (MVICs) $\geq$ 3.0 and 3.1 N·m·kg^{-1}, respectively, was associated with better knee-related function. Pietrosimone et al[2] observed that approximately 31% of participants met the 3.1-N·m·kg^{-1} strength-sufficiency cutoff at a mean of 37.04 $\pm$ 36.7 months post-ACLR.[2] Therefore, more than two-thirds of individuals with ACLR returned to activities of daily living with quadriceps strength that did not meet these strength-sufficiency cutoff scores. Nevertheless, whether the inability to attain sufficient quadriceps strength would be associated with aberrant gait biomechanics after ACLR remains unclear.

Furthermore, quadriceps weakness is linked with the development of knee osteoarthritis, which is a substantial contributor to long-term disability after ACLR.[8] Individuals with weaker quadriceps have demonstrated a higher incidence of idiopathic knee osteoarthritis.[9] Quadriceps weakness was associated with deleterious changes in femoral articular cartilage composition within the first 6 months after ACLR[4] and radiographic tibiofemoral joint space narrowing 4 years after ACLR.[5] The mechanisms linking quadriceps weakness to increased risk of PTOA development after ACLR are not fully understood; however, researchers[10,11] have hypothesized that poor quadriceps function contributes to the development and perpetuation of altered gait biomechanics after ACL injury and ACLR. Quadriceps weakness has been linked to a stiffened-knee gait strategy, characterized by a more extended knee in early stance, less knee excursion throughout stance, and smaller peak internal knee-extension moments (KEMs) during the first half of the stance phase of gait, in various musculoskeletal conditions including ACLR.[10,12] Individuals with weaker quadriceps may exhibit a stiffened-knee strategy partly due to the inability to generate adequate eccentric quadriceps contractions, which may alter the loading of joint tissues during gait.[11,13] Proper quadriceps function has been proposed to assist in controlling knee flexion during dynamic activities after ACLR[11]; this controlled knee flexion has been suggested as leading to better energy attenuation at the knee during gait.[11] Individuals with ACLR and less quadriceps strength symmetry exhibited smaller peak knee-flexion angles (KFAs) in the first half of stance,[14] and those with a lower rate of quadriceps torque development had greater vertical ground reaction force (vGRF) loading rates during gait shortly after heel strike.[15] The effects of quadriceps weakness on gait biomechanics in other portions of stance, such as mid- and late stance, are less commonly evaluated. Impaired quadriceps function also likely influences gait biomechanics during mid- and late stance, because individuals accelerate the center of mass upward and forward, resulting in impaired physical performance during activities of daily living. Therefore, the purpose of our study was to compare gait biomechanics (KFA, KEM, and vGRF) throughout the stance phase of gait between participants after ACLR with strong (conservatively defined[7] as $\geq$3.0 N·m·kg^{-1}) or weak ($<$3.0 N·m·kg^{-1}) quadriceps. We hypothesized that individuals with weak quadriceps would demonstrate less KFA and KEM throughout stance than those with strong quadriceps. We also expected those with weak quadriceps to show greater vGRF after heel strike, suggesting an impaired ability to attenuate energy in the lower extremity. Understanding the biomechanical differences between individuals who are and those who are not able to meet the 3.0-N·m·kg^{-1} cutoff will provide further justification for developing rehabilitation guidelines to optimize quadriceps strength sufficiency in patients after ACLR.

METHODS

Study Design

We recruited a sample of convenience into a retrospective comparison-control study from an ongoing cross-sectional study (N = 147; Table). Individuals' height and mass were collected using standard means. All participants performed a gait analysis and an MVIC during a single session. They were retrospectively assigned to either the strong ($\geq$3.0 N·m·kg^{-1}) or weak ($<$3.0 N·m·kg^{-1}) quadriceps group based on strength scores.[7] Each person provided written informed consent, and the study was approved by the Biomedical Institutional Review Board at the University of North Carolina at Chapel Hill.

Participants

Participants were between the ages of 18 and 35 years and had sustained a unilateral ACL injury. Gait biomechanics may undergo changes in both the ACL-injured and contralateral limb during the first 12 months post-ACLR[16]; therefore, we only included individuals who underwent ACLR $>$12 months before testing (Table). We did not exclude potential participants based on a maximum time limit post-ACLR but did exclude any with a history of any other lower extremity orthopaedic surgery, ACLR revision surgery, multiligament reconstruction at the time of ACLR, physician-diagnosed knee osteoarthritis (ie, diagnosed radiographically or based on the clinical examination), balance or neuromuscular disorder, or an orthopaedic injury to either limb between ACLR and the time of testing.[17] Recruits were solicited from the university's health system orthopaedic clinics, club sport teams, varsity athletics, and general community. The International Knee Documentation Committee Index was collected from each participant before testing.

A moderate effect (mean difference = 0.06 body weight [BW]; d = 0.60)[16] occurred between symptomatic and asymptomatic individuals with ACLR for the largest magnitude difference in vGRF throughout stance using the functional waveform gait analysis approach. We used measures consistent with previously published literature[7] to define the calculation of mean differences between individuals with strong and those with weak quadriceps as well as variability estimates across the waveforms using 5 gait trials from each participant. Hence, we estimated that a minimum of 9 people would be needed to detect a statistically significant moderate mean difference between groups across the waveform, assuming similar intertrial variability as previously reported (2-tailed $\alpha = 0.05$; $1-\beta = 0.8$; G*Power Statistical Power Analysis Software,[18] version 3.1). We

Table. Participant Characteristics and Outcome Measures

Characteristic	Total (N = 147)	Group: Strong (n = 31)	Group: Weak (n = 116)
	No.		
Sex, females/males	98/49	11/20	87/29[a]
Anterior cruciate ligament graft type			
Patellar tendon autograft	103	3	80
Hamstrings/gracilis autograft	36	7	29
Quadriceps tendon autograft	4	1	3
Allograft	4	0	4
	Mean ± SD		
Age, y	21.2 ± 3.3	22.4 ± 3.4	20.7 ± 3.2[b]
Body mass index	24.4 ± 3.6	24.7 ± 3.8	24.4 ± 3.8
Time since anterior cruciate ligament reconstruction, mo	29.3 ± 27.3	37.5 ± 31.3	27.2 ± 25.8
Maximal voluntary isometric contraction, $N \cdot m \cdot kg^{-1}$	2.43 ± 0.75	3.48 ± 0.4	2.15 ± 0.55[b]
Walking speed, m/s	1.29 ± 0.15	1.29 ± 0.14	1.29 ± 0.15
International Knee Documentation Committee Index, normalized to 100	84 ± 10	86 ± 9	83 ± 10

[a] Indicates a greater percentage of women in the weak than the strong group.
[b] Indicates a lower value than that of the strong group ($P < .05$).

enrolled a much larger cohort to ensure that the strong and weak groups would each contain >9 individuals.

Quadriceps Strength Testing

Quadriceps strength was assessed using a HUMAC Norm dynamometer (CSMi). Torque signal was sampled at either at 600 Hz or 2000 Hz and low-pass filtered at 50 Hz using a zero-phase-shift, fourth-order Butterworth filter. All participants were positioned on the dynamometer with the hips and knees flexed to 85° and 90°, respectively.[19] The pelvis and torso were secured to the chair using adjustable straps, and the upper limbs were folded across the chest to isolate the contribution of the quadriceps musculature. The dynamometer arm was secured to the leg 3 cm proximal to the lateral malleolus and adjusted so the knee-joint center was aligned with the dynamometer axis of rotation.

Individuals were instructed to push into the lever arm as fast and as hard as possible and to maintain maximal effort for approximately 2 seconds. Research assistants provided standardized oral encouragement, and real-time visual feedback of torque production was displayed on a computer monitor in front of the dynamometer.[20] A series of submaximal graded "warm-up" isometric contractions was performed between 25% and 75% of participant-perceived maximal effort. Three to 5 practice trials were conducted to ensure production of maximal effort; practice trials were continued until the torque measurements ceased to increase within 10% of the previous trial.[20] The peak torques from the final 2 maximal-effort practice trials were averaged and used as a minimal torque threshold for the subsequent trials. Two maximal-effort trials, in which the peak amplitude of the MVIC met or exceeded the torque threshold from the practice trials, were averaged and used in the final data analysis. Peak torque was normalized to body mass (newton meters per kilogram), and individuals with peak torque $\geq$3.0 $N \cdot m \cdot kg^{-1}$ were considered *strong*.

Walking Gait Biomechanics

All participants were outfitted with 25 retroreflective markers and a rigid cluster of 3 retroreflective markers placed over the sacrum.[21,22] Marker positions were quantified using a 10-camera, 3-dimensional motion-capture system (VICON; Nexus). Participants walked barefoot at a self-selected speed over 2 force plates (model FP406010; Bertec Corp) embedded in a 6-m walkway such that the entire stance phase could be collected from only the ACLR limb.[21,23] they were instructed to walk at a comfortable self-selected speed, look straight ahead, and maintain a constant speed through 2 sets of timing gates (model TF100; Trac Tronix) centered on the force plates. Five practice trials were performed to familiarize the participants with the gait task as well as to determine the mean walking speed for the test trials. Five test trials were performed and considered *acceptable* if (1) the ACLR limb made contact with a single force plate for the entirety of stance, (2) a forward gaze was maintained, (3) consistent gait speed (±5%) was maintained, and (4) gait was not visibly altered during the trial.[17,21,23]

Kinematics were sampled at 120 Hz and low-pass filtered at 10 Hz using a fourth-order recursive Butterworth filter, whereas force data were sampled at 1200 Hz and low-pass filtered at 10 Hz using a fourth-order recursive Butterworth filter. *Stance* was defined as the interval between heel strike (vGRF > 20 N) and toe off (vGRF < 20 N). The knee-flexion angle was calculated as the angle of the shank relative to the thigh using Euler angles,[21] and KEM was calculated using a standard inverse-dynamics approach. Vertical ground reaction force was normalized to BW for each participant, and KEM was normalized to the product of participant BW and height (meters).[17,21,23]

Statistical Analyses

Before our primary analyses, we evaluated potential differences in discrete characteristic variables (Table) between the strong and weak groups using independent-samples t tests for continuous variables and χ^2 tests for dichotomous variables with $\alpha \leq .05$ (SPSS, version 19.0; IBM Corp). We conducted separate functional waveform gait analyses[24] for each biomechanical outcome (KFA, KEM, and vGRF) in both groups. Before analysis, vGRF (BW), KFA, and internal KEM (newton meters per

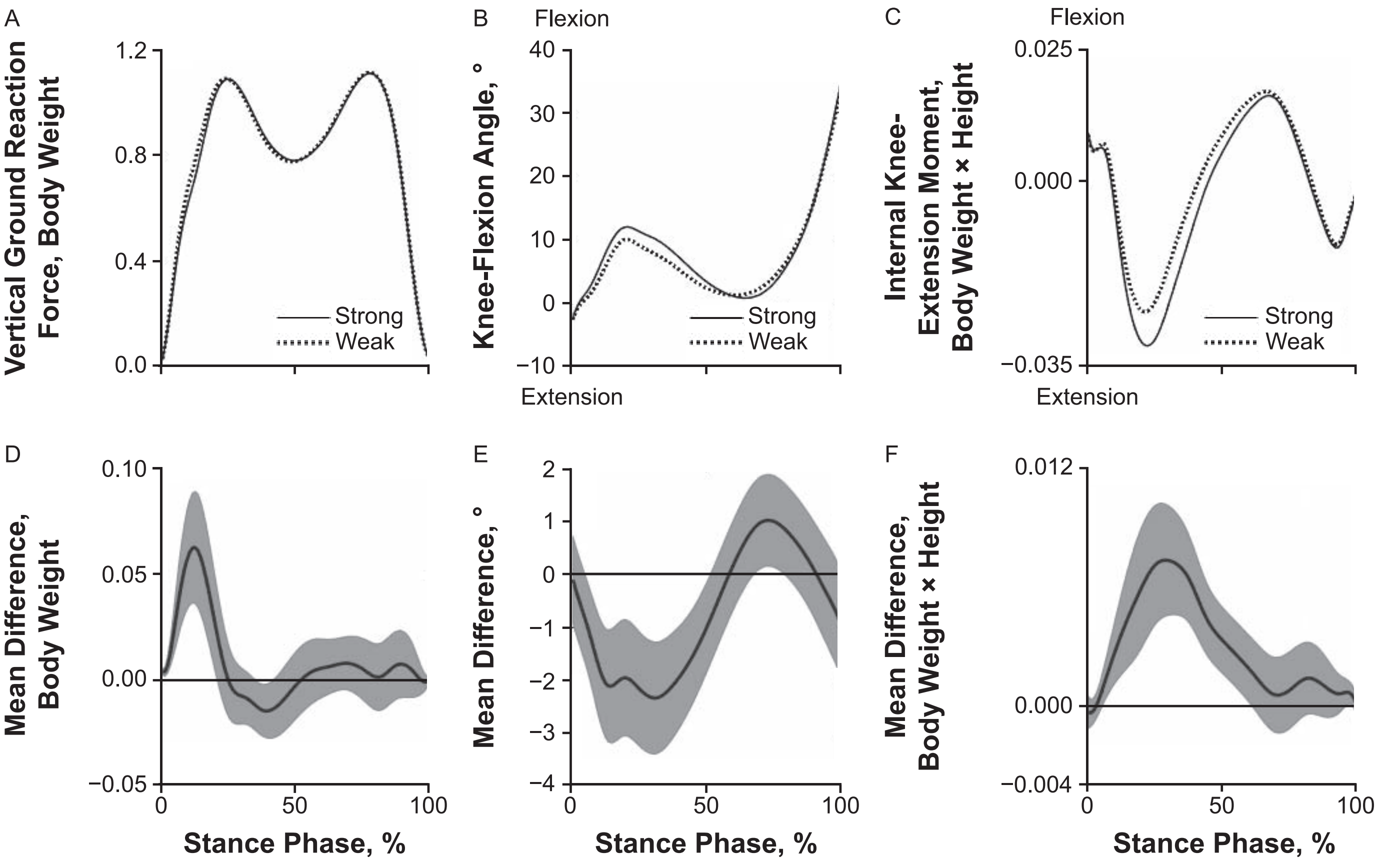

Figure. Gait differences in individuals with anterior cruciate ligament reconstruction who had strong (strong group) or weak (weak group) quadriceps. A–C, Mean ensemble waveforms for the strong and weak groups plotted over the stance phase of walking for mean vertical ground reaction force normalized to body weight (BW), knee-flexion angle, and knee-extension moment normalized to body weight (BW) × height (m). D–F, pairwise comparisons depicting mean differences and corresponding 95% CIs for the ensemble waveforms displayed in A, B, and C, respectively. For D–F, biomechanical variables were considered different for portions of the stance phase when 95% CIs did not overlap zero. (Although we excluded this information to increase clarity, the precise mean difference is always exactly between the upper and lower 95% CI.) Greater vertical ground reaction force was found in the weak group for the first 22% of stance (A and D). The weak group demonstrated less knee-flexion angle between 6% and 52% of stance and more knee flexion between 68% and 79% of stance (B and E). Knee-extension moment was greater in the strong group between 7% and 62% of stance (C and F).

kilogram) data from each of the 5 trials were extracted and time normalized to 101 data points using custom algorithms in MATLAB (version R2017A; The MathWorks).[24] We present ensemble waveforms for each biomechanical outcome in the strong and weak groups in Figure A–C. Next, separate pairwise functional comparisons were conducted by calculating the mean differences between the strong and weak group waveforms throughout stance phase using the functional data-analysis package in R statistical computing software (version 3.4.3; The R Foundation). Mean differences between the strong and weak groups are represented with a solid line throughout stance phase in Figure D–F. Finally, 95% CIs were constructed around the mean differences for the strong and weak group waveforms using a shaded band (Figure D–F). The strong and weak groups were considered different at any percentile of the stance phase in which mean differences and corresponding 95% CIs did not cross zero (Figure D–F).[24]

RESULTS

The weak group demonstrated a greater frequency of females ($\chi^2 = 17.19$, $P > .001$), lower MVIC ($t_{145} = 12.5$, $P > .001$), and younger mean age ($t_{145} = 2.31$, $P > .001$; Table). The weak group displayed greater vGRF than the strong group in the first 22% of stance (peak between-groups difference = 6.2% BW) and less vGRF between 36% and 43% of stance (peak between-groups difference = 1.4% BW; Figure A and D). The weak group exhibited less KFA than the strong group between 6% and 52% of stance (peak between-groups difference = 2.3°) but greater KFA between 68% and 79% of stance (peak between-groups difference = 1.0°; Figure B and E). Knee-extension moment was greater in the strong group between 7% and 62% of stance (peak = 0.007 BW·height; Figure C and F).

DISCUSSION

In agreement with our hypothesis, the weak group showed more of a stiffened-knee walking strategy in the ACLR limb than the strong group. The weak group had less overall knee range of motion through most of stance, a product of less knee flexion in the first half of stance and less knee extension in the second half of stance, than did the strong group. Less knee range of motion in most of the first half of stance was accompanied by less KEM (7%–62% of stance), suggesting the strong group was able to generate a greater quadriceps-related moment during the weight-acceptance phase of gait than the weak group. Greater vGRF was found in the first 22% of stance in the weak group, indicating that the weak group experienced higher loads applied to the ACLR extremity during early stance. Pietrosimone et al[2] observed that individuals capable of producing knee-extension torque $>$3.1 N·m·kg^{-1} had better patient-reported outcomes after ACLR. Our results suggested that the weak group, which was not capable of producing quadriceps strength $\geq$3.0 N·m·kg^{-1}, demonstrated a stiffened-knee strategy during gait and greater lower extremity loading immediately after heel strike. Overall, our study provides further evidence to support the goal of achieving quadriceps strength sufficiency $\geq$3.0 N·m·kg^{-1} after ACLR. Individuals who achieve this cutoff also had gait biomechanics linked to better lower extremity force attenuation during gait[6] as well as better patient-reported outcomes.[2,7]

Many of the previous evaluations[13,14,25] of the associations between strength and gait biomechanics in participants with ACLR focused on peak biomechanics, often within the first half of stance. A compelling advantage of functional waveform gait analysis is the ability to detect differences between the groups at any point in stance without requiring researchers to predefine a specific discrete variable.[24] However, we cannot assume that every between-groups difference at any point of stance is equally important in influencing patient function or lower extremity joint health. Differences in knee flexion between the strong and weak groups existed for approximately 67% of the gait cycle, resulting in not only a less-flexed knee for much of the first half of stance (6%–52% of stance) but also a less-extended knee in the second half of stance (68%–79% of stance). Less sagittal-plane knee excursion during gait may decrease the tibiofemoral contact area engaged during each step, which may negatively influence tissue mechanics and lead to deleterious changes in joint tissue health.[26] In addition, less knee extension in the second half of stance may influence propulsion of the body forward. Future investigators should determine whether a small change in knee extension (approximately 1°) between 68% and 79% of stance influences gait speed,[27] given that slower gait speed is also associated with joint tissue metabolism[28] and composition[29] consistent with cartilage breakdown in individuals with ACLR. Furthermore, the weak group displayed less KEM between 7% and 62% of stance, suggesting that they had developed strategies to avoid larger quadriceps-related moments during the portions of stance phase when the limb is undergoing greatest loading. Also, quadriceps weakness may influence energy-attenuation strategies before and after peak loading occurs in the first half of stance. Vertical ground reaction force was greater in the first 22% of stance in the weak group, which corresponds to the periods of stance when vGRF loading rates are highest after heel strike. Greater impulsive loading after heel strike in weaker individuals may be associated with less KFA and KEM in early stance, because a stiffened-knee strategy may contribute to the inability to attenuate forces immediately after heel strike. The weak group exhibited greater unloading of the vGRF for a small percentage of midstance (36%–43% of stance); however, whether less loading during midstance for individuals with weak quadriceps is part of an overall gait strategy seeking to reduce vGRF impulse in response to greater vGRF in the first 22% of stance remains unknown.

Only 21% of our sample met the 3.0-N·m·kg^{-1} strength cutoff and were classified as strong. Conversely, all individuals in our study were at least 12 months post-ACLR at the time of testing, cleared for return to unrestricted physical activity, and free of any subsequent knee injury after the primary, unilateral ACLR. Currently, no evidence suggests that achieving a 3.0-N·m·kg^{-1} or 3.1-N·m·kg^{-1} strength cutoff decreases the risk of secondary ACL injury or should be a criterion for returning a person with ACLR to unrestricted participation in physical activity. Kuenze et al[7] showed that achieving a strength-sufficiency cutoff of 3.0 N·m·kg^{-1} was effective for distinguishing uninjured individuals without knee-related symptoms from patients with ACLR who had knee-related symptoms. Similarly, researchers[2] determined that achieving a strength-sufficiency cutoff of 3.1 N·m·kg^{-1} was associated with greater odds of reporting a score of $\geq$90% using the International Knee Documentation Committee subjective knee evaluation form in patients with ACLR. According to our data, individuals who achieved the cutoff score were less likely to present with a stiffened-knee strategy. Therefore, it is possible that achieving the 3.0-N·m·kg^{-1} threshold may assist in identifying patients with ACLR who will achieve acceptable long-term outcomes. Yet the evidence is not clear that improving quadriceps strength will directly influence gait biomechanics in other clinical populations with knee conditions, such as idiopathic knee osteoarthritis.[30] Our study was cross-sectional, and as a result, we were unable to determine causality between quadriceps weakness and aberrant gait biomechanics. Future authors should explore if improving strength to $>$3.0 N·m·kg^{-1} in individuals with ACLR alters gait biomechanics.

Only 11% of the females in our sample were classified as strong, compared with approximately 41% of the males (Table). This finding is consistent with the work of Kuenze et al,[31] who identified that women who were $>$12 months post-ACLR had greater quadriceps weakness than men who were $>$12 months post-ACLR. No consensus exists as to the influence of sex on changes in gait biomechanics after ACLR; however, Di Stasi et al[32] reported that knee-flexion excursion decreased in female participants during gait between a preoperative time point and 6 months post-ACLR, which was not the case in their male counterparts across the same timeframe. Unfortunately, the functional analysis of variance we used does not lend itself to adjusting the shape of the waveforms to account for covariates, such as sex. Collectively, these data justify the need to further characterize the influence of sex on the association between achieving strength-sufficiency cutoffs and gait biomechanics.

Our study was the first to compare KFA, KEM, and vGRF between individuals with strong and those with weak quadriceps after ACLR using a previously developed[7] clinical strength cutoff score of 3.0 N·m·kg^{-1}. However, some limitations should be addressed to inform future work. We did not evaluate the influence of concomitant injury or graft type on biomechanical differences in the strong and weak groups. Researchers should conduct larger studies to assess whether the effects of concomitant injury and graft selection alter the association between quadriceps strength and gait biomechanics. In addition, time post-ACLR seems to influence the association between gait biomechanics and patient-reported function[22]; therefore, future authors should determine the influence of time post-ACLR on the association between quadriceps strength and gait biomechanics. Finally, we focused on sagittal-plane knee biomechanics and lower extremity loading of the ACLR limb. It will be important to evaluate whether quadriceps strength cutoffs influence the contralateral limb or other biomechanical outcomes in other lower extremity joints or other planes of motion. In addition, determining how differences in gait biomechanics influence energy absorption in individuals who meet strength-sufficiency cutoff scores after ACLR will be useful.[33]

CONCLUSIONS

Individuals with weak quadriceps who did not meet the 3.0-N·m·kg^{-1} strength cutoff demonstrated less overall knee flexion between 6% and 52% of stance and less knee extension between 68% and 79% of stance than did individuals with strong quadriceps. The weak group also demonstrated less KEM between 7% and 62% of stance and greater vGRF in the first 22% of stance than did individuals with ACLR who met the 3.0-N·m·kg^{-1} strength cutoff. These biomechanical differences indicate greater use of a stiffened-knee strategy compared with that of individuals with strong quadriceps, which may be related to PTOA development.

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