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PATELLOFEMORAL AND TIBIOFEMORAL JOINT LOADING DURING A SINGLE-LEG FORWARD HOP FOLLOWING ACL RECONSTRUCTION

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Author contributions statement: The overall series of studies of biomechanics after ACLR, of which this is a part, was conceived by LGP, AGC, KMC and ALB. HGM and TSW were the surgeons who performed the ACL reconstruction for injured participants. LGP, KMC and ALB designed and undertook the hopping gait experiments. AGC and LGP were responsible for collecting and quantifying patient demographic data, patient reported outcome measures and analyses of patient imaging data. PS developed and implemented the musculoskeletal modelling pipeline, including patellofemoral and tibiofemoral joint models, and also undertook all statistical analyses. PS and AGS wrote the initial drafts of this manuscript, however, all authors contributed to its development, and have read and approved the final submitted manuscript.

ABSTRACT

Altered biomechanics are frequently observed following anterior cruciate ligament reconstruction (ACLR). Yet, little is known about knee-joint loading, particularly in the patellofemoral-joint, despite patellofemoral-joint osteoarthritis commonly occurring post-ACLR. This study compared knee-joint reaction forces and impulses during the landing phase of a single-leg forward hop in the reconstructed knee of people 12-24 months post-ACLR and uninjured controls. Experimental marker data and ground forces for 66 participants with ACLR (28±6 years, 78±15 kg) and 33 uninjured controls (26±5 years, 70±12 kg) were input into scaled-generic musculoskeletal models to calculate joint angles, joint moments, muscle forces, and the knee-joint reaction forces and impulses. The ACLR group exhibited a lower peak knee flexion angle (*mean difference:* -6 degrees; *95% confidence interval:* [-10,-2] degrees), internal knee extension moment (-3.63 [-5.29,-1.97] percentage of body weight * participant height (BW*HT)), external knee adduction moment (-1.36 [-

2.16,-0.56] %BW*HT) and quadriceps force (-2.02 [-2.95,-1.09] body weights (BW)). The ACLR group also exhibited a lower peak patellofemoral-joint compressive force (-2.24 [-3.31,-1.18] BW), net tibiofemoral-joint compressive force (-0.74 [-1.20, 0.28] BW), and medial compartment force (-0.76[-1.08,-0.44] BW). Finally, only the impulse of the patellofemoral-joint compressive force was lower in the ACLR group (-0.13 [-0.23,-0.03] body weight-seconds). Lower compressive forces are evident in the patellofemoral- and tibiofemoral-joints of ACLR knees compared to uninjured controls during a single-leg forward hop-landing task. Our findings may have implications for understanding the contributing factors for incidence and progression of knee osteoarthritis after ACLR surgery.

Keywords: ACL injury, contact force, knee osteoarthritis, post-traumatic, musculoskeletal modelling

INTRODUCTION

Anterior cruciate ligament (ACL) injury is a well-known risk factor for knee osteoarthritis[1]. Degenerative changes associated with osteoarthritis are observed in both the tibiofemoral-joint (TFJ) and patellofemoral-joint (PFJ) in up to 90% of young adults within 10 years of ACL injury, irrespective of surgical ACL reconstruction (ACLR) or nonoperative management[2,3]. The mechanobiological processes associated with the initiation and progression of knee osteoarthritis post-ACLR are still unclear, but altered movement patterns and mechanical loading within the joint are thought to play critical roles[4-7].

Considerable differences between the ACLR limb compared to the contralateral limb and/or uninjured controls in joint kinematics and moments have been observed for tasks such as walking[8], running[9,10], single-leg drop landing[11] and forward hop-landing[12-15]. The most evident and consistent differences for the ACLR limb are in the sagittal plane[8], including a lower peak knee flexion angle and peak external knee flexion moment[8], and increased overall leg stiffness[12]. These differences are apparent from 6 months after ACLR[7,13], but persist for at least another 8 years[16], suggesting lower-limb biomechanical function never fully recovers.

Diminished loading may be a salient biomechanical feature following ACLR. Compared to uninjured knees, lower PFJ contact forces post-ACLR have been found during running[9,10], while lower TFJ compressive forces were reported for walking[7,17], running and side-stepping[17]. Furthermore, lower TFJ anterior shear forces, but greater TFJ compressive forces, were observed in the ACLR knee during drop-landings[11], indicating a possible task-dependency of loading following ACLR. While many studies have investigated joint kinematics and moments after ACLR, only a few studies have explicitly reported knee-joint forces. Most notably, very few investigations of PFJ contact forces have been undertaken, even though the PFJ frequently displays rapid early cartilage loss and radiographic changes after ACLR[18-20].

The single-leg forward hop-for-distance is a challenging but routine task to evaluate function after ACLR, and to assess readiness for return to sport[21]. It is an effective surrogate for explosive/ballistic sagittal-plane movements commonly undertaken during sport, and may reflect the upper limits of lower-limb joint loading experienced. During the landing phase of this task, individuals must coordinate their trunk and lower-limb muscles to arrest their forward motion while simultaneously resisting collapse due to gravity[12]. The landing phase imparts sudden high-impact loading on the lower-limb, with rapidly applied ground forces rising to more than double the magnitude experienced at take-off[22]. Peak PFJ reaction forces during landing may be as high as 8.6 body weights (BW) in uninjured individuals[23], but TFJ loading has not been reported to our knowledge. Moreover, PFJ loading during a forward hopping task following ACLR remains unexplored. Thus, understanding knee-joint loading during high impact tasks after ACLR may yield new insights into the risk of re-injury and the development of post-traumatic osteoarthritis.

Our objective was to quantify the reaction forces and impulses in the ACLR knee during the landing phase of a forward hop for individuals 12-24 months postoperatively, and to compare results with those from uninjured controls. We hypothesised that peak compressive forces, as well as compressive impulses, in the PFJ and TFJ following ACLR would be lower than in uninjured controls.

METHODS

Study design. Case-control study (Level of Evidence III).

Study participants. Sixty-six adults 12-24 months following ACLR and 33 uninjured controls were included. Experimental data from these cohorts were used previously[12], with one additional participant in each group for the present study. Participants with ACLR were part of a larger cohort ($n = 111$) of consecutively eligible patients with a primary hamstring-tendon autograft ACLR from one of two orthopaedic surgeons[21,24,25]. At the time data collection, participants were aged 18-50 years and were involved in physical activity 1-3 times per week (Tegner Activity Scale ≥ 4). The study was approved by the University of Melbourne Human Research Ethics Committee (ID: 1136167). Exclusion criteria were previous injury or surgery to the contralateral knee, grade III injury to other knee ligaments, subsequent injury or surgery to the ACLR knee, or any other condition affecting physical function. Sports involvement was assessed using the Sports Activity Classification[26]. Quadriceps strength was measured as the maximum isometric knee extension moment, with participants seated at 90 degrees hip flexion and 60 degrees knee flexion, using a

KinCom dynamometer (Chattecx Corp, TN, USA)[27]. Objective knee function was assessed using hop-for-distance and side-hop tasks, while patient-reported function was assessed with the Cincinnati Knee Rating Scale[28].

Biomechanical data collection. The detailed experimental procedure was recently published[6,12]. Briefly, biomechanical testing was undertaken at the Movement Research Laboratory, Centre for Health, Exercise and Sports Medicine, University of Melbourne. Participants walked forwards three steps in time with a metronome set at 100 beats per minute and performed a single forward hop, taking off and landing on the same leg. The ACLR group hopped using their reconstructed leg, while controls used their right leg. Hop distance was standardised (with marks on the floor) to equal 100% of the participant's leg length (i.e., from greater trochanter to the floor). Participants were barefoot and folded their arms across their chest. A trial was deemed successful if the participant landed on their mark, and comfortably maintained their balance for at least two seconds after foot-strike. A maximum of 10 trials were attempted, but the number of successful trials varied between participants, ranging from 4 to 7 (median: 5). No participants reported any knee pain while undertaking the task.

During each hop trial, the spatial trajectories of 32 retro-reflective markers placed on the torso, pelvis and both lower-limbs of the participants[29] were collected using a 12-camera Vicon motion analysis system (Oxford Metrics, Oxford, UK) at 120 Hz. A static standing trial was recorded initially, to determine participant anthropometry for subsequent musculoskeletal modelling. For this purpose, an additional 8 markers were placed on the lower-limb which were removed prior to undertaking dynamic trials. A single force plate (AMTI Watertown, MA, USA) was used to measure ground reaction force (GRF) data at 1080 Hz. Kinematic and GRF data were filtered at 15 Hz using a 4th-order, zero-lag, recursive Butterworth filter.

Musculoskeletal modelling. Musculoskeletal modelling was undertaken using OpenSim 4.0[30] via the API in MATLAB (The Mathworks Inc., Natick, MA). For each participant, a whole-body three-dimensional musculoskeletal model was created by scaling the inertial properties and musculotendon geometry of a reference model, adapted from Arnold et al.[31], using anatomical measurements estimated from the static trial. The reference model consisted of 14 rigid segments, 23 degrees-of-freedom and 94 Hill-type muscle-tendon units. The head, arms and torso were combined into a single rigid body, articulating with the pelvis via a ball-and-socket. Each hip-joint was modelled as a ball-and-socket, and each knee, ankle and subtalar-joint was modelled as a hinge. For each leg, the motions of each tibia and patella relative to the femur were prescribed with respect to the knee flexion angle, and a passive inelastic patellar tendon articulated with the tibia and patella. The metatarsophalangeal joints, modelled as hinges, were locked at the reference position.

Joint angles were calculated using inverse kinematics, which minimised the sum-of-squares of distances between experimental marker trajectories and the equivalent virtual markers on the model[32]. Joint moments were calculated using inverse dynamics. Muscle forces were estimated from joint moments using static optimisation, by minimizing the sum-of-squares of activations subject to constraints imposed by each muscle's force-length-velocity properties[33].

PFJ forces. A simple model of the PFJ was implemented in MATLAB to calculate PFJ reaction forces. A full mathematical description of the model is presented in Sritharan et al.[10]. Briefly, it consisted of a rigid patella, and the three-dimensional musculotendon geometry of the vastus laterialis, vastus intermedius, vastus medialis, rectus femoris and the patellar tendon derived directly from each participant's OpenSim model. For each trial, at each time step, the instantaneous vasti and rectus femoris forces were applied to the patella (Figure 1A), acting along each respective muscle's instantaneous line-of-action, calculated from the OpenSim model's instantaneous pose. The sum of these muscle force vectors is the quadriceps force, $\mathbf{F}^{Q}$. Subsequently, the three-dimensional patellar tendon force vector, $\mathbf{F}^{PT}$, was calculated by equating the components of the quadriceps force and patellar tendon force tangent to the surface of the femur at that time step. Finally, by assuming negligible patellar mass and zero sliding friction of the patella along the trochlear groove, the net PFJ reaction force vector, $\mathbf{F}^{PFJ}$, is simply equal and opposite to the sum of the quadriceps and patella tendon forces:

$$\mathbf{F}^{PFJ} = -(\mathbf{F}^{Q} + \mathbf{F}^{PT}) = \begin{bmatrix} F_x^{PFJ} \\ 0 \\ F_z^{PFJ} \end{bmatrix} \tag{1}$$

Where F_x^{PFJ} is the component of the net PFJ reaction force perpendicular to the surface of the femur at that time step, and F_z^{PFJ} is the component acting mediolaterally. The magnitude of the *PFJ compressive force* experienced by the patella acting against the surface of the trochlear groove is the Euclidian norm of the *x*- and *z*-components of the net PFJ reaction force:

$$F^{PFJ} = |\mathbf{F}^{PFJ}| = \sqrt{\left(F_x^{PFJ}\right)^2 + \left(F_z^{PFJ}\right)^2} \tag{2}$$

TFJ forces and medial/lateral compartment forces. For each trial, the TFJ reaction forces and moments were calculated at each time step by applying the joint angles, muscle forces, the GRFs, and the calculated patellar tendon forces, to the musculoskeletal model, and solving for the instantaneous three-dimensional joint reaction forces and moments. The three-dimensional TFJ reaction force and reaction moment together represent the net loads borne by the knee, that balance the forces and moments applied by the knee-spanning muscles and external loads. For this study, the component of the net TFJ reaction force acting axially along the tibia was defined as the *TFJ compressive force*, while the fore-aft component with respect to the tibia was defined as the *TFJ anterior shear force*.

The TFJ compressive force was further partitioned into medial and lateral tibiofemoral compartment forces, given by $\mathbf{F}^{MED}$ and $\mathbf{F}^{LAT}$ respectively, by considering a simple force-moment balance in the frontal-plane (Figure 1B). Firstly, we calculated the sum of reaction moments in the frontal-plane about the lateral compartment centre-of-pressure (point *O*), $\mathbf{M}_O$, which has magnitude:

$$M_O = M_X^{TFJ} + r_{OA} F_Y^{TFJ} \tag{3}$$

Where M_X^{TFJ} and F_Y^{TFJ} are the magnitudes of the frontal-plane TFJ reaction moment and the TFJ compressive force, respectively; and r_{OA} is the horizontal distance from point *O* to the knee-joint centre (point *A*). Assuming static equilibrium, $\mathbf{F}^{MED}$, acting at the medial tibiofemoral compartment centre-of-pressure (point *B*), produces a moment about *O* that balances $\mathbf{M}_O$. The magnitude of $\mathbf{F}^{MED}$ is therefore:

$$F^{MED} = \frac{M_O}{r_{OB}} \tag{4}$$

Where r_{OB} is the horizontal distance from points *O* to *B*. Finally, the magnitude of $\mathbf{F}^{LAT}$ is simply given by:

$$F^{LAT} = F_Y^{TFJ} - F^{MED} \tag{5}$$

The medial and lateral compartment centres-of-pressure were assumed to always be located at the midpoint of the medial and lateral tibial plateaus in the frontal-plane, respectively. Reference dimensions for each plateau were obtained from Yoshioka et al.[34], and scaled to each participant's anthropometry.

Data analysis. Demographic differences between the ACLR and uninjured control groups were compared using Welch's *t*-tests, Mann-Whitney *U*-tests or Chi-square tests, as appropriate based on distribution of data. We analysed biomechanical data from the *landing phase* only, defined as the time period from foot-strike to peak knee flexion angle, because the lower-limb experiences the greatest loading during this phase[22]. The impulse of each joint force of interest, i.e. the PFJ and TFJ compressive forces, the TFJ anterior shear force, and the medial and lateral compartment forces, was defined as the time integral of that force over the landing phase.

Group means (with standard deviations), mean differences (with 95% confidence intervals), as well as Cohen's *d* effect sizes were calculated for peak joint angles, joint moments, quadriceps forces, knee-joint reaction force peaks and impulses, as well as the timing of their respective peaks. For each variable, the group mean peak values were obtained by first calculating mean peak values for each participant (by averaging the discrete peak values from that participant's successful trials), and then averaging these participant means for each group.

All biomechanical variables were normally distributed and analysed using Welch's *t*-tests with significance set *a priori* at $\alpha = 0.05$. Between-group differences in body weight were accounted for by expressing knee-joint reaction force peaks and impulses in normalised units (body weights, BW, and body weight-seconds, BWs, respectively). All analyses were undertaken in RStudio (RStudio, Inc. Boston, MA) using R v4.0.2.

RESULTS

The ACLR group was tested 17±3 months post-ACLR, were, on average, 2 years older, significantly heavier and had inferior self-reported and objective knee function (Table 1). More than two-thirds of the participants from each group were involved in

Level I or II sport at the time of testing. There was no between-group difference in quadriceps strength.

Peak values for knee flexion angle, internal knee extension moment, quadriceps force and external knee adduction moment were all lower in the ACLR group (Table 2). There were no between-group differences in the timing of each peak, except for the knee adduction moment, with controls peaking slightly later (Table 2; Figure 2). The ACLR group had a smaller knee flexion angle at initial contact, but a similar range of motion (Figure 2A).

The temporal patterns of the internal knee extension moment, quadriceps force and knee adduction moment revealed qualitative between-group differences in slope during the first half of landing (Figures 2B, 2C and 2E, respectively), with the ACLR group demonstrating a lower loading rate. Hamstrings forces were small for all participants, but on average were slightly higher in the ACLR group, who also showed substantial variability in waveform pattern and magnitude (Figure 2D).

The timing of the peaks of the PFJ and TFJ compressive forces were similar between groups (Table 2; Figures 3A and 3B). However, the magnitudes of the peaks of both the PFJ and TFJ compressive forces were significantly lower in the ACLR group (Table 2; Figures 3A and 3B). Qualitatively, the temporal pattern of the PFJ compressive force for the ACLR group was less steep during the first half of landing compared to controls (Figure 3A), indicating a lower rate of force development. The TFJ anterior shear force was not different between groups (Table 2; Figure 3C). The impulse of the PFJ compressive force was significantly lower in the ACLR group (Table 2), whereas the impulses of the TFJ compressive and anterior shear forces were not different.

The peak medial tibiofemoral compartment force was lower in the ACLR group (Table 2; Figure 4A), whereas the peak lateral compartment force was not different (Table 2; Figure 4B). There were no between-group differences in the respective impulses (Table 2).

DISCUSSION

Our objective was to quantify and compare the reaction forces and impulses in the PFJ and TFJ during the landing phase of a single-leg forward hop task in people 12-24 months after ACLR surgery and in uninjured controls. Both peak PFJ and TFJ compressive forces were lower in the ACLR group, however, only the PFJ compressive impulse was smaller, therefore our hypothesis of lower reaction forces and impulses in the ACLR knee was only partially supported. Our findings extend the emerging data showing diminished knee-joint loading after ACLR, and contribute to understanding the long-term health of the ACLR knee in physically-active people.

Our estimates of the peak PFJ compressive forces for both groups are within the range of recently published results for uninjured participants, however there is a paucity of data for robust comparison. Only one study has reported PFJ compressive force for hopping. In that study, uninjured participants performed multiple hops at a fixed distance of 80 cm (approximately l00% leg length, similar to the present study) at various cadences, with a reported peak PFJ compressive force of 8.6 BW[23]. While this value is approximately 20% lower than that for our control group, disparities may be attributable to differences in task execution and modelling paradigms. Our peak PFJ compressive force for both groups approximated values reported for vertical drop landing (7.9-9.2 BW)[35], and travelling dance jump landing (8.9-10.4 BW)[36], but greatly exceeded values reported for running at moderate speeds (4.5-6.7 BW)[9,10,37].

There is also limited data for comparison of TFJ compressive forces in hop-landing tasks. Our peak TFJ compressive forces for both groups, exceeded that reported for running[9,17], but approximated that reported for drop landings[11]; whereas our peak TFJ anterior shear forces for both groups agree with previous data for drop landings[11]. Finally, our peak knee flexion angles and peak internal knee extension moments are comparable with data for healthy[38] and ACLR[14] groups during hop-landing, reinforcing the validity of our calculated TFJ compressive and anterior shear forces.

Our study adds to a growing body of evidence suggesting that diminished compressive forces in the knee are a salient feature of biomechanics post-ACLR. Previously, we reported that people post-ACLR experienced lower peak PFJ compressive forces in their reconstructed knee compared to their uninjured contralateral limb during running[10]. The present study extends these data, showing that these quantities are also lower compared to individuals without knee injury. Furthermore, our findings of a lower peak PFJ compressive force during a hop-landing 12-24 months after surgery concurs with Bowersock et al.[9], who investigated running in individuals 4.5 years post-ACLR. We found that individuals with ACLR demonstrated a similar knee range of motion to controls, but had a smaller knee flexion angle at initial contact. Our previous work using this cohort[12] showed the ACLR group also landed with smaller ankle plantar flexion angles but greater lumbar forward flexion. This straighter-legged but more “bent-over” strategy shifted the body centre-of-mass more anteriorly, reducing the moment arm of the GRF about the knee in the sagittal-plane, and therefore quadriceps forces. Thus, the smaller knee flexion angle and reduced quadriceps force explains the lower PFJ compressive forces in the ACLR group. All other things being equal, this landing strategy would tend to accentuate the compressive force contribution to the TFJ by the quadriceps and other knee-spanning muscles, and hypothetically lead to greater TFJ compressive forces.

However, the peak quadriceps forces in our ACLR group were so diminished that their peak TFJ compressive force was, in fact, slightly lower than controls.

Our results, together with those reported for walking[7], running[9,17] and side-stepping[17] post-ACLR correspond with reduced knee-joint loading and altered movement patterns in individuals with PFJ pain[39], as well as moderate TFJ osteoarthritis in older individuals[40]. In these latter studies, knee pain likely influenced gait mechanics. However, at the time of testing, none of the participants with ACLR in our study reported any pain, and over two-thirds had returned to sport. Therefore, the reasons underpinning the reduction in compressive forces in both the PFJ and TFJ after ACLR remain unclear, but are probably related to the tendency of individuals with ACLR to land with a smaller knee flexion angle at initial contact, which may be motivated to some extent by psychological factors such as feelings of instability in the knee or fear of re-injury[41]. Furthermore, it is possible that our reported differences reflect habits formed and retained post-surgery, when the reconstructed knee may have initially been symptomatic.

Despite the lower peak compressive forces for the ACLR group, only the impulse of the PFJ compressive force was found to be different, i.e. lower in the ACLR group. This outcome may simply reflect our models, where the PFJ forces impulses are largely modulated by the knee flexion angle and quadriceps forces alone. Significantly lower quadriceps forces and knee flexion angles both contributed to a large between-group difference in peak PFJ compressive force (>2 BW), and therefore a sizeably lower impulse in the ACLR group. However, in the TFJ, which is loaded through more complex muscular contributions, the magnitudes of the differences in peak compressive forces (both the TFJ compressive force and the compartment forces) were markedly smaller (<1 BW), and any associated differences in impulses were less likely to be significant. Yet, previous running studies have reported lower compressive impulses in both the PFJ[10] and TFJ[9] after ACLR. Therefore, altered movement patterns and muscular coordination post-ACLR may not always lead to reductions in compressive impulses in the TFJ for some types of tasks, even if they reduce peak compressive forces.

Diminished peak compressive forces in the ACLR knee may be associated with the accelerated development of post-traumatic osteoarthritis. Preliminary evidence from a study of 22 individuals after ACLR showed that a subset of 7 participants who had developed TFJ osteoarthritis 5 years post-surgery had also demonstrated lower medial tibiofemoral compartment forces at 6 months post-surgery compared to those who did not develop osteoarthritis[7]. Furthermore, a study of 100 individuals with ACLR 2-3 years after surgery showed that greater TFJ compressive forces were associated with larger tibial cartilage volumes[42]. Thus, the markedly lower peak PFJ compressive forces and impulses observed in the ACLR group might be related to the high rates of post-traumatic PFJ osteoarthritis (more so than TFJ osteoarthritis) observed on MRI[25] and radiographs[18,43].

Despite this association, the causal link between early-onset osteoarthritis after ACLR and diminished compressive forces and/or impulses in the knee is not known. Understanding this link is further complicated by internal biological factors specific to the ACLR knee, such as the body's inflammatory responses to injury, increased sensitisation to load and localised changes in stress-strain distributions within the reconstructed joint. Sensitisation may be compounded by higher body mass[44], which was significantly different between groups, and is a strong predictor of post-traumatic osteoarthritis[45]. These factors may be as important as net joint loads in the ACLR knee's long-term response to loading[46]. Crucially, with our limited understanding of altered cartilage mechanobiology after ACLR, we cannot infer from our findings that, if diminished knee-joint forces after ACLR increase the risk of post-traumatic osteoarthritis, then interventions designed to elevate joint loading will counter that risk.

Our study is not without limitations. Firstly, we calculated muscle forces using a standard static optimisation approach that minimised the total activation of the lower-limb muscles, and hence could not completely account for subject-specific knee flexor-extensor muscle co-contraction behaviour. Specifically, electromyographic studies have shown elevated hamstrings-quadriceps muscle activation after ACLR surgery[11,47], which can increase the overall magnitudes of the applied muscle forces and therefore also knee-joint reaction forces. In the present study, hamstrings forces were small, but elevated in the ACLR group (Figure 2D), and showed considerable variability, a finding we also reported previously with a different musculoskeletal model[12]. Nevertheless, we acknowledge that hamstrings and quadriceps forces in the ACLR group may be underestimated. Although our TFJ compressive force and medial compartment force are both significantly lower in the ACLR group, the differences are small in magnitude (<1 BW). Thus hypothetically, when additional hamstrings-quadriceps co-contraction is considered, the true TFJ compressive forces of the ACLR group might potentially be equivalent to, or greater than, controls. However, given the large between-group difference in PFJ compressive forces in the present study (>2 BW), we believe that the overall finding of lower PFJ forces in the ACLR group would likely be unchanged. Ultimately, while static optimisation is a robust method of calculating muscle forces, future studies of single-leg forward hop-landing using EMG-informed approaches may better elucidate joint loading in the ACLR knee. As an ancillary note, our peak quadriceps forces are approximately 2

BW greater than those we reported in Sritharan et al.[12], which we attribute to more rigorous model scaling in our present work, as well as differences in muscle model parameters and musculoskeletal geometry between reference models.

Secondly, our simple static equilibrium-based force-moment models used to calculate the PFJ compressive force, and the tibiofemoral compartment forces, did not include ligaments or other tensile soft tissue that may resist unloading/destabilising forces and moments due to muscles and external loads. Notably, the medial and lateral collateral ligaments, as well as the ACL, are naturally pretensioned in the knee such that they can apply combined passive forces of 0.5-1 BW on the compartments of the TFJ[48]. While we acknowledge that our results for tibiofemoral compartment forces may be underestimated, our modelling approach (Figure 1B) has been widely applied to studies of knee-joint loading[49], while our PFJ model (Figure 1A) improves upon simpler sagittal-plane approaches previously used[10].

Thirdly, we did not collect hop-landing data for the uninjured limb of the ACLR group. A study of hop-landing biomechanics between the ACLR limb, uninjured contralateral limb and controls found differences across all pairwise comparisons[15], due to adaptations specifically associated with the ACLR limb, rather than aberrant landing strategies inherent to the individual and more likely to manifest in both limbs. We similarly found biomechanical differences between the ACLR and uninjured contralateral limbs during running due to adaptations associated with the ACLR limb[10], hence are confident that analogous between-limb differences would also be observed for hop-landing.

Finally, we did not exclude participants with ACLR who underwent concomitant partial meniscectomy. It is known that partial meniscectomy is associated with aberrant movement patterns and elevated risk of future osteoarthritis[50]. A subgroup analysis of our ACLR cohort, of whom 24 underwent concomitant partial meniscectomy, showed no group differences in peak TFJ compressive or anterior shear forces, PFJ and TFJ impulses, or peak lower-limb joint angles and moments. However, those who underwent concomitant meniscectomy had a slightly lower peak PFJ compressive force (mean difference: 0.82 BW, $p = 0.044, d = -0.51$), hinting that future targeted studies of knee-joint loading after ACLR with concomitant meniscectomy may be able to uncover novel and important findings with respect to post-traumatic osteoarthritis.

In summary, our study of single-leg forward hop-landings found that lower peak compressive forces in the knee may be a salient biomechanical feature 12-24 months after ACLR surgery. Further research is needed to better understand the role that altered knee-joint loading plays in the initiation and/or progression of osteoarthritis after ACLR in physically-active people.

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REFERENCES

1. Risberg MA, Oiestad BE, Gunderson R, et al. 2016. Changes in Knee Osteoarthritis, Symptoms, and Function After Anterior Cruciate Ligament Reconstruction: A 20-Year Prospective Follow-up Study. Am J Sports Med 44:1215-1224.

2. Culvenor AG, Cook JL, Collins NJ, et al. 2013. Is patellofemoral joint osteoarthritis an under-recognised outcome of anterior cruciate ligament reconstruction? A narrative literature review. Br J Sports Med 47:66-70.

3. Lohmander LS, Englund PM, Dahl LL, et al. 2007. The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis. Am J Sports Med 35:1756-1769.

4. Andriacchi TP, Koo S, Scanlan SF. 2009. Gait mechanics influence healthy cartilage morphology and osteoarthritis of the knee. J Bone Joint Surg Am 91 Suppl 1:95-101.

5. Culvenor AG, Schache AG, Vicenzino B, et al. 2014. Are knee biomechanics different in those with and without patellofemoral osteoarthritis after anterior cruciate ligament reconstruction? Arthritis Care Res (Hoboken) 66:1566-1570.

6. Culvenor AG, Perraton L, Guermazi A, et al. 2016. Knee kinematics and kinetics are associated with early patellofemoral osteoarthritis following anterior cruciate ligament reconstruction. Osteoarthritis Cartilage 24:1548-1553.

7. Wellsandt E, Gardinier ES, Manal K, et al. 2016. Decreased Knee Joint Loading Associated With Early Knee Osteoarthritis After Anterior Cruciate Ligament Injury. Am J Sports Med 44:143-151.

8. Hart HF, Culvenor AG, Collins NJ, et al. 2016. Knee kinematics and joint moments during gait following anterior cruciate ligament reconstruction: a systematic review and meta-analysis. Br J Sports Med 50:597-612.

9. Bowersock CD, Willy RW, DeVita P, et al. 2017. Reduced step length reduces knee joint contact forces during running following anterior cruciate ligament reconstruction but does not alter inter-limb asymmetry. Clin Biomech (Bristol, Avon) 43:79-85.

10. Sritharan P, Schache AG, Culvenor AG, et al. 2020. Between-Limb Differences in Patellofemoral Joint Forces During Running at 12 to 24 Months After Unilateral Anterior Cruciate Ligament Reconstruction. Am J Sports Med 48:1711-1719.

11. Tsai LC, McLean S, Colletti PM, et al. 2012. Greater muscle co-contraction results in increased tibiofemoral compressive forces in females who have undergone anterior cruciate ligament reconstruction. J Orthop Res 30:2007-2014.

12. Sritharan P, Perraton LG, Munoz MA, et al. 2020. Muscular Coordination of Single-Leg Hop Landing in Uninjured and Anterior Cruciate Ligament-Reconstructed Individuals. J Appl Biomech:1-9.

13. Oberlander KD, Bruggemann GP, Hoher J, et al. 2014. Knee mechanics during landing in anterior cruciate ligament patients: A longitudinal study from pre- to 12 months post-reconstruction. Clin Biomech (Bristol, Avon) 29:512-517.

14. Orishimo KF, Kremenic IJ, Mullaney MJ, et al. 2010. Adaptations in single-leg hop biomechanics following anterior cruciate ligament reconstruction. Knee Surg Sports Traumatol Arthrosc 18:1587-1593.

15. Kotsifaki A, Whiteley R, Van Rossom S, et al. 2021. Single leg hop for distance symmetry masks lower limb biomechanics: time to discuss hop distance as decision criterion for return to sport after ACL reconstruction? Br J Sports Med.

16. Erhart-Hledik JC, Chu CR, Asay JL, et al. 2018. Longitudinal changes in knee gait mechanics between 2 and 8 years after anterior cruciate ligament reconstruction. J Orthop Res 36:1478-1486.

17. Saxby DJ, Bryant AL, Modenese L, et al. 2016. Tibiofemoral Contact Forces in the Anterior Cruciate Ligament-Reconstructed Knee. Med Sci Sports Exerc 48:2195-2206.

18. Culvenor AG, Lai CC, Gabbe BJ, et al. 2014. Patellofemoral osteoarthritis is prevalent and associated with worse symptoms and function after hamstring tendon autograft ACL reconstruction. Br J Sports Med 48:435-439.

19. Culvenor AG, Eckstein F, Wirth W, et al. 2019. Loss of patellofemoral cartilage thickness over 5 years following ACL injury depends on the initial treatment strategy: results from the KANON trial. Br J Sports Med 53:1168-1173.

20. Culvenor AG, Crossley KM. 2016. Patellofemoral Osteoarthritis: Are We Missing an Important Source of Symptoms After Anterior Cruciate Ligament Reconstruction? J Orthop Sports Phys Ther 46:232-234.

21. Patterson BE, Crossley KM, Perraton LG, et al. 2020. Limb symmetry index on a functional test battery improves between one and five years after anterior cruciate ligament reconstruction, primarily due to worsening contralateral limb function. Phys Ther Sport 44:67-74.

22. Augustsson J, Thomee R, Linden C, et al. 2006. Single-leg hop testing following fatiguing exercise: reliability and biomechanical analysis. Scand J Med Sci Sports 16:111-120.

23. Ristow A, Besch M, Rutherford D, et al. 2019. Patellofemoral Joint Loading During Single-Leg Hopping Exercises. J Sport Rehabil:1-6.

24. Culvenor AG, Collins NJ, Guermazi A, et al. 2016. Early Patellofemoral Osteoarthritis Features One Year After Anterior Cruciate Ligament

Reconstruction: Symptoms and Quality of Life at Three Years. Arthritis Care Res (Hoboken) 68:784-792.

25. Culvenor AG, Collins NJ, Guermazi A, et al. 2015. Early knee osteoarthritis is evident one year following anterior cruciate ligament reconstruction: a magnetic resonance imaging evaluation. Arthritis Rheumatol 67:946-955.

26. Grindem H, Eitzen I, Moksnes H, et al. 2012. A pair-matched comparison of return to pivoting sports at 1 year in anterior cruciate ligament-injured patients after a nonoperative versus an operative treatment course. Am J Sports Med 40:2509-2516.

27. Perraton L, Clark R, Crossley K, et al. 2017. Impaired voluntary quadriceps force control following anterior cruciate ligament reconstruction: relationship with knee function. Knee Surg Sports Traumatol Arthrosc 25:1424-1431.

28. Barber-Westin SD, Noyes FR, McCloskey JW. 1999. Rigorous statistical reliability, validity, and responsiveness testing of the Cincinnati knee rating system in 350 subjects with uninjured, injured, or anterior cruciate ligament-reconstructed knees. Am J Sports Med 27:402-416.

29. Schache AG, Baker R. 2007. On the expression of joint moments during gait. Gait Posture 25:440-452.

30. Delp SL, Anderson FC, Arnold AS, et al. 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng 54:1940-1950.

31. Arnold EM, Ward SR, Lieber RL, et al. 2010. A model of the lower limb for analysis of human movement. Ann Biomed Eng 38:269-279.

32. Lu TW, O'Connor JJ. 1999. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. J Biomech 32:129-134.

33. Anderson FC, Pandy MG. 2001. Static and dynamic optimization solutions for gait are practically equivalent. J Biomech 34:153-161.

34. Yoshioka Y, Siu DW, Scudamore RA, et al. 1989. Tibial anatomy and functional axes. J Orthop Res 7:132-137.

35. Cho J-H, Kim K-H, Moon G-S, et al. 2011. The differences in patellofemoral compression force with different height. Korean Journal of Sport Biomechanics 21:335-343.

36. Simpson KJ, Jameson EG, Odum S. 1996. Estimated patellofemoral compressive forces and contact pressures during dance landings. Journal of applied biomechanics 12:1-14.

37. Lenhart RL, Thelen DG, Wille CM, et al. 2014. Increasing running step rate reduces patellofemoral joint forces. Med Sci Sports Exerc 46:557-564.

38. van der Harst JJ, Gokeler A, Hof AL. 2007. Leg kinematics and kinetics in landing from a single-leg hop for distance. A comparison between dominant and non-dominant leg. Clin Biomech (Bristol, Avon) 22:674-680.

39. Brechter JH, Powers CM. 2002. Patellofemoral joint stress during stair ascent and descent in persons with and without patellofemoral pain. Gait Posture 16:115-123.

40. Astephen JL, Deluzio KJ, Caldwell GE, et al. 2008. Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. J Orthop Res 26:332-341.

41. Kvist J, Ek A, Sporrstedt K, et al. 2005. Fear of re-injury: a hindrance for returning to sports after anterior cruciate ligament reconstruction. Knee Surg Sports Traumatol Arthrosc 13:393-397.

42. Young People With Old Knees Research T, Saxby DJ, Bryant AL, et al. 2017. Relationships Between Tibiofemoral Contact Forces and Cartilage Morphology at 2 to 3 Years After Single-Bundle Hamstring Anterior Cruciate Ligament Reconstruction and in Healthy Knees. Orthop J Sports Med 5:2325967117722506.

43. Frobell RB, Roos HP, Roos EM, et al. 2013. Treatment for acute anterior cruciate ligament tear: five year outcome of randomised trial. BMJ 346:f232.

44. Griffin TM, Guilak F. 2005. The role of mechanical loading in the onset and progression of osteoarthritis. Exerc Sport Sci Rev 33:195-200.

45. Li RT, Lorenz S, Xu Y, et al. 2011. Predictors of radiographic knee osteoarthritis after anterior cruciate ligament reconstruction. Am J Sports Med 39:2595-2603.

46. Herzog W, Clark A, Wu J. 2003. Resultant and local loading in models of joint disease. Arthritis Rheum 49:239-247.

47. Hall M, Stevermer CA, Gillette JC. 2015. Muscle activity amplitudes and co-contraction during stair ambulation following anterior cruciate ligament reconstruction. J Electromyogr Kinesiol 25:298-304.

48. Moir L, Piovesan D, Schmitz A. 2017. Cartilage Stiffness and Knee Loads Distribution: A Discrete Model for Landing Impacts. Journal of Computational and Nonlinear Dynamics 12.

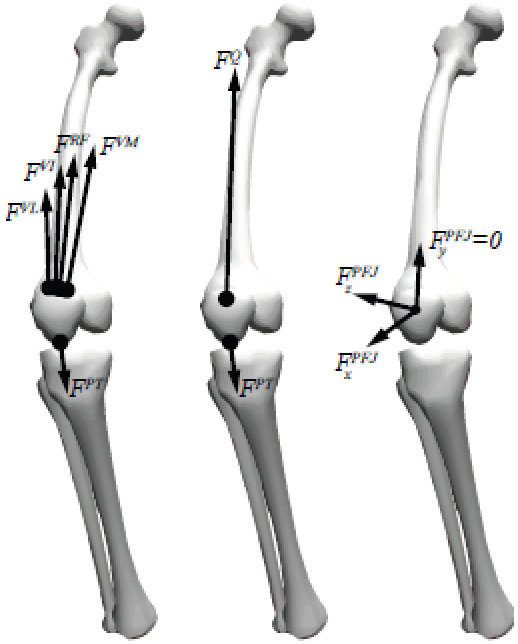
49. Winby CR, Lloyd DG, Besier TF, et al. 2009. Muscle and external load contribution to knee joint contact loads during normal gait. J Biomech 42:2294-2300.

50. Capin JJ, Khandha A, Buchanan TS, et al. 2019. Partial medial meniscectomy leads to altered walking mechanics two years after anterior cruciate ligament reconstruction: Meniscal repair does not. Gait Posture 74:87-93.

FIGURES

Figure 1. (A) Schematic diagram of the model used to calculate PFJ reaction forces. The vector sum of the vastus medius, vastus intermedius, vastus lateralis and rectus femoris muscle forces, which have magnitudes F^{VM}, F^{VI}, F^{VL}, F^{RF} respectively, is the quadriceps force vector, which has magnitude F^{Q}. The three-dimensional quadriceps force vector and the position and orientation of the patella in space are used to calculate the three-dimensional patella tendon force vector, which has magnitude F^{PT}. Using static equilibrium, and assuming zero friction (setting the component of the net PFJ reaction force acting tangent to the femur surface in the sagittal-plane to zero, i.e. $F_{y}^{PFJ} = 0$), the non-zero components of the net PFJ reaction force are the component acting normal to the instantaneous surface of the femur, and the component acting mediolaterally, which have magnitudes F_{x}^{PFJ} and F_{z}^{PFJ} respectively. (B) Schematic diagram of the tibial plateau model used to calculate tibiofemoral-joint compartment forces. F_{Y}^{TFJ} and M_{X}^{TFJ} are the known magnitudes of the TFJ compressive force and frontal-plane reaction moment, respectively, acting at the knee-joint centre, point *A*. F^{MED} and F^{LAT} are the magnitudes of the unknown medial and lateral compartment forces acting at their respective centres-of-pressure, points *O* and *B*. r_{OA} and r_{OB} are the distances from *O* to *A* and *B*, respectively. M_{O} is the magnitude of the net moment acting about *O*, due to F_{Y}^{TFJ} and M_{X}^{TFJ}.

A. Schematic diagram for calculation of patellofemoral-joint reaction forces



B. Schematic diagram for calculation of tibiofemoral-joint compartment forces

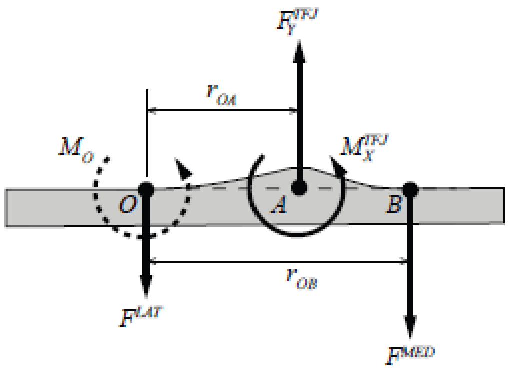


Figure 2. Time histories of the (A) knee flexion angle; (B) internal knee extension moment; (C) quadriceps force; (D) hamstrings force; and (E) external knee adduction moment, during the landing phase, for ACLR (red) and uninjured control (blue) groups. Landing phase (0-100%) is the time between foot strike and maximum knee flexion angle.

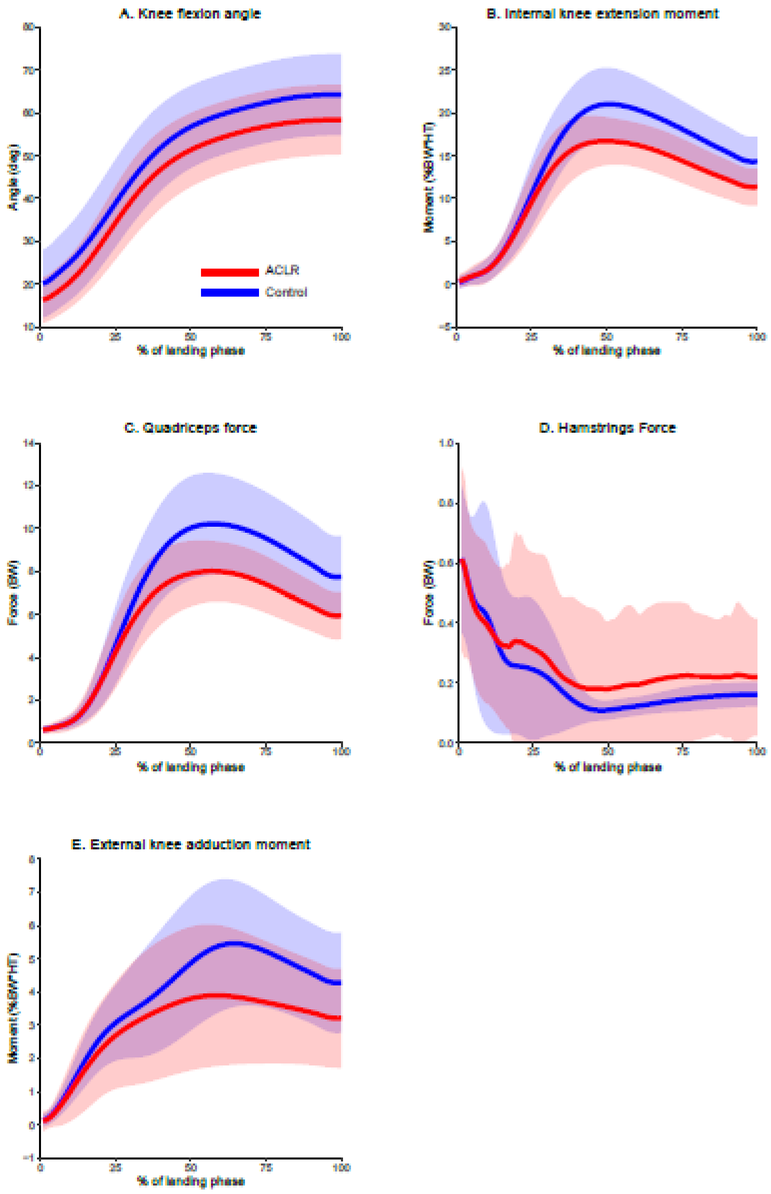


Figure 3. Time histories of the (A) PFJ compressive force; (B) TFJ compressive force; and (C) TFJ anterior shear force, for the ACLR (red) and uninjured control (blue) groups. Landing phase (0-100%) is the time between foot strike and maximum knee flexion angle.

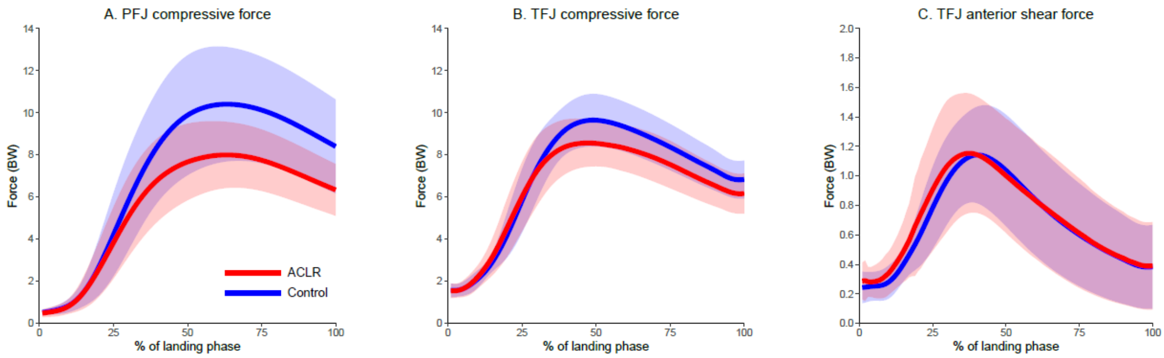
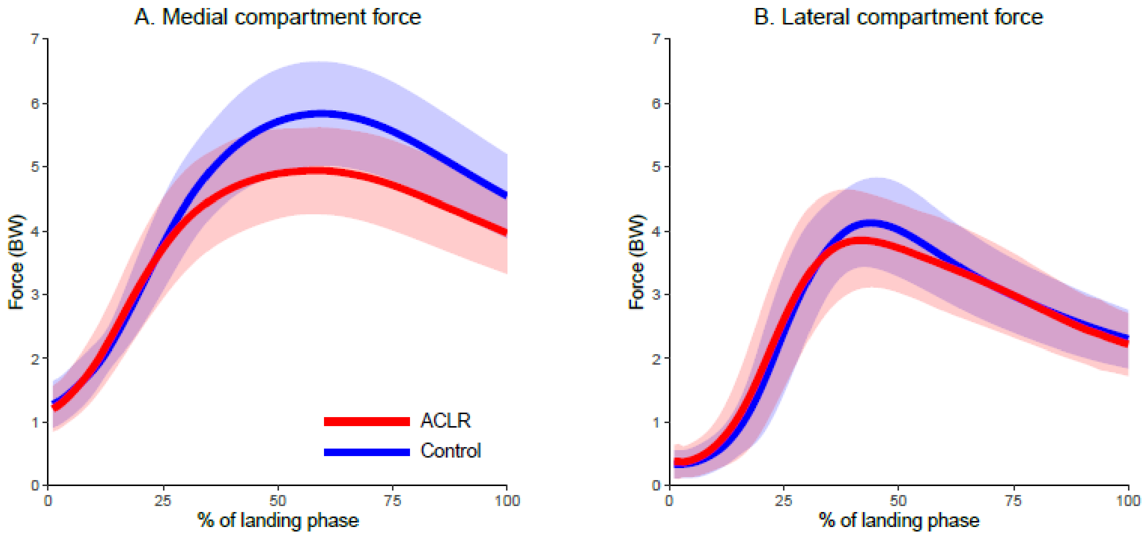


Figure 4. Time histories of the (A) medial tibiofemoral compartment forces; and (B) lateral tibiofemoral compartment forces, for the ACLR (red) and uninjured control (blue) groups. Landing phase (0-100%) is the time between foot strike and maximum knee flexion angle.



Tables

Table 1. Participant characteristics

	ACLR (*n* = 66)	Control (*n* = 33)	*P*
Age[a], years	28±6	26±5	0.050
Body mass, kg	78±15	70±12	*0.005*
Height, m	1.75±0.10	1.71±0.08	*0.044*
Maximum hop distance, m	1.20±0.23	1.32±0.28	*0.038*
Leg length, m (% of max hop distance)	0.88±0.06 (76%)	0.86±0.04 (68%)	0.052
Women[b], n (% of participants in group)	24 (36%)	16 (48%)	0.247
Quadriceps strength[c], Nm/kg	1.84±0.71	2.00±0.67	0.291
Time after ACLR surgery, months	17±3	-	-
Concomitant partial meniscectomy, n (%)	24 (36%)	-	-
Level I/II sports participation[b,d], n (%)	46 (69%)	22 (66%)	0.759
Tegner Activity Scale[e], /10	6 (3)	6 (4)	0.958
Knee function and hop performance			
Cincinnati Knee Rating Scale[e], %	90 (14)	100 (0)	< *0.001*
Hop for distance LSI[e], %	96 (12)	102 (7)	< *0.001*
Side hop LSI[e], %[¶]	90 (38)	106 (25)	< *0.001*
Landing phase time[f], seconds	0.18±0.04	0.18±0.03	0.366

ACLR: anterior cruciate ligament reconstruction; n: number of participants; LSI: limb symmetry index. Note: Unless indicated otherwise below, values are presented as mean ± standard deviation with *P*-values calculated using Welch's *t*-test. Significance levels for all tests were set *a priori* at $\alpha = 0.05$.

[a]One participant from the ACLR group was over 45 years old, no controls were over this age.

[b]Groups compared using Chi-square tests.

[c]Quadriceps strength measured as the maximum isometric knee extension torque measured using an isokinetic dynamometer where participants were seated with 90 degrees hip flexion and 60 degrees knee flexion[35].

[d]Sports activity classification as defined by Grindem et al.[22]. Level I: Jumping, cutting, pivoting (e.g. soccer, basketball); Level II: Lateral movements, less pivoting then Level I (e.g. racket sports, alpine skiing).

[e]Values presented as median (interquartile range) with groups compared using Mann-Whitney *U*-tests.

[f]Landing phase defined as time from foot strike to maximum knee flexion angle.

Table 2. Differences in peak knee-joint biomechanical variables between ACLR and control groups during the landing phase.

	Mean ± SD		**Mean diff (95% CI)**	***P***	***d***
	ACLR	**Control**			
Knee flexion angle (degrees)	59±8	64±10	-6 (-10, -2)	*0.002*	-0.67
Internal knee extension moment (%BW*HT)	18.24±2.70	21.86±4.34	-3.63 (-5.29, -1.97)	*< 0.001*	-1.09
External knee adduction moment (%BW*HT)	4.65±1.95	6.01±1.76	-1.36 (-2.16, -0.56)	*0.001*	-0.72
Quadriceps force (BW)	8.56±1.44	10.60±2.45	-2.02 (-2.95, -1.09)	*< 0.001*	-1.10
Knee-joint reaction forces (BW)					
PFJ compressive	8.56±1.64	10.80±2.81	-2.24 (-3.31, -1.18)	*< 0.001*	-1.07
TFJ anterior shear	1.35±0.43	1.25±0.32	0.10 (-0.06, 0.27)	0.229	0.26
TFJ compressive	9.35±1.02	10.10±1.19	-0.74 (-1.20, -0.28)	*0.002*	-0.69
Medial compartment	5.31±0.71	6.08±0.77	-0.76 (-1.08, -0.44)	*< 0.001*	-1.05
Lateral compartment	4.44±0.69	4.48±0.63	-0.04 (-0.31, 0.23)	0.765	-0.06
Knee-joint reaction impulse (BWs)					
PFJ compressive	0.76±0.17	0.89±0.26	-0.13 (-0.23, -0.03)	*0.012*	-0.64
TFJ anterior shear	0.13±0.07	0.12±0.03	0.02 (-0.01, 0.04)	0.230	0.26
TFJ compressive	1.21±0.26	1.26±0.19	-0.04 (-0.15, 0.06)	0.385	-0.19
Medial compartment	0.73±0.16	0.79±0.14	-0.06 (-0.12, 0.00)	0.068	-0.38

Lateral compartment	0.48±0.14	0.46±0.07	0.02 (-0.02, 0.06)	0.365	0.16
Time of peak (% of landing phase)					
External knee adduction moment	56±15	62±12	-6 (-12, -0)	*0.038*	-0.45
Quadriceps force	56±9	58±8	-2 (-6, 2)	0.280	-0.23
PFJ compressive force	62±11	64±9	-2 (-6, 3)	0.476	-0.15
TFJ anterior shear force	39±9	40±6	-1 (-5, 2)	0.416	-0.17
TFJ compressive force	48±11	49±8	-1 (-5, 3)	0.588	-0.10
Medial compartment force	59±15	60±11	-1 (-6, 4)	0.661	-0.08
Lateral compartment force	43±11	43±8	0 (-3, 4)	0.836	0.04

PFJ: patellofemoral-joint; TFJ: tibiofemoral-joint; BW: body weights; BWs: body weights-seconds; %BW*HT: percentage of body weight * subject height; SD: standard deviation; CI: confidence interval. Quadriceps force is the sum of vastus medialis, vastus intermedius, vastus lateralis and rectus femoris muscle forces. *P*-values were calculated using Welch's *t*-tests with significance set *a priori* at $\alpha = 0.05$. Significant differences are shown in *italics* and shaded grey. Effect sizes were calculated using Cohen's *d*.