HHS Public Access
Author manuscript
J Orthop Res. Author manuscript; available in PMC 2018 December 28.

Published in final edited form as:
J Orthop Res. 2017 March ; 35(3): 651–656. doi:10.1002/jor.23408.

Predictors of Knee Joint Loading After Anterior Cruciate Ligament Reconstruction

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Abstract

Anterior cruciate ligament (ACL) injury results in altered knee joint mechanics which frequently continue even after ACL reconstruction. The persistence of altered mechanical loading of the knee is of concern due to its likely role in the development of posttraumatic osteoarthritis (OA). Joint contact forces are associated with post-traumatic OA development, but evaluation of factors influencing the magnitude of contact forces after ACL injury is needed to advance current strategies aimed at preventing post-traumatic OA. Therefore, the purpose of this study was to identify predictive factors of knee joint contact forces after ACL reconstruction. Thirty athletes completed standard gait analysis with surface electromyography 6 months after ACL reconstruction. An electromyographic-driven musculoskeletal model was used to estimate joint contact forces. External knee adduction moment was a significant predictor of medial compartment contact forces in both limbs, while vertical ground reaction force and co-contraction only contributed significantly in the uninvolved limb. The large influence of the knee adduction moment on joint contact forces provides mechanistic clues to understanding the mechanical pathway of post-traumatic OA after ACL injury.

Keywords

ACL; OA; gait; cartilage; contact force

Anterior cruciate ligament (ACL) rupture is the most common internal knee lesion experienced by young individuals during sports activities.[1] ACL injury results in altered knee joint mechanics which frequently continue even after ACL reconstruction.[2–4] The persistence of altered mechanical loading of the knee is of concern due to its likely role in

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AUTHORS' CONTRIBUTIONS
AK and KM assisted with musculoskeletal model development and data interpretation. MA complete ACL reconstruction for all included subjects. TB assisted with model development and data interpretation along with substantial contributions to the research design. LS-M also contributed significantly to research design and analysis of data. All authors were involved with critical revision of this manuscript. All authors have read and approved the final submitted manuscript.

Conflicts of interest: None.

the development of post-traumatic osteoarthritis (OA).[4–6] Post-traumatic knee OA is currently an irreversible disease which affects many after ACL injury but has few effective treatment options for young individuals.[7] The prevalence of post-traumatic knee OA 10–20 years after ACL injury ranges from 10% to 90% in individuals despite reconstruction.[5] The urgency to develop preventative strategies becomes apparent when considering the negative health consequences of OA in young athletes. Those with a history of knee joint injury during youth sports demonstrated symptoms consistent with OA including lower reported knee function, worse functional hop scores, and being 3.75 times more likely to be overweight or obese 3–10 years after injury compared to their uninjured counterparts.[8] The impact of post-traumatic OA may not only influence participation in functional activities and physical activity in young adulthood but broaden into negative physical, emotional, and financial consequences later in life.

An improved understanding of knee joint loading factors which influence post-traumatic OA development is needed in order to shape effective preventative strategies. Knee joint contact forces provide a more specific understanding of knee joint loading than kinematics or moments by representing forces encountered by the articular cartilage. Lower knee joint contact forces 6 months after ACL reconstruction have been linked to the subsequent development of post-traumatic knee OA 5 years after surgery.[9] Improving knee joint contact forces may be the most malleable factor directly affecting degenerative processes occurring in the articular cartilage after ACL injury.

Joint contact forces are influenced by many elements, including both external moments acting upon the knee as well as internal forces generated by soft tissue dynamics such as muscle co-contraction.[10] Knee joint moments and muscle co-contraction patterns after ACL injury are modifiable through rehabilitation programs incorporating neuromuscular training.[11,12] However, it is unknown how factors such as joint moments and muscle co-contraction combine to influence knee joint contact forces. The poor understanding of joint loading factors in posttraumatic OA development after ACL injury has impeded the advancement of optimal preventative treatment strategies. Insight into the contributions of each of these factors to joint contact forces after ACL injury will provide a framework for designing effective post-traumatic OA preventative strategies. Therefore, the purpose of this study was to determine if ground reaction forces, knee joint moments, and muscle co-contraction predict knee joint contact forces 6 months after ACL reconstruction.

METHODS

Subjects

Thirty subjects between the ages of 14 and 51 with a complete, unilateral ACL injury within the previous 7 months were included in this study. All were part of a larger randomized control trial (Level 1 evidence) of 55 patients exploring the effects of neuromuscular training implemented prior to ACL reconstruction[13] and were regular participants in level I cutting and pivoting activities such as soccer or basketball or level II activities such as downhill skiing or tennis prior to injury.[14,15] All included subjects demonstrated poor knee function and/or episodes of giving way (noncopers) prior to ACL reconstruction.[16] Exclusion criteria included a concomitant grade III injury to other knee ligaments, repairable meniscus injury, or full-thickness articular cartilage lesion >1cm^2 diagnosed prior to ACL reconstruction.

All subjects resolved initial impairments (knee joint effusion, knee range of motion, pain, and obvious gait impairments) after completing a rehabilitation protocol described by Hurd et al.[17] prior to study enrollment. Study approval was granted by the Institutional Review Board at the University of Delaware and all patients provided written informed consent. After study enrollment subjects completed an additional pre-operative rehabilitation program aimed at restoring lower extremity strength and neuromuscular control.[13] Anatomic ACL reconstruction was completed by a single, board-certified orthopedic surgeon using either a four-bundle semitendinosus–gracilis autograft or soft tissue allograft. Criterion-based post-operative rehabilitation was completed by all subjects utilizing a protocol described by Adams et al.[18]

Three-Dimensional Gait Analysis

All 30 subjects completed biomechanical gait analysis with electromyography (EMG) 6 months after ACL reconstruction. Retro-reflective markers were placed on bony landmarks of each lower extremity and rigid shell clusters at the pelvis, thighs, and shanks.[3] Patients walked at a self-selected speed which was maintained (±5%) throughout the testing session. Stance phase kinematics and ground reaction forces were measured using an 8-camera system (VICON, Oxford Metrics Ltd., London, UK) sampled at 120Hz and one force platform (Bertec Corporation, Worthington, OH) sampled at 1,080Hz. Joint angles and joint moments calculated using inverse dynamics were determined within commercial customized software (Visual 3D, C-Motion, Germantown, MD). Kinematics and kinetic data were low-pass filtered at 6Hz and 40Hz, respectively. Moments were normalized to mass (kg) and height (m).

Surface EMG electrodes were placed on seven muscles on each limb (rectus femoris, medial and lateral vasti, semitendinosus, long head of biceps femoris, medial and lateral gastrocnemii). Surface EMG was collected at 1,080Hz (MA-300 EMG System, Motion Lab Systems, Baton Rouge, LA). Maximal voluntary isometric contractions were used to normalize EMG amplitude during subsequent walking trials. Raw EMG data was high-pass filtered (2nd order Butter-worth, 30Hz), rectified, and low-pass filtered (2nd order Butterworth, 6Hz) to create linear envelopes. The linear envelopes of each muscle was normalized to the maximum value found during maximum voluntary isometric or walking trials. EMG for the vastus intermedius was set to the average of the vastus medialis and lateralis linear envelopes. EMG for the semimembranosus was set to the linear envelope of the semitendinosus and EMG for the short head of the biceps femoris was set to the linear envelope of the long head of the biceps femoris.

EMG-Driven Modeling

Stance phase kinematic, kinetic, and EMG data collected during gait analysis were used as inputs to a patient-specific musculoskeletal model to estimate medial compartment joint contact forces.[19] This model has been validated previously by accurately and reliably predicting in vivo joint contact forces collected from an instrumented knee prosthesis during gait.[20] It has also previously been used within the ACL population to characterize the loading profiles of individuals with ACL injury, and estimated joint contact forces of the model have effectively differentiated patients with low levels of knee function and an increased risk of post-traumatic OA after ACL injury.[3,21]

Modeling steps included anatomic scaling, EMG-driven model calibration, muscle force prediction, and contact force calculation. A patient-specific lower extremity anatomic model was created using SIMM software (SIMM 4.0.2, Musculographics, Inc., Chicago, IL) to characterize musculoskeletal geometry using subject anthropometric measurements collected during gait analysis.[22] The model included foot, shank, thigh and pelvis segments driven by 10 musculotendinous actuators across the knee (rectus femoris, medial and lateral vasti, vastus intermedius, semimembranosus, semitendinosus, short and long head of biceps femoris, medial and lateral gastrocnemii). Stance-phase kinematics were used to obtain patient-specific musculotendon lengths and muscle moment arms for each trial.

An activation dynamics model was used to transform the neural EMG signal of each muscle into muscle activation. A contraction dynamics Hill-type muscle model completed transformation of muscle activation into muscle force. Model calibration aimed to seek the optimal solution to minimize the difference of the internal net moment of the knee flexors and extensors (sum of individual muscle moments [muscle force x moment arm]) to the external moment computed using inverse dynamics within the sagittal plane. Modifiable muscle characteristics including optimal fiber length, tendon slack length, electromechanical delay, nonlinear shape factor, and two recursive filter coefficients were allowed to vary with physiological bounds.[19,23] Calibration occurred until the squared difference between the sagittal plane internal moment curve and the external moment curve was minimized.

After calibration, muscle forces for three novel gait trials were estimated using optimized muscle parameters, kinematic data, kinetic data, and EMG data from gait analysis. The external knee adduction moment was expressed about a contact point in the lateral compartment fixed at the midpoint of the lateral compartment width. The lateral compartment was defined as 50% of the tibial plateau width estimated from markers placed at the medial and lateral knee. The external knee adduction moment was balanced by the internal adduction moment (calculated from estimated muscle forces) about the lateral contact point and the medial compartment contact force acting at the contact point of the medial compartment. This process allowed for the estimation of the unknown medial compartment contact force. Peak medial compartment contact force during the first half of stance was the variable of interest in this study, which was normalized to body weight (BW).

Biomechanical Variables

Variables of interested within this study included peak medial compartment contact force (normalized to body weight [BW]), vertical ground reaction force (measured at peak medial compartment contact force) (BW), peak external knee flexion moment (Nm/kg · m), peak external knee adduction moment (Nm/kg · m), and co-contraction measured by surface EMG between the knee flexors (hamstrings, gastrocnemii) and knee extensors (quadriceps) (measured at peak medial compartment contact force). Peak medial compartment contact force during the first half of stance was used due to the occurrence of peak knee flexion moment and peak knee adduction moment during this same weight acceptance period. The vertical ground reaction force was included in this analysis in addition to the knee flexion moment and knee adduction moment due to inconsistent correlational relationships between them in the current sample (vertical ground reaction force and peak knee flexion moment: Involved: *r*: 0.648, *p*:<0.001; Uninvolved: *r*: 0.179, *p*: 0.344; vertical ground reaction force and peak knee adduction moment: Involved: *r*: 0.355, *p*: 0.054; Uninvolved: *r*: 0.089, *p*: 0.639). Co-contraction was calculated using a method developed by Rudolph et al.[24] This method estimates the magnitude of co-contraction by dividing the activity of the more active muscle group by the activity of the less active muscle group, and multiplying this ratio by the sum of the activity in both groups (sum of knee flexors and knee extensors). All reported variables were the average across three walking trials.

Statistical Analysis

Statistical analyses were completed using PASSW 23.0 software (SPSS, Inc., Chicago, IL). Hierarchical linear regression analyses were used to determine the contributions of the independent variables (vertical ground reaction force, knee moments, and muscle co-contraction) to the dependent variable (peak medial compartment contact force) with separate models completed for each limb. Each variable was entered separately, beginning with knee flexion moment and followed by knee adduction moment, ground reaction force, and co-contraction, respectively, for each limb. A priori statistical significance was set at ≤ 0.05.

RESULTS

Baseline characteristics of the 30 subjects who completed gait analysis 6 months after ACL reconstruction are provided in Table 1. The mean peak medial compartment contact force for all 30 subjects 6 months after ACL reconstruction was 2.75 ± 0.66 BW for the involved limb and 2.89 ± 0.65 BW for the uninvolved limb. Overall sample means for peak knee flexion moment, peak knee adduction moment, vertical ground reaction force and co-contraction are provided in Table 2. The peak knee flexion moment, peak knee adduction moment, vertical ground reaction force, and co-contraction of the involved limb explained 38.0% of the variance in peak medial compartment contact force. Only the addition of peak knee adduction moment significantly improved the model (Table 3), and it was also the only significant contributor to the final regression model (knee flexion moment: β: 0.074, *p*: 0.727; knee adduction moment: β: 0.445, *p*: 0.016; ground reaction force: β: 0.264, *p*: 0.268; co-contraction: β: —0.137, 0.442). For the uninvolved limb, the four variables explained 79.0% of the variance in peak medial compartment contact force. Knee adduction moment, ground reaction force, and co-contraction each significantly improved the regression model when added at each of their respective steps (Table 3) and each were also significant contributors to the final model (knee flexion moment: β: 0.030, *p*: 0.750; knee adduction moment: β: 0.535, *p*: <0.001; ground reaction force: β: 0.666, *p*: <0.001; co-contraction: β: — 0.205, 0.043).

DISCUSSION

The findings of this study indicate that musculoskeletal model inputs of knee joint moments, ground reaction forces, and muscle co-contraction had stronger relationships and explained over twice as much of the variability in medial compartment loads for the uninvolved limb compared to the involved limb 6 months after ACL reconstruction. The knee adduction moment was the only significant predictor of joint contact forces in the medial compartment for both limbs. The frontal plane moment is used within the model to determine the load sharing between the medial and lateral knee compartments; thus, the strong relationship between the knee adduction moment and medial compartment contact force during weight acceptance of stance phase is not surprising and has been previously reported.[25,26] Knee adduction moment influences the pathogenesis of both primary OA and post-traumatic OA after ACL injury.[9,27–29] The current findings further substantiate the knee adduction moment's strong influence on compressive loading within the knee's medial compartment. Targeted methods using surgical or rehabilitation approaches to alter the frontal plane moment and articular cartilage loading may be essential in preventing the majority of patients with ACL injury experiencing the devastating consequences of knee OA.

The knee adduction moment is influenced both by the magnitude of the vertical ground reaction force and the lever arm upon which the ground reaction force creates an external moment about the knee. The lever arm reflects the distance between the knee joint's center of rotation with respect to the vector of the ground reaction force in the frontal plane. The lever arm has been found to have a greater impact on the magnitude of the knee adduction moment than the ground reaction force in patients with knee osteoarthritis.[30] Thus, the knee adduction moment can be large or small with an accompanying small or large ground reaction force, respectively, if the lever arm is of sufficient magnitude.[30] Lower limb alignment can influence the size of the lever arm by moving the joint center of rotation of the knee, thus influencing the magnitude of the knee adduction moment and, in turn, medial compartment contact forces. Knee adduction angle dynamically reflects the alignment of the lower limb and predicts almost three times as much variance in the first peak of the knee adduction moment compared to the vertical ground reaction force.[31] The weak correlation between the knee adduction moment and ground reaction force for the uninvolved limb within our sample (r: 0.089, p: 0.639) allows the ground reaction force to be an independent predictor of medial compartment contact forces. However, this correlation is stronger at the involved limb (r: 0.355, p: 0.054) which may prevent ground reaction force from significantly contributing to joint contact forces for the involved limb.

Joint contact forces in the medial compartment during walking at 6 months after ACL reconstruction were dependent on muscle co-contraction in the uninvolved limb but not the involved limb. Antagonistic muscles contracting simultaneously across a joint cause joint compression and increase joint contact force.[10] However, caution must be used before directly relating greater co-contraction magnitudes to greater joint contact loads. Co-contraction in the current study was calculated using surface EMG data collected during gait analysis. EMG is a measure of the neural command to a muscle and does not directly correlate to muscle force. Muscle weakness is common after ACL reconstruction, predominantly in the involved limb.[32] Although co-contraction values were not different between limbs of ACL-injured individuals, the resultant muscle forces across the knee may not have been equal between limbs due to weakened contractile elements in the knee flexors and extensors of the involved limb. Thus, a co-contraction index in the uninvolved limb of similar magnitude to the involved limb could result in a greater contribution to uninvolved joint contact forces if uninvolved limb muscles are stronger and higher muscle forces exist.

Co-contraction was inversely related to joint contact force in both limbs (Involved: β: — 0.137; Uninvolved: β: — 0.205). The negative correlation between these two variables may initially be unexpected. Muscle co-contraction is recognized as an attempt to stiffen or stabilize the knee joint after ACL injury.[24,33,34] Movement through greater and more normal knee range requires isolated muscle activation with low levels of co-contraction. However, walking with greater knee angles will also result in larger joint moments and contact forces. The product of this interaction may be the co-existence of either low levels of co-contraction with large magnitudes of joint contact force present with movement through large knee excursions, or high levels of co-contraction with large magnitudes of joint contact forces during stiffened knee gait strategies. Although not predictive of joint contact forces in the ACL-reconstructed limb in the current study, excessive muscle co-contraction between the knee flexors and extensors in the involved limb has previously been demonstrated both before and after ACL reconstruction.[24,35] Further study is warranted to verify the role of muscle co-contraction on tibiofemoral compressive forces and joint degeneration.

Compressive loading of the articular cartilage is thought to play a significant role in the development and progression of knee osteoarthritis.[36] Although the onset of post-traumatic OA after ACL injury is early and its prevalence is high, the understanding of mechanical mechanisms initiating the degeneration of articular cartilage is limited. The estimation of joint contact forces using a subject-specific EMG-driven musculoskeletal model as used within the current study provides unique insight to the loading environment of the knee through noninvasive means by incorporation estimated muscle forces derived from surface EMG data. Direct measurement of contact forces within the knee is difficult and not currently possible without placement of an instrumented knee joint prostheses.[20] Not only have asymmetric knee joint contact forces early after ACL reconstruction (estimated through musculoskeletal modeling) been linked to the subsequent initiation of post-traumatic OA,[9] but also appear to be related to an athlete's functional performance and readiness to return to sports activities after surgery.[21]

The predictors chosen within the study failed to explain the full variability in medial compartment contact forces after ACL reconstruction, particularly in the involved limb. Variables such as additional patient specific anthropometric variables (e.g., knee joint geometry obtained from CT scans) or more detailed kinematics profiles (e.g., tibiofemoral joint translation obtained from fluoroscopy) may be further determinants in resultant knee joint contact forces.[37] In addition, co-contraction was calculated using muscle activation measured from surface EMG data. Although this method provides a less direct relationship to joint contact forces than the use of estimated muscle forces from the musculoskeletal model, it was chosen as it is more readily measurable within a biomechanical laboratory setting.

In conclusion, external knee adduction moment was a significant predictor of medial compartment contact forces for both limbs during gait 6 months after ACL reconstruction, while ground reaction force and cocontraction only contributed significantly on the uninvolved limb. The large influence of the knee adduction moment on joint contact forces may provide a critical clue to understanding the mechanical pathway of post-traumatic OA after ACL injury. Further work is needed to identify additional driving factors of joint loading in the ACL-injured limb and develop treatment strategies to avert the onset and deleterious consequences of post-traumatic OA.

ACKNOWLEDGMENTS

The authors acknowledge Drs. Wendy Hurd, Erin Hartigan, and Emily Gardinier for their assistance with data collection and processing, the University of Delaware Physical Therapy Clinic for the excellent care provided to our patients, and the Delaware Rehabilitation Institute Research Core for their assistance with subject retention. This trial has been registered at clinicaltrials.gov/(NCT01773317). None of the authors have professional or financial affiliation that may be perceived to have biased the presentation.

Grant sponsor: National Institutes of Health; Grant numbers: R01 AR048212, R01 AR046386, P30 GM103333, R01 HD087459.

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Statement of Clinical Significance:

This study provides critical information in improving the understanding of mechanisms influencing the development of post-traumatic OA after ACL injury. Further work is needed to identify additional driving factors of joint loading in the ACL-injured limb and develop treatment strategies to avert the deleterious consequences of post-traumatic OA.

Table 1.

Baseline Characteristics of All 30 Subjects Who Completed Gait Analysis at 6 Months

Age (baseline) (yrs)	30.5 (11.1)
Sex	19 M, 11 F
Body mass index (kg/m^2)	26.7 (4.0)
Pre-injury activity level	17 level 1, 13 level 2
Time from ACL injury to ACL reconstruction (wks)	16.3 (11.0)
Time from ACL reconstruction to 6-month gait analysis (wks)	26.7 (2.8)
Graft type	19 allograft, 11 autograft
Walking velocity during 6-month gait analysis (m/s)	1.6 (0.1)

Parentheses represent 1 standard deviation. Autograft indicates a four-bundle semitendinosus–gracilis autograft.

Table 2.

Mean Gait Characteristics of All 30 Subjects During Gait Analysis at 6 Months

	Involved	Uninvolved
Peak medial compartment contact force (BW)	2.75 (0.66)	2.89 (0.65)
Peak knee flexion moment (Nm/kg · m)	0.38 (0.13)	0.47 (0.16)
Peak knee adduction moment (Nm/kg · m)	0.27 (0.09)	0.29 (0.09)
Vertical ground reaction force (BW)	1.13 (0.11)	1.17 (0.13)
Co-contraction	0.13 (0.06)	0.13 (0.08)

Parentheses represent 1 standard deviation.

Table 3.

Regression Analysis of Biomechanical and Co-Contraction Variables to Medial Compartment Contact Force 6 Months After ACL Reconstruction

	R^2	*p*-Value (Overall Model)	R^2 Change	*p*-Value (R^2 Change)
Involved				
pKFM	0.072	0.150	0.072	0.15
pKFM + pKAM	0.343	0.003	0.271	0.002***
pKFM + pKAM + GRF	0.365	0.007	0.022	0.357
pKFM + pKAM + GRF + CoC	0.380	0.015	0.015	0.442
Uninvolved				
pKFM	0.036	0.319	0.036	0.319
pKFM + pKAM	0.396	0.001	0.361	<0.001***
pKFM + pKAM + GRF	0.751	<0.001	0.355	<0.001***
pKFM + pKAM + GRF + CoC	0.790	<0.001	0.038	0.043***

pKFM, peak external knee flexion moment; pKAM, peak external knee adduction moment; GRF, vertical ground reaction force; CoC, co-contraction of the knee flexors and knee extensors.

*** $p \leq 0.05$.