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Anterior cruciate ligament reconstruction and cartilage contact forces – a 3D computational simulation

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Abstract

Background—Clinical outcome studies showed a high incidence of knee osteoarthritis after anterior cruciate ligament reconstruction. Abnormal joint kinematics and loading conditions were assumed as risking factors. However, little is known on cartilage contact forces after the surgery.

Methods—A validated computational model was used to simulate anatomic and transtibial single-bundle anterior cruciate ligament reconstructions. Two graft fixation angles (0° and 30°) were simulated for each reconstruction. Biomechanics of the knee was investigated in intact, anterior cruciate ligament deficient and reconstructed conditions when the knee was subjected to 134N anterior load and 400N quadriceps load at 0°, 30°, 60° and 90° of flexion. The tibial translation and rotation, graft forces, medial and lateral contact forces were calculated.

Findings—When the graft was fixed at 0°, the anatomic reconstruction resulted in slightly larger lateral contact force at 0° compared to the intact knee while the transtibial technique led to higher contact force at both 0° and 30° under the muscle load. When graft was fixed at 30°, the anatomic

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Conflict of interest

None of the authors have any financial or personal relationships with other people or organizations that could have inappropriately influenced or biased the work.

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reconstruction overstrained the knee at 0° with larger contact forces, while the transtibial technique resulted in slightly larger contact forces at 30°.

Interpretation—This study suggests that neither the anatomic nor the transtibial reconstruction can consistently restore normal knee biomechanics at different flexion angles. The anatomic reconstruction may better restore anteroposterior stability and contact force with the graft fixed at 0°. The transtibial technique may better restore knee anteroposterior stability and articular contact force with the graft fixed at 30° of flexion.

Keywords

Anterior cruciate ligament reconstruction; Biomechanics; Computational model; Contact force

1. Introduction

Anterior cruciate ligament (ACL) reconstruction has been routinely performed to treat patients with a ruptured ACL (Lyman et al., 2009). While numerous studies have reported that contemporary single-bundle ACL reconstruction is efficient to restore knee stability under an anterior tibial load (Imhauser et al., 2013; Kato et al., 2013; Li et al., 2006), many recent studies have reported prevalent radiographic osteoarthritis (OA) 10 years after ACL reconstruction (Ajuied et al., 2014; Claes et al., 2013; Lohmander et al., 2007).

Understanding the causative mechanisms of OA development after ACL reconstruction is critical for improvement of the surgery to prevent long-term joint degeneration (Dare and Rodeo, 2014).

Mechanical factors including abnormal kinematics and consequent abnormal loading within the joint often initiate knee cartilage degeneration (Felson et al., 2000; Guilak et al., 2014; Wen and Lohmander, 2014). Recent mechanobiological studies reported that above-threshold dynamic compression of cartilage in the presence of inflammatory cytokines can induce and accelerate cartilage degradation (Li et al., 2013). In-vivo patient study also demonstrated that contemporary ACL reconstruction could cause higher cartilage contact deformation compared to the uninjured contralateral side (Hosseini et al., 2012). Several in-vitro cadaveric studies have indicated that single-bundle ACL reconstructions restored knee stability with abnormal graft forces (Cuomo et al., 2007; Li et al., 2006; Yamamoto et al., 2004) and higher cartilage contact pressure (Imhauser et al., 2013; Morimoto et al., 2009). However, due to the difficulties of experimental measurement technology, the intrinsic biomechanics of the knee after ACL reconstruction, including the graft forces and articular contact forces in response to various external conditions, is still not well understood.

Therefore, the objective of this study is to investigate the effect of two widely used single-bundle ACL reconstructions (the conventional transtibial technique and the anatomic anteromedial portal ACL reconstruction technique (Bedi and Altchek, 2009)) on the graft forces and cartilage contact forces when the knee is subjected to simulated physiological loads. We simulated the anatomic and the transtibial ACL reconstructions when the grafts were fixed at 0° or 30° of flexion to successfully restore the anterior stability of the knee using a validated 3D computational knee joint model (Li et al., 1999).

2. Method

2.1. Knee model

The 3D computational knee model was constructed using MR images of a cadaveric knee. The contours of the tibia and femur bones and cartilage were digitized from the MR images to construct the 3D bony models and cartilage geometry of the knee. The insertion area of each ligament was digitized from the specimen and positioned on the bony model (Li et al., 1999). The digitization system has an accuracy of 0.2 mm. The model included tibial and femoral bones, cartilage layers, ligaments and meniscus elements (Fig. 1).

This computational knee joint model was established and validated in our previous study (Li et al., 2002). The bony structures were modeled as rigid bodies. The cartilage was modeled as a layer of linearly elastic material with a Young's modulus of 5 MPa and a Poisson's ratio of 0.45 (Blankevoort et al., 1991; Li et al., 1999). The ligaments were represented by functional bundles in the model, where the ACL and the posterior cruciate ligament (PCL) were represented by four bundles, the medial collateral ligament (MCL) by five bundles and the lateral collateral ligament (LCL) by three bundles (Suggs et al., 2003). The ligament bundles were simulated using nonlinear tensile springs (Blankevoort and Huiskes, 1996) with the axial moduli adapted from previous investigations (Andriacchi et al., 1983; Blankevoort and Huiskes, 1996; Butler et al., 1986). The medial and lateral menisci were each simulated using three compression springs that were connected to the tibia and femur, and distributed in the anterior–posterior (AP) and medial–lateral directions (Fig. 1).

In establishment of this model, the native knee joint was tested under various external loads on a robotic testing system and the kinematic responses were measured (Li et al., 2002). These experimental data was then used to validate the model predictions under the same external loads (Li et al., 2002). This validation indicated that the knee kinematics and ligament forces predicted by the model were consistent with the experimental data over a wide range of loading conditions and flexion angles. This model has been used to investigate the effect of ACL injury on knee function and the effect of graft stiffness on knee biomechanics after ACL reconstruction (Li et al., 2002; Suggs et al., 2003).

2.2. Simulation of ACL reconstruction

The ACL deficient case was simulated by removing the ACL elements from the model. To simulate a single bundle anatomic ACL reconstruction (Bedi and Altchek, 2009) and a single bundle transtibial ACL reconstruction (Gill and Steadman, 2002), the intact ACL was replaced with a graft that was also modeled using a single nonlinear spring element. The stiffness of the graft was determined using the reported axial moduli of a bone-patellar tendon-bone (BPTB) graft (10 mm in diameter) (Butler et al., 1986; Cooper et al., 1993) for both reconstructions. For each ACL reconstruction, 10 mm tunnels were made based on the insertions of the ACL on the tibia and femur. In the anatomic ACL reconstruction, the tunnels were made with centers located at the centers of ACL insertion on the tibia and femur (Figs. 2A & 2B). For the transtibial technique, the tunnels were centered at the tibial posterolateral (PL) insertion and the femoral high anteromedial position (Figs. 2C & 2D). Two experienced sports medicine surgeons (Authors 2 and 4) placed the BPTB graft

insertion centers on the tunnel inlets on the tibia and femur for the two ACL reconstruction techniques. To simulate graft fixations using interference screws, all grafts were rigidly fixed at the tunnel inlets on the tibia and femur in this study.

Surgical graft fixation was simulated at two flexion angles (0° and 30° of flexion) for both ACL reconstruction techniques. At each fixation angle, the graft's initial tension (or reference length) was adjusted until the normal anterior tibial translation of the knee was restored under a 134 N anterior tibial load (Imhauser et al., 2013; Markolf et al., 2009). The reference length of the graft was then used to calculate knee biomechanics at other flexion angles along the flexion path. In this study, a simulated 134 N of anterior load and a 400N quadriceps load (half body weight of the donor) were applied to the tibia when the knee was at 0°, 30°, 60° and 90° of flexion. The anterior load was applied at the knee center, which was defined as the midpoint of the transepicondylar line. The vector of quadriceps load was determined by the orientation of the inferior pole of the patella with respect to the tibial tubercle using radiographs of the cadaver knee (Suggs et al., 2003). In computational simulation, the femur was fixed in space. The tibia was fixed in flexion, but could move in other 5 degrees of freedom. A large sliding contact algorithm was adopted to analyze the tibiofemoral articulation. The model was analyzed in 4 conditions: intact ACL, ACL deficient, anatomic ACL reconstruction and transtibial ACL reconstruction. For each knee condition, the anterior tibial translation, internal tibial rotation, ACL/graft forces and the resultant contact forces at the medial and the lateral compartments were calculated. The patellofemoral joint was not included in this model and friction was also not modeled.

3. Results

3.1. Knee kinematics under the anterior tibial load

Under the 134N anterior load, the anterior tibial translation of the intact knee increased from 0° to 30° of flexion and then decreased slightly from 30° to 90° of flexion. The anterior tibial translations of the ACL deficient knee had the same trend, but the magnitudes were 30–40% larger than the intact knee. The internal tibial rotations of the ACL deficient knee were 16–52% larger than intact knee.

When the graft was fixed at 0°, the anatomic ACL reconstruction showed 19–32% higher anterior tibial translation and 53–88% lower graft forces than the intact knee at higher flexion angles (Table 1). The internal tibial rotation was 25% and 29% larger than intact knee at 30° and 90° respectively. However, the transtibial ACL reconstruction resulted in 11–22% lower anterior tibial translations and 68–87% higher graft forces at other flexion angles. The internal tibial rotation was 17% and 23% less than intact knee at 30° and 90° respectively.

When the graft was fixed at 30°, the anatomic reconstruction resulted in a 34% less anterior tibial translation and 42% less internal tibial rotation at 0° of flexion, but approximately 20% higher tibial translation at higher flexion angles (Table 1). The graft force was 84% higher than the native ACL at 0°, but 49% and 68% lower at 60° and 90° of flexion respectively. The transtibial ACL reconstruction resulted in 14% higher anterior tibial translations at 0° and slightly less anterior tibial translation at higher flexion angles. The

internal tibial rotation was 19% larger than intact knee at 0° and 11% smaller at 90° of flexion. Except at 0°, the graft forces were 24–43% higher than the native ACL at other flexion angles.

3.2. Knee biomechanics under the quadriceps load

The intact ACL and grafts were only affected by the quadriceps muscle load at 0° and 30° of flexion. The quadriceps load did not affect the ACL function at 60 and 90° in this knee model. Compared to intact knee, the anterior tibial translations of the ACL deficient knee were 33% and 43% larger at 0° and 30° of flexion respectively while the medial and lateral contact forces were 10–16% smaller (Table 2).

3.2.1 Graft fixation at 0°—At 0° of flexion, the anatomic ACL reconstruction resulted in a similar anterior tibial translation, internal tibial rotation and graft force as the native knee (Table 2). At 30°, the tibial translation was 6.0 mm which was 22% larger than that of the native knee, the internal tibial rotation was increased by 24%, and the graft force was 50% lower than that of the native ACL. The articular contact forces were similar between the native knee and anatomic ACL reconstructed knee at the medial compartment at 0° of flexion, but 3% higher than the native knee at the lateral compartment. The contact forces of the ACL reconstructed knee were lower than the native knee at 30° of flexion (Table 3).

For the transtibial ACL reconstruction, the anterior tibial translation was 3.7 mm and the internal tibial rotation was 5°, similar to the native knee at 0° of flexion (Table 2). The graft force was 107.2 N, 16% higher than the native ACL. At 30°, the tibial translation was 4.5 mm that was 8% lower than that of the native knee. The internal tibial rotation was reduced by 6%. The graft force was 142.8 N, 46% larger than the native ACL. The articular contact forces of the ACL reconstructed knee were 6% and 5% higher than the native knee at the medial and the lateral compartments, respectively at 0° of flexion (Table 3). The contact forces of the ACL reconstructed knee were over 9% and 12% higher than the native knee at the medial and the lateral compartments, respectively at 30° of flexion.

3.2.2 Graft fixation at 30°—At 0° of flexion, the anatomic ACL reconstruction resulted in a 2.3 mm anterior tibial translation which was 36% less than the native knee (Table 2). The internal tibial rotation was reduced by 40%. The graft force was 171.9 N, 86% higher than the native ACL. At 30°, the anterior tibial translation (4.7mm), internal tibial rotation (6.5°) and graft force (97.1 N) were similar to the native knee. The articular contact force of the reconstructed knee was 6% higher at 0°, but 3% lower at 30° of flexion compared to the native knee at the medial compartment (Table 3). At the lateral compartment, the contact forces of the ACL reconstructed knee were 17% higher at 0° and 1% higher at 30° of flexion than the native knee.

For the transtibial technique, the anterior tibial translation was 19% and 8% higher than the native knee at 0° and 30° flexion, respectively (Table 2). The internal tibial rotation was 19% (at 0°) and 10% (at 30°) larger than those of the native knee. The graft force was 57.8 N, –37% lower than the native ACL at 0° of flexion. At 30°, the graft force was 104.1 N, similar as the native ACL. The articular contact forces of the ACL reconstructed knee were 2% and 5% lower than the native knee at the medial and the lateral compartments,

respectively at 0° of flexion (Table 3). At 30° of flexion, the contact forces of the ACL reconstructed knee were 4% higher than the native knee at both the medial and the lateral compartments.

4. Discussion

In this study, we simulated two widely used single-bundle ACL reconstructions using a previously validated 3D computational knee model (Li et al., 2002). Biomechanics of the knee was calculated under a simulated quadriceps load. The data indicated that when the ACL graft was fixed at 0° of flexion, the anatomic ACL reconstruction could lead to a larger tibial translation at 30° of flexion (a loose knee). The articular contact forces were only slightly increased compared to the native knee at 0°. The transtibial reconstruction technique resulted in an over-constrained knee at 30° of flexion, accompanied with increased graft forces and articular contact forces at both 0° and 30° of flexion. When the graft was fixed at 30°, the anatomic reconstruction resulted in an over-constrained knee at 0° of flexion with an increased graft force and articular contact forces. However, the transtibial reconstruction resulted in a loose knee at 0° of flexion, but a slightly loose knee with larger contact forces at 30° of flexion.

In literature, anatomic reconstruction has been shown to lead to increased anterior tibial translations from 0° to 60° of flexion (Imhauser et al., 2013), i.e., a loose knee joint as flexion angle increases. When the graft was fixed to restore anterior laxity at 30° of flexion, anatomic reconstruction was shown to over-constrain the knee at full extension (Imhauser et al., 2013; Yamamoto et al., 2004). The graft forces of the anatomic ACL reconstruction decreased with knee flexion (Cuomo et al., 2007) and the knee was shown to be under-constrained at higher flexion angles (Loh et al., 2003; Yamamoto et al., 2004). In addition, Loh et al also reported smaller graft forces and larger anterior tibial translation at 60° and 90° of flexion (Loh et al., 2003) for an anatomic ACL reconstruction. Our study showed similar trends compared to these experimental data when the knee was subjected to an anterior tibial load.

Our computational simulation of the ACL reconstructions revealed interesting biomechanics of the knee. For the anatomical ACL reconstruction, the knee is always relatively constrained at low flexion angles, but loose at high flexion angles. Therefore, to avoid over-constraining the knee, the graft may be fixed at low flexion angles in an anatomic ACL reconstruction. There are studies reporting that anatomic ACL reconstructions resulted in 10 to 20% graft failure within five years of surgery (Aglietti et al., 2010; Suomalainen et al., 2012). The higher graft forces at low flexion angles could over-stretch the graft. Our recent in-vivo ACL kinematics study revealed an increasing stretching of the ACL at hyper-extension of the knee (Yong Feng, 2015). However, our simulation indicated that the anatomic ACL reconstruction could not provide sufficient constrain at high flexion angles of the knee. For the transtibial reconstruction, the knee is in general loose at low flexion angles, but overstrained at high flexion angles. This is consistent in trend with cadaveric studies reported in literature (Li et al., 2006; Yagi et al., 2002). Fixation of the graft at 30° caused less increases in graft forces and cartilage contact forces under the quadriceps load than fixing the graft at 0° of flexion.

These data indicated that both ACL reconstruction methods could only partially restore the knee joint biomechanics under the simulated muscle load. The cartilage contact forces cannot be consistently restored at different knee positions. The results showed that the magnitude of the graft force is related to the knee kinematics and contact forces. A larger graft force could result in less knee motion and larger contact forces. In addition to the magnitude of the graft force, the orientation of the graft force may also affect the knee kinematics and contact force after ACL reconstruction (Appendix). The anteroposterior component of the graft force resists anteroposterior translation of the knee, while the proximal-distal component is associated with the articular contact forces. It is interesting to note that the graft forces for both reconstructions were similar to the intact ACL at 30° of flexion when graft was fixed at 30° of flexion. However, the proximal component of the graft force of the transtibial reconstruction was larger, along with a larger articular contact force than the anatomic reconstruction. Therefore, restoration of both magnitude and orientation of the native ACL force is important to ensure normal kinematics and contact force.

The computational simulation data may have interesting clinical relevance. In current ACL reconstruction studies, post-operative OA has been widely reported (Fithian et al., 2005; Kessler et al., 2008; Li et al., 2011a). While various factors have been assumed to lead to joint degeneration, abnormal knee kinematics and subsequent increases in cartilage contact forces have been widely cited as a causative factor. A previous in-vivo patient follow up study by Hosseini et al. (Hosseini et al., 2012) revealed that the cartilage contact deformation of the ACL reconstructed knee could not be restored to the level of uninjured contralateral side. Further, several studies have revealed that the T1ρ and T2 relaxation times of the knee after ACL reconstruction are changed compared to normal or uninjured contralateral knee (Bae et al., 2015; Li et al., 2011b; Theologis et al., 2014). Our computational simulation indicated that while both ACL reconstruction techniques could be efficient to restore knee kinematics under the simulated muscle load, the ACL graft forces and the articular cartilage contact forces could not be consistently restored to those of the native knee. However, to further clarify the biomechanical mechanisms of post-operative cartilage degeneration, a longitudinal patient follow up investigation is necessary that will correlate the changes in cartilage contact pressure after ACL reconstruction with the cartilage degeneration development measured using cartilage thickness changes, or T1ρ or T2 values changes.

It should be noted that in our simulation of ACL reconstruction, the anatomic ACL reconstruction locates both tunnel centers at the centers of ACL insertions on the tibia and femur. However, the graft is at eccentric positions on the tibial and femoral tunnels in actual ACL reconstruction patients (Clancy et al., 1982; Higano et al., 2013). Therefore, this anatomic ACL reconstruction is not an actual center-center tunnel ACL reconstruction. The graft positioning of this study results in decreased graft forces during knee flexion under the anterior tibial load, causing either tight at full extension or slack at higher flexion angles (Imhauser et al., 2013; Yamamoto et al., 2004). To achieve actual center-center ACL graft reconstruction, eccentric placement of the tunnels for more accurate placement of the graft should be used in future studies (Clancy et al., 1982; Lee et al., 2015).

This study is a simulation of ACL reconstruction using a 3D computational knee model. A series of assumptions have been adopted in establishment of the model. For example, the graft was assumed to be rigidly fixed at the bony surfaces to represent a perfect graft fixation. The graft was simulated using a straight line. Further, this model simulates the time zero behavior of the knee. No graft/tunnel healing or remodeling was considered. Associated meniscus injuries have been known to affect the development of knee osteoarthritis after ACL reconstruction (Claes et al., 2013), but meniscus injuries were not included in this study. Despite these various assumptions, this model was previously validated and optimized using cadaveric experimental data (Li et al., 1999). Therefore, the data could provide useful references for future in-vitro or in-vivo experimental designs that are aimed to improve ACL reconstruction techniques to better restore intrinsic knee biomechanics and prevent long term cartilage degeneration.

5. Conclusion

This computational simulation indicated that neither the anatomic nor the transtibial ACL reconstructions can consistently restore normal knee biomechanics. The anatomic technique may better restore anteroposterior stability and cartilage contact forces at low flexion angles when the graft was fixed at 0° than the transtibial technique, but may result in a loose knee beyond 30° of flexion. The transtibial technique may better restore anteroposterior stability and cartilage contact forces when the graft was fixed at 30° than the anatomic technique. An optimal ACL reconstruction that can consistently restore normal knee kinematics and cartilage contact pressure is yet to be developed.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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APPENDIX

Table A1

Posterior and proximal components of the ACL and graft forces in intact and ACL reconstructed conditions of the knee in response to a 134N anterior tibial load.

Flexion angle	Intact	Graft fixation at 0°		Graft fixation at 30°	
		Anatomic	Transtibial	Anatomic	Transtibial
<u>Posterior</u> components of ACL/graft forces (N)					
0°	57.4	58.5 (2%)	58.2 (1%)	101.3 (76%)	36.6 (−36%)
30°	66.9	32.8 (−51%)	92.5 (38%)	66.0 (−1%)	70.2 (5%)
60°	67.8	17.1 (−75%)	99.3 (46%)	35.6 (−47%)	78.9 (16%)
90°	61.9	7.2 (−88%)	106.6 (72%)	19.6 (−68%)	82.7 (34%)
<u>Proximal</u> components of ACL/graft forces (N)					
0°	58.1	58.0 (0%)	86.7 (49%)	106.7(84%)	52.8 (−9%)
30°	65.2	28.0 (−57%)	132.0 (102%)	59.7 (−8%)	95.5 (46%)
60°	52.1	12.0 (−77%)	110.1 (111%)	24.7 (−53%)	84.5 (62%)
90°	32.8	3.8 (−88%)	83 (153%)	9.8 (−70%)	62.2 (90%)

Two fixation methods were simulated (match ATT at 0° and 30°). The percentages were the differences when compared to the intact knee.

Table A2

Posterior and proximal components of the ACL and graft forces in intact and ACL reconstructed conditions of the knee in response to a 400N quadriceps load.

Flexion angle	Intact	Graft fixation at 0°		Graft fixation at 30°	
		Anatomic	Transtibial	Anatomic	Transtibial
<u>Posterior</u> components of the ACL/graft forces (N)					
0°	62.8	67.5 (7%)	59.8 (−5%)	112.1 (79%)	33.0 (−47%)
30°	66.3	34.9 (−47%)	81.5 (23%)	67.7 (2%)	61.4 (−7%)
60°	0	0	0	0	0
90°	0	0	0	0	0
<u>Proximal</u> components of the ACL/graft forces (N)					
0°	62.3	65.3 (5%)	86.0 (38%)	115.8 (86%)	45.8 (−26%)
30°	63.9	28.8 (−55%)	113.1 (77%)	60.0 (−6%)	81.0 (27%)
60°	0	0	0	0	0
90°	0	0	0	0	0

Two fixation methods were simulated (match ATT at 0° and 30°). The percentages were the differences when compared to the intact knee.

Highlights

1. Neither the anatomic nor the transtibial anterior cruciate ligament reconstruction technique can completely restore normal biomechanics and articular contact force along the flexion path.
2. The anatomic anterior cruciate ligament reconstruction may better restore anteroposterior stability and articular contact force with the graft fixed at 0° of flexion.
3. The transtibial technique may better restore anteroposterior stability and articular contact force with the graft fixed at 30° of flexion.

Fig. 1.

The 3D knee model includes the femur, tibia, patella, cartilage layers, ligaments and menisci, where the menisci are simulated as equivalent resistant springs. At the lateral side a-s represents the anterior compressive spring, p-s posterior spring, and s-l the lateral spring. There are similarly three springs used at the medial side. Reprinted, with permission of the original publisher, Elsevier, from Jeremy Suggs et al. (Suggs et al., 2003)

Fig. 2.

Tunnel positions and graft positions for anatomic single-bundle anterior cruciate ligament reconstruction (A&B) and transtibial technique (C&D) using bone-patellar tendon-bone graft. For the anatomic ACL reconstruction, the tunnels were centered at the centers of ACL insertion on the tibia and femur. For the transtibial technique, the tunnels were centered at the tibial posterolateral (PL) insertion and the femoral high anteromedial position

Table 1

Anterior tibial translation, internal tibial rotation and ACL/graft force of the knee in intact, ACL deficient, anatomic ACL reconstruction and transtibial technique in response to a 134N anterior load.

Flexion angle	Intact	ACL Deficient	Graft fixation at 0°		Graft fixation at 30°	
			Anatomic	Transfibial	Anatomic	Transfibial
Anterior tibial translation (mm)						
0°	3.5	4.7 (34%)	3.5 (0%)	3.5 (0%)	2.3 (−34%)	4.0 (14%)
30°	5.3	7.2 (36%)	6.3 (19%)	4.7 (−11%)	5.3 (0%)	5.3 (0%)
60°	5.0	7.0 (40%)	6.6 (32%)	4.1 (−18%)	5.9 (18%)	4.7 (−6%)
90°	4.9	6.4 (31%)	6.2 (24%)	3.8 (−22%)	5.9 (20%)	4.4 (−10%)
Internal tibial rotation (deg.)						
0°	3.6	5.1(42%)	3.6(0%)	3.8(6%)	2.1(−42%)	4.3(19%)
30°	5.2	7.9(52%)	6.5(25%)	4.3(−17%)	5.1(−2%)	5.4(4%)
60°	3.2	3.7(16%)	3.4(6%)	3.0(−6%)	3.1(−3%)	3.2(0%)
90°	5.6	7.4(32%)	7.2(29%)	4.3(−23%)	6.7(20%)	5.0(−11%)
ACL/graft force (N)						
0°	84.8	0 (−100%)	87.1 (3%)	106.4 (25%)	155.7 (84%)	65.6 (−23%)
30°	97.5	0 (−100%)	45.7 (−53%)	163.8 (68%)	93.9 (−4%)	120.5 (24%)
60°	89.9	0 (−100%)	22.2 (−75%)	150.9 (68%)	46.1 (−49%)	117.6 (31%)
90°	72.9	0 (−100%)	8.7 (−88%)	136.5 (87%)	23.4 (−68%)	104.5 (43%)

Two fixation methods were simulated (match anterior stability at 0° and 30°). The percentages were the differences compared to intact knee.

Table 2

Anterior tibial translation, internal tibial rotation and ACL/graft force of the knee in intact, ACL deficient, anatomic ACL reconstruction and transtibial technique in response to a 400N quadriceps load.

Flexion angle	Intact	ACL Deficient	Graft fixation at 0°		Graft fixation at 30°	
			Anatomic	Trans tibial	Anatomic	Trans tibial
Anterior tibial translation (mm)						
0°	3.6	4.8 (33%)	3.5 (−3%)	3.7 (3%)	2.3 (−36%)	4.3 (19%)
30°	4.9	7.0 (43%)	6.0 (22%)	4.5 (−8%)	4.7 (−4%)	5.3 (8%)
60°	2.1	2.1 (0%)	2.1 (0%)	2.1 (0%)	2.1 (0%)	2.1 (0%)
90°	−2.5	−2.5 (0%)	−2.5 (0%)	−2.5 (0%)	−2.5 (0%)	−2.5 (0%)
Internal tibial rotation (deg.)						
0°	4.8	6.3(31%)	4.7(−2%)	5.0(4%)	2.9(−40%)	5.7(19%)
30°	6.8	10.0(47%)	8.4(24%)	6.4(−6%)	6.5(−4%)	7.5(10%)
60°	4.5	4.5(0%)	4.5(0%)	4.5(0%)	4.5(0%)	4.5(0%)
90°	−4.3	−4.3(0%)	−4.3(0%)	−4.3(0%)	−4.3(0%)	−4.3(0%)
ACL/graft force (N)						
0°	92.4	0 (−100%)	100.0 (8%)	107.2 (16%)	171.9 (86%)	57.8 (−37%)
30°	97.7	0 (−100%)	48.5(50%)	142.8 (46%)	97.1 (1%)	104.1 (7%)
60°	0	0	0	0	0	0
90°	0	0	0	0	0	0

Two fixation methods were simulated (match anterior stability at 0° and 30°). The percentages were the differences compared to intact knee.

Articular contact forces of the medial and lateral compartments in intact, ACL deficient, anatomic ACL reconstruction and transtibial technique in response to a 400N quadriceps load.

Table 3

Flexion angle	Intact	ACL Deficient	Graft fixation at 0°		Graft fixation at 30°	
			Anatomic	Transfibial	Anatomic	Transfibial
Medial contact force (N)						
0°	217.7	191.4 (−12%)	215.3 (−1%)	229.9 (6%)	230.4 (6%)	214.0 (−2%)
30°	234.5	210.2 (−10%)	219.8 (−6%)	255.2 (9%)	228.2 (−3%)	243.0 (4%)
60°	213.0	213.0 (0%)	213.0 (0%)	213.0 (0%)	213.0 (0%)	213.0 (0%)
90°	210.5	210.5 (0%)	210.5 (0%)	210.5 (0%)	210.5 (0%)	210.5 (0%)
Lateral contact force (N)						
0°	252.9	216.6 (−14%)	258.7 (3%)	264.9 (5%)	295.8 (17%)	240.5 (−5%)
30°	242.3	202.4 (−16%)	221.0 (−9%)	272.3 (12%)	244.5 (1%)	251.2 (4%)
60°	203.1	203.1 (0%)	203.1 (0%)	203.1 (0%)	203.1 (0%)	203.1 (0%)
90°	209.8	209.8 (0%)	209.8 (0%)	209.8 (0%)	209.8 (0%)	209.8 (0%)

Two fixation methods were simulated (match anterior stability at 0° and 30°). The percentages were the differences compared to intact knee.