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Effects of the posterior cruciate ligament reconstruction on the biomechanics of the knee joint: a finite element analysis

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Abstract

Background. Previous experimental studies have been conducted to evaluate the biomechanical effects of posterior cruciate ligament reconstruction; but no consensus has been reached on the preferred method of reconstruction.

Methods. The 3D finite element mesh of a knee joint was reconstructed from computed tomography and magnetic resonance images. The ligaments were considered as hyperelastic materials. The tibiofemoral and patellofemoral joints were modeled with large sliding contact elements. The 3D model was used to simulate knee flexion from 0° to 90° in four cases: a knee with a "native" posterior cruciate ligament, a resected posterior cruciate ligament, a reconstructed single graft posterior cruciate ligament, and a reconstructed double graft posterior cruciate ligament.

Findings. A resected posterior cruciate ligament induced high compressive forces in the medial tibiofemoral and patellofemoral compartments. The pressures generated in the tibiofemoral and patellofemoral compartments were nearly the same for the two reconstruction techniques (single graft and double graft). The single graft resulted in lower tensile stresses inside the graft than for the double graft.

Interpretation. Firstly, a resected posterior cruciate ligament should be replaced to avoid excessive compressive forces, which are a source of cartilage degeneration. Secondly, the two types of posterior cruciate ligament reconstruction techniques partially restored the biomechanics of the knee in flexion, e.g. contact pressures were restored for pure flexion of the knee. The reconstruction techniques therefore partially restore the biomechanics of the knee in flexion. A double graft reconstruction is subjected to the highest tensile stresses.

Keywords: Biomechanics; Knee; Posterior cruciate ligament; Reconstruction

1. Introduction

The posterior cruciate ligament (PCL) is composed of anterolateral (AL) and posteromedial (PM) fiber bundles (Harner and Höher, 1998; Harner et al., 2001). Its rupture is a source of laxity, which may induce abnormal function of the knee joint. In the long-term, abnormal cartilaginous wear might occur and then premature knee arthritis. PCL injuries are less common than anterior cruciate ligament (ACL) injuries, and they often go unrecognized. Despite its relative importance, the PCL has received much less attention regarding its anatomical and biomechanical roles in knee joint function. This lack of biomechanical information may in part explain the poorer clinical outcomes following PCL injury and surgery (Harner et al., 2001). Basically, the replacement of the deficient ligament is expected to restore the mechanical stability of the knee. However, long-term

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clinical results are still inconclusive and knee joint stability is unfortunately not always restored.

After a PCL injury, the choice between an operative and a conservative treatment remains controversial. For a surgical treatment, the surgeon might choose between single graft and double graft reconstructions. But, the long-term results of this treatment are unclear. The main subjects of studies in previous literature were focused on the kinematics effects of PCL deficiency and of PCL reconstruction. To evaluate the effectiveness of surgical treatment, few experimental studies have been conducted to measure the tibiofemoral and patellofemoral forces after PCL surgery (Kanamori et al., 2000; Skyhar et al., 1993).

Gill et al. (2004) measured the contact pressures induced by "native", deficient and reconstructed PCL in the patellofemoral joint for different angles of knee flexion, by using a thin film transducer. They observed that contact pressures were higher for a PCL-deficient knee and for a reconstructed knee, which might contribute to the long-term degeneration observed in both nonoperatively treated and PCL-reconstructed knees. Moreover, Gill et al. (2003) have conducted an in vitro biomechanical study to evaluate the effects of PCL reconstruction on the kinematics of the knee. They found that PCL reconstruction does not restore the six degrees of freedom knee kinematics.

Two techniques are commonly adopted for PCL reconstruction: (a) single graft reconstruction replacing the AL fiber bundle, and (b) double graft reconstruction replacing the AL and PM fiber bundles. The single graft reconstruction is thought to control the posterior translation over the entire range of knee flexion, but abnormal posterior translation frequently appears in the long term due probably to graft elongation (Harner et al., 2000a; Harner et al., 2000b). Double graft reconstructions were then investigated to improve the function of the knee.

These PCL reconstruction techniques (single graft and double graft reconstructions) have been compared in previous experimental studies (Bergfeld et al., 2001; Hagemeister et al., 2002; Mannor et al., 2000; Race and Amis, 1996; Race and Amis, 1998; Stahelin et al., 2001). Harner et al. (2000b) have shown that the kinematics behavior of a knee with a "native" PCL and a knee with double graft reconstructed PCL were similar. The posterior tibia translation with double graft did not significantly differ from the intact knee, while a significant difference of behavior was found with single graft reconstruction. Double graft reconstruction also restored in situ forces in the graft better than did single graft reconstruction.

Despite the improvement of PCL reconstruction with double graft, the long-term biomechanical behavior of PCL reconstruction remains unknown. Further biomechanical analysis is still needed for a rational choice between these two techniques.

This inconsistency of PCL replacement motivated the present study. A more complete knowledge of PCL biomechanics may bring new insights for better surgical reconstruction. PCL replacement may be improved if biomechanical behavior of the knee joint is better controlled. To improve current surgical results, it is essential to look further into the interaction of the PCL with other joint elements. Namely, the knowledge of stress inside the soft structures and joint bearing forces (tibiofemoral, patellofemoral) is required to better understand the biomechanical behavior of the joint. The experimental measurement of forces, strains and tissue interactions is extremely difficult. This difficulty is enhanced by the knee joint morphology, which is highly individual; a small shape change of any element of the knee joint may induce significant differences not only in the kinematics but also in the biomechanics of the joint.

The main goal of this study was to develop a numerical model, which is able to evaluate the effect of PCL resection and the different PCL reconstruction techniques on the biomechanics of the knee. The *first step* was to generate the numerical model of an individual healthy knee joint with a "native" PCL including bones (femur, tibia, patella, fibula) and major soft tissues (ligaments, patellar tendons, cartilage and menisci). In the *second step*, the PCL was resected for comparison to a knee with "native" PCL. The *third step* consisted of evaluation of the influence of surgical reconstruction techniques on the forces generated in the tibiofemoral and patellofemoral compartments, and on the stresses induced inside the PCL and inside the grafts by using the previously developed model. Three cases were simulated: a knee with (a) a resected PCL, (b) single graft reconstructed PCL, and (c) double grafts reconstructed PCL.

2. Methods

2.1. Data acquisitions

Image acquisition (magnetic resonance images, computed tomography images) was performed at the Department of Diagnostic and Interventional Radiology, Centre Hospitalier Universitaire Vaudois, Lausanne, Switzerland.

Data acquisitions were performed on the right knee of a volunteer. The knee was immobilized in full extension inside a plaster cast avoiding any movement during magnetic resonance (MR) and computed tomography (CT) scanner image acquisition. Eight points of references were placed on the lower limb (4 points on the femur and 4 points on the tibia) in order to match the 3D geometrical models of bone and soft structures reconstructed from CT-scanner and MR images.

Magnetic resonance (MR) images were used for the 3D reconstruction of soft structures (ligaments, menisci, tendons, and cartilage). The scanner used for acquisition was on MR scanner (Siemens, model Magnetom Symphony, Germany). The optimal size of the pixels (pixel spacing) was 0.39 mm, with a resolution of 512×512 pixels. Sections were 3 mm thick. The images were taken in the sagittal, coronal and transverse planes.

CT-scanner images were used for 3D reconstruction of bone structures (tibia, femur, patella). The femur and the tibia were sectioned transversely parallel to the bicondylar plane. Sections were 1.25 mm thick and spaced every 5 mm.

2.2. Reconstruction of bones and soft structures

Amira 3.0 software was used for semi-automatic segmentation of MR and CT-scanner slices. The external contour of bones and soft tissues were then accurately defined on each CT and MR slices with a digitization less than 0.8 mm (2 pixels).

The amount of error in this study was estimated to vary from 1 to 3 pixels (0.8 ± 0.4 mm), due to patient movement during examination.

Bones and soft structures were matched by using reference points fixed on the femur and tibia during image acquisition (Fig. 1).

The curves obtained from Amira were transferred to the Patran software (MacNeal-Schwendler, South Coast Metro, California, USA) and used for the 3D reconstruction of bones and soft tissue structures. The 3D meshes of different structures were then generated with Patran (Fig. 2).

Bone structures were meshed with rigid surface elements due to their small strain compared to soft structures. Soft structures were meshed with 3D hexahedral elements.

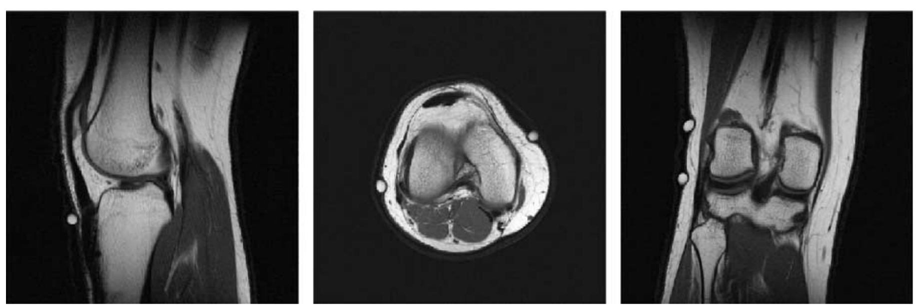

Fig. 1. Magnetic resonance images with reference points in the sagittal, coronal, and frontal planes.

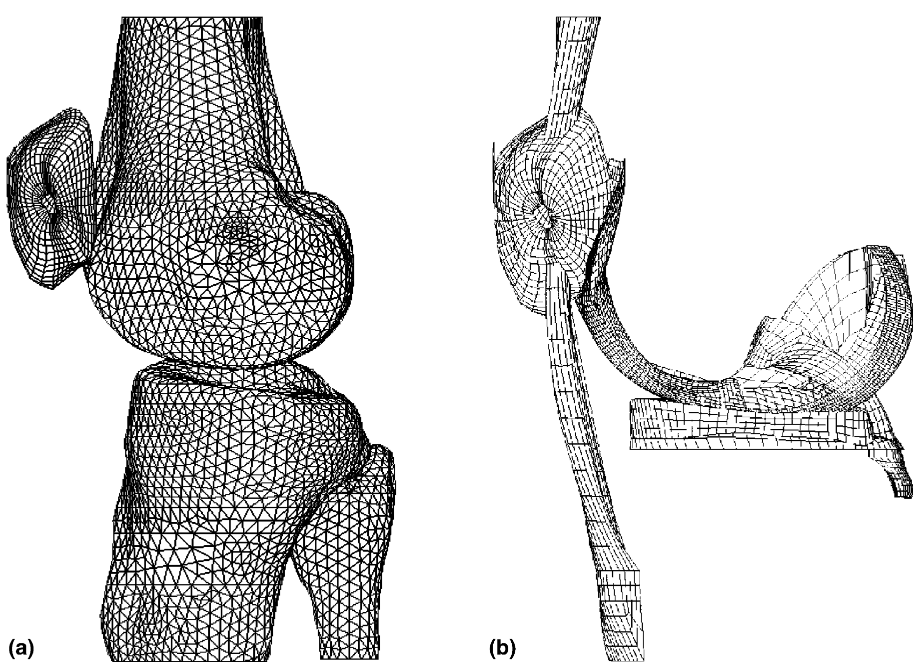

Fig. 2. (a) Three-dimensional finite element meshes of bone structures with tetrahedral elements (femur, tibia, patella, fibula). (b) Three-dimensional finite element meshes of soft structures (cartilage layers, ACL, PCL, LCL, MCL, menisci, patellar tendons).

The medial collateral ligament (MCL), the lateral collateral ligament (LCL), the ACL, and the PCL were taken into account. These ligaments and the patellar tendon were modeled with a non-linear hyperelastic law corresponding to the strain energy (Pioletti et al., 1998; Pioletti and Rakotomanana, 2000):

$$W_e = \alpha \exp\left[\beta(I_1 - 3) - \frac{\alpha\beta}{2}(I_2 - 1)\right]$$

α and β are material constants, and I_1 and I_2 are the strain invariants.

$$I_1 = \mathrm{tr}[C]$$

$$I_2 = \frac{1}{2}\left([\mathrm{tr}\ C]^2 - \mathrm{tr}\ [C]^2\right)$$

$C = F^{\mathrm{T}}F$ being the (right Cauchy-Green) material metric tensor, where $F = \frac{\partial y}{\partial x}$ is the gradient deformation tensor.

The mean values of α and β were obtained from experimental measurements (Pioletti et al., 1998) and are reported in Table 1.

The cartilage layers of the tibia, femur and patella were considered as homogeneous isotropic materials (Young's modulus: 12 MPa, Poisson's ratio: 0.45) (Moglo and Shirazi-Adl, 2003). The mechanical properties of the quadriceps were obtained from literature (Staubli et al., 1999). The mechanical properties of the medial and lateral collateral ligaments were obtained from (Woo et al., 1986). Mechanical properties of patellar tendon were used for grafts (Pioletti et al., 1998).

2.3. Contact surface modeling

The tibiofemoral, patellofemoral and meniscofemoral joints were modeled with frictional discontinuous unilateral contacts elements allowing large slip. Coulomb friction was used for the tangential contact law. The coefficient of friction was set to 0.1 (McCutchen, 1962). These interfaces allowed computing of the stress transfer.

2.4. Loading conditions

The loading condition was flexion from full extension to 90° of flexion. Flexion of 90° was obtained by applying forces in the directions of the biceps femoris and the semi-tendinosus muscles. The femur was fixed and the tibia was free in 6 degrees of freedom. The quadriceps muscles were represented with 80 linear springs fixed in their upper section, corresponding to their attachment at the proximal femur. All ligaments were considered to be free of stress at the full extension position.

Table 1
Mean values of α and β (Pioletti et al., 1998)

	α [MPa]	β
Anterior cruciate ligament	0.30	12.20
Posterior cruciate ligament	0.18	17.35
Patellar tendon	0.09	66.96

3. Numerical implementation

The numerical simulations were performed with ABAQUS/Standard 6.3 software (Hibbit, Karlsson and Sorensen Inc, Pawtucket, Rhode Island, USA). The model was applied to calculate the total force due to contact pressure in the tibiofemoral and patellofemoral compartments, and the tensile stress inside the PCL.

Four cases were simulated, a knee with:

(a) Native PCL,
(b) No PCL,
(c) Reconstructed PCL with single graft: The femoral and tibial attachments of the bundle graft were located within the native AL bulk of the PCL insertion sites.
(d) Reconstructed PCL with double graft: The attachments of the bundle grafts were located within the native AL and PM bulks of the PCL insertion sites.

4. Results

4.1. Medial tibiofemoral compartment

The compressive forces in the medial tibiofemoral compartment during knee flexion were calculated in the four cases (native PCL, no PCL, single graft reconstruction and double grafts reconstruction) as shown in Fig. 3. It was observed that the maximal values of the compressive force in the medial compartment occurred at 65° of flexion. The maximal values are reported in Table 2. It has been shown (Skyhar et al., 1993) that no PCL induced a compressive force 30% higher than with a "native" PCL.

The contact zones in the medial and lateral compartments at different angles of flexion with a "native" PCL are shown in Fig. 4.

4.2. Lateral tibiofemoral compartment

The compressive force in the lateral tibiofemoral compartment was also calculated during knee flexion (Fig. 5). No PCL induced a lower compressive force in the lateral tibiofemoral compartment. The maximal values are reported in Table 3. The behaviors of native PCL and reconstructed PCL were similar; no significant difference was found with the native PCL and reconstructed PCL at different angles of flexion. The

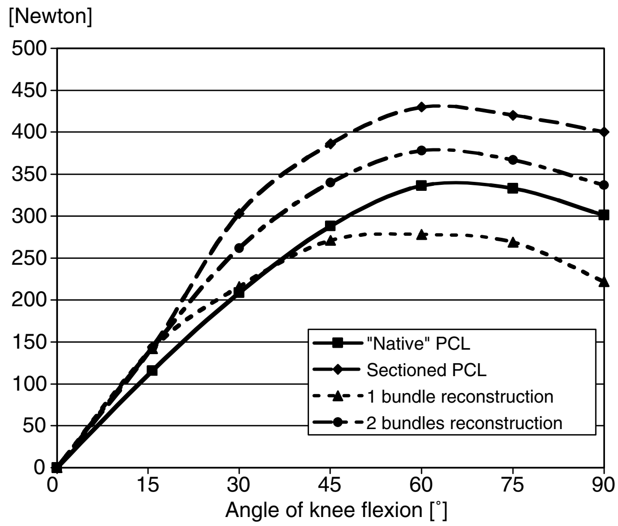

Fig. 3. Evolution of compressive force during a knee flexion in the medial tibiofemoral compartment with (a) native PCL, (b) a resected PCL, (c) single graft reconstruction, and (d) two bundles reconstruction.

Table 2
Maximal values of joint bearing forces in the medial tibiofemoral compartment at 65° of knee flexion

	Compressive force [N]
A "native" PCL	338
A sectioned PCL	445
One bundle reconstructed PCL	311
Two bundles reconstructed PCL	378

distribution of compressive forces in the lateral tibiofemoral compartment is reported in Fig. 4.

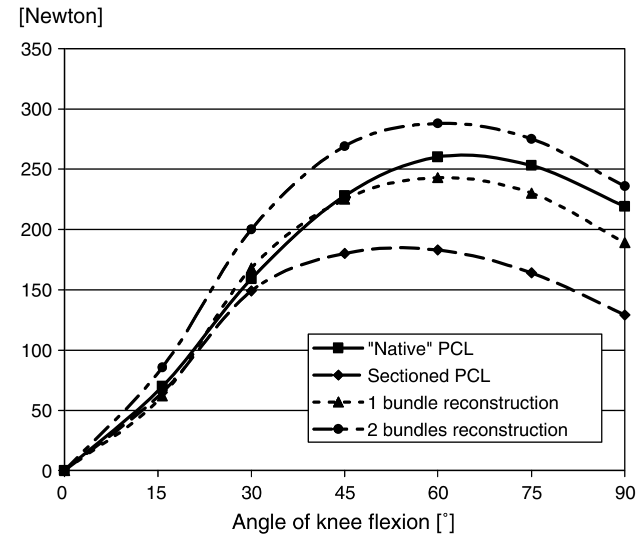

Fig. 5. Evolution of compressive force during a knee flexion in the lateral tibiofemoral compartment with (a) native PCL, (b) a resected PCL, (c) single graft reconstruction, and (d) two bundles reconstruction.

4.3. Patellofemoral compartment

The force generated in the patellofemoral compartment by knee flexion was also calculated (Fig. 6). The distribution of the contact pressure at the cartilage layer of the patella was reported at different angles of flexion in Fig. 7. The maximal values are reported in Table 4. No PCL induced a higher force. The value of patellar contact force increased with knee flexion angle. No significant difference was found between the compressive force induced with a native and reconstructed PCL.

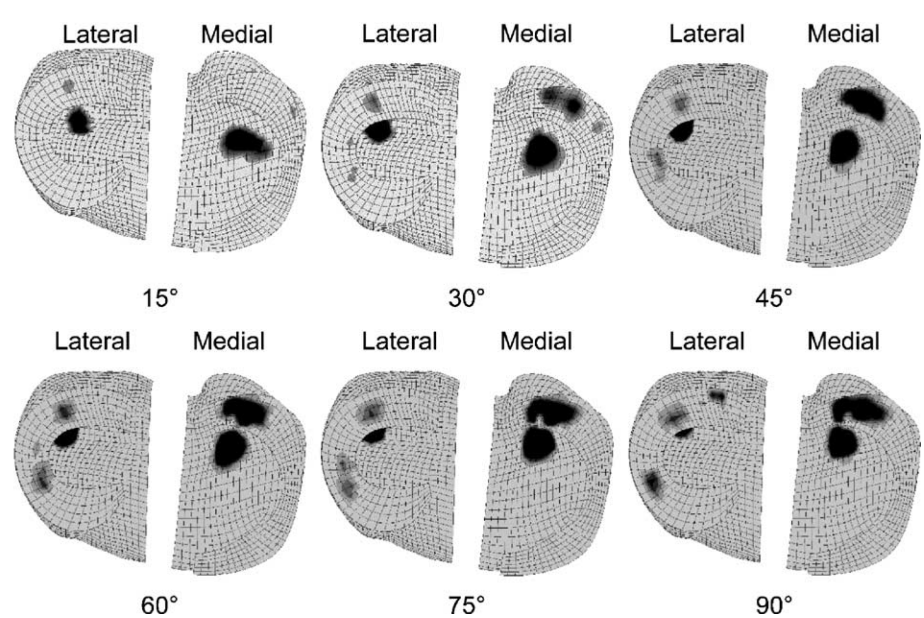

Fig. 4. Distribution of the compressive stress at the cartilage layer of the tibial compartments during a knee flexion. The contact surface in the tibiofemoral compartment is composed of the part of the tibia cartilage uncovered by the menisci and the part covered by the menisci.

Table 3
Maximal values of joint bearing forces in the lateral tibiofemoral compartment during a knee flexion

	Compressive force [N]
A "native" PCL	255
A sectioned PCL	183
One bundle reconstructed PCL	238
Two bundles reconstructed PCL	288

Table 4
Maximal values of joint bearing forces in the patellofemoral compartment at 65° of knee flexion

	Compressive force [N]
A "native" PCL	398
A sectioned PCL	440
One bundle reconstructed PCL	402
Two bundles reconstructed PCL	398

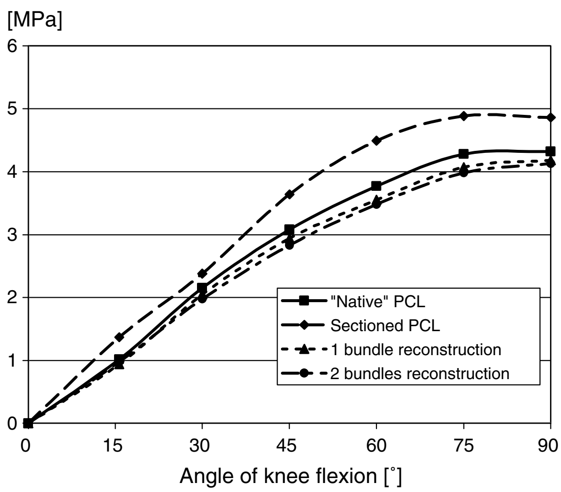

Fig. 6. Evolution of compressive stress during a knee flexion in the patellofemoral compartment with (a) native PCL, (b) a resected PCL, (c) single graft reconstruction, and (d) two bundles reconstruction.

4.4. Tensile stress in the PCL

The maximal values of tensile stress inside the native and the reconstructed PCL, during knee flexion, were calculated (Fig. 8). The tensile stress inside the PCL increased with the angle of knee flexion. The maximal values were obtained at 90° of flexion.

The single graft reconstruction induced a lower tensile stress inside the graft compared with the double grafts reconstruction.

5. Discussion

No numerical calculations were previously conducted to evaluate the biomechanical effects of PCL deficiency and replacement. The present numerical study might therefore bring new insights to better understand the effectiveness of PCL reconstructions. The main goal of our study was to evaluate the biomechanical effects of a resected PCL and its replacement. To this end, we have *firstly* developed a numerical model of an intact knee with a "native" PCL as a reference. In a *second step*, the PCL was resected. In a *third step*, the PCL was replaced with a one-bundle reconstructed graft replacing the AL fiber of the PCL. In a *fourth step*, the PCL was replaced with a double graft; the insertion zones of the bundles were located at the insertion zones of the native PCL. The graft was modeled as a hyperelastic material corresponding to the constitutive law

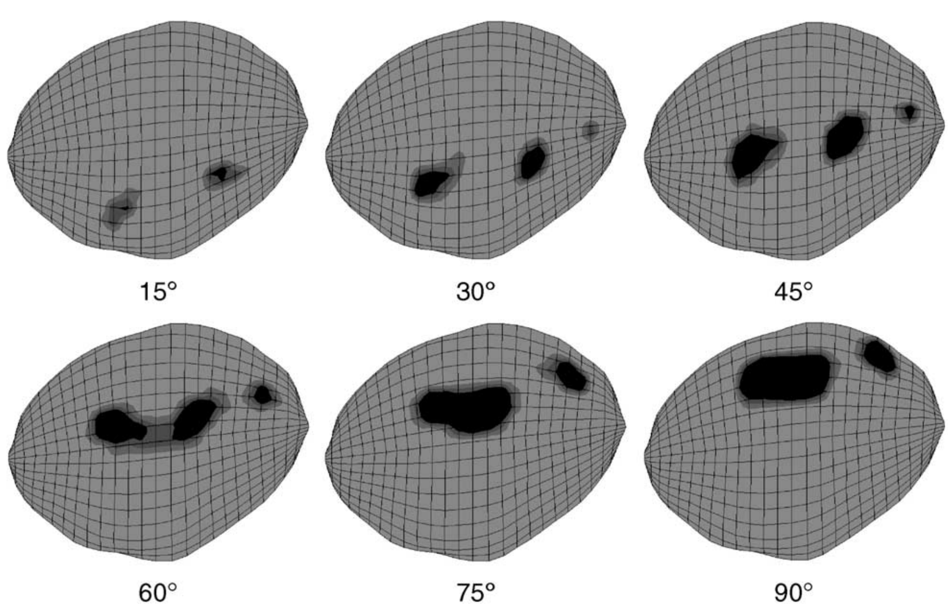

Fig. 7. Distribution of the compressive stress at the cartilage layer of the patella during the knee flexion.

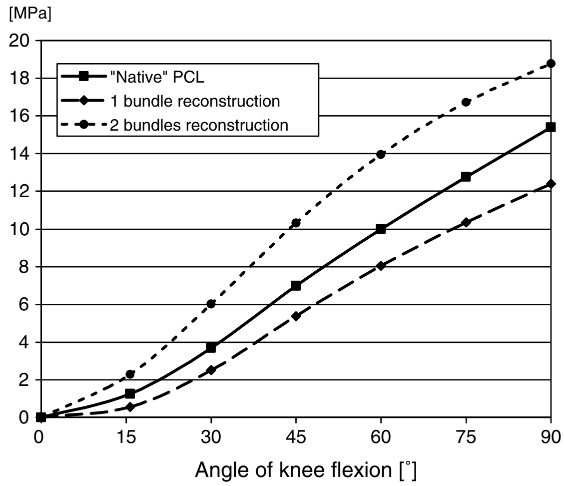

Fig. 8. Evolution of peak of tensile stress inside the PCL during the knee flexion in the (a) native PCL, (b) single graft reconstruction, and (c) double grafts reconstruction.

of patellar tendon. This law allowed for finite deformation of the soft tissues. The model included the major bony and soft structures of the knee, mainly the ligaments, the cartilage layers, the menisci and the patellar tendon.

Previous studies (Bendjaballah et al., 1997; Blankevoort and Huiskes, 1996; Li et al., 1999; Mommersteg et al., 1996) have been conducted to measure the kinematics and the evolution of joint contact areas by using 3D numerical models. Li et al. (1999) developed a 3D finite element tibiofemoral joint model (FEM) of a human knee validated by experimental data. The joint model geometry was reconstructed from MR images of a cadaverous knee specimen. Knee kinematics data under anterior–posterior tibia loads were obtained. They calculated the joint kinematics and in situ forces in the ligaments in response to axial tibia moments. However, in their model, ligaments were modeled with non-linear elastic springs, and menisci were simulated by equivalent-resistance springs. This model did not allow calculation of stresses in these soft tissues as accurately as desired.

Bendjaballah et al. (1997) used a non-linear 3D finite element model of the human tibiofemoral joint to investigate the mechanics of the knee under drawer forces. They confirmed that the PCL and ACL were the primary restraints to femoral anterior and posterior drawer forces respectively. They also found that a resection of one of the cruciate ligaments increased drastically the joint anterior–posterior motion. Moreover, resection of cruciate ligaments (PCL and ACL) increased the compressive force on the tibia plateau transmitted through the menisci.

Blankevoort and Huiskes (1996) have developed a mathematical model of the knee. Their model was simplified by considering ligaments as multiple straight-line elements and not with 3D geometries of the ligaments (Mommersteeg et al., 1996a). They did not take into account the stabilizing effects of the menisci. The model has been improved in (Mommersteeg et al., 1996b) by using multi-bundle structures with non-uniform mechanical properties and zero force lengths as ligaments, but the patellofemoral joint was not taken into account.

An analytical model of the knee in the sagittal plane was developed by Zheng et al. (1998) to estimate the forces at the knee during exercise. They determined tibiofemoral compressive forces and cruciate ligament tensions during knee extension, leg press and squat by using the resultant force and torque at the knee, muscle forces, and orientation and moment arms of the muscles and ligaments. In their model, forces in the sagittal plane only were analyzed.

In our model, the tibiofemoral, patellofemoral and meniscofemoral joints were modeled with frictional discontinuous unilateral contact elements allowing large slip. The three dimensional geometries of the ligaments, cartilage layers, menisci and the patellar tendon were reconstructed from MR images. The ligaments and patellar tendon were modeled with a hyperelastic law allowing large deformations. The strain energy of the constitutive law was experimentally determined and identified.

5.1. Medial femorotibial and patellofemoral compartments

The medial tibiofemoral compressive force was the first biomechanical parameter calculated in our study. Our results have shown, that a resected PCL induced a slightly high compressive force in the medial tibiofemoral and patellofemoral compartments at 65° of flexion. The high compressive force with a resected PCL, calculated in this study, in the medial tibiofemoral and patellofemoral compartments could be related to the high pressure measured with a PCL-deficiency in (Kanamori et al., 2000; Singerman et al., 1999; Skyhar et al., 1993), and might explain the occurrence of osteoarthritis developed in long term in these regions with nonoperatively treated PCL injuries (Harner et al., 2001). Keller et al. (1993) have suggested that despite a good short-term result of non-operative treatment of the PCL, late knee arthritis might occur.

The single graft reconstructed PCL induced a slightly lower pressure than the "native" and the double graft reconstructed PCL. This result could be related to the work of Harner et al. (2000a), which suggested that the tibiofemoral contacts and the posterolateral structures might share the load in the case of single graft reconstruction.

Abdel-Rahman and Hefzy (1998) developed a complex mathematical model of an intact knee joint to calculate the tibiofemoral contact pressure and the ligament forces during a knee flexion. As a loading condition, they applied a posterior force at the center of mass of the tibia. They found that the contact pressure in the medial tibiofemoral joint increased until 60° and then decreased until 90°. By comparing our results with their findings, it was found that the behavior of the contact pressure in the medial compartment was similar to our results. They found that the force in the AL bulk of the PCL was maximal around 60° of knee flexion. However, in their model, the ligamentous structures were represented with spring elements and the tibiofemoral contact surfaces were considered as planar surfaces.

5.2. Lateral femorotibial compartment

The second variable calculated, in this study, was the force generated in the lateral femorotibial compartment. In the four cases, we have found that the contact pressure in the lateral compartment reached a maximum around 65° of flexion and then decreased until 90°. A ''native'' PCL, a resected PCL, and the double graft reconstructed PCL induced nearly the same force in the lateral compartment. The single graft reconstruction induced the lowest pressure. This observation seems to confirm the experimental findings in previous studies (Skyhar et al., 1993; Singerman et al., 1999) where the rupture of the PCL did not affect the lateral tibiofemoral compartment. Our results on the pressure in the lateral tibiofemoral compartment were different from that of Abdel-Rahman and Hefzy (1998) which found that the contact force decreased from full extension to 90° of flexion. We suggest that this difference in their results and our results might be due to the difference in the geometry of the tibial plateau and in the representation of the ligaments, and to the difference of loading conditions.

5.3. Tensile stress inside the PCL and grafts

The last biomechanical variable calculated was the tensile stress inside the PCL. We have calculated the tensile stress generated inside the PCL by the movement of the tibia due to the simulated forces of the biceps and the semitendinosus muscles as described in the loading conditions. For a native PCL and reconstructed PCL, the high values of tensile stress are located at the femoral insertion zones of the graft that are zones of PCL ruptures as observed in some clinical cases. We have found that the tensile stresses for a native PCL and double graft reconstructed PCL were slightly greater than that of the single graft reconstructed PCL as seen in the work of Harner et al. (2000a).

5.4. Limitations of the model

There are several limitations of our model. First, we have simplified the model by considering only as ligaments the ACL, the PCL and the collateral ligaments. We have not taken into account the posterolateral structures and the capsular ligamentous structures as in previous experimental studies (Skyhar et al., 1993; Singerman et al., 1999).

Second, knee flexion only was the loading condition considered. Other loading conditions such as internal/external rotation or varus/valgus loading should be tested to better understand the biomechanical behavior of the knee with a ''native'', a resected and reconstructed PCL. Moreover, as seen in experimental measurements by other authors, the posterior tibial loading in different angle of the knee flexion should be simulated to test the laxity of the knee with a resected PCL and after reconstruction of the PCL. This study does not take into account graft preconditioning and remodeling. Consolidation at the attachment site was also not considered.

5.5. Perspectives

Our results were similar with experimental measurements of previous studies. In order to improve our model, a more sophisticated model should be used; the posterolateral and the capsular ligamentous structures that have an important role in knee stability should be included. Moreover, an anatomic biomechanical study should be conducted to investigate optimal graft pretension during PCL reconstruction.

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