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Partial Meniscectomy Changes Fluid Pressurization in Articular Cartilage in Human Knees

M. Kazemi [1], L.P. Li [1*], M.D. Buschmann [2], P. Savard [2]

[1] Department of Mechanical and Manufacturing Engineering, University of Calgary,
2500 University Drive, N.W., Calgary, Alberta, Canada T2N 1N4
[2] Institut de Génie Biomédical, École Polytechnique de Montréal, C.P. 6079,
succ. Centre-ville, Montréal, Québec, Canada H3C 3A7

*Corresponding author:
LePing Li, Ph.D., P.Eng.
Department of Mechanical and Manufacturing Engineering
University of Calgary
2500 University Drive, N.W.
Calgary, Alberta, Canada T2N 1N4

Phone: 1 403 210 7537; Fax: 1 403 282 8406
Email: Leping.Li@ucalgary.ca

Abstract

Partial meniscectomy is believed to change the biomechanics of the knee joint through alterations in the contact of articular cartilages and menisci. Although fluid pressure plays an important role in the load support mechanism of the knee, the fluid pressurization in the cartilages and menisci has been ignored in the finite element studies of the mechanics of meniscectomy. In the present study, a 3D fibril-reinforced poromechanical model of the knee joint was used to explore the fluid flow dependent changes in articular cartilage following partial medial and lateral meniscectomies. Six partial longitudinal meniscectomies were considered under relaxation, simple creep and combined creep loading conditions. In comparison to the intact knee, partial meniscectomy not only caused a substantial increase in the maximum fluid pressure but also shifted the location of this pressure in the femoral cartilage. Furthermore, these changes were positively correlated to the size of meniscal resection. While in the intact joint, the location of the maximum fluid pressure was dependent on the loading conditions, in the meniscectomized joint the location was predominantly determined by the site of meniscal resection. The partial meniscectomy also reduced the rate of the pressure dissipation, resulting in even larger difference between creep and relaxation times as compared to the case of the intact knee. The knee joint became stiffer after meniscectomy because of higher fluid pressure at knee compression followed by slower pressure dissipation. The present study indicated the role of fluid pressurization in the altered mechanics of meniscectomized knee.

Keywords**:** Articular cartilage mechanics, Creep, Stress relaxation, Finite element analysis, Fluid pressure, Knee joint mechanics, Meniscectomy

1. Introduction

The menisci in the knee play an important role in joint function during daily activities, such as improving the joint congruency, joint stability and lubrication [1-3]. The menisci support 45-75% of the joint load, depending on the loading and health condition of the knee [4]. The load support mechanism of the menisci is facilitated by its wedge shape, which reduces up to 90% of the total contact area between the femoral and tibial cartilages [5].

Traumatic injuries commonly occur in the menisci and may change the joint mechanics. If the tear occurs in the avascular zone, i.e. the inner peripheral region, the chance of healing is very low due to lack of blood supply there; and so partial meniscectomy is a possible treatment [6]. Previous studies indicate that the removal of the inner third of the meniscus decreased the contact area by 10% and increased the local peak contact stresses by 65% [7]. The long-term consequences of partial meniscectomy include decreased tensile properties of cartilage [8], cartilage damage [9], and increased occurrences of osteoarthritis [10,11]. The increase in the size of meniscus resection was reported to increase the chance of ostearthritis and accelerate its progression [10].

In parallel to the experimental investigations, finite element modeling has been generally accepted to study the impact of partial meniscectomy on the knee joint biomechanics. Considering linear elastic behavior for the cartilages and menisci, an increase in contact pressure was reported after partial medial meniscectomy [12]. Using a fiber reinforced model, considerable alterations in stress distributions were predicted for a partial meniscectomized joint [13]. Assuming linear elastic and isotropic behavior for cartilage and menisci, the maximal stress in articular cartilage in the meniscectomized knee was predicted about double of that in a healthy joint [14]. Partial meniscectomy on either side, combining with frontal plane knee alignment, increased the contact stresses and strains in the cartilage with greater increase in the lateral meniscectomy. In this latter case, isotropy and transversely isotropy were assumed for the cartilages and menisci, respectively [15].

Previous studies have been focused on the impact of partial meniscectomy on the contact stresses of the knee. Little information is known about changes in fluid pressurization in cartilages and menisci, although fluid flow and pressure are believed to play an important role in proper load transfer in the knee joint [16-18]. Possible changes in fluid pressurization in the cartilages and menisci following meniscectomy may not only alter the load distribution in the tissues, but also affect their nutrient transport and lubrication. Therefore, information about changes in fluid pressurization in articular cartilage following partial meniscectomy can be useful in better understanding of the tissue degeneration and onset of osteoarthritis. Considering the difficulties associated with the measurements of fluid pressure in cartilage and menisci, computational models have been increasingly used to investigate the fluid pressurization in the tissues in vitro [19-22]. Recently, an MRI derived knee joint model was proposed to account for the fluid pressurization in the cartilaginous tissues in situ [23]. The creep behavior of total meniscectomized knee joints (both menisci were removed) was compared to that of intact joints [24].

The objective of the present study was to investigate the impact of partial meniscectomy on the fluid pressurization in the knee cartilage. A 3D model of the knee joint including bones, cartilages, menisci and four major ligaments was used for the present study. A fibril-reinforced material model was used for the cartilages and menisci to simulate the poromechanical, nonlinear and anisotropic behavior of these tissues. To compare the responses of the intact and meniscectomized knees, three different loading protocols were considered: stress relaxation, simple creep and combined creep loadings.

2. Methods

2.1 Finite Element Modeling of the Intact and Meniscectomized Joints

The geometry of a tibio-femoral joint was taken from a right leg, which was previously reconstructed from magnetic resonance imaging of a 26-year-old healthy man [25]. The solid model included femur, tibia, fibula as well as their adjacent cartilages, medial and lateral menisci and four major ligaments (Fig. 1). The finite element mesh of the knee was generated using ABAQUS v6.8-2 (Simulia Inc., Providence, RI). Rigid elements were used to model the bones due to their higher stiffness compared to the other tissues. Only bone surfaces needed to be meshed using triangular elements, resulting in 80% reduction in element number as compared to the elastic element representation. A total of 37250 elements were used for the whole model discretization, of which 12829 elements were used to mesh the bones. Continuum hexahedral porous elements were used for the cartilages and menisci, and continuum solid elements were used to mesh the ligaments. Second order of displacement was considered for the femoral cartilage (20-node hexahedral elements) and linear displacement was considered for the tibial cartilages, menisci and ligaments (8-node hexahedral). The 20-node element usually gives better results for the stress. The use of lower order of elements for the tibial cartilages and menisci was to improve contact convergence. The contact approach formulated in ABAQUS experiences slow convergence when the contact surface is formed by 20-node elements.

Since the longitudinal tear is the most common type of meniscus lesion [26,27], this partial meniscectomy was considered in the present study. Variations of length and location of the meniscectomy were also taken into account to simulate clinical cases ranging from small to nearly the whole circumference of the meniscus [28]. Both medial and lateral meniscectomies were investigated in order to address the reported differences [29]. In total, six cases of partial longitudinal meniscectomies were simulated (Fig. 1): anterior lateral (AL), central lateral (CL), extended lateral (EL), posterior medial (PM), central medial (CM) and extended medial (EM) meniscectomies. In each case, the corresponding elements from the avascular region of the meniscus was removed circumferentially (Fig. 1).

2.2 Tissue Modeling and Material Properties

Cartilages and menisci were modeled as fully saturated porous media to account for the fluid pressurization in these tissues. A fibril-reinforced model was used for cartilages and menisci to consider the role of collagen fibers. For the femoral cartilage, the orientation of collagen fibers (type II) was assumed based on split-line pattern [30]. In the case of the menisci, the direction of the primary collagen fibers (type I) was oriented in circumferential direction. For the tibial cartilage, random orientation of fibers was assumed due to lack of data about the orientation. Anterior and posterior cruciate ligaments, as well as medial and lateral collateral ligaments, were modeled as fibril-reinforced solid material. The fiber direction was aligned in the longitudinal direction for the ligaments. The initial strains in ligaments were not considered in the present study since the reported values are normally beyond the small strain range [31].

Material properties of different tissues were obtained from the literature [32-35] and are summarized in Table 1. The non-fibrillar solid matrices of cartilages, menisci and ligaments were assumed as linearly elastic. Collagen fiber nonlinear properties were considered using quadratic stress-strain relationship simulating the mechanical properties in the toe region. The viscoelastic stresses of the fibrillar network were described in a previous study [36] and are included here for the convenience of reading. The stress is presented in terms of the hereditary integral:

$$\sigma_x^f(t) = \sigma_x^f(0) + \int_0^t G_x(t-\tau) E_x^f(\varepsilon_x) \dot{\varepsilon}_x d\tau \qquad (1)$$

where $\sigma_x^f(t)$ is the tensile stress in the x or primary fiber direction, $G_x(t)$ is the relaxation function, E_x^f is the fibrillar modulus and ε_x is the tensile strain. In the present study with small deformation assumption, the instantaneous fibrillar modulus was used:

$$G_x(t)E_x^f(\varepsilon_x) = E_x^0 + E_x^\varepsilon \varepsilon_x \tag{2}$$

where E_x^0 and E_x^ε are elastic constants in the primary fiber direction (Table 1). The same equations with different coefficients were used for the two perpendicular directions, y and z. Orthotropic fibrillar properties were also considered for the menisci (Table 1). The permeability in the cartilages and menisci was orthotropic with respect to the fiber direction. A user-defined FORTRAN subroutine was used to implement the fiber properties [36].

2.3 Loads, Boundary Conditions and Contact Interactions

Three loading protocols were considered for the current study: 1) *stress relaxation* with a ramp displacement of 0.2 mm; 2) *simple creep* of a ramp load of 300N, which was approximately half of the body weight of the subject; and 3) *combined creep* of 300N and internal torque of 500 N.mm. In all cases, the displacement or force was ramp compressive and applied in the proximal-distal direction with knee in full extension. All loadings were applied in one second at constant rate and then held constant up to 10,000 seconds. These loadings were limited to small magnitudes due to the use of small deformation theory.

Zero fluid pressure was set for the free surface of the femoral cartilage that was not in contact with its mating surface. Since small deformation was considered in this study, the prescribed pressure boundary condition was assuemd to be unchanged during loading. The femur was constrained in varus-valgus and flexion-extension rotations [37] but free in all other directions.

The tibia and fibula were completely fixed. These geometrical boundary conditions were used to avoid large deformation and large displacement.

The ends of both lateral and medial menisci were constrained to the tibial plateau using the TIE option in ABAQUS to approximate the meniscal horns-tibia attachments. The ends of the ligaments were fixed to the corresponding bones using the TIE option as well. To simulate the mechanical contacts between the cartilaginous tissues, six contact pairs were defined with three on the lateral and three on the medial sides: femoral cartilage-meniscus, femoral cartilage-tibial cartilage and meniscus-tibial cartilage. The surface to surface frictional hard contact based on penalty method was used for the simulations. The coefficient of friction between all tissues was set to 0.02 [38].

2.4 Solution Method

The *Soil Consolidation* procedure in ABAQUS/Standard was used to determine the fluid flow in the tissues. Newton method was used to solve the nonlinear equations. In each time increment, the convergence criterion of force, moment (if any) and volumetric flux was set to 0.5% of the overall time-averaged value of the spatial force, moment and flux, respectively. In the last iteration of each time increment, the largest correction of the nodal displacement, rotation (if any) and pore fluid pressure was set to be less than 1% of the largest increment of the nodal displacements, rotations and pore pressures, respectively. Satisfying all the above criteria, a time increment was considered convergent if the maximum increment of pore pressure in the last iteration was less than 0.003 MPa for the relaxation and 0.01 MPa for the creep loadings.

3. Results

3.1 Stress Relaxation

For the case of the intact knee, high fluid pressure regions were observed in both medial and lateral condyles with slightly higher pressure in the medial condyle (Fig. 2a, t = 10s). As the relaxation developed further, the fluid pressure decreased monotonically: the maximum pressure was reduced by 18%, 43%, 88% and 93% at t = 10s, 100s, 2000s and 4000s, respectively. The maximum pressure remained higher in the medial than in the lateral condyle (Figs. 2a and 3a). The fluid pressure gradient in each condyle was rather small as compared to any meniscectomy case (Fig. 2a vs. b-g).

With the anterior lateral (AL) meniscectomy, the fluid pressure increased substantially in the lateral condyle as compared to the case of intact knee (Fig. 2b vs. 2a, Fig. 3b vs. 3a and Table 2). The high-pressure centered in the lateral condyle, in the region that was in direct contact with the tibial cartilage (Fig. 2b). The increase in the maximum pressure due to meniscectomy was about 35%, 43%, 98% and 116% at t = 10, 100s, 2000s and 4000s, respectively (Fig. 4a). During the entire relaxation, the lateral condyle was more pressurized than the medial condyle (Figs. 2b and 3b and Table 2). The surface maximum pressure in the lateral condyle at t = 400s was about 1.8 times of that in the medial condyle (Table 2, AL).

Similarly, for the case of the central lateral (CL) meniscectomy, the lateral condyle, which was supported by the injured meniscus, encountered greater pressure than the medial condyle (Figs. 2c and 3c and Table 2). For instance, at t= 400s, the maximum pressure in the lateral condyle was about 1.94 times of that in the medial condyle (Table 2, CL). Approximately 42%, 65%, 88% and 74% of increase in the peak fluid pressure were observed, respectively, at t = 10s, 100s, 2000s and 4000s, as compared to the intact joint (Fig. 4a).

For the case of the extended lateral (EL) meniscectomy, the increase in the maximum pressure was further greater: approximately 49%, 78%, 162% and 184% higher than that in the intact joint at *t*= 10s, 100s, 2000s and 4000s, respectively (Fig. 2d vs. 2a, Fig. 3d vs. 3a and Fig. 4a). The high-pressure region was extended in the central region of the lateral condyle during early relaxation (Figs. 2d and 3d). The maximum pressure in the lateral condyle was remarkably higher than that in the medial condyle (about 2.12 times higher at *t* = 400s, Table 2).

With the medial meniscectomies (PM, CM and EM), the higher pressure region was observed in the medial condyle, which remained true during relaxation (Figs. 2e-g and 3e-g). Considerable increase in the fluid pressure was observed for all PM, CM and EM cases (Fig. 4b, and Table 2), with the highest increase for the EM case.

3.2 Simple Creep

For the intact knee, the maximum fluid pressure in the femoral cartilage occurred in the medial condyle (Fig. 5a and Table 2), which was similar to the case of stress relaxation. Again the pressure gradient was rather small. As creep developed, the fluid pressure decreased and the high-pressure regions stabilized at the anterior sites of the medial and lateral condyles (Fig. 5a, 2000s). The decrease in the maximum fluid pressure was about 13%, 16%, 48% and 62% of the maxima at *t*= 10s, 100s, 2000s and 4000s, respectively.

For the lateral meniscectomies (AL, CL and EL), the pressure in the femoral cartilage increased remarkably with respect to the intact knee, and the high-pressure region shifted to the lateral side, i.e. the side with the injured meniscus (Figs. 5b and 6a and Table 2). In general, the highest increase in pressure was observed in the EL case (Fig. 6a and Table 2). Moreover, a very low rate of decay in the fluid pressure was observed in the AL and EL cases (Fig. 6a). As creep developed, the high-pressure region stabilized in the part of femoral cartilage that coincided with the meniscectomy site (e.g. Fig. 5b, 2000s).

The fluid pressure was observed to increase with all medial meniscectomies with the higher-pressure in the medial condyle (Figs. 5c vs. 5a), again at the side with injured meniscus. At t = 400s, the maximum pressure in the medial condyle was about 2.2, 3.1 and 3.2 times of that in the lateral condyle for the PM, CM and EM cases, respectively (Table 2).

3.3 Combined Creep Force and Torque

For the intact knee, the maximum fluid pressure occurred in the lateral condyle (Fig. 7a). The high-pressure regions moved towards the anterior side as creep developed (Fig. 7a, 2000s vs. 10s). The maximum pressure decreased monotonically with creep: the decrease was about 12%, 27%, 55% and 67% of the maxima at t = 10s, 100s, 2000s and 4000s, respectively.

For all lateral meniscectomies (AL, CL and EL), the high-pressure region was observed at the central part of the lateral condyle (Fig. 7b, shown for the EL case only). In the case of the AL meniscectomy, the high-pressure region extended towards anterior-lateral corner as creep developed (not shown). For the CL and EL meniscectomies, the high-pressure region extended along anterior-posterior direction (Fig. 7b, shown for the EL case only). Similar to the two previous loading conditions, the EL meniscectomy experienced the highest increase in the fluid pressure among three lateral meniscectomies (Fig. 8a). A very low rate of pressure dissipation was also observed for the EL and CL cases (Fig. 8a).

For the PM and CM meniscectomies, the high-pressure region was observed in the medial condyle (not shown). For the EM meniscectomy, the maximum pressure occurred in the central region of the medial condyle (Fig. 7c, 10s). The location of high-pressure region changed little during creep (Fig. 7c, 2000s vs. 10s). The pressure decayed very slowly in this case (Fig. 8b).

The maximum pressure was decreased by 6%, 35% and 54%, at t =500s, 2000s and 4000s, respectively. The pressure was still high at 10000s (Fig. 8b).

4. Discussion

The objective of this study was to investigate the impact of partial meniscectomy on the fluid pressurization in the articular cartilage. An MRI derived model of the knee joint including site-specific nonlinear fibril reinforcement in the cartilages and menisci [23, 24] was employed for this purpose. The fluid pressurization in the femoral cartilage was analyzed for six cases of longitudinal meniscectomies, each under three relaxation or creep load conditions. Using the intact knee as reference, the major findings on the consequence of the partial meniscectomy were: 1) substantial increase in the fluid pressure, 2) shift of high-pressure region within a condyle and/or between the lateral and medial condyles, and 3) positive correlation between the increase in the fluid pressure and the size of meniscal resection.

Remarkable increase in the fluid pressure level was observed in all six meniscectomy cases as compared to the intact joint (Table 2 and Figs. 4, 6 and 8). We focused on presenting the fluid pressure results to complement the literature. Obviously, the overall contact area between the articular surfaces decreased and the cartilage-cartilage contact area slightly increased because of the meniscal resection (not shown). This was in agreement with the published experimental or numerical data on the contact areas [12, 39, 40].

The fluid pressure in the femoral cartilage of the intact joint was more evenly distributed in the two condyles. There was also smaller difference between the maximum pressures in the medial and lateral condyles, as compared to the meniscectomy cases (Figs. 2 and 5). The maximum pressure was observed in the medial condyle of the intact knee under the loading conditions of stress relaxation and simple creep (Table 2). This result was in agreement with the experimental data that showed higher stresses in the medial compartment of the knee [41-43]. For all results presented here, the femur was not constrained in the internal rotation about its vertical axis but was constrained against varus-valgus and flexion-extension rotations. The sizes of high-pressure regions in medial and lateral condyles of the intact knee were more or less comparable (e.g. Fig. 5a). We also tested the case of no internal rotation for the intact knee under simple creep loading, and found an obviously larger high-pressure region in the medial compartment than the lateral compartment (not shown). This finding was consistent with the published data about the effect of internal rotation on the contact pressure [37].

The maximum fluid pressure always occurred in the condyle that was in contact with the resected meniscus in all meniscectomies (Table 2 and Figs. 2, 5 and 7). This result indicated that a lateral meniscectomy shifted the maximum pressure from the medial to the lateral condyle, because the maximum pressure occurred in the medial condyle in the intact knee. The pressure difference between the lateral and medial condyles was much greater in the meniscectomized knee than that in the intact knee (Figs. 2, 3, 5 and 7). Using the EM meniscectomy as an example, when the knee was subjected to the simple creep loading, the peak pressure in the superficial layer of the medial condyle was about 3.2 times of that in the lateral condyle (Table 2). Such a great contrast in the results for the two condyles was not reported in the literature when elastic models were used. However, this result may depend on the loading condition and knee flexion. In the present study, the joint was constrained in flexion-extension and varus-valgus rotations. A study on the sensitivity of the results to loading and boundary conditions may further shed light on the subject.

The high-pressure region also shifted within the condyle in all meniscectomies. In general, the high-pressure region shifted to the central part of the condyle, in the region that was in direct contact with the tibial cartilage (Figs. 2, 3, 5 and 7). Similar results in contact stresses have been reported [34, 37, 39]. We further modeled creep and relaxation. Our results also indicated the shift of high-pressure region with time. Noting that the meniscectomy could also lower the rate of the fluid pressure disipation, the relative increase in the fluid pressure due to meniscectomy could be extremely high in some regions at late stage of creep/relaxation. With the simple creep loading, the increase was more than 500% at t = 2000s in the center of the lateral condyle for the EL meniscectomy (Fig. 5b vs. 5a); it was about 670% at t = 2000s in the center of the medial condyle for the EM meniscectomy (Fig. 5c vs. 5a).

The increase in the fluid pressure and shift of the high-pressure regions were observed to escalate with the size of meniscal resection. As compared to the other cases, the pressure increase was more remarkable in the EL and EM meniscectomies; the dissipation of the fluid pressure was also much slower in the two cases (Figs. 4, 6 and 8). Similar results in the contact forces/stresses were reported in the literature [12,15]. In addition, our study showed considerably altered pressure distribution in the extended meniscectomies (Figs. 5 and 7 and Table 2). Consequently, the fluid pressure can substantially increase in the central regions of *both* condyles after an extended meniscectomy on *one* side. In other words, the extended medial meniscectomy not only increased the fluid pressure in the medial condyle remarkably, but could also increase the pressure in certain regions of the lateral condyle at a smaller scale. Published results obtained from elastic modeling indicated minimal effect on the intact meniscus and no change in the location of peak contact pressure [12]. This discrepancy may indicate the change of knee contact mechanics by the fluid flow in the tissues, which was not considered in the literature.

Much higher increase in stress was predicted in the lateral than medial meniscectomy when fluid pressure was not considered [44]. The present results did not support the reported prediction. It is not currently clear whether this discrepancy was produced by fluid pressure considered here or large deformation considered in the reference [44]. Moreover, the discrepancy could have partially been produced by different rotational constraints used: in the present study the knee was constrained in varus-valgus rotation while in the reference [44] the knee is free in this rotation. Controversial results were also seen in the literature regarding the lateral vs. medial meniscectomies. For instance, based on a 10-year follow up performed by the French Arthroscopic Society, no significant difference between medial and lateral meniscectomies was found [45], while some other studies reported higher risks for medial [46] or lateral meniscectomies [29, 44]. We will further investigate this issue when large deformation is incorporated in our modeling.

While the loading condition was an important factor in the fluid pressurization in the knee joint, some qualitative findings were the same for the three loading conditions investigated: the location of the maximum pressure was determined by the site of meniscal resection; the extended meniscectomy caused the highest pressure increase; the high pressures occurred in the tibial-femoral cartilage contact regions (as discussed earlier). In contrast with other cases, the CM meniscectomy under creep loading demonstrated the highest increase in peak pressure at early times (Fig. 5b). However, the pressure in the CM case then decayed faster than the EM case, and became lower as creep developed further (Fig. 6b). For the intact knee, the medial condyle was more pressurized in compression during both creep and relaxation, but the maximum pressure was shifted to the lateral condyle by additional torsion (Table 2). The major difference between creep and relaxation was much slower pressure decay in creep than in relaxation (Figs. 4 vs. 6). This difference was previously confirmed in experiments for the tissue explants [21].

Several limitations existed in this study. The small deformation theory was used, which helped us investigate the nonlinear creep and relaxation of the knee contact mechanics without experiencing too much numerical difficulties. For the intact knee under combined creep loading, the compressive strain in the contact region was generally around 1% and the highly localized maximum strain was less than 5%. In addition, only knee in full extension was considered in the current study. The fluid pressurization in the knee could be different in different flexion angles. We will study the poromechanics of the knee joint with physiological loadings and large deformation after we have understood the mechanics with standard analytical or simple experimental loadings. We will also confirm whether the results presented here are subject-specific.

In summary, we have investigated the fluid pressurization in the femoral cartilage within the knee for the combinations of seven meniscal states (intact + 6 meniscectomies) and three loading conditions. Partial meniscectomies were observed to change the fluid pressurization in the articular cartilage in a complex fashion, including the increase in the maximum fluid pressure, the case and time-dependent shift of high-pressure regions.

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Table 1. Material properties of the non-fibrillar solid matrix and the fibrillar matrix of soft tissues (Modulus: MPa; Permeability: mm^4/Ns).

	Non-fibrillar solid matrix				Fibrillar moduli, Eq. (2)			
	Elastic modulus	Poisson's ratio	Permeability		Primary fiber direction: *x*		Perpendicular: *y* or *z*	
			Primary fiber direction: *x*	Perpendicular: *y* or *z*	E^0	E^f	E^0	E^f
Femoral cartilage	0.26	0.36	0.002	0.001	3	1600	0.9	480
Tibial cartilage	0.26	0.36	0.002	0.001	2	1000	2	1000
Meniscus	0.50	0.36	0.002	0.001	28	0	5	0
Ligament	1.0	0.30	N/A		10	14000	0	0

Table 2. Maximum fluid pressure (MPa) at 400s in the surface layer of each condyle. The greater in the two condyles is identified in bold for each of the 21 cases (7 meniscal states × 3 loading conditions).

	Medial condyle			Lateral condyle		
	Relaxation	Simple creep	Combined loading	Relaxation	Simple creep	Combined loading
Intact	**0.089**	**0.47**	0.43	0.075	0.38	0.43
AL	0.090	0.44	0.38	**0.16**	**0.74**	**0.82**
CL	0.095	0.44	0.39	**0.18**	**0.85**	**0.93**
EL	0.10	0.40	0.35	**0.22**	**0.85**	**0.96**
PM	**0.16**	**0.79**	**0.85**	0.075	0.36	0.39
CM	**0.21**	**1.02**	**1.01**	0.073	0.33	0.38
EM	**0.24**	**0.94**	**1.02**	0.077	0.29	0.35

FIGURE CAPTIONS

Figure 1. Finite element model of the right knee developed for different longitudinal meniscectomies: a) anterior lateral (AL), b) central lateral (CL), c) extended lateral (EL), d) posterior medial (PM), e) central medial (CM) and f) extended medial (EM). Shaded elements were removed in each case to simulate the corresponding meniscectomy.

Figure 2. Fluid pressure (MPa) in the deep layer of the femoral cartilage at t=10s of the stress relaxation loading. (a) Intact knee, (b) AL, (c) CL, (d) EL, (e) PM, (f) CM, and (g) EM meniscectomized knees. The fluid pressure was obtained for the centroids of the elements located at the 3/4 depth from the articular surface. The dashed lines indicate the position of the (remaining) menisci if they were attached.

Figure 3. Fluid pressure (MPa) in the deep layer of the femoral cartilage at t=2000s of the stress relaxation loading. (a) Intact knee, (b) AL, (c) CL, (d) EL, (e) PM, (f) CM, and (g) EM meniscectomized knees. The fluid pressure was obtained for the centroids of the elements located at the 3/4 depth from the articular surface. The dashed lines indicate the position of the (remaining) menisci if they were attached.

Figure 4. Normalized maximum fluid pressure in the femoral cartilage of the knee when subjected to stress relaxation. (a) Lateral meniscectomies (AL, CL and EL), (b) medial meniscectomies (PM, CM and EM). The results were normalized with respect to the maximum pressure of the intact knee at 10s. The fluid pressure was obtained for the layer of the femoral cartilage located at the 3/4 depth from the articular surface.

Figure 5. Fluid pressure (MPa) in the deep layer of the femoral cartilage of the knee when subjected to simple creep loading. The left column is the results at t= 10s and the right column corresponds to t= 2000s. (a) Intact knee, (b) EL and (c) EM meniscectomized knees. The fluid

pressure was obtained for the centroids of the elements located at the 3/4 depth from the articular surface. The dashed lines indicate the position of the (remaining) menisci if they were attached.

Figure 6. Normalized maximum fluid pressure in the femoral cartilage of the knee when subjected to simple creep loading. (a) Lateral meniscectomies (AL, CL and EL), (b) medial meniscectomies (PM, CM and EM). The results were normalized with respect to the maximum pressure of the intact knee at 10s. The fluid pressure was obtained for the layer of the femoral cartilage located at the 3/4 depth from the articular surface.

Figure 7. Fluid pressure (MPa) in the deep layer of the femoral cartilage of the knee when subjected to combined creep loading. The left column is the results at t= 10s and the right column corresponds to t= 2000s. (a) Intact knee, (b) EL and (c) EM meniscectomized knees. The fluid pressure was obtained for the centroids of the elements located at the 3/4 depth from the articular surface. The dashed lines indicate the position of the (remaining) menisci if they were attached.

Figure 8. Normalized maximum fluid pressure in the femoral cartilage of the knee when subjected to combined creep loading. (a) Lateral meniscectomies (AL, CL and EL), (b) medial meniscectomies (PM, CM and EM). The results were normalized with respect to the maximum pressure of the intact knee at 10s. The fluid pressure was obtained for the layer of the femoral cartilage located at the 3/4 depth from the articular surface.