Changes in Knee Joint Mechanics after Medial Meniscectomy Determined with a Poromechanical Model

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ABSTRACT

The menisci play a vital role in the mechanical function of knee joint. Unfortunately, meniscal tears often occur. Meniscectomy is a surgical treatment for meniscal tears, however, mechanical changes in the knee joint after meniscectomy is a risk factor to osteoarthritis. The objective of this study was to investigate the altered cartilage mechanics of different medial meniscectomies using a poromechanical model of the knee joint. The cartilaginous tissues were modeled as nonlinear fibril-reinforced porous materials with full saturation. The ligaments were considered as anisotropic hyperelastic and reinforced by a fibrillar collagen network. A compressive creep load of ¾ body weight was applied in full extension of the right knee during 200 seconds standing. Four finite element models were developed to simulate different meniscectomies of the joint using the intact model as the reference for comparison. The modeling results showed a higher load support in the lateral than medial compartment in the intact joint, and the difference in the load share between the compartments was augmented with medial meniscectomy. Similarly, the contact and fluid pressures were higher in the lateral compartment. On the other hand, the medial meniscus in the normal joint experienced more loading than the lateral one. Furthermore, the contact pressure distribution changed with creep, resulting in a load transfer between cartilage and meniscus within each compartment while the total load born by the compartment remained unchanged. The present study has quantified the altered contact mechanics on the type and size of

meniscectomies, which may be used to understand meniscal tear or support surgical decisions.

Keywords: Finite element analysis; Knee joint mechanics; Creep; Fluid pressure; Meniscectomy; Meniscal lesion

1. Introduction

The meniscal tears are approximate 30% of possible tears in the knee joint and complex tears require medical attention [1]. While meniscectomy is a surgical treatment for complex meniscal tears, it is also a risk factor to osteoarthritis (OA) through changes in the contact mechanics of menisci and articular cartilages [2-7]. Before 1950, surgeons preferred to perform total meniscectomy for all meniscal tears, but later, began to perform partial meniscectomy to preserve as much meniscus as possible [8-11]. This is due to the fact that total meniscectomies caused more damage than partial meniscectomies [12-14] and long-term follow-up studies have shown a high rate of reoperations after total meniscectomies [15]. Especially, when the tear occurs in the avascular zone that includes less blood supply than in other zone of menisci, partial meniscectomy becomes standard treatment [16]. Both lateral and medial meniscectomies have become conventional. However, medial meniscectomies are more common than lateral meniscectomies due to higher occurrence of injuries in medial menisci associated with the anatomy [2, 17-20].

Computational modeling can provide vital information for the prediction of disease progression and of the potential for therapeutic interventions [21]. Finite element methods (FEM) have been used to investigate contact behavior of the knee joint to understand meniscectomy induced OA [2, 3, 5, 7, 22]. The FEM may be conveniently used to simulate various meniscal resections while the alteration in an experimental setup may be difficult. Although the FEM is advantageous in this regard, achieving reliable results requires a realistic numerical model. A realistic finite element model of knee joint depends on its numerical procedure, boundary conditions, and material modeling [22-25].

The single-phase elastic, i.e. no fluid, material model has been extensively used to investigate the effect of meniscectomy on articular cartilage mechanics. The effect of location and extent of the medial meniscectomy on tibial cartilage was investigated using elastic finite element models [26]. The medial partial meniscectomies with varying degree of resections were previously modeled with linearly elastic and transversely isotropic tissue properties, and it was reported that as the size of meniscus resection increased, the joint contact pressure increased [2, 27]. The effects of medial meniscectomy on OA were investigated undergoing suitable boundary conditions using single-phase models [17]. The contact pressure distributions on cartilage following partial meniscectomy were also examined with the fiber reinforced model [28]. Considering the articular cartilage and menisci as isotropy and transversely isotropy, respectively, it was revealed that knee alignment and force distribution played a significant role in the altered joint mechanics after both total and partial meniscectomy [29]. In all these studies, the fluid pressurization in the soft tissues was ignored, assuming instantaneous load response of the knee joint. Because articular cartilages and menisci have shown substantial poromechanical behavior, the fluid pressure and flow may play an important role in the mechanical behavior at the joint level [3, 5, 30-32]. It may be necessary to consider the fluid pressurization induced changes in the load distribution and lubrication in the joint to understand the consequence of meniscectomy [3, 33].

In a previous study of six partial meniscectomies at various locations, the impact of fluid pressure on femoral cartilage was investigated using a poromechanical model [3]. The intact and meniscectomized knee FE models were compared under a simple creep load, taking into account the fluid pressurization [5]. In another study, an axisymmetric biphasic model was used to understand whether articular cartilage damage or subchondral fracture more likely caused the onset of OA [34]. The role of collagen network in the arthritic and repaired cartilage was also examined through an axisymmetric fibril-reinforced model [35]. Small deformation was assumed in these studies due to difficulties of numerical convergence when taking the fluid pressurization into account. However, the knee joint subjects to large deformation by the high ground reaction force arising from body weight (BW) in daily activities. For example, the reaction force was found to be, respectively, 346%, 316%, 259%, 253%, 246%, 225% and 107% of the body weight in stair descending, stair ascending, one legged stance, knee bending, standing up, sitting down and two legged stance [36]. The contact behaviors in the healthy and meniscectomized tibiofemoral joints were explored through large deformation using single phase elastic models only [37]. The impact of meniscectomy on articular cartilages of knee joint was evaluated in large deformation with only short-term loadings, while the menisci were modeled as transversely isotropic and elastic, the articular cartilages as fibril-reinforced viscoelastic [38, 39].

Few studies on meniscectomy have focused on the fluid-pressure induced mechanical response of the knee joint undergoing large deformation in the soft tissues. Therefore, the objective of the present study was to investigate the impact of medial meniscectomy on the poromechanics of the knee joint with large deformation in the cartilaginous tissues. This may help understand cartilage homeostasis and OA onset.

2. Methods

2.1 Meniscectomy model and mesh

The joint model was previously reconstructed from the right knee joint of a female participant with normal leg alignment and no history of leg injury. The 24-year-old 59 kg participant was in supine position with the knee joints at complete extension for a 3T MRI scan [40]. The 3D model consisted of bones: femur, tibia, fibula, and soft tissues: femoral and tibial cartilages, menisci, four major ligaments, i.e. anterior cruciate (ACL), posterior cruciate (PCL), medial collateral (MCL), and lateral collateral (LCL) ligaments (Fig. 1). The finite element mesh of the knee joint was generated using IA-FEMesh (University of Iowa, Iowa City, IA). The bones were considered as rigid because they have much higher stiffness than the cartilaginous tissues. The mesh consisted of quadrilateral elements (R3D4 in ABAQUS, Simulia, France) for bony structures, hexahedral porous elements (C3D8P) for cartilages and menisci, and continuum solid elements (C3D8) for ligaments. First order linear pure hexahedral elements were especially preferred for the cartilaginous tissues for faster convergence than tetrahedral elements for contact analysis when fluid pressurization is considered [41]. A total of 32579 elements were used for the whole intact model. The femoral and tibial cartilages were meshed with four layers of elements through the tissue thickness while the menisci were discretized using five layers of elements in the thickness direction. This mesh produced not only convergent results but also smooth depth variations of stresses and fluid pressures according to the mesh sensitivity analysis performed in the previous studies with the same constitutive model for cartilages and menisci as used in the present study [42].

Only medial meniscectomies were explored in this study, because medial meniscal lesion is more common than the lateral one [17-20]. Four clinical cases were simulated and compared with the case of intact knee (Fig. 1): Partial25, Partial50, Sub-total and Total medial meniscectomies, which included the size and location variations of the meniscal resections. Partial25 and Partial50 modeled, respectively, approximate 25% and 50% of the resections in the longitudinal direction after longitudinal tears that are the most frequent type of meniscal lesions [43-45]. Sub-total modeled the removal of 50% of the anterior medial meniscus. The lateral meniscus was considered intact in all cases.

2.2 Soft tissue properties

The fibril-reinforced constitutive model previously developed was employed in the present study for modeling soft tissues [46], where the solid-fluid interaction was modeled by Darcy's law. Articular cartilages and menisci were modeled as nonlinear fibril-reinforced and fully saturated porous media. Nonlinear, anisotropic fibril-reinforced solid modeling was employed for the ligaments because fluid pressurization may be neglected in tensile mechanical behavior of ligaments. It was reported that the collagen fibers play an important role in fluid pressurization in cartilages [33]. The orientations of collagen fibers were incorporated using measured split-line pattern for femoral cartilage [47], while the fiber orientation for tibial cartilage was assumed to align in the local *x*-axis direction (coordinate system at the element level) due to data unavailability. The primary fiber orientation was assumed circumferential for the menisci. Finally, for the ligaments, primary collagen fiber direction was aligned in the longitudinal direction [33].

Material properties of tissues (Table 1) were obtained from the literature [5, 48-53]. The non-fibrillar solid matrices of all soft tissues were modeled as isotropic, hyperelastic and defined with the Neo-Hookean hyperelasticity. For the collagen network, quasi-linear viscoelasticity was assumed. The Young's modulus of the fibrillar matrix was considered orthotropic (Table 1). Due to the fact that large deformation theory was used in the present study, the initial strains were considered for the ligaments. These values were obtained from previous studies: 2% in MCL and LCL, 2.5% in ACL, and nil in PCL [37, 52, 54]. These strain values are converted to stress values as initial conditions [54]. The knee joint model was previously numerically tested for large deformation [55]. A user defined FORTRAN subroutine in ABAQUS was used to implement the material model previously developed [55].

2.3 Load and boundary conditions

A compressive creep load of 390N, which was nearly ¾ body weight of the female participant, was used in the poromechanical finite element modelling. The 390N force was maximum ground reaction force obtained during the dual fluoroscopy measurement of the same participant under a creep protocol in our previous study [40]. The moderate force was applied on the knee, but large deformation theory was used to evaluate long-term response of the soft tissues in the knee joint. The load was ramped in one second and held constant up to 600sec in the measurement but 200sec in the present study. This was due to our intention to focus on the effect of the meniscectomy under simple creep and compare the modeling result of the present study with experimental results [56-61] in the literature. The load was applied on the femur in the proximal-distal direction. The femur was unconstrained in all translations but fixed in all rotations. Tibia and fibula were constrained in all degrees of freedom. These boundary conditions were chosen to be as close as possible to that of the experimental studies used for comparison [56-61].

Femoral and tibial cartilages were attached to the femur and tibia, respectively, using the TIE contact, which means the two issues experience no relative motion between the contact surfaces. The ends of the ligaments were constrained to the corresponding bones using TIE contacts as well. The meniscal horns were tied to the tibial plateau to simulate the meniscal horns-tibia attachments. Six contact pairs were defined in the finite element model to simulate the behavior of mechanical contacts between the cartilaginous tissues. Three were defined for each of the medial and lateral sides: femoral cartilage-tibial cartilage, menisci-femoral cartilage, and menisci-tibial cartilage. The six contact pairs reduced to four pairs in the case of total medial meniscectomy.

The pore pressure was set to zero for any uncontact/free surface to allow fluid exudation from the cartilaginous tissues. Since large deformation was considered in this study, geometrical nonlinearities were included in the finite element analysis using NLGEOM option in ABAQUS.

2.4 Numerical modeling and solution

The finite sliding option was used for each contact pair to consider the effect of large deformation. The nonlinear surface-to-surface hard contact option in ABAQUS was selected for contact modelling. For each contact pair, one surface was defined as the master surface and the other one as the slave surface [41]. The linear penalty method was used to avoid slave surface penetration into the master surface, and the contact pressure was calculated at each surface node. The frictional coefficient between the cartilaginous tissues was set to 0.2 [62]. The soil consolidation method in ABAQUS Standard was used to implement Darcy's law that correlates the fluid pressure gradients to the fluid velocity relative to the solid, where the fluid pressure is considered as a nodal variable in the FE formulation [41]. The maximum pore pressure increment was set to 0.05MPa in each numerical iteration. Twelve parallel core processors and up to 370 GB of RAM were used to run the ABAQUS simulation (University of Calgary).

3. Results

The maximum contact pressure decreased with creep in the *lateral tibial plateau* for all cases (Fig. 2), although the average contact pressure was virtually constant during creep for each plateau due to little change in the contact area while total force remained unchanged: the average contact pressures during creep were 0.97 and 0.51 MPa, respectively, for the lateral and medial tibial cartilages of the intact joint. The corresponding contact center did not shift with creep due to meniscectomies. On the *medial plateau*, however, the variations of contact pressure and contact center depended on the type of meniscectomy (Fig. 3): while the maximum contact pressure increased with creep for the intact knee and longitudinal meniscectomies (partial25 and partial50), it decreased for sub-total meniscectomized knee; the contact center slightly shifted within the medial side, as creep developed, for the intact knee and longitudinal meniscectomies, but did not shift for the sub-total and total medial meniscectomies (figure for total meniscectomy is not included).

The maximum contact pressure in all cartilages at 1 sec was increased with the size of meniscal resection, except for sub-total meniscectomy (Fig. 4). For the sub-total meniscectomy, the contact pressure was decreased by 10% and 8%, respectively, in the lateral and medial cartilages, but increased by 43% in the femoral cartilage, as compared to the intact knee (Fig. 4); the highest contact pressure, 3.22 MPa, occurred in the medial condyle. While the contact pressures in the femoral cartilage were distributed very differently across the condyles prior to creep (Fig. 5a), it distributed more equally between the condyles with creep time (Fig. 5b).

At the maximum compressive force 390N (1sec), the vertical loads born by the lateral and medial tibial cartilages are shown in Table 2. The lateral side was subjected to 54%, 77%, 118%, 81%, and 142% higher total load than the medial side, respectively, in the intact, partial25, partial50, sub-total and total meniscectomized joints. For the lateral side, the total load was increased by 6%, 13%, 6% and 17%, respectively, with the partial25, partial50, sub-total and total meniscectomies, as compared to that in the intact knee. For the medial side, the total load was decreased by 8%, 20%, 10% and 26%, respectively, with the partial25, partial50, sub-total and total meniscectomies.

The fluid pressure in articular cartilage was generally consistent with the contact pressure variations. As expected, the maximum fluid pressure decreased rather fast during creep (Fig. 6). The maximum fluid pressures in the lateral cartilage were altered more with medial meniscectomy (Fig. 6). The maximum fluid pressure (at 1sec) in the lateral tibial cartilage was increased by 5%, 12%, and 16%, respectively, with the partial25, partial50, and total meniscectomies, but decreased by 10% for the sub-total meniscectomy. Accordingly, the maximum fluid pressure in the medial tibial cartilage was decreased by 17%, 14%, 9%, and 13%, respectively, with the partial25, partial50, sub-total, and total meniscectomies (Fig. 7). The highest fluid pressure occurred in the lateral tibial cartilage for all cases except for the sub-total meniscectomy; it was 3.04 MPa in the case of total meniscectomy (Fig. 7). As creep developed, the location of the maximum fluid pressure in articular cartilages shifted to the central region and deeper layer for all cases (not shown).

At the peak fluid pressurization (at 1sec), the maximum fluid pressure in the femoral cartilage was increased by 5%, 10%, 27%, and 14%, respectively, with the partial25, partial50, sub-total, and total meniscectomies as compared to that in the intact knee (Fig. 8). While the locations of both maximum fluid and contact pressures did not change with the longitudinal and total meniscectomy, they shifted from the lateral to medial condyle after sub-total meniscectomy.

The directions of principal stresses in the cartilaginous tissues were aligned approximately in the fiber directions assigned in the FE model (not shown). The first principal stress in a location of the medial tibial cartilage was initially similar in the intact and total medial meniscectomized knee joints. However, it became substantially different with increased loading and creep; the stress was reduced considerably slower in the total meniscectomized knee than in the intact knee (Fig. 9).

The portion of load born by the lateral compartment varied with medial meniscectomy, but remained constant during creep (Fig. 10a), indicating no load transfer between two compartments during creep of standing stance. Over 50% knee compression developed during creep (Fig. 10b, equilibrium not reached at 200s).

4. Discussion

The objective of the present study was to investigate the changes in articular cartilage mechanics following medial meniscectomy using large deformation theory. The cartilaginous tissues were modeled as poromechanical to consider the effect of fluid pressurization on the joint mechanics. Four models were developed to mimic various resections of medial meniscus (Fig. 1). The intact model was used as the reference model. The main findings from the present study were: 1) the peak contact and fluid pressures were approximately two times that obtained previously in similar studies; 2) the fluid pressure dissipation was faster than the previously predicted for creep loading; 3) the contact pressure in the lateral side of the intact knee joint was higher than that in the medial side and this difference was augmented with medial meniscectomy; the contact pressure redistributed within each compartment during creep.

Our computational results for the intact knee compare reasonably well with those from the literature (Tables 3 & 4). When 390 N was applied in 1 sec, the displacement of the femur in the proximal-distal direction was 0.62 mm, which is compatible with the experimental results obtained from the cadaver joints or independent FE results (Table 4) [56-58, 63]. The maximum contact pressure in the intact model occurred in the lateral tibial cartilage, which was consistent with the experimental results (Table 3). In general, our predicted contact area shows similar tendency with the experimental results when considering the difference in loading magnitudes (Table 3). Moreover, Table 2 also indicates good numerical results, because the total contact force in the joint, as calculated from contact pressures and areas, was consistent with the applied force, 390N.

Both the size and type of meniscectomy may adversely affect the joint contact mechanics. Sub-total meniscectomy exhibited the largest alteration in the contact pressure, while the contact pressure was generally increased with the size of meniscal resection in all other meniscectomies considered in the present study. The fluid pressure center was shifted from the lateral condyle to the medial condyle (Fig. 8), and the vertical displacement of the joint was surprisingly the smallest (Fig. 10b) for the sub-total medial meniscectomy, possibly due to uneven tissue compression after the meniscus lost its crescent shape. The first principal stress in the medial tibial plateau was also substantially altered with Total meniscectomy (Fig. 9). These dramatic changes in Sub-total and Total meniscectomies may cause new tears in the tissues. Our results support the surgical decision to keep the crescent shape of the medial meniscus in partial meniscectomies whenever possible [64, 65]. Furthermore, we have provided additional information on the fluid pressure changes following meniscectomy, which can be correlated to the altered tensile loading in the collagen network that potentially leads to cartilage degeneration. Most published meniscectomy models used single-phase material laws and thus the contact mechanics associated with the fluid pressurization was not investigated [2, 17, 26-28].

The lateral compartment supported more loading than the medial one in the intact joint, and this difference in load share between the two compartments was augmented with medial meniscectomy (Fig. 10, Table 2). In other words, the load increased in the lateral but decreased in the medial compartment with medial meniscectomy, because additional loading was transferred from the medial side to the lateral side due to meniscal loss in the medial side. This, together with a smaller contract area on the lateral side, explains why the contact pressure in the lateral side was substantially higher than that in the medial side (Fig. 4). Although the lateral compartment supported more total loading, the meniscus to cartilage contact in the medial side experienced 48% more force than that in the lateral side in the intact knee (Table 2). This may explain why the medial meniscus lesion is more common than lateral meniscus lesion [17-20].

The contact pressure redistributed with creep within the lateral or medial compartment (Figs. 2 & 3) while there was no load transfer from one compartment to another during creep of standing stance (Fig. 10), indicating the necessity to model the time-dependent contact pressure greatly influenced by fluid pressures in the cartilaginous tissues. The maximum contact pressure was initially higher in the lateral than medial cartilage, but was decreased with creep in the lateral cartilage (Fig. 2 vs Fig. 3) that was supported by the intact meniscus. This was consistent with the literature [3, 5, 55] because, as creep develops, more loading is transferred from articular cartilage to menisci. Therefore, the loading in the lateral tibial cartilage decreased (Fig. 2) as it increased in the intact meniscus. However, the scenario was rather different in the medial compartment, especially for the longitudinal meniscectomies. In parallel with the increased contact pressure in the medial meniscus, the contact pressure in the medial tibial cartilage increased with creep as well (Fig. 3a, b, and c). This result may have been partially produced by the 3D translation of the femur: the displacements in the proximal-distal, lateral-medial, and posterior-anterior directions reached, respectively, 0.62mm, 0.86mm, and 1.18mm under the 390 N compressive force applied in 1sec, noting that the horizontal translation was larger than the vertical one. Because of sliding in the joint, the contact surface between the rest of medial meniscus and the corresponding articular cartilages were decreased, especially for the inner tissue of the medial meniscus (Fig. 3a, b, and c). The contact pressure in this area, therefore, increased with creep. Another reason may have been the geometric structure of the articular cartilage. The thickness of tibial cartilage decreases from center to medial or lateral side. The contact center shifted to the medial side, where cartilage is thinner (Fig. 3a, b, and c). Donahue et al. [24] revealed that the geometries of the tissues may be primary interest to evaluate the onset of OA and a decrease in 10% in meniscus thickness resulted in 20% increase in the contact pressure on the tibial cartilage.

This study employed a large deformation theory. The contact pressure during early creep was distributed rather differently across the condyles (Fig. 5), as compared to that reported in the published studies using small deformation theory with similar force and boundary conditions [3, 5]. This discrepancy may stem from different contact definitions used. Small sliding contact was chosen for the case of small deformation whereas finite sliding was allowed for large deformation. Both maximum contact and fluid pressures predicted in this study were approximately two times that in the previous studies of knee joint under 300N compressive load applied in 1 sec [3, 5]. In addition, it was observed that fluid pressure dissipation was faster here than that in the previous studies. However, this difference in the results could partially be caused by subject-specific knee joint properties.

There are several limitations in the present study. First, different human subjects have been used when comparing the studies [3, 5] used small deformation theory with the present study considered large deformation theory. Therefore, it is not clear how much differences in the results were caused by the subject-specific joint geometry. It may be interesting to create an average virtual knee model [66] that may improve the reproducibility of the results expected by the research community [25]. Second, a simple compressive creep load was applied on the femur in full extension with all rotations constrained. The fluid pressurization and contact pressure in the knee joint could vary with the rotations of femur. Third, lateral meniscectomy was found to cause more stress changes in the joint than medial meniscectomy [67]. However, it was not the scope of the present study to compare lateral and medial meniscectomies. Finally, the present poromechanical model may not be considered as validated because in-situ fluid pressures in the joint have not been experimentally measured [21]. Nevertheless, the fibril-reinforced constitutive model has been validated against multiple measurements [3, 5] and some of the present results compared well with published results. We concur this great insight on modeling: “the power of computational models lies in their ability to be used to investigate scenarios beyond those that can be experimentally examined” [21].

In summary, we have determined the altered cartilage mechanics for four medial meniscectomy cases under a simple creep loading, using the intact knee joint as a reference. For an intact heathy knee joint in standing stance, the lateral compartment supports more loading than the medial compartment. However, the medial meniscus is actually subjected to more loading than the lateral meniscus, which may be one reason why medial meniscal lesions are more common than the lateral ones. Medial meniscectomies further shifted more loading from the medial compartment to the lateral one, resulting in further increased contact pressure in the lateral side. Moreover, a small shift of loading from the medial to lateral side could considerably increase the contact pressure on the lateral side because of its less congruence of joint contact as compared to the medial side [68]. This increase in contact pressure on the lateral side due to meniscectomy may have an implication in the onset of OA. The sub-total meniscectomy exhibited the most distinct mechanical alterations among the cases considered, showing a substantial load shift to the cartilage-cartilage contact within the medial compartment after the loss of crescent shape of the meniscus. Finally, the present study has shown substantial influence of fluid pressure on the load redistribution and tissue deformation in the joint. The fluid-induced creep causes different levels of load transfer between cartilage and meniscus with different meniscectomies, which cannot be understood with an elastic model.

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Figure Captions

Fig. 1 Finite element model of the right knee joint reconstructed for medial meniscectomy: (a) intact medial meniscus (Intact), (b) approximate 25% inner tissue resection (Partial25), (c) approximate 50% inner tissue resection (Partial50), (d) approximate 50% tissue resection in the anterior (Sub-total), and (e) total medial meniscectomy (Total; superior view). The dark areas indicate the location of meniscectomy. The lateral meniscus was intact in all cases. The mesh consists of 10754, 11680, 6220 and 3925 elements, respectively, for the bones, articular cartilages, full menisci and ligaments.

Fig. 2 Maximum contact pressure in the lateral tibial cartilage with creep loading for the normal and medial meniscectomized joints. The contact pressures during creep averaged over contact areas were 0.97 and 0.51 MPa, respectively, for the lateral and medial tibial cartilages of the intact joint.

Fig. 3 Contact pressure (MPa) distributions on the medial tibial cartilage for the cases of: (a) Intact, (b) partial25, (c) partial50, and (d) sub-total meniscectomized knees (superior view showing the remaining meniscus). The contact pressures were shown at 1sec (when creep just began) and 200sec (during late creep).

Fig. 4 Maximum contact pressure in articular cartilages (lateral, medial tibial, and femoral cartilages) at 1sec when the force reached to maximum and creep began. The maximum contact pressure in the femoral cartilage was increased by 5%, 12%, 43%, and 15%, respectively, with the partial25, partial50, sub-total, and total meniscectomies.

Fig. 5 Contact pressure (MPa) distributions on the articular surface of femoral cartilage in the intact model at (a) 1sec, and (b) 200sec. Inferior view.

Fig. 6 Maximum fluid pressure: (a) in the lateral, and (b) in the medial tibial cartilages. Location of maximum fluid pressure may change with time.

Fig. 7 Maximum fluid pressures in the lateral, medial tibial, and femoral cartilages at 1sec when the force reached to maximum and creep began.

Fig. 8 Fluid pressure (MPa) in the layer at approximately 3/8 depth from the articular surface of femoral cartilage at 1sec for: (a) Intact knee, and (b) sub-total meniscectomy. Inferior view; lateral condyle on the left.

Fig. 9 First principal stress obtained from the same site in the deepest layer of the medial tibial cartilage in the intact and total meniscectomized knee joints. The site was located beneath the medial meniscus in the medial and anterior corner when the meniscus was intact.

Fig. 10 Load and displacement of intact and meniscectomized knee joints for the creep of 390 N compressive force applied in 1 sec. (a) the portion of load born by the lateral compartment, and (b) the vertical displacement associated with the compression of cartilage and meniscus.

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Table 1 Material properties of the fibrillar and non-fibrillar matrices of the soft tissues.

	Collagen fibrillar network $\sigma = A\varepsilon + B\varepsilon^2$				Non-fibrillar Matrix (isotropic)	
	Primary fiber direction (x) [MPa]		Perpendicular directions (y,z) [MPa]		Young's modulus [MPa]	Poisson's ratio
	A	B	A	B		
Femoral Cartilage	1.38	367.14	0.41	110.14	1.0	0.47
Tibial Cartilage	0.92	229.46	0.92	229.46	1.0	0.47
Menisci	12.84	0	2.30	0	5.0	0.42
Ligaments	46.47	1118.60	0	0	15.0	0.46
Weight constants (g^m), characteristic times (λ^m) [55]	$G(t) = 1 + \sum_m g^m \, exp\left(\frac{-t}{\lambda^m}\right)$				g^1=0.870; λ^1=10 g^2=0.036; λ^2=100 g^3=0.273; λ^3=1000	
Permeability [mm^4/Ns]	Darcy's law: $\phi^f v_x = -k_x p_{,x}$				k_x =0.002; k_y = k_z =0.001	

The primary fiber direction was incorporated in the x-axis of the local xyz coordinate system. Orthotropic permeability was higher in the x, i.e. fiber, direction [5, 46].

Table 2 Vertical contact force in each contact pair of the knee joint models (t = 1sec).

Model	Contact force on lateral side (N)		Contact force on medial side (N)		Total contact force in joint (N)
	Cartilage to cartilage	Meniscus to cartilage	Cartilage to cartilage	Meniscus to cartilage	
Intact	206	31	108	46	391
Partial25	218	32	110	31	391
Partial50	234	34	113	10	391
Sub-total	214	37	126	13	390
Total	242	34	114	N/A	390

The forces were obtained from contact pairs containing the tibial surface. Noting that the total reaction force should be 390N, the error indicated in the last column is negligible.

Table 3 Comparison of the predicted contact pressures (at 1sec) and contact areas with the experimental results for the intact knee joint.

	Method	Load (N)	CPLTC (MPa)	CPMTC (MPa)	CAL (mm^2)	CAM (mm^2)
Present model	FEA	390	2.36	1.53	262.78	316.62
Fukubayashi and Kurosawa [59]	Experimental	200	–	–	270 ± 0.5	420 ± 1.5
Fukubayashi and Kurosawa [59]	Experimental	500	–	–	530 ± 1.5	420 ± 0.6
Brown and Shaw [60]	Experimental	500	–	–	1125 ± 180 (CAL + CAM)	
Fukubayashi and Kurosawa [59]	Experimental	1000	4	3	640 ± 1.8	510 ± 0.8
Brown and Shaw [60]	Experimental	1000	–	–	1250 ± 100 (CAL + CAM)	
Morimoto et al [61]	Experimental	1000	5.24±1.0	4.76±1.2	443.83 ±107.3	595.12 ±154.7

CPLTC: contact pressure in the lateral tibial cartilage; CPMTC: contact pressure in the medial tibial cartilage; CAL: contact area in the lateral compartment; CAM: contact area in the medial compartment. The predicted contact pressure from the present study was the maximum for each tissue, while the measured was averaged over the contact area.

Table 4 Comparison of the predicted vertical displacement with previously published experimental and computational results for the intact knee joint.

	Method	Load (N)	Vertical displacement (mm)
Present FE model	No creep Late creep (200s)*	390 390	0.62 1.52
Beidokhti et al [58]	Experimental	106	0.36
Mesfar and Shirazi-Adl [63]	Elastic FE model	411	0.8
Kurosawa et al [56]	Experimental	650	0.76
Shrive et al 1978 [57]	Experimental	1000	1.2

(*) Creep was not completed as seen from Fig. 10b.