Creep behavior of the intact and meniscectomy knee joints

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The mechanical functions of the menisci may be partially performed through the fluid pressurization in articular cartilages and menisci. This creep behavior has not been investigated in whole knee joint modeling. A three-dimensional finite element knee model was employed in the present study to explore the fluid-flow dependent creep behaviors of normal and meniscectomy knees. The model included distal femur, tibia, fibula, articular cartilages, menisci and four major ligaments. Articular cartilage or meniscus was modeled as a fluid-saturated solid matrix reinforced by a nonlinear orthotropic and site-specific collagen network. A 300 N compressive force, equal to half of body weight, was applied to the knee in full extension followed by creep. The results showed that the fluid pressurization played a substantial role in joint contact mechanics. Menisci bore more loading as creep developed, leading to decreased stresses in cartilages. The removal of menisci not only changed the stresses in the cartilages, which was in agreement with published studies, but also altered the distribution and the rate of dissipation of fluid pressure in the cartilages. The high fluid pressures in the femoral cartilage moved from anterior to more central regions of the condyles after total meniscectomy. For both intact and meniscectomy joints, the fluid pressure level remained considerably high for thousands of seconds during creep, which lasted even longer after meniscectomy. For the femoral cartilage, the maximum principal stress was generally in agreement with the fiber direction, which indicated the essential role of fibers in load support of the tissue.

1. Introduction

The human knee joint performs its mechanical function through multiple contacts between articular surfaces of cartilage or meniscus. The cartilaginous tissues are highly viscoelastic because of the interstitial fluid flow and possibly intrinsic viscoelastic properties of the tissues. It is noteworthy that the viscoelasticity is essential for the knee joint mechanical functioning (Daniels, 1990; Fung, 1993; Mak, 1986). These tissues appear very soft to facilitate joint motion when not pressurized. However, they can be highly pressurized under fast knee compression to support and redistribute joint loads (Li et al., 2008; Mow et al., 1980; Spilker et al., 1992). Finite element analysis has been extensively used to understand the mechanics of the knee (Bendjallah et al., 1995; Li et al., 1999a; Peña et al., 2006). Comparing experimental
studies, advantages of finite element modeling include easy control of loading and boundary conditions, convenient parametric studies and efficiency in time and cost. Certain important mechanical parameters, such as stress and fluid pressure in articular cartilage in situ, can be adequately determined by modeling, but may be very difficult to measure experimentally. Moreover, inverse finite element methods can be used to estimate the mechanical properties of cartilages and menisci when the measurement data are used as the model input (Cao et al., 2006).

The finite element modeling of the fluid flow in the articular cartilage and menisci has been quite comprehensive with simple tissue explant geometries (Ateshian et al., 1994; Li et al., 1999b; Wilson et al., 2004). The fluid flow has also been successfully modeled in a 3D fiber-reinforced model of the temporomandibular joint (Pérez del Palomar and Doblaré, 2007). However, the relevant results have not been extended to the knee tissues in the joint contact configuration. In the case of knee joint, the fluid pressure was normally ignored to avoid numerical difficulties associated with 3D contact modeling (Haut Donahue et al., 2002; Penrose et al., 2002; Shirazi et al., 2008). Consequently, articular cartilage was often simplified as a single-phase linearly elastic material in the whole joint modeling (Bendjaballah et al., 1995; Li et al., 1999a; Peña et al., 2006). Menisci were commonly modeled as isotropic and linearly elastic (Peña et al., 2005, 2006), and in some cases were considered as transversely isotropic (Haut Donahue et al., 2002) or a linearly elastic solid reinforced by collagen fibers (Bendjaballah et al., 1995; Penrose et al., 2002; Shirazi et al., 2008). These simplifications yielded low computational cost, fast convergent, and yet invaluable results on some aspects of joint mechanics. For example, meniscectomy was found to decrease the total contact area, and consequently, increase the stresses in the knee joint (Bendjaballah et al., 1995). The maximal stress in articular cartilage in meniscectomy joint was about double of that in an intact joint (Peña et al., 2005). The results in stresses were used to determine the modes of cartilage damage. For example, in a total meniscectomy joint, a failure criterion based on relative shearing stresses predicted damage patterns in closer agreement with clinical results than the one based on absolute maximum shearing stress. The damage predicted for partial meniscectomies was less severe than the total meniscectomy (Peña et al., 2008). Significant changes in joint contact mechanics, such as stress distributions, were also predicted for a partial meniscectomy joint (Shirazi and Shirazi-Adl, 2009).

The fluid-flow dependent viscoelastic responses of the normal and meniscectomy knees were not determined previously. This missing information may be needed to understand joint disease and injury since the knee functions as a central load-bearing mechanism. Clinical studies have demonstrated cartilage damage in a few years after the total or partial meniscectomy (Crevoisier et al., 2001; Jackson, 1968; McNicholas et al., 2000). Fibrillation, flattening and formation of femoral cartilage, especially close to the sites of meniscus removal, were among the changes that might have caused cartilage degeneration leading to osteoarthritis (DiStefano, 1980; Roos et al., 1998; Tapper and Hoover, 1969).

Furthermore, meniscectomy might have altered the joint alignment, which increased the joint moments and changed the load distribution (Allen et al., 1984; Bai et al., 2001; Burks et al., 1997). This change in load distribution could initiate or advance osteoarthritis in cartilages (Cerejo et al., 2002; Fairbank, 1948; Levy et al., 1989). The change in fluid pressurization in the tissues might have played an important role in these scenarios.

The objective of the present study was to determine and compare the creep behavior of normal and total meniscectomy knees associated with the fluid pressurization in the tissues. A fibril-reinforced model with fluid pressurization was employed for articular cartilages and menisci to investigate the viscoelastic response of the knee joint under simple compression. The four major ligaments were also considered as fiber-reinforced, and bones were considered as rigid.

2. Methods

2.1. Geometry and finite element mesh

The knee geometry was previously constructed using magnetic resonance imaging of a healthy male’s right leg (age 26, height 1.74 m and weight 70 kg) (Cheung et al., 2005). The current finite element model (Fig. 1) contains distal femur, tibia, fibula, femoral, tibial and fibular cartilages, medial and lateral menisci as well as four major ligaments: anterior cruciate (ACL), posterior cruciate (PCL), lateral collateral (LCL) and medial collateral ligaments (MCL).

Since bones are much stiffer than their adjacent soft tissues, rigid body representation was used for femur, tibia and fibula. A total of 12,829 triangular surface elements were used to represent the bone surfaces, including 5668 elements for femur and 5385 for tibia. Each bone was positioned by a reference node that had six degrees
of freedom. Consistent with literature, the reference point was chosen to be located approximately at the mass center of each bone (Bendjaballah et al., 1995). The global frame of reference was oriented to approximate the anatomical directions: the X-axis in the anterior–posterior direction and the Y-axis in the distal–femoral direction (Fig. 1).

Hexahedral elements were used to mesh the cartilages, menisci and ligaments. A total number of 24,421 elements were used to mesh the soft tissues. Four layers of hexahedral elements were generated along the thickness of femoral cartilage with an average element aspect ratio of 2.5. The average aspect ratio was about 3.4 for the menisci elements and 1.8 for the tibial cartilage elements. Quadratic elements were used for the femoral cartilage while trilinear elements were employed for other soft tissues. A linear variation of fluid pressure was assumed in these elements. The aspect ratios and element types were chosen so that the numerical computations could be completed in a reasonable time, typically about one week on a computer with four 2.5 GHz CPUs and 16 GB of RAM.

2.2. Material properties

A nonlinear fibril-reinforced model was employed for cartilages and menisci, which were assumed as fully saturated porous media. Linear elasticity was assumed for the non-fibrillar proteoglycan matrix while quadratic stress–strain behavior was considered for the collagen matrix. The intrinsic viscoelasticity of the collagen fibers was formulated (Li et al., 2009) but was omitted in the present study for simplicity. The Young’s modulus of the fibrillar matrix was taken to be direction dependent and vary linearly with tensile strain but to be zero for compression (Li et al., 2009), e.g.:

\[
E_f^f(\varepsilon_f) = E_f^0 + E_f^\varepsilon \varepsilon_f
\]

where \(E_f^f\) is the Young’s modulus in the primary fiber direction, and \(E_f^0\) and \(E_f^\varepsilon\) are elastic constants. The same function was also used for the \(y\) and \(z\) directions or transverse directions. The orthotropic properties of the tissues were thus incorporated by orienting the \(x\)-axis of the local xyz coordinate system to the primary fiber direction. Orthotropic permeability was assumed for cartilages and menisci in reference to the local coordinate system as well.

For the femoral cartilage, collagen fibers were assumed to follow the split-line pattern (Below et al., 2002). The \(x\) direction coincided with the split-line direction. The fibrillar tensile properties were \(E_f^f = 1600\) MPa and \(E_f^0 = 3\) MPa, which were determined with reference to the average values of previous modeling (Li et al., 2003). The fibrillar tensile properties in the \(y\) and \(z\) directions were taken to be 0.3 of the values in the \(x\) direction to approximate the tensile test data (Woo et al., 1976). These tensile properties in the \(y\) and \(z\) directions simulated the tensile stiffness of the collagen network produced by the cross-links. The non-fibrillar matrix was assumed isotropic with Young’s modulus of \(E_m = 0.26\) MPa and Poisson ratio of \(v_m = 0.36\). The hydraulic permeabilities of all cartilaginous tissues were \(k_x = 0.002\) mm/s/NS and \(k_y = k_z = 0.001\) mm/s/NS (Li et al., 2009).

For the tibial cartilage, fibers were assumed randomly oriented with same value of Young’s modulus in all directions: \(E_x^f = E_y^f = E_z^f = 1000\) MPa and \(E_x^0 = E_y^0 = E_z^0 = 2\) MPa. These were average values over the orthogonal directions, where fibers were not oriented in a particular direction in the articular surface. The properties of non-fibrillar matrix were assumed to be the same as that for the femoral cartilage.

Similar to the cartilages, menisci were also considered as fiber-reinforced. The primary fibers were oriented circumferentially (Aspden et al., 1985), and the fibrillar modulus in this direction was \(E_f^0 = 28\) MPa. The Young’s modulus in the radial and axial directions was \(E_f^0 = E_y^0 = 5\) MPa. These values were compatible with those in the literature (Shirazi et al., 2008), when small deformation was considered. The non-fibrillar matrix of menisci was assumed to be isotropic with \(E_m = 0.5\) MPa and \(v_m = 0.36\).

The fibers in the ligaments were assumed only in the longitudinal direction with uniform material properties over the length. The fibrillar properties in the longitudinal direction were \(E_f^f = 14\) 000 MPa and \(E_f^0 = 10\) MPa, which yielded a fibrillar modulus of 150 MPa at 1% strain. This value was reasonable for the toe region of experimental data (Hansen et al., 2006; Johnson et al., 1994; Louis-Ugbo et al., 2004). The initial strain in ligaments was not considered here since it is normally beyond the small strain range assumed in the present study (Grood and Hefzy, 1982). The non-fibrillar properties were taken to be \(E_m = 1.0\) MPa and \(v_m = 0.3\) (Hirokawa and Tsuruno, 2000).

The tissue properties were implemented by a user-defined FORTRAN subroutine previously incorporated in ABAQUS.
Both intact and total meniscectomy knee joints were analyzed for comparison. The creep behavior of 6000 s was simulated. The implicit finite element technique in ABAQUS Standard was used to solve the contact problem. The nonlinear surface to surface contact discretization based on the hard contact constraint was chosen. The linear penalty method was used for the contact constraint enforcement. For the intact knee, six contact pairs were considered with 3 on the medial side and 3 on the lateral side: femoral cartilage and meniscus, femoral and tibial cartilages, and meniscus and tibial cartilage. For the meniscectomy knee, only two contact pairs remained accordingly. For each of the contact pairs, one surface was defined as the master surface and the other one as the slave surface. Each slave node always interacts with the same set of master nodes during the surface mechanical contact. A kinematic constraint on contact overclosure is approximated so that the nodes on the slave surface do not penetrate into the master surface. This is done by using the linear penalty method, which provides the value of contact pressure at each surface node. The coefficient of surface friction was 0.02, a value in the normal range of friction of human articular joints (Mow et al., 1993).

Newton method was used to solve the nonlinear problem. Because the contact pairs were 3D fluid filled shapes with variable curvatures, the solution convergence criteria were defined precisely. For each time increment, two types of iterations were performed: severe discontinuity iterations (SDI) and equilibrium iterations. With maximum 50 iterations in each increment, SDI was performed until the open–close changes in contact and slip–stick changes in friction were zero or very small. Equilibrium iterations then started with maximum 16 iterations per increment. In both types of iterations, convergence was judged based on five criteria. First, the largest residual nodal force must be less than a fraction of overall time-averaged value of the spatial force (averaged force of the whole structure). This fraction was set to 0.5% for this analysis. If the first criterion was satisfied, the second one to check was the largest correction of last iteration of the nodal displacement to be less than a fraction of the largest increment of the nodal displacements. This fraction was taken as 1%. Force equilibrium was achieved if these two criteria were satisfied. The third and fourth criteria were based on the equilibrium of pore fluid volumetric flux and pore pressure, respectively. Similar to force residual and displacement equilibrium, the volumetric flux was considered as converged if the largest residual volumetric flux was less than 0.5% of overall spatial flux and the largest correction of pore pressure was less than 1% of the largest pore pressures increment. If equilibrium of force and volumetric flux were achieved, the last criterion was to check the maximum increment of pore pressure in the last iteration to be less than a given value, which was set to 0.02 MPa. A converged time increment would achieve when all five criteria were satisfied.
Fig. 4 – Fluid pressure in a deep layer of femoral cartilage in the meniscectomy knee at (a) 10 s, and (b) 2000 s. The fluid pressure was obtained for the centroids of the elements which are approximately at 7/8 depth from the articular surface.

Consistent with the contact pressure, the maximum fluid pressure occurred in the medial condyle at early times (Fig. 3(a), 10 s). The central regions of the medial and lateral condyles, which were in contact with the tibial cartilage, encountered higher pressures than the outer regions which were in contact with menisci (Fig. 3(a)). The centers of high pressure, however, shifted with time (Fig. 3(b) vs. 3(a)). The maximum fluid pressure in femoral cartilage decayed slowly with time and was still considerable at 6000 s (Fig. 5). For instance, the maximum fluid pressure at 2000 s was about 52% of the maxima that occurred at 1 s.

The maximum fluid pressure for the meniscectomy knee also occurred in the medial condyle of the femoral cartilage; the pressure was still remarkably high at 2000 s, about 66% of the peak pressure (Fig. 4). Unlike the intact joint, the highly pressurized regions in the meniscectomy knee remained in the central regions of the two condyles (Figs. 4 vs. 3). The central region of the medial condyle encountered maximum of 335% increase in fluid pressure after meniscectomy (Figs. 4(b) vs. 3(b), at 2000 s). The corresponding value was about 281% for the central region of the lateral condyle.

In general, the fluid pressure decreased significantly slower after the menisci were removed (Fig. 5). For instance, only 3% decrease in fluid pressure was observed from 1 s to 100 s after meniscectomy. The fluid pressure at a given point in the central part of lateral condyle decreased monotonically with creep in the intact knee, but experienced a small increase for about 100 s in the meniscectomy knee (Fig. 5). The maximum pressure in the femoral cartilage of the meniscectomy knee was about 12%, 13%, 19%, 53% and 74% higher than that in the intact knee for t = 10 s, 1000 s, 2000 s, 4000 s and 6000 s, respectively. The increase in fluid pressure in the center of lateral condyle due to meniscectomy was about 7.5%, 223%, 385% and 1170% at t = 10 s, 1000 s, 2000 s and 4000 s, respectively (Fig. 5).

At 10 s, the maximum principal stress in the solid was observed in the medial condyle (Fig. 6). The direction of maximum principal stress generally agreed with the fiber direction (Fig. 7, for clarity, only medial condyle is shown). Similarly, the directions of principal stresses in meniscus were somewhat aligned with the fiber directions (not shown).

Consistent with the fluid pressure change, the principal stresses increased after meniscus removal (Fig. 8). Unlike the intact joint, the stress level in the central loaded regions remained almost constant for the meniscectomy joint because there were no menisci to redistribute the stresses.
Fig. 6 – First principal stress (tensile stress) in the solid of femoral cartilage for the intact knee at 10 s, shown for the deep layer that is approximately at 7/8 depth from the articular surface.

Fig. 7 – Principal stress directions in the solid of the surface layer of femoral cartilage elements in the medial condyle of the intact knee (t = 10 s).

(Fig. 8). The maximum principal stress in meniscectomy joint was about 15%, 32%, 66%, 115% and 148% higher than that in the intact knee for t = 10 s, 1000 s, 2000 s, 4000 s and 6000 s, respectively.

The loading was gradually transferred from the cartilages to the menisci during creep (Fig. 9(b) vs. 9(a)). This led to a significant decrease in cartilage stress level (Fig. 8). Moreover, it was observed that the maximum stress occurred in the lateral menisci (Fig. 9) which led to a low stress level for the lateral condyle of the femoral cartilage (Fig. 6).

4. Discussion

As a step toward accurate finite element modeling of the knee joint, a previously developed 3D fibril-reinforced knee model (Gu and Li, 2011) was employed to account for the

Fig. 8 – Maximum first principal stress in the matrix of femoral cartilage as a function of time for both intact and meniscectomy knees, calculated for the deep layer that is approximately at 7/8 depth from the articular surface.
fluid pressure of the soft tissues in situ. Nonlinear fiber properties were considered for the cartilages, menisci and ligaments. We attempted to determine the influence of the fluid pressurization in the cartilaginous tissues on the joint mechanics, after much progress has been made in knee modeling using elastic models (Bendjaballah et al., 1995; Li et al., 1999a; Peña et al., 2006).

The loading and boundary conditions were chosen to be as close as possible to those in experimental studies of cadaveric joints (Ahmed and Burke, 1983; Kurosawa et al., 1980; Shrive et al., 1978). Under 300 N of loading, the femoral axial displacement and contact pressure range were in good agreement with the published data (Bendjaballah et al., 1995; Kurosawa et al., 1980; Shirazi et al., 2008). The maximum contact pressure was 0.27 and 0.47 MPa, respectively, under 200 and 500 N compressive forces (Kurosawa et al., 1980).

Using an elastic finite element model, the maximum contact pressure was predicted to be 0.37 and 0.58 MPa, respectively, under 200 and 400 N compressive forces (Shirazi et al., 2008). The contact pressures predicted by the present modeling were in the reasonable ranges, although they were time dependent (Fig. 2).

The predicted rate of dissipation in fluid pressure was very low in creep, which was consistent with the high viscoelastic time constant of the cartilage (Armstrong et al., 1984). At 2000 s, the maximum fluid pressure was about 52% of the maxima (Normal knee, Fig. 3(b)). This may indicate that the fluid pressurization plays an important role not only in short-term responses but also in long-term creep behavior of the joint. The fluid pressure dissipated much faster in a relaxation loading protocol calculated previously using the same knee model (Gu and Li, 2011). This characteristic of creep versus relaxation behaviors is similar to what found in creep and relaxation tests using cartilage disks (Li et al., 2008).

The orientation of maximum principal stress in the femoral cartilage was generally in agreement with fiber direction, which is consistent with the role of collagen network in load support and stress distribution of the tissue.

The present study demonstrated how the menisci gradually redistributed the loadings in the joint. Menisci bore more loading as creep developed (Fig. 9), which led to a decrease in the stresses in cartilage. Menisci removal led to remarkable changes in contact mechanics of the joint, such as much slower dissipation of the fluid pressure at long times (Fig. 5). This is also consistent with the function of fluid pressure in cartilage load-bearing mechanism. In particular, the central regions of the condyles, which were in direct contact with the tibial cartilage, encountered great pressure increase as creep developed if the menisci were removed (Figs. 4(b) vs. 3(b)). The increase in the fluid pressure produced extra tension in the collagen network, which might trigger cartilage degeneration or tissue damage. Therefore, cartilage degeneration or lesion could be initiated from the central regions of the condyles after meniscectomy, although this hypothesis needs to be tested in future studies.

The limitations of the present study include small deformation assumption, simple elastic formulation for the tissue matrices and compressive loading with knee in full extension. These simplifications made it easier for us to obtain some results when the time dependence or viscoelasticity of the problem was considered. The constitutive relationship highlights the nonlinear fibril reinforcement, but simplifies the mechanical properties of the non-fibrillar matrix (Li et al., 2003, 2008). It is necessary to investigate the creep behavior of the knee under general loading conditions after we have gained experiences in the complex numerical problem. In spite of these limitations, the present results may inspire further studies on the viscoelastic response of the knee.

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Conflict of interest

The authors have no conflict of interest to disclose.

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