Computational Poromechanics of Human Knee Joint

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Abstract. Extensive computer modeling has been performed in the recent decade to investigate the mechanical response of the healthy and repaired knee joints. Articular cartilages and menisci have been commonly modeled as single-phase elastic materials in the previous 3D simulations. A comprehensive study considering the interplay of the collagen fibers and fluid pressurization in the tissues in situ remains challenging. We have developed a 3D model of the human knee accounting for the mechanical function of collagen fibers and fluid flow in the cartilages and menisci. An anatomically accurate structure of the human knee was used for this purpose including bones, articular cartilages, menisci and ligaments. The fluid pressurization in the femoral cartilage and menisci under combined creep loading was investigated. Numerical results showed that fluid flow and pressure in the tissues played an important role in the mechanical response of the knee joint. The load transfer in the joint was clearly seen when the fluid pressure was considered.

1. Introduction
An essential mechanism in the load support of the knee joint is the multiple mechanical contacts between the hydrated cartilaginous tissues, which are articular cartilages and menisci. These tissues can be highly pressurized with the interstitial fluid under knee compression and therefore the load response of the knee is influenced by the fluid flow consequently produced by the gradients of the fluid pressure. In the normal joint, articular cartilages provide smooth surfaces for the relative movements of the bones and minimize stress concentrations in the knee. The menisci are located between the tibial and femoral cartilages and minimize the direct contact between the cartilages [8]. Depending on the loading conditions, some joint load is supported by the four ligaments, i.e. anterior cruciate, posterior cruciate, medial collateral and lateral collateral ligaments. These ligaments also provide the joint stability by restricting its motion in the corresponding directions.

Articular cartilages and menisci are often considered as biphasic materials. The solid phase is mainly composed of proteoglycan matrix and the collagen network. The fluid phase is the most prevalent component of the cartilaginous tissues [9]. These knee tissues exhibit viscoelastic behavior, e.g. creep and stress relaxation, under external loadings. The viscoelasticity is mainly due to the effect of interstitial fluid flow and intrinsic viscoelasticity of the collagen fibers.

Computational models of the knee joint and its individual tissues (articular cartilages, menisci and ligaments) have been extensively used in the recent decades to investigate the load response of the knee in different loading or healthy conditions. The possible medical applications of these models include understanding the mechanics of cartilage damage, osteoarthritis, meniscectomy, ligament...
reconstruction, joint abnormality and alignment [1-3]. Invaluable information on the mechanical functions of the knee joint can be obtained from such studies. The past computational studies used either simplified geometrical models with fairly complex constitutive relationships [4,5] or real three-dimensional (3D) knee geometries with simple elastic solid constitutive behavior only [6,7]. Biphasic models have been used extensively to study the viscoelasticity of the cartilages and menisci using two-dimensional (2D) and axisymmetric geometries, or experimental explant geometries [4]. However, cartilages and menisci have been modeled as single-phase materials when anatomically accurate 3D geometries of the knee joint were used in the finite element modeling [3,7].

A 3D model of the knee joint with fluid pressurization in the soft tissues remains challenging because of the multiple mechanical contacts involving viscoelastic response. Recently, we have proposed a 3D model that includes fluid pressurization in the cartilages and menisci, as well as proper fiber orientations in these tissues [10,11]. In the present study, our recent 3D finite element model was used to investigate knee mechanics under combined creep loading. The model includes femur, tibia and fibula as bony structures, cartilages and menisci as fluid-saturated tissues and the four major ligaments as elastic tensile elements.

2. Methods

The geometry of the knee joint was previously constructed using MRI data of a male's right knee. The subject had no previous records of knee injury or surgical operations. The finite element model was generated using ABAQUS software v6.8-2 (Simulia Inc., Providence, RI, USA). The bony structures of the knee (femur, tibia, and fibula) were considered as rigid. This assumption was based on the fact that bones are much stiffer as compared to the soft tissues. In order to account for the effect of collagen fibers in cartilages, menisci and ligaments, these tissues were modeled as fibril-reinforced materials. Cartilages and menisci were modeled using porous elements to account for fluid pressurization in these tissues. Solid elastic elements were used for the ligaments. The same finite element meshes that were tested previously were used in the present study [10].

The material model for the cartilages and menisci was previously developed [14]. It consists of a fluid-saturated matrix that is reinforced by a fibrillar matrix (fibril-reinforced modelling). The fluid pressure and flow are described by a linear Darcy law, while the permeability of the tissue can be dependent on the void ratio which changes with tissue compression. The fibrillar matrix is considered nonlinear and orthotropic.

The fiber orientation in different tissues was chosen based on available data in the literature. The site specific orientation based on split line patterns [12] was used for the femoral cartilage and the depth dependency of the collagen orientations was ignored. Circumferential direction was taken to be the primary fiber direction in the menisci [13]. Fibers in ligaments were aligned in the longitudinal direction. Due to lack of data about the tibial cartilages, random orientation was used for these tissues.

The required mechanical properties for each tissue are: the Young's modulus and Poisson's ratio of the proteoglycan matrix (non-fibrillar part of the solid phase), void ratio and permeability of the solid phase and the Young's modulus of the collagen network. It was assumed that the collagen fibers only support tension and have zero stiffness in compression. Material properties of different tissues were obtained from the literature and can be found in our recent study [11]. A user defined FORTRAN subroutine was used to implement the fiber properties into the ABAQUS software.

The loading protocol was a combined creep loading of 300-N compressive force and 1500-N.mm valgus torque. Both the force and torque were simultaneously applied in one second and remained constant thereafter. Loads were applied to the femur, while tibia and fibula were constrained in all directions. Femur was constrained in lateral-medial and anterior-posterior translations as well as flexion-extension and internal-external rotations. The implicit finite element using transient Soil Consolidation method was used for the simulations. The Newton method was used to solve the nonlinear equations. The maximum time increments were controlled manually and in the meantime by
given convergence criteria. They were chosen so that the numerical results we obtained did not undergo significant changes when the values of the maximum time increments were reduced further. The mechanical contact between cartilaginous tissues was modeled using small sliding frictional contact between femoral cartilage-menisci, menisci-tibial cartilages, and femoral cartilages-tibial cartilages. For each contact pair, one surface is selected as the master surface and the other one as the slave. The contact constraint can be enforced using different methods. In the present study, the linear penalty method was selected for the contact enforcement. Using penalty method, the contact constraint is approximately enforced and very small penetrations of the master surface into the slave surface may occur in some areas. This method results in less convergence difficulties as compared to strict enforcement of the contact constraint.

![Contact Pressure](image1.png)

**Figure 1.** Contact pressure on the articulating surface of the femoral cartilage for (a) $t = 1s$ and (b) $t = 100s$, respectively. The lateral condyle is on the left and the anterior site on the top of the figure (the inferior view of the right knee).

![Fluid Pressure](image2.png)

**Figure 2.** Fluid pressure in the deep layer of the elements of the femoral cartilage for (a) $t = 1s$ and (b) $t = 100s$, respectively (the inferior view of the right knee).
3. Results
For the combined loading considered, the maximum contact pressure on the articulating surface of the femoral cartilage was observed in the lateral condyle (Figure 1a, $t = 1s$). As time increased or creep developed, the contact pressure in the femoral cartilage decreased (Figure 1b, $t = 100s$) and more loading was transferred to the menisci (figures follow). The maximum fluid pressure in the femoral cartilage was also observed in the lateral condyle (Figure 2a). As creep developed, the fluid pressure decreased and the high pressure regions moved towards the anterior sites of the femoral condyles (Figure 2b). During the entire creep, the lateral meniscus was much more pressurized as compared to the medial meniscus (Figure 3). At early times, the highly pressurized regions were observed in the central regions of the inner third of the lateral meniscus (Figure 3a). As time increased, the high pressure regions concentrated at the anterior part of the inner third (Figure 3b).

\[ \text{Fluid Pressure (MPa), } t=1s \quad \text{Fluid Pressure (MPa), } t=100s \]

\[ \text{Max Principal Stress (MPa), } t=1s \quad \text{Max Principal Stress (MPa), } t=100s \]

**Figure 3.** Fluid pressure in the menisci for (a) $t = 1s$ and (b) $t = 100s$, respectively. The lateral meniscus is on the right and the anterior site on the top of the figure (the top view of the right knee).

**Figure 4.** The first principal stress in the deep layer of the elements of the femoral cartilage for (a) $t = 1s$ and (b) $t = 100s$, respectively (the inferior view of the right knee).

The first principal stress (i.e. the greatest principal stress that was tensile in this case) in the femoral cartilage seemed to have two maxima in the same range in both condyles at early times (Figure 4a). However, the lateral compartment encountered significantly more stresses as creep developed (Figure
4b). The high stress regions in both condyles moved toward the anterior sites as creep developed (Figure 4b vs. 4a) and the stress values decreased as time increased. The lateral meniscus was more stressed as compared to the medial meniscus (Figure 5). The latter was almost stress-free under the given small loading (Figure 5). The high stress region was observed at the central part of the inner third of the meniscus and this location remained almost stationary during the creep period considered. Furthermore, the stress level in the menisci increased with creep, which was in contrast to the fluid pressurization in the femoral cartilage.

![Figure 5](image)

**Figure 5.** The first principal stress in the menisci for (a) \( t = 1s \) and (b) \( t = 100s \), respectively (the top view of the right knee).

4. Discussions

A fibril-reinforced poromechanical model of the cartilages and menisci was used to study the tibiofemoral contact configuration under a combined creep compression and valgus torque. The results of the healthy and meniscectomy joints under simple compression have been obtained earlier [11]. As compared to the joint under simple compression, the valgus torque significantly changed the stress/fluid pressure patterns of the cartilages and menisci. In particular, the lateral compartment supported more stresses/pressures compared to the medial compartment (Figures 1 to 3). While in the case of knee joint under simple compression [11], the medial compartment supported more loading. Due to effect of fluid flow and pressure, the mechanical responses were time-dependent. For instance, the contact pressure in the femoral cartilage changed as time passed. In particular, the maximum value of the contact pressure decreased as creep developed (Figure 1). This was due to two main reasons: increase in the contact area, and more load transfer from the femoral cartilage to the menisci. Comparison between the time-dependent first principal stress in the cartilage and menisci (Figure 4 vs. 5) indicates the latter reason. The stress in the cartilage decreased as creep developed (Figure 4). However, the stress in the menisci increased with creep, which showed that the menisci supported more loading as the contact area increased (the applied force/torque were held constant after \( t = 1s \), i.e. creep after 1s).

The fluid pressure distribution in the femoral cartilage was also time-dependent. Due to fluid exudation and changes in contact configuration, the fluid pressure decreased as creep developed, which was expected. During the entire creep, the maximum values of fluid pressure were observed at the lateral condyle of the cartilage, which was because of the effect of valgus torque and constrained motion of the femur in lateral-medial and posterior-anterior directions. However, the location of the high pressure region shifted within the condyle as creep developed. As time increased, the high pressure regions moved from the central area of the lateral condyle to the anterior sites (Figure 2b).
The fluid pressure distributions in the medial and lateral menisci were considerably different. During the entire creep, the lateral meniscus was highly pressurized, especially in the central areas of the inner third (Figure 3). The medial meniscus, however, was almost pressure-free during the entire creep. This was due to the effect of valgus torque that imposed more loading on the lateral compartment and less on the medial side. The location of the high pressure region in the lateral meniscus was time-dependent. It moved from the central areas at early times to the anterior site as time increased.

The maximum first principal stress in the lateral condyle of the femoral cartilage decreased more rapidly as compared to that in the medial condyle (Figure 4). Again, this was due to the fact that the lateral condyle was pushed into lateral meniscus by both compressive load and the valgus torque. This established a better contact condition between the lateral contact pairs as compared to the loose contact between the medial pairs (i.e. medial condyle and the medial meniscus). However, such an effect should not be seen under higher physiological loadings in the presence of pre-stresses in ligaments. The distribution pattern of the first principal stress in the menisci remained virtually unchanged during the entire creep (Figure 5). The lateral meniscus was more stressed as compared to the medial one and its stress level increased as creep developed (Figure 5).

This study further showed the feasibility of a fibril-reinforced poromechanical model to describe the time-dependent response of the cartilages and menisci under different loadings. It has also been observed how the loading in the joint transfers from one tissue to another tissue. The major limitation of this modelling was small deformation assumption, which prevented us from analyzing the joint under higher physiological loadings. However, using small deformation assumption the complex contact situation of the tissues can be studied more comprehensively. The large deformation study will be performed after we get better insight about the convergence difficulties associated with the contact of the cartilages and menisci as well as the fluid pressurization. The model must be finally validated with new experimental results. One possibility is to use cadaveric knees to measure the reaction forces and contact pressures in the joints. Another possibility is to measure the tissue deformation in the knee using MRI, when the knee is loaded with a non-metal device. In either case, the validation will remain challenging.

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