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# Influence of Meniscectomy and Meniscus Replacement on the Stress Distribution in Human Knee Joint

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**Abstract**—Studying the mechanics of the knee joint has direct implications in understanding the state of human health and disease and can aid in treatment of injuries. In this work, we developed an axisymmetric model of the human knee joint using finite element method, which consisted of separate parts representing tibia, meniscus and femoral, and tibial articular cartilages. The articular cartilages were modeled as three separate layers with different material characteristics: top superficial layer, middle layer, and calcified layer. The biphasic characteristic of both meniscus and cartilage layers were included in the computational model. The developed model was employed to investigate several aspects of mechanical response of the knee joint under external loading associated with the standing posture. Specifically, we studied the role of the material characteristic of the articular cartilage and meniscus on the distribution of the shear stresses in the healthy knee joint and the knee joint after meniscectomy. We further employed the proposed computational model to study the mechanics of the knee joint with an artificial meniscus. Our calculations suggested an optimal elastic modulus of about 110 MPa for the artificial meniscus which was modeled as a linear isotropic material. The suggested optimum stiffness of the artificial meniscus corresponds to the stiffness of the physiological meniscus in the circumferential direction.

**Keywords**—Knee mechanics, Meniscectomy, Artificial meniscus.

## INTRODUCTION

Osteoarthritis, which is one of the five leading causes of physical disability in elderly men and women, has long been associated with mechanical insult.[26–28,59] Analysis of the stress distribution in human knee joint can help immensely in understanding the underlying mechanisms and causes of cartilage degeneration and have direct implication in prevention of pathological degeneration of the cartilage and injury and in tissue engineering.[17,32,44,47,49,59,60] Load-induced articular cartilage damage is generally classified into; Type 1: Damage without disruption of the underlying bone or calcified cartilage layer, and Type 2: Subchondral fracture with or without damage to the overlying articular cartilage.[5] Several theoretical models predicted that the maximum level of shear stress in the articular cartilage of the knee joint occurs at the bone–cartilage interface.[19,25,61] These approaches, which are mainly based on modeling the articular cartilage as a homogenous isotopic material, have been successful in explaining type 2 damage of the articular cartilage. Considering the tibial and femoral articular cartilage material as transversely linear isotropic predicts that the maximum shear stress occur at the interface of the cartilages,[18,25,61] explaining the underlying mechanisms and causes of type 1 damage of the articular cartilage. It is clear from these studies that the predicted stresses in the knee joint are sensitive to the constitutive equations of the materials and constants adopted for the mechanical properties of the articular cartilage. The intricate structure of the articular cartilage which varies significantly through its thickness, in addition to the complexity of the loading environment and connections between different knee compartments are some of the key challenges in developing theoretical and numerical models for knee mechanic analysis. To study the mechanics of human knee joint, we have constructed an axisymmetric model of the human knee joint (section "Details of the Computational Model"). In section "The Role of Meniscectomy on the Stress Distribution in the Articular Cartilage", the proposed computational model is employed to investigate the role of articular cartilage material characteristics on the distribution of stresses in a healthy knee joint and a knee joint after complete meniscectomy (removal of the meniscus) under a simple loading history associated with the standing posture. Moreover, a parametric

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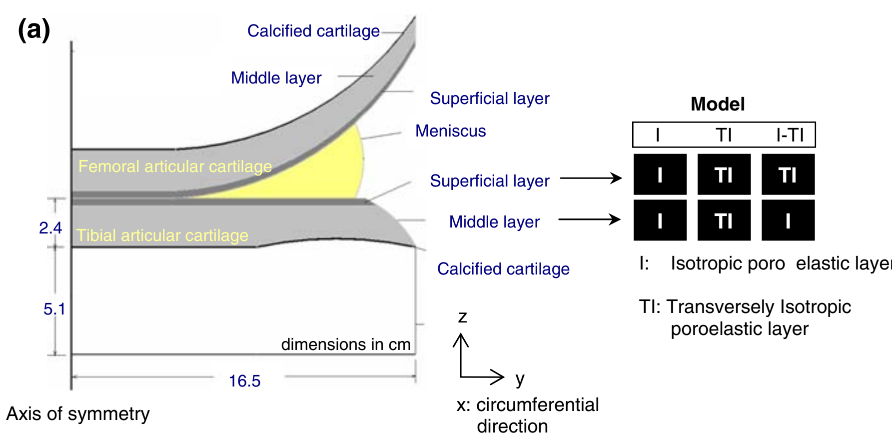
study was carried out in section "Developing Artificial Meniscus" to gain some insight into the mechanics of a knee joint with an artificial (synthetic) meniscus. The conclusions are provided in the final section, where some limitations of this work are discussed and suggestions for future research are provided.

## DETAILS OF THE COMPUTATIONAL MODEL

Following the work of Wilson *et al.*[61] an axisymmetric finite element model of the knee joint was developed using a commercially available finite element software, Adina 7.5 (Adina R & D, Inc., Watertown, MA, USA), Fig. 1a. The model includes different components representing tibia, meniscus and femoral, and tibial articular cartilages. The cartilage itself comprises three layers: the superficial tangential layer, middle and deep zones as one layer, and the calcified part of the cartilage. The top superficial-tangential zone consisting of sheets of randomly oriented fibril network lying parallel to the cartilage surface is generally considered as transversely isotropic material and has a considerably different material properties compared to the middle and deep zones.[3] The maximum thickness of the cartilage was taken to be 2.4 mm at the center of the model.[1,4,15,21,44] The thickness of the superficial tangential layer varies between 10% and 20% of the total thickness of the cartilage.[43] In our model the superficial layer and calcified part of the cartilage accounted for 15% and 5% of the total thickness of the total cartilage thickness, respectively.[46] Recent studies have shown that the location and severity of cartilage damage after mechanical overloading highly depends on its thickness, with thinner cartilage being more vulnerable to damage than thicker samples.[43] In this study, no attempt was made to study the role of cartilage thickness on the mechanics of the knee joint. To study the influence of the material model of articular cartilage on the stress distribution in knee joint, three separate sets of numerical models were developed (see Fig. 1):

*Isotropic cartilage (I model)*: In this model, the superficial layer, middle, and deep zones were all modeled as a single proelastic isotropic layer.
*Transverse isotropic cartilage (TI model)*: In this model, the superficial layer, middle, and deep zones were modeled as a single proelastic transversely




(b)

| | Material constants | Reference(s) |
|---|---|---|
| **Cartilage layer:** Isotropic poroelastic | $E = 0.69$ MPa, $\nu = 0.018$ $k = 3{*}10^{-15}$ m$^4$/N.s, $\Phi_m = 0.25$ | 14 |
| **Cartilage layer:** Transversely Isotropic poroelastic | $E_x = E_y = 5.8$ MPa, $E_z = 0.46$ MPa $\nu_{xy} = 0.0002$, $\nu_{yz} = 0$, $G_{xz} = 0.37$ MPa $k = 5.1{*}10^{-15}$ m$^4$/N.s, $\Phi_m = 0.25$ | 14 |
| **Meniscus:** Transversely Isotropic poroelastic | $E_x = 100$ MPa, $E_y = E_z = 0.075$ MPa $\nu_{xy} = 0.0015$, $\nu_{yz} = 0.5$, $G_{xy} = 0.025$ MPa $k = 1.26{*}10^{-15}$ m$^4$/Ns, $\Phi_m = 0.75$ | 12,24,58 |
| **Meniscus:** Transversely Isotropic elastic | $E_x = 140$ MPa, $E_y = E_z = 20$ MPa $\nu_{xy} = 0.2$, $\nu_{yz} = 0.3$, $G_{xy} = 50$ MPa | 62 |
| **Calcified Cartilage** | $E = 10$ MPa, $\nu = 0.499$ | 42 |
| **Bone** | $E = 400$ MPa, $\nu = 0.3$ | 59 |

**FIGURE 1. (a) Axisymmetric model of the knee joint and the models considered for the tibial and femoral cartilages. (b) Material properties assigned to the components of the computational model: cartilage, meniscus, calcified cartilage, and bone ($E$ = Elastic modulus, $\nu$ = Poisson ratio, $k$ = Permeability, $G$ = Shear modulus and $\Phi_m$ = Solid volume fraction) and the associated references.**

isotropic layer, which is stiffer in the plane parallel to the cartilage surface than in the normal direction.

*Isotropic cartilage with transversely isotropic superficial layer (I-TI model)*: In this model, the superficial layer was considered as transverse isotropic layer with stiffer plane parallel to the cartilage surface. The middle and deep zones were modeled as a separate proelastic and isotropic layers, respectively.

To investigate the role of meniscectomy on the stress distribution in the human knee joint, two models were constructed: A model for a healthy knee joint as presented in Fig. 1a and a model for a knee joint after complete meniscectomy by excluding the part representing the meniscus in Fig. 1a. In the model of the healthy knee joint, the meniscus was modeled as a proelastic transversely isotropic material, which is stiffer in the circumferential direction. This higher stiffness in the circumferential direction is associated with the collagens oriented primarily in the circumferential direction of the meniscus. The material constants used for modeling the meniscus, articular cartilage layers, and bone are shown in Fig. 1b. The biphasic characteristics of the meniscus as well as articular cartilages were taken into account in all the calculations. A limited set of results is also presented by modeling the meniscus as a transversely isotopic linear elastic material as suggested in Ref. 44. In section ''Developing Artificial Meniscus'', where the role of an artificial meniscus on the stress distribution in the knee joint is discussed, the hypothetical artificial meniscus has identical geometry as the meniscus in Fig. 1a and is modeled as isotopic linear elastic material.

Here, we assumed that the lateral and medial parts of the knee joint carry the same amount of load in standing posture, independent of the knee joint condition. The load, applied in the form of a uniform pressure to the bottom face of the tibia, was increased linearly to 0.17 MPa in one second and then kept constant till 60 s. This loading corresponds to half of the body weight of a 60 kg person.[61] The stress distributions in the models were calculated using finite element contact analysis. Our preliminary calculations indicated that having more than one deformable contact pair in the model can cause serious computational problems due to the complex geometry of the model and the material properties of the articular cartilage and meniscus. To overcome this problem, a thin flexible membrane with a very low stiffness was incorporated between the cartilage and the meniscus, which was rigidly connected to the meniscus in the model of the healthy joint. This elastic thin layer separates two contact surfaces with a nonlinear properties and convergence is achieved much faster. Free sliding (i.e., no friction) condition was assumed for contact between the articular cartilages and at the meniscus–cartilage interfaces.[20] A limited set of calculations was carried out by including the femur in the finite element model revealing its minimal influence on the stress distribution in the knee joint under the considered loading. In all the calculations presented in this study, the femur was not included in the calculations and a clamped boundary condition was applied at the interface of the cartilage with femur.

## THE ROLE OF MENISCECTOMY ON THE STRESS DISTRIBUTION IN THE ARTICULAR CARTILAGE

Type 1 and type 2 articular cartilage damage have been associated with excessive shear stresses at the cartilage–cartilage interface and the bone–cartilage interface, respectively.[20,31] In this section, the distribution of the maximum shear stress in the articular cartilage of the human knee joint is analyzed for different models of the articular cartilage discussed in section ''Details of the Computational Model''. Figure 2 displays the time histories of the maximum shear stress at the cartilage–cartilage interface and bone–cartilage interface for all the cartilage models considered when the knee joint is subject to the applied uniform pressure history, which mimics the standing posture. The calculations were carried out for the models of the healthy knee joint and the knee joint after meniscectomy by excluding the part representing the meniscus. All the models suggest a significant increase in the maximum shear stress after meniscectomy at the cartilage–cartilage interface, Fig. 2a. For example, the I-TI model predicts that the maximum shear stress in the interface of cartilages elevates by a factor of about four after meniscectomy. The increase in the maximum level of shear stress at the bone–cartilage interface due to meniscectomy is approximately 150% for all the models, Fig. 2b. The maximum shear stress at the cartilage–bone interface decreases during the constant load interval (1–60 s) in the model of the knee joint after meniscectomy with a trend that depends on the material model of the articular cartilage. On the other hand, the predicted time histories of the maximum shear stress at the bone–cartilage interface are slightly sensitive to the adopted material model for the articular cartilage, Fig. 2b. In contrast, the time histories of the maximum shear stress at the cartilage–cartilage interface are highly dependent on the material model adopted for the articular cartilage in the computational model, Fig. 2a. The calculations based on the I-TI model suggest remarkably higher

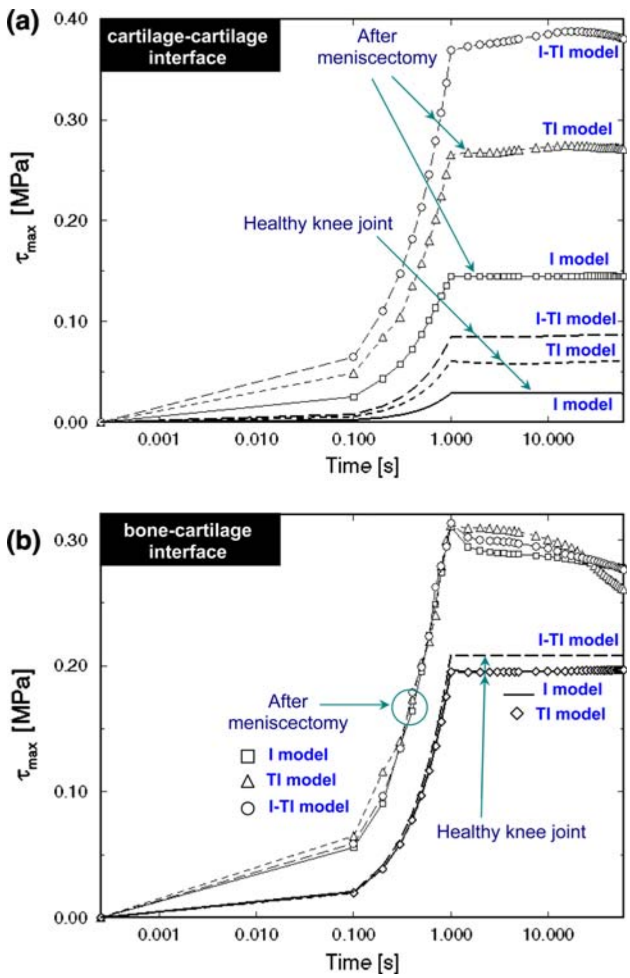



**FIGURE 2. Variation of the maximum shear stress at (a) at the cartilage–cartilage interface and (b) at the cartilage–bone interface versus time for different material models for the articular cartilage. The results are presented for the healthy knee joint and knee joint after meniscectomy. The meniscus is modeled as a transversely isotropic poroelastic material.**

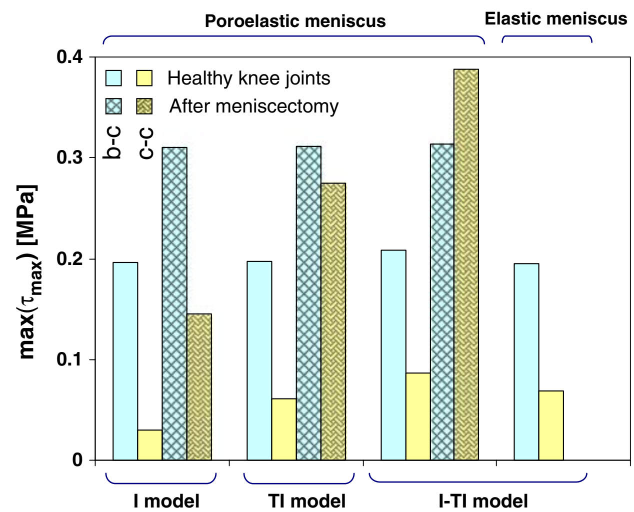



**FIGURE 3. Maximum shear stresses at the bone–cartilage interface (denoted in this figure by b-c) and cartilage–cartilage interface (denoted in this figure by c-c) for different material models of the articular cartilage during the 60 s period of calculations. The results are presented for the healthy knee joint and knee joint after meniscectomy. For the I-TI model, the results are presented for two different material models of the meniscus.**

level of maximum shear stress at the cartilage–cartilage interface; approximately 260% and 290% higher than the prediction based on the I model of the healthy knee joint and the knee joint after meniscectomy. Our numerical simulations complements previous experiments on the knee joint of human cadavers, which indicated an increase in the peak contact pressures of approximately 235%[7] and experiments based on animal models[36,37] and synthetic models.[30]

To gain further insight into the role of meniscectomy on the stress distribution in the articular cartilage of the knee joint, we have summarized the predictions from each of the models of the articular cartilage in Fig. 3. For the isotopic cartilage model, the maximum shear stress in the articular cartilage is at the cartilage–bone interface in both models of a healthy knee joint and the joint after meniscectomy. The location of the maximum shear stress moves closer to the center of the model (axis of symmetry in Fig. 1a) by excluding the meniscus. This is consistent with previous studies which assumed the articular cartilage as a homogenous isotopic material.[61] For the transversely isotropic model (TI model), the location of the maximum shear stress is at the bone–cartilage interface similar to that in the I model. The level of maximum shear stress at the cartilage–cartilage interface after complete meniscectomy is comparable with the maximum shear stresses calculated at the bone–cartilage interface and is significantly higher than the predicted values by the I model. Studying the evolution of the stress distribution in the knee joint shows that for the model of knee joint with complete meniscectomy, upon loading, the maximum shear stress occurs at the bone–cartilage interface in the first 35 s and then it shifts to the interface of the two cartilages due to the biphasic material characteristic of the meniscus and cartilage. Note that the values shown in Fig. 3 are the maximum values of the shear stress during the 60 s time interval. The time histories of maximum shear stress are presented in Fig. 2 and discussed before. This model suggests that the knee joint becomes susceptible to both type 1 and type 2 damage after meniscectomy, as also described by Wilson *et al.*[61] On the other hand, it is well known that the structure of the cartilage undergoes remodeling under mechanical stimuli.[6] The structural remodeling can enhance the strength of the articular cartilage at the sites of high local stresses. These pathological observations may suggest that Type 1 cartilage damage is more likely to happen after complete meniscectomy, as the maximum shear stress at the cartilages interface is increased by a factor of 4.5, compared to only 1.5-fold increase in the maximum shear stress at the bone–cartilage interface. Further clinical and theoretical studies are required to examine the validity of this hypothesis. Finally for the I-TI model, the location of maximum shear stress shifts to the interface of the cartilage–cartilage upon removal of the meniscus. Complete meniscectomy causes the maximum shear stresses at the bone–cartilage interface and at the cartilage interfaces to increase by 150% and 460%, respectively. The excessive increase in the shear stresses at the cartilage–cartilage interface after meniscectomy may lead to type 1 cartilage damage. In general, the level of stresses predicted by the I-TI model is considerably higher than the other two models considered, Figs. 3 and 4a. The contours of the shear stress in the articular cartilage are predicted to be significantly different from those in the I models. Figure 4b shows the shear stress distribution for the I-TI model at two instants: 1 s and 60 s for both the model of the healthy knee joint and that of the knee after complete meniscectomy. The stress distributions in the articular cartilage calculated using this model are in general in close agreement with those obtained by considering the biphasic characteristics of the meniscus. A selected set of results for the I-TI model with transversely isotopic linear elastic meniscus is provided in Fig. 3. Moreover, a set of parametric calculations was carried out by modeling the meniscus as a homogenous isotopic elastic material and systematically varying its stiffness. The results will be discussed in section ''Developing Artificial Meniscus''.

## DEVELOPING ARTIFICIAL MENISCUS: PRELIMINARY STUDY FOR DETERMINING THE OPTIMUM MATERIAL CHARACTERISTICS

The meniscus is known to have a very limited regenerative capability. Injury to the meniscus, which is amongst the most frequent injuries in orthopedic practice, and the consequent meniscectomy leads to excessive stresses in the articular cartilage and therefore cartilage degeneration as observed by several clinical observations[22] and quantified for the case of complete meniscectomy in section ''The Role of Meniscectomy on the Stress Distribution in the Articular Cartilage''. New treatment modalities for repair of the injured meniscus have been recently offered by advances in the surgical procedures and tissue engineering. Various arthroscopic repair techniques have been proposed to avoid meniscectomy.[8,16,41,52] Furthermore, developments in the field of tissue engineering and regenerative medicine could lead to emergence of novel replacement techniques for the whole meniscus in a near future.[2,10,29,55] Recently, several biomaterials have been considered for constructing artificial menisci, and the preliminary clinical studies have indicated their potential in aiding in treatment of the load-induced cartilage degeneration due to meniscectomy.[13,34,35,54,56,57] Here, we use the computational model developed in section ''Details of the Computational Model'' to study the role of the mechanical properties of an artificial meniscus on the mechanics of the knee joint. The results of our study are applicable for both synthetic biocompatible materials and tissue-engineered constructs which are under consideration as replacements for the meniscus.

In the subsequent calculations, the artificial meniscus was taken to have identical geometry as the meniscus in the model of the healthy knee joint presented in Fig. 1a. In the calculations, the meniscus was taken as a linear elastic material and its Young's modulus was varied systematically in the range of 10 MPa to 1.6 GPa. Two values of Poisson ratio were considered for the artificial meniscus, 0.3 and 0.45. In all the calculations, the I-TI model was employed to represent the articular cartilages of the knee joint. The distribution of shear stresses in the articular cartilage was studied in each calculation. The results of this study are summarized in Fig. 5a. The present model in general predicts higher values of shear stress at the interface of cartilages compared to the calculations presented in section ''The Role of Meniscectomy on the Stress Distribution in the Articular Cartilage''. By increasing the stiffness of the artificial meniscus to about 200 MPa, the maximum shear stress at the cartilage–cartilage interface decreases considerably, while further stiffening of the meniscus in the model has no significant effect on the value of this stress. In contrast, the maximum shear stress at the interface of the bone–cartilage is not sensitive to the elastic modulus of the artificial meniscus. We have also studied the dependence of the maximum shear stress at the cartilage–meniscus interface on the stiffness of the artificial meniscus. Increasing the stiffness of the artificial meniscus up to 90 MPa decreases the level of stress at the cartilage–meniscus interface, while further increase in the stiffness of the artificial meniscus elevates the level of the stress. These calculations suggest that the optimum stiffness of the artificial meniscus, which may minimize the susceptibility of the articular cartilage to degeneration due to excessive shear stresses, is in the range of 100–120 MPa which corresponds to the stiffness of the physiological meniscus in the circumferential direction.[58,62] The results presented here are

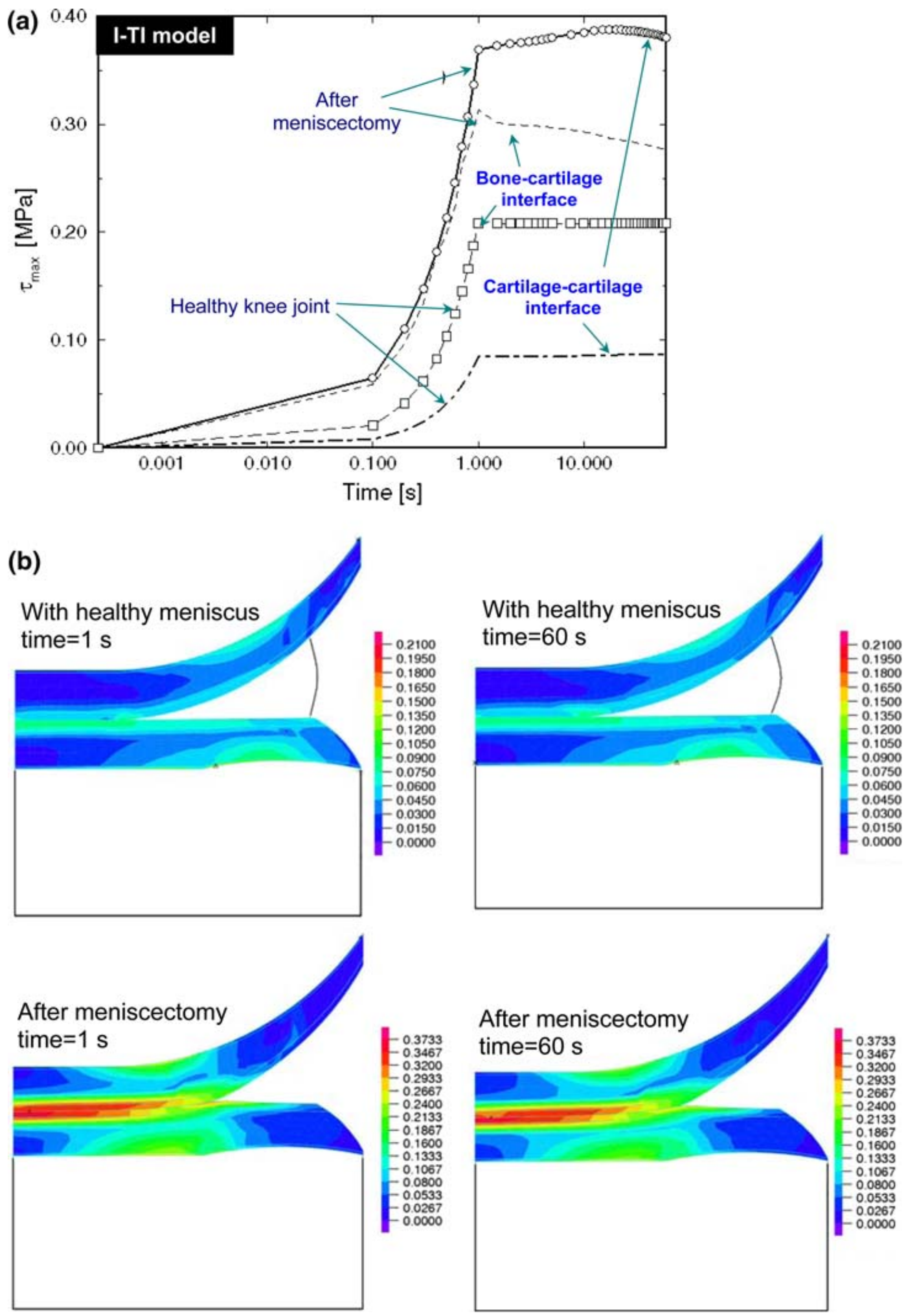



**FIGURE 4. (a) Variation of the maximum shear stress at the cartilage–cartilage interface and the bone–cartilage interface versus time for the I-TI model with biphasic meniscus model. The results are presented for the healthy knee joint and knee joint after meniscectomy. (b) Distribution of maximum shear stress in the articular cartilages of I-TI Model with and without meniscus at time = 1 and 60 s.**

slightly sensitive to the Poisson ratio of the artificial meniscus in the computational model. The maximum sensitivity to the artificial meniscus Poisson ratio is observed for the shear stress at the cartilage–cartilage interface, where decreasing the Poisson ration from 0.45 to 0.3 lead to maximum 10% reduction in maximum shear stress.

Polyvinyl alcohol-hydrogel (PVA-H), which has been recently proposed as a possible candidate for constructing artificial menisci,[33,34] in general has a considerably lower stiffness than the optimum range suggested above. This lack of sufficient mechanical properties is a major obstacle in their use for developing artificial meniscus and cartilage.[34,53] Due

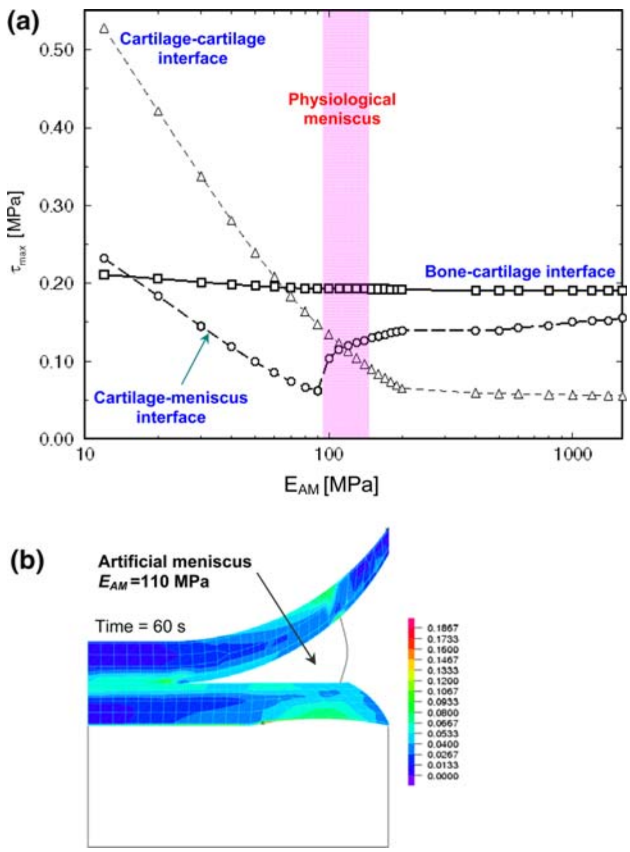



**FIGURE 5. (a) Influence of the Young's modulus of the artificial meniscus, dented by $E_{AM}$, on the maximum shear stresses at the interfaces of cartilage–cartilage, cartilage–bone, and cartilage meniscus. A range for the stiffness of physiological meniscus in the circumferential direction of the knee is also shown. (b) Shear stress distribution in the auricular cartilages with artificial meniscus with $E_{AM}$ = 110 MPa. The Poisson ratio of the artificial meniscus is equal to 0.45 in this set of calculations.**

to its unique structure, the healthy meniscus has a significantly higher stiffness in the circumferential direction. A potential solution to this deficiency is incorporating biocompatible fibers in constructing the artificial menisci to enhance their stiffness in the circumferential direction of the knee joint. We have performed an additional set of calculations by considering the artificial meniscus as a transversely isotopic material with elastic modulus of 100 MPa in the circumferential direction and 20 MPa in the other two directions (as suggested in Ref. 44 for the pathological meniscus, see Fig. 1b). No significant difference was observed by reducing the stiffness of the artificial meniscus in directions normal to the circumferential direction by a factor of 5. This result suggests that the artificial meniscus stiffness in the circumferential direction is a determinant factor in preventing cartilage degeneration due to excessive stresses.

## CONCLUDING REMARKS

We presented a set of results to study the role of material characteristics of the articular cartilage on the stress distribution in the human knee joint using finite element method. Two models were considered corresponding to the healthy knee joints and the knee joint after complete meniscectomy, which provided insight into the initiation of both type 1 and type 2 cartilage damage in secondary osteoarthritis. Additional sets of calculations were carried out to study the role of an artificial meniscus on the mechanics of the knee joint. The success of an artificial meniscus in the body depends on several factors including its material characteristics, geometry, biocompatibility, the technique used by the surgeon, the condition of the patient, and his/her activities which determine the loading applied to the knee joint. In this study, we investigated the role of material characteristics of a hypothetical artificial meniscus on the load-induced stresses in the articular cartilages of the knee joint under a loading associated with standing posture. This parametric study suggests an optimum stiffness which is very close to the stiffness of the pathological meniscus in the circumferential direction and has direct implications in designing artificial meniscus using synthetic biomaterials or tissue engineering. In addition to the load bearing and shock absorption characteristics of the meniscus, it is also postulated to have key roles in proprioception, joint lubrication, and nutrition of the articular cartilage.[40,41,51,63] Development of an artificial meniscus should be proceeded by careful consideration of the meniscus function, which calls for a rigorous interdisciplinary collaboration between researchers and clinicians.

Despite the simplification and assumptions made in constructing the computational model, this study elucidates some of the important aspects of load-induced cartilage degeneration after complete meniscectomy and addresses some of the key challenges in developing artificial menisci, while emphasizing the need for further systematic studies by considering more realistic geometry of the knee joint and using three-dimensional models, as well as other loading conditions. This includes the exact three-dimensional geometry of the meniscus as well as it alteration during joint function, which may immensely affect the knee mechanics. Thus, a key step towards a better understanding of the mechanics of human knee joint in health and disease is development of detailed three-dimensional computational models. Examples of such computational models are limited[9,11,23,38,39,45,48–50] and applications in understanding the state of human health and disease are still at their infancy. Development of such computational models is also an inevitable step in studying the human knee mechanics under physiological loading conditions that occur during everyday activity. Our current research effort is focused on developing 3D models of the knee joint based on MRI data. However, 3D computational models are extremely complex and need a great deal of memory for convergence. Thus, 2D computational models may still provide valuable insight into knee mechanics and initiation and progression of osteoarthritis.

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