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Model for analyzing the mechanical behavior of articular cartilage under creep indentation test

Mohammad Mehdi Elhamian, a) Hossein Karami, Mansour Alizadeh, Mahmood Mehrdad Shokrieh, and Alireza Karimib)

School of Mechanical Engineering, Iran University of Science and Technology, Tehran 16846, Iran; Tissue Engineering and Biological Systems Research Laboratory, School of Mechanical Engineering, Iran University of Science and Technology, Tehran 16846, Iran; and Composites Research Laboratory, Center of Excellence in Experimental Solid Mechanics and Dynamics, School of Mechanical Engineering, Iran University of Science and Technology, Tehran 16846, Iran

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In this study, an innovative depth dependent biphasic transversely isotropic model (DBT) was proposed to study the mechanical behavior of Articular Cartilage (AC). To find a more precise model to address the mechanical behavior of AC, the vital role of collagen fibers in all zones of the AC has been taken into account and depth dependent elasticity mechanical properties of cartilage are calculated as a function of collagen fibers orientation and volume fraction. Material parameters of permeability function were calculated in such a way that the variations of indenter displacement with time predicted by Finite Element Method (FEM) simulation for creep indentation test of the AC sample based on DBT model. In addition, the test was simulated by an isotropic-biphasic model to compare the capabilities of these two models and difference in mechanical behaviors of biphasic-isotropic and depth dependent transversely isotropic materials. According to the calculations, the presence of collagen fibers triggers increasing of stresses in fibers direction and decreasing of stresses perpendicular to fiber direction in the superficial and deep zones of AC. The findings of this study may have implications not only for calculating stress distributions in AC components but also for developing progressive damage model of AC for predicting osteoarthritic cartilage behavior in different cartilage-related diseases. © 2014 AIP Publishing LLC.

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I. INTRODUCTION

Articular Cartilage (AC) is a permeable, fluid-filled, and mechanically poroviscoelastic connective tissue, which has a key asset in synovial joints.1,2 Its unique mechanical behavior and poor regenerative capacities make it a highly specialized material. Cartilage covering the articulating surfaces of jointed bones provides a low-friction (nearly in frictionless), wear-resistant surface results in an optimal transmission, bearing, distribution, and absorption of applied loads across diarthrodial joints (i.e., hip, knee, etc).3,4

AC consists of three phases:5,6 a fluid phase, a solid phase, and an ionic phase. Ionic phase is attributed to electrolytes existing in fluid phase. The fluid phase is composed of interstitial water (60%–80% of wet weight). Solid phase includes collagen fibers (mainly type II), proteoglycan (PG) matrix, and the other multi-adhesive glycoproteins.7 Fluid and solid phases’ interactions contribute to a biphasic poroviscoelastic behavior of cartilage.8 Low metabolic activity of cartilage is related to the chondrocyte cells, embedded within an extensive extracellular matrix and each one is surrounded by a narrow region called the precellular matrix.9 Collagens work as the main structural elements in solid phase as a three dimensional network. Mechanical properties of the AC are predominantly attributed to the microstructural properties of proteoglycan matrix and collagen fibers. Interaction between collagen content and proteoglycan matrix is an important factor that controls the stress-strain, and fluid flow in the articular cartilage.

It has been shown that composition and structure of cartilage constituents vary with depth from surface to the other end side of cartilage. Cartilage components exhibit distinctively different anisotropies, which vary as a function of depth.10 To study the depth-dependency of this hyaline tissue, it is divided into three morphological zones based on the collagen fiber density and orientation: the superficial zone (also called tangential zone), which is characterized by highly dense, packed layers of uniform collagen fibers orienting parallel to joint surface and lower content of proteoglycan; transitional zone in which fibers are arranged oblique or randomly organized and PG content increases; deep zone in which fibers are located perpendicular to the joint having the most PG content, having their largest diameters and are sparse.11 Moreover, interstitial fluid fraction decreases along with increasing distance from surface. Depth-dependency of composition and structural arrangement make the cartilage mechanical properties depth-dependent, anisotropic, and heterogeneous.12

Different analytical and numerical studies have been employed to study the cartilage response to different types of loading in static and dynamic conditions considering cartilage one phasic to three phasic media. Some studies take
cartilage as an isotropic material in spite of its intrinsic features resulting reasonable output in some modeling.\textsuperscript{13,14} Other works have shown that considering cartilage as a transversely isotropic leads to a better and more accurate results fitting experimental data than isotropic models. AC is usually studied by three types of experimental tests: confined compression, unconfined compression, and indentation test.

In this study, a creep indentation test of an AC sample has been simulated based on the presented depth dependent biphasic transversely (DBT) isotropic model, and predicted mechanical behavior was compared to calculations of isotropic-biphasic model and experimental data. In addition, the role of collagen fibers in strengthening of AC and stress distribution in different zones has been studied.

\section{II. MATERIALS AND METHODS}

\subsection{A. Definition of depth-dependent transversely isotropic model}

A DBT model was employed to simulate an indentation creep test. The duty of collagen fibers is transduction of any kind of applied forces through articular cartilage. The proposed DBT model has the ability to consider the role of collagen fibers in all zones of AC. Collagen fibers are parallel to surface in the superficial zone, perpendicular to it in the deep zone and their angle toward cartilage surface $\phi$ in the transition zone nonlinearly varies from 0° at the top to 90° at the end of this zone. Therefore, AC has depth dependent mechanical properties. For the sake of simplicity in the numerical simulation, the heterogeneous, anisotropic cartilage was approximated as transversely isotropic model. The axis $z$ (perpendicular to the cartilage surface) was assumed to be symmetric and plane $\tau-\theta$ isotropic.\textsuperscript{15}

\begin{equation}
E_{\text{CSM}} = \frac{E_L^2 + 4E_L G_{LT} \Delta + 2E_L E_T + 8v_{LT} E_T G_{LT} \Delta - 4v_{LT} E_T^2 + 4E_T G_{LT} \Delta + E_T^2}{\Delta(3E_L + 2v_{LT} E_T + 3E_T + 4G_{LT} \Delta)},
\end{equation}

where $\Delta = 1 - v_{LT} v_{TL}$, $E_L$ and $E_T$ are the longitudinal and transverse moduli of fictitious unidirectional layer with equal volume fraction of fibers parallel to the cartilage surface. $V_{fz}$ and $V_{fr}$ are equivalent fibers volume fraction perpendicular and parallel to the cartilage surface, respectively, in which volume fraction of collagen fibers is equal to sum of these volume fractions ($V_f = V_{fr} + V_{fz}$).

Volume fractions of equivalent horizontal and vertical fibers could be determined as a function of fibers orientations $\phi$ and collagen fibers volume fraction $V_f$.

\begin{equation}
V_{fr} = \frac{V_f}{1 + \tan \phi},
\end{equation}

Depth dependent mechanical properties of solid phase of the AC were calculated by multidirection composite micromechanics model as a function of cartilage components elasticity mechanical properties, collagen fibers volume fraction, and angle toward cartilage surface as below

\begin{equation}
E_z = E_f \times V_{fz} + \left[\frac{E_f \times E_m}{V_{fr} \times E_m + (1 - V_{fz}) \times E_f}\right] \times (1 - V_{fz}),
\end{equation}

\begin{equation}
E_{\phi} = \frac{E_{\text{CSM}} \times E_f}{V_{fz} E_{\text{CSM}} + (1 - V_{fz}) E_f},
\end{equation}

\begin{equation}
\frac{1}{G_{rz}} = \frac{1}{G_{r\phi}} = \frac{V_{fz} + \left(\frac{1 - V_{fz}}{G_{f}}\right)}{V_{fr} + \left(\frac{1 - V_{fr}}{G_{m}}\right)} \times (1 - V_{fz}) + \frac{V_{fr} + \left(\frac{1 - V_{fr}}{G_{m}}\right)}{E_{\text{CSM}}},
\end{equation}

\begin{equation}
v_{z\phi} = \frac{v_{fr} \times V_{fz} + \left(1 - V_{fz}\right)}{V_{fr} \times v_{fr} + \left(1 - V_{fr}\right)} v_{\text{CSM}},
\end{equation}

\begin{equation}
G_{r\phi} = \frac{E_f}{2 \times (1 + v_m)},
\end{equation}

where $E_m$, $G_m$, and $\nu_m$ are elastic modulus, shear modulus, and Poisson’s ratio of PG matrix and $E_f$, $G_f$ and $\nu_f$ are elastic modulus, shear modulus, and Poisson’s ratio of collagen fiber, respectively. $E_{\text{CSM}}$ and $v_{\text{CSM}}$ are elastic modulus and Poisson’s ratio of continuous strand mat composite and could be determined as\textsuperscript{16}

\begin{equation}
V_{fz} = \frac{V_f \times \sin \phi}{\cos \phi + \sin \phi},
\end{equation}

This model predicts depth dependent mechanical properties of solid phase of the AC as a function of the mechanical properties of collagen fibers and PG matrix, volume fraction of fibers, and their angle toward cartilage surface $\phi$.

Variation of collagen fibers volume fraction and angle through cartilage depth were calculated for Schinagl \textit{et al.}\textsuperscript{17} sample as below

\begin{equation}
V_f = 0.22 - 0.02 \times \zeta,
\end{equation}

\begin{table}[h]
\centering
\caption{Elastic properties of collagen fiber and proteoglycan matrix.}
\begin{tabular}{ll}
\hline
 & Young’s modulus (MPa) & Poisson ratio \\
\hline
Collagen fiber & 10 & 0.30 \\
Proteoglycan matrix & 0.09 & 0.23 \\
\hline
\end{tabular}
\end{table}
\[
\varphi(z) = \begin{cases} 
0 & \text{if } \xi \leq 0.090 \\
5.97 - 127.52x + 988.23x^2 - 3977.59x^3 + 11180.49x^4 - 16470.40x^5 + 9111.64x^6 & \text{if } 0.09 < \xi < 0.818, 
\end{cases}
\]

where \( \xi \) is the normalized depth \((\xi = z/H)\) in which \( \xi = 0 \) at the top of superficial zone and it reaches to 1 at the end of deep zone.

Wide range of the mechanical properties of collagen fibers and PG matrix have been reported.\textsuperscript{17–20} Table I shows experimental measurements of these mechanical properties.\textsuperscript{21–23}

The DBT model considers AC as a transversely isotropic material in which its mechanical properties are a function of cartilage depth. According to Figure 1, AC was divided into 10 layers. The calculated mechanical properties of each layer have been reported in Table II.

Permeability of this biphasic model was assumed as a function of cartilage void ratio “e” as below:\textsuperscript{24}

\[
k = k_0 \left( \frac{e}{e_0} \right)^{K} \exp \left( \frac{M}{2} \left( \frac{1 + e}{1 + e_0} \right)^2 - 1 \right),
\]

where \( k_0 \) and \( e_0 \) are the initial permeability and initial void ratio, respectively, and \( M \) and \( k \) are the non-dimensional permeability coefficients. These values were calculated for creep indentation test by fitting the model response to the experimental measurements.

An isotropic-biphasic model with same permeability and average elasticity mechanical properties of DBT model was created to compare the abilities of these two models and mechanical behavior of biphasic isotropic and depth dependent transversely isotropic materials. Mechanical properties of isotropic model are reported in Table II.

B. FEM simulation of creep test by DBT model

A creep test on the cylindrical cartilage sample with 1.15 mm thickness and 10 mm diameter based on Keenan \textit{et al.} experiment was simulated in which an axisymmetric sample was created. The thickness of sample was divided into 10 layers and elasticity mechanical properties and permeability of each layer was defined using the calculations of DBT model.

The 316 sintered stainless steel (\( E = 190 \text{ GPa}, \nu = 0.28 \)) indenter was considered as a porous material with 50% porosity and 2 mm diameter. Constant permeability of indenter is 5 orders greater than that of the cartilage initial permeability. The friction coefficient between indenter and AC was assumed to be 0.26. First step of loading protocol includes making a static contact between indenter and cartilage by a small tare load as indenter weight (0.015 N), afterward indenter linearly ramped up to 0.35 N in 12 s and applied force was maintained for about 4230 s. A free draining boundary condition was applied to the lateral side of indenter and cartilage and upper surface of cartilage which is not under indenter. The lateral side of cartilage was free for radial displacement but the bottom line of cartilage was fixed. The schematic representation of model along with applied boundary conditions is shown in Figure 1. Since the highest deformations were observed in the superficial layers, and in order to prevent distortion of elements in layer one due to edge point of indenter, the mesh considered more refined in the superficial zone especially in the layer 1 compared to the deep layers. Layer 1 has 230 rows of pore-fluid/stress element, layers 2–5 have two and layers 6–10 have one row. Figure 2 shows the mesh applied to the cartilage.

III. RESULTS AND DISCUSSIONS

By fitting the overall response of the model in creep indentation simulation to experimental data, the material parameters were calculated as \( K_0 = 1.7 \times 10^{-15} \text{ m}^4/\text{N s,} \).

<table>
<thead>
<tr>
<th>Layer</th>
<th>( E_{zz} ) (MPa)</th>
<th>( E_{xx} ) (MPa)</th>
<th>( v_{xy} )</th>
<th>( v_{yx} )</th>
<th>( \eta_{xy} )</th>
<th>( G_{zz} ) (MPa)</th>
<th>( G_{xx} ) (MPa)</th>
<th>( G_{yy} ) (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Layer 1</td>
<td>0.829</td>
<td>0.115</td>
<td>0.829</td>
<td>0.245</td>
<td>0.320</td>
<td>0.034</td>
<td>0.047</td>
<td>0.314</td>
</tr>
<tr>
<td>Layer 2</td>
<td>0.805</td>
<td>0.178</td>
<td>0.805</td>
<td>0.165</td>
<td>0.320</td>
<td>0.036</td>
<td>0.047</td>
<td>0.305</td>
</tr>
<tr>
<td>Layer 3</td>
<td>0.795</td>
<td>0.192</td>
<td>0.795</td>
<td>0.154</td>
<td>0.319</td>
<td>0.037</td>
<td>0.046</td>
<td>0.301</td>
</tr>
<tr>
<td>Layer 4</td>
<td>0.723</td>
<td>0.422</td>
<td>0.723</td>
<td>0.079</td>
<td>0.317</td>
<td>0.046</td>
<td>0.046</td>
<td>0.274</td>
</tr>
<tr>
<td>Layer 5</td>
<td>0.655</td>
<td>0.630</td>
<td>0.655</td>
<td>0.057</td>
<td>0.316</td>
<td>0.055</td>
<td>0.046</td>
<td>0.249</td>
</tr>
<tr>
<td>Layer 6</td>
<td>0.662</td>
<td>0.582</td>
<td>0.662</td>
<td>0.061</td>
<td>0.316</td>
<td>0.053</td>
<td>0.046</td>
<td>0.252</td>
</tr>
<tr>
<td>Layer 7</td>
<td>0.624</td>
<td>0.686</td>
<td>0.624</td>
<td>0.053</td>
<td>0.315</td>
<td>0.058</td>
<td>0.046</td>
<td>0.237</td>
</tr>
<tr>
<td>Layer 8</td>
<td>0.515</td>
<td>1.011</td>
<td>0.515</td>
<td>0.038</td>
<td>0.311</td>
<td>0.074</td>
<td>0.045</td>
<td>0.197</td>
</tr>
<tr>
<td>Layer 9</td>
<td>0.113</td>
<td>2.101</td>
<td>0.113</td>
<td>0.013</td>
<td>0.247</td>
<td>0.244</td>
<td>0.046</td>
<td>0.045</td>
</tr>
<tr>
<td>Layer 10</td>
<td>0.112</td>
<td>2.082</td>
<td>0.112</td>
<td>0.013</td>
<td>0.247</td>
<td>0.244</td>
<td>0.046</td>
<td>0.045</td>
</tr>
<tr>
<td>Isotropic</td>
<td>0.3</td>
<td>0.3</td>
<td>0.3</td>
<td>0.10</td>
<td>0.10</td>
<td>0.10</td>
<td>0.137</td>
<td>0.137</td>
</tr>
</tbody>
</table>


\[ e_0 = 7.13 \text{ (void ratio is defined as the fracture of fluid volume to solid phase volume), } M = 7, \text{ and } \kappa = 0.08, \text{ which is in range of reported values for the AC.}^{17} \] 
The correlation coefficient between the model results and experimental data was \( R^2 = 0.998. \)

Figure 3 shows variations of permeability by alterations of the void ratio based on Eq. (13) and computed permeability coefficients. This permeability function was used to define the permeability of DBT and isotropic-biphasic models.

The computed variations of indenter displacement by time for FEM simulation of the DBT model, the isotropic-biphasic model, and experiment data are compared and indicated in Figure 4. According to this figure, isotropic-biphasic model cannot accurately predict the mechanical behavior of AC. The indenter position reaches faster to steady condition and shows a higher displacement value in each time in isotropic-biphasic model than experimental data. The reason of this difference is followed in the role of pore pressure for calculation of indenter displacement.

Figures 5 and 6 compare the calculated variations of void ratio and pore pressure of DBT and isotropic-biphasic models for different points of AC, respectively. These points are shown in Figure 2. Point 1 is located in the middle of layer 1 at the superficial zone and under the indenter. Point 2 is in the middle of cartilage thickness at transition zone and point 3 is at the top of last layer in the deep zone.

According to Figure 5(a) and based on lower axial elastic modulus of cartilage in DBT model in the superficial zone (point 1), calculated void ratio by DBT model is always...
lower than that of the isotropic-biphasic model. However, as it is shown in Figure 5(b) and 5(c), a higher axial elastic modulus of cartilage in DBT model is observed compared to the isotropic model in point 2 of the transient zone as well as in point 3 of the deep zone. In addition, the void ratio calculated by DBT model is always higher than that of the isotropic model. The DBT model predicts that transition and deep zones of cartilage radially expand in applying force period, hence, void ratio becomes greater than its initial value in this period but in the creep period the superficial zone compressed these zones so void ratio decreases.

Figure 6 compares variations of calculated pore-pressure by DBT and isotropic-biphasic models. Pore-pressure calculated in point 1 at the superficial zone and point 2 at the transition zone by DBT model is higher than that of the isotropic model (Figures 6(a) and 6(b)). Nonetheless, pore-pressure calculated in point 3 at the deep zone by DBT model is lower than that of the isotropic model. In addition, isotropic model in all zones sooner reaches to its maximum pore-pressure value in loading period and becomes equal to DBT model at steady value in creep period.

Figure 7 compares the calculated radial stress $S_{rr}$ values of different points of sample by the DBT and isotropic models. According to Figure 7(a), in loading period a radial tensile stress created in bout models in the superficial zone at point 1 and by increasing the applied load the radial stress increases. By starting the creep period, radial stress drops and reaches to a compressive value at the steady condition in the end of creep period for bout models. At all test period, radial stress calculated by the DBT model in point 1 is approximately three times greater than isotropic model. According to Figure 7(b), the tensile radial stress calculated in the transition zone at point 2 from both methods is approximately the same and according to Figure 7(c) tensile radial stress in deep zone at point 3 calculated by isotropic-biphasic model is almost always five times greater than that of DBT model calculations.

The results revealed that the existence of collagen fibers parallel to the cartilage surface in the superficial zone, three times increases the radial stresses generated in this zone. However, as the ultimate strength of collagen fibers is 4-5 times greater than AC, the existence of the collagen fibers in the radial direction reinforces ultimate strength of
the cartilage. Moreover, the presence of collagen fibers perpendicular to tidemark in the deep zone profoundly reduces the radial stresses to a fifth of radial stresses generated in isotropic-biphasic model.

Figure 8 compares the calculated axial stress $S_{zz}$ values of different points of sample by DBT and isotropic-biphasic models. According to Figure 8(a), the axial stress calculated in the superficial zone of point 1 by DBT model is always lower than that of the isotropic model. Figure 8(b) shows that the axial stress induced at point 2 in the transition zone is approximately as same as for bout models during the test period. However, the axial stress created in DBT model at point 3 in the deep zone is approximately 15% higher than isotropic model during test period (Fig. 8(c)).

Due to the orientations of collagen fibers in the transition zone, the stresses induced in all directions are approximately as same as the isotropic-biphasic model. Nevertheless, the results indicated that the existence of collagen fibers in the superficial zone due to their orientations in this zone reduces induced axial stresses, which should be supported by PG matrix which has much less ultimate strength than collagen fibers. In addition, the existence of collagen fibers in the deep zone increases axial stresses in this zone, which should be mainly supported by collagen fibers.

Although an innovative depth dependent biphasic transversely isotropic model has been used in this study to address the mechanical properties of the AC, it is the author’s belief that a more complicated and precise model can be presented using hyperelastic material models, such as Ogden,27–29 Mooney-Rivlin,30–34 Neo-Hookean,35–38 and Yeoh.39–48

IV. CONCLUSIONS

In this study, the material coefficients of permeability function of an AC in creep test were calculated by an innovative depth dependent biphasic transversely isotropic model. This model by considering the role of collagen fibers in all cartilage zones as a function of their volume fractions and orientations has the ability to predict stresses in all directions and depths of the cartilage. It was shown that the presence of collagen fibers increases reaching time to the maximum deflection of indentor, pore-pressure, and all induced stresses.
in all directions and zones in creep indentation test. Although the existence of collagen fibers increases stresses in fiber direction in the superficial and deep zones, they reduce induced stresses perpendicular to the fiber direction in these zones, which supports with PG matrix with much less ultimate strength than the collagen fibers. Since most of stresses in fiber direction support by collagen fibers in the superficial and deep zones, thus existence of collagen fibers increases the ultimate strength of the articular cartilages. Isotropic-biphasic models could not correctly calculate the stress distribution of AC and shall not be used for predicting damage modeling of AC. But the proposed DBT model may have implications not only for developing progressive damage model of AC but also for potential of predicting osteoarthritic cartilage behavior in different cartilage-related diseases.

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