An innovative three-dimensional biphasic-laminated composite model for articular cartilage tissue

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An innovative three-dimensional biphasic-laminated composite model for articular cartilage tissue

MM Elhamian¹,²,³, M Alizadeh¹,², MM Shokrieh¹,³ and A Karimi¹,²

Abstract
Articular cartilage is a three-dimensional (3D) biphasic, fiber-reinforced composite material in which collagen fibers' have key asset in tolerating the applied stresses to cartilage by reinforcing the proteoglycan matrix of articular cartilage, especially in the superficial zone. This study is aimed to establish an innovative model based on 3D microstructure of articular cartilage for anticipating the stress distribution in the cartilage constituents. To do this, articular cartilage is modeled as a 3D biphasic-laminated composite in the superficial zone. Finite element simulation is carried out for further study on the articular cartilage behavior under unconfined indentation test. The reaction force and depth-dependent stress distribution of the presented model are compared with the classic isotropic biphasic model. Although isotropic biphasic model has been basically used to characterize the mechanical behavior of articular cartilage under different loading conditions, the results reveal that it fails to calculate stress distribution in all directions. The results of this study indicate the importance of collagen fibers’ role in reducing the stresses applied to proteoglycan matrix in all cartilage depth and increasing articular cartilage strength. Presented model has the ability to be developed for studying progressive damage of articular cartilage and the effect of collagen fibers orientations in the superficial zone on mechanical behavior of the articular cartilage.

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Keywords
Articular cartilage, 3D-laminated composites, finite element analysis, stress distribution

Introduction
Articular cartilage is a permeable, fluid-filled, and mechanically poroviscoelastic connective tissue which plays a very important role in synovial joints. 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, wear-resistant surface that results in an optimal transmission, bearing, distribution, and absorption of applied loads across the diarthrodial joints (i.e. hip, knee, etc.).

Cartilage contains a large amount of interstitial fluid (70–90 wt%) surrounded in a porous solid media. The solid media is consisted of a progressive structure, including collagen type II fibers (10–20 wt%) and negatively charged proteoglycans (5–10 wt%). The collagen fibers are the fundamental structural elements which constitute a three-dimensional (3D) network that is adjusted to the mechanical loading of the joint, and their orientation varies with depth. Since collagen fibers in the solid phase have different and distinctly directional material properties, the articular cartilage is considered as a composite with anisotropic material properties. An intricate structure of the fibers is being active just under tensile loading, therefore, it invokes an inhomogeneous material with an anisotropic time-dependent behavior.

Based on the collagen fiber arrangement, articular cartilage can be divided into three distinct, morphological zones. Superficial zone starts from cartilage surface and it covers 10–20% of cartilage thickness approximately. Collagen fibers are parallel to surface in this zone and they are randomly oriented and distributed in surface plane. Transition zone is in the middle of cartilage thickness. In this zone, collagen fibers have no preferential orientations. At the top of this zone, collagen fibers’ angle is near 0° toward superficial surface, and this angle reaches to 90° at the end of transition zone. Last zone of articular cartilage is deep zone in which collagen fibers are perpendicular to cartilage surface and tidemark. This zone covers 30% of cartilage thickness approximately and it has the largest aggregate modulus than other zones.

Various theoretical models have been carried out to explain the mechanical behavior of the articular cartilage. The first models were isotropic and elastic and could be applied only to characterize the instantaneous or equilibrium responses of cartilage after a step load application. As a next step, the isotropic biphasic solution could also explain the time dependence of the cartilage mechanical behavior. In addition to the isotropic solution, transversely isotropic biphasic, poroviscoelastic, fibril-reinforced poroelastic or conewise linear elasticity biphasic, random-oriented-laminated composite and depth-dependent transversely isotropic biphasic models exist to improve the accuracy of the theoretical models to predict the cartilage response.

In this study, an innovative 3D biphasic-laminated composite model is presented for predicting mechanical behavior of the articular cartilage. This model has the ability to
consider the role of collagen fibers orientation of the superficial zone on their load bearing in asymmetric geometrics. Mechanical behavior of the articular cartilage under unconfined indentation test is predicted by this model and compared with calculations of isotropic biphasic model. In addition, the role of collagen fibers of superficial zone on load bearing of cartilage is studied.

Materials and methods

3D biphasic-laminated composite method

Collagen fibers in the superficial zone of articular cartilage play dominant mechanical roles in cartilage response under transient compression. 3D biphasic-laminated composite method considers the superficial zone of articular cartilage as a 3D-laminated composite with proper amount of equivalent unidirectional laminas in which collagen fibers in each unidirectional lamina are parallel to cartilage surface and each with a different orientation from 0° to 180° as collagen fibers in laminated composite covers all directions in superficial zone.

Mechanical properties of equivalent unidirectional laminas should be calculated by 3D micromechanics models of laminated composites as a function of elasticity mechanical properties of collagen fibers and proteoglycan matrix and volume fraction of fibers. By considering cartilage surface as \( r - \theta \) plane and perpendicular to cartilage plane as \( z \) direction, the mechanical properties of each equivalent lamina can be calculated as:

\[
E_L = (E_f) \cdot V_f + E_m \cdot V_m
\]

\[
\frac{1}{E_T} = \frac{V_f}{E_f} + \frac{V_m}{E_m}
\]

\[
\nu_{LT} = \nu_f \cdot V_f + \nu_m \cdot V_m
\]

\[
\frac{1}{G_{LT}} = \frac{G_f \cdot G_m}{G_m \cdot V_f + G_f \cdot V_m}
\]

where \( E_L \) and \( E_T \) are the elastic modules of an equivalent unidirectional lamina in longitudinal and transverse to collagen fibers direction, respectively, and \( \nu_{LT} \) and \( G_{LT} \) are Poisson’s ratio and shear modulus in surface plane. \( V_f \) and \( V_m \) are fibers and matrix volume fraction, \( E_f, G_f, \) and \( \nu_f \) are elastic modulus, shear modulus, and Poisson’s ratio of collagen fiber, and \( E_m, G_m, \) and \( \nu_m \) are elastic modulus, shear modulus, and Poisson’s ratio of proteoglycan matrix.

To calculate the Poisson’s ratio perpendicular to fibers plane \( \nu_{TT} \), we have to consider the following issues. By knowing that total cartilage deformation perpendicular to fiber direction is equal to sum of deformation of fibers and matrix and by using the definition of Poisson’s ratio in perpendicular to fibers plane \( \nu_{TT} \), and in addition, by knowing that total cartilage deformation parallel to fiber direction equals to deformation of fibers and
matrix and by using definition of Poisson’s ratio in parallel to fibers plane \( \nu_{LT} \), we can write the following equations:

\[
\nu_{TT} = \frac{V_f \cdot \nu_f \cdot \nu_t + V_m \cdot \nu_m^2}{V_f \cdot \nu_f + \nu_m \cdot \nu_m}
\]  

(5)

By knowing that the Poisson’s ratio in perpendicular to fibers plane, shear modulus in \( r - \theta \) plane would be calculated from Hashin’s equation:\(^48\)

\[
G_{TT} = \frac{E_T}{2(1 + \nu_{TT})}
\]  

(6)

Mechanical properties of collagen fibers and proteoglycan matrix have been reported by several authors.\(^49,50\) Table 1 shows experimental measurements of mechanical properties of collagen fibers and proteoglycan matrix employed for numerical analysis.\(^51\)

Collagen fibers’ volume fraction in the superficial zone was considered as 0.3 and permeability in all zones of cartilage was assumed as 2.787 mm/s.\(^52\) Table 2 shows calculated mechanical properties of equivalent 3D unidirectional lamina. Since collagen fibers are under compression due to their orientation in the transition and radial zones and fibers just have the ability to withstand the tensile stresses, as a result, mechanical properties of these zones were set to proteoglycan matrix.

**FEM simulation of indentation test by presented model**

A 3D sample based on presented model was created to simulate an unconfined indentation test on cylindrical cartilage sample with 0.5 mm thickness and 5 mm diameter. The dimensions of sample and boundary conditions are exactly the same as\(^53\) simulation in which a regular isotropic biphasic model was used. This model also was created in this article to compare the predicted mechanical behavior of two models.

Thickness of the superficial zone of the presented model was assumed as 10% of total cartilage thickness,\(^52\) this zone was divided into 16 layers with equal thicknesses. Each layer was assumed as a 3D unidirectional lamina in which collagen fibers angle in each layer increases 45° counterclockwise than its below layer. Figure 1 shows the schematic of simulated sample and layers of superficial zone. Table 3 shows the direction of collagen fibers in each layer.

The 316L sintered stainless steel \( (E = 190 \text{ GPa}, \nu = 0.28) \) indenter was considered as a porous material with 0.5 mm diameter and a constant permeability which is five orders greater than cartilage initial permeability with 50% porosity.\(^54\) The friction coefficient between indenter and articular cartilage was assumed 0.002. The indenter was prescribed a displacement history as in the first step (1 s), it linearly compress the cartilage with constant speed of 0.05 mm/s. In the second step, indenter stays in its new position for 4 s.

A free draining boundary condition was applied to lateral side of indenter and cartilage and upper surface of cartilage which is not under indenter. Lateral side of
cartilage was free for radial displacement but the bottom line of cartilage was fixed. Since highest deformations take place in 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 layer one compared with deep layers.  

Table 1. Elastic properties of collagen fiber and proteoglycan matrix.

<table>
<thead>
<tr>
<th></th>
<th>Young’s modulus (MPa)</th>
<th>Poisson ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Collagen fiber</td>
<td>10</td>
<td>0.30</td>
</tr>
<tr>
<td>Proteoglycan matrix</td>
<td>0.2</td>
<td>0.05</td>
</tr>
</tbody>
</table>

Table 2. Calculated mechanical properties of different zones of articular cartilage.

<table>
<thead>
<tr>
<th></th>
<th>$E_L$ (MPa)</th>
<th>$E_T$ (MPa)</th>
<th>$\nu_{LT}$</th>
<th>$\nu_{TT}$</th>
<th>$G_{LT}$ (MPa)</th>
<th>$G_{TT}$ (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superficial zone layers</td>
<td>3.1</td>
<td>0.28</td>
<td>0.125</td>
<td>0.23</td>
<td>0.135</td>
<td>0.115</td>
</tr>
<tr>
<td>Transition and radial zones</td>
<td>0.2</td>
<td>0.2</td>
<td>0.05</td>
<td>0.05</td>
<td>0.095</td>
<td>0.095</td>
</tr>
</tbody>
</table>

$E_L$: the elastic module of an equivalent unidirectional lamina in longitudinal to collagen fibers direction; $E_T$: the elastic modules of an equivalent unidirectional lamina in transverse to collagen fibers direction; $\nu_{LT}$: Poisson’s ratio of surface plane; $G_{LT}$: shear modulus in surface plane; $G_{TT}$: shear modulus in r–θ plane; $\nu_{TT}$: Poisson’s ratio of fibers plane.

Table 3. Collagen fibers orientations in each layer.

<table>
<thead>
<tr>
<th>Layers number</th>
<th>Fibers orientation (degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0 45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>2</td>
<td>45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>3</td>
<td>90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>4</td>
<td>135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>5</td>
<td>0 45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>6</td>
<td>45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>7</td>
<td>90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>8</td>
<td>135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>9</td>
<td>0 45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>10</td>
<td>45 90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
<tr>
<td>11</td>
<td>90 135 0 45 90 135 0 45 90 135 0 45 90 135</td>
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<tr>
<td>12</td>
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<td>13</td>
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<tr>
<td>15</td>
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</tr>
<tr>
<td>16</td>
<td>135 0 45 90 135 0 45 90 135 0 45 90 135</td>
</tr>
</tbody>
</table>

Figure 1. Schematic of simulated sample and layers of superficial zone.
Results and discussions

For validating presented model and our calculations, the variations of normalized reaction force of indenter with time predicted by this model were compared with the predictions of isotropic biphasic model\(^5\) and analytical calculations\(^6\) as shown in Figure 2. According to this figure, presented 3D biphasic-laminated composite method predicted the same mechanical behavior as analytical calculations and finite element method simulations based on isotropic biphasic model. But presented model besides predicting mechanical behavior of articular cartilage, by considering the role of collagen fibers and their orientations in superficial zone, has the ability to predict the 3D-distributed stresses in this zone.

According to symmetry of simulated sample geometry and boundary conditions, predicted stress distributions in all layers of superficial zone have the same behavior but the value of stresses in all directions decreases by moving from the superficial surface to cartilage depth. Figure 3 shows stresses distribution along cartilage diameter on the surface of articular cartilage at the end on first step calculated by presented model. According to this figure, maximum stress in all directions established under indenter edge. As according to Figure 3(a), maximum stress in fiber direction \(\sigma_{rr}\) is 0.21 MPa, according to Figure 3(b), maximum stress in \(\theta\) direction \(\sigma_{\theta\theta}\) is 0.02 MPa, and according to Figure 3(c), maximum stress perpendicular to cartilage surface \(\sigma_{zz}\) is 0.175 MPa.

Figure 4 shows stress distribution along cartilage diameter on the surface of articular cartilage at the end on the first step calculated by isotropic biphasic model. Since this model does not see the role of collagen fibers it considers articular cartilage as an isotropic and biphasic material. According to Figure 4(a), maximum stress in fiber direction \(\sigma_{rr}\) calculated 0.06 MPa and according to Figure 4(c), maximum stress perpendicular to cartilage surface \(\sigma_{zz}\) calculated 0.320 MPa.

**Figure 2.** Comparison of normalizes reaction force predicted by presented model, isotropic biphasic model, and analytical calculations.
Comparing the results of these two models, existence of collagen fibers in the superficial zone of articular cartilage increases created stress in fiber directions so that \( \sigma_{rr} \) calculated by 3D biphasic-laminated composite model is 3.5 times greater than \( \sigma_{rr} \) calculated by isotropic biphasic model. But existence of collagen fibers in the superficial

**Figure 3.** Stresses distribution along cartilage diameter on the surface of articular cartilage at the end on first step calculated by presented model (a) \( \sigma_{rr} \), (b) \( \sigma_{\theta\theta} \), and (c) \( \sigma_{zz} \). \( \sigma_{rr} \): maximum stress in fiber direction; \( \sigma_{\theta\theta} \): maximum stress in \( \theta \) direction; \( \sigma_{zz} \): maximum stress perpendicular to cartilage surface.

Comparing the results of these two models, existence of collagen fibers in the superficial zone of articular cartilage increases created stress in fiber directions so that \( \sigma_{rr} \) calculated by 3D biphasic-laminated composite model is 3.5 times greater than \( \sigma_{rr} \) calculated by isotropic biphasic model. But existence of collagen fibers in the superficial
zone reduces created stress perpendicular to cartilage surface so that $\sigma_{zz}$ calculated by presented model is only 55% of $\sigma_{zz}$ calculated by isotropic biphasic model. Since ultimate strength of collagen fibers are four to five times greater than articular cartilage,\textsuperscript{56,57} so existence of collagen fibers in radial direction reinforce ultimate strength of cartilage.

**Figure 4.** Stresses distribution along cartilage diameter on the surface of articular cartilage at the end on first step calculated by isotropic biphasic model (a) $\sigma_{rr}$, (b) $\sigma_{\theta\theta}$ and (c) $\sigma_{zz}$. $\sigma_{rr}$: maximum stress in fiber direction; $\sigma_{\theta\theta}$: maximum stress in $\theta$ direction; $\sigma_{zz}$: maximum stress perpendicular to cartilage surface.
Figure 5 shows pore pressure created along cartilage diameter on the surface of articular cartilage at the end on first step calculated by presented model. According to this figure, maximum pore pressure creates at the center of cartilage disc and is 0.042 MPa which is much less than stresses created in solid phase.

Figure 6 compares variations of stress in different directions of articular cartilage in test period at the edge of indenter on cartilage surface.

Figure 5. Pore pressure created along cartilage diameter on the surface of articular cartilage at the end on first step calculated by presented model.

Figure 6. Variations of stress in different directions of articular cartilage in test period at the edge of indenter on cartilage surface.

Figure 5 shows pore pressure created along cartilage diameter on the surface of articular cartilage at the end on first step calculated by presented model. According to this figure, maximum pore pressure creates at the center of cartilage disc and is 0.042 MPa which is much less than stresses created in solid phase.

Figure 6 compares variations of stress in different directions of articular cartilage in test period at the edge of indenter on cartilage surface. According to this figure, stress created in fiber directions $\sigma_{rr}$ and perpendicular to cartilage surface $\sigma_{zz}$ is approximately
the same during the test period but cartilage is always under tension in fibers direction and is always under compression in perpendicular to cartilage surface direction. Also stresses in all directions reaches to their maximum values at the end of first step period which is when the indenter reaches to its maximum displacement.

Since in indentation test, indenter cross section is 1/100 of cartilage surface it acts as a concentrated force. By moving through cartilage depth, stresses created in all directions of cartilage decreases. Table 4 compares calculated perpendicular stress to cartilage surface $\sigma_{zz}$ by the two models in different layers of articular cartilage at the end of first step period and under the indenter edge.

### Conclusions

3D biphasic-laminated composite method was presented in this article to study the mechanical behavior of articular cartilage. By considering the role of collagen fibers in superficial zone of articular cartilage in this model, stress distribution in different directions of articular cartilage could be study as a function of mechanical properties of cartilage constitutes, collagen fibers volume fraction, and their orientations. Presented model shows that existence of collagen fibers in superficial zone reduces approximately 45% the created stresses in perpendicular to cartilage surface but increases 3.5 times the stresses created in fibers direction, under unconfined indentation test. Since ultimate

<table>
<thead>
<tr>
<th>Distance from cartilage surface ((\mu m))</th>
<th>Presented model</th>
<th>Isotropic biphasic model</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.562</td>
<td>-0.15</td>
<td>-0.26</td>
</tr>
<tr>
<td>4.687</td>
<td>-0.145</td>
<td>-0.26</td>
</tr>
<tr>
<td>7.812</td>
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<td>-0.25</td>
</tr>
<tr>
<td>10.937</td>
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<td>-0.24</td>
</tr>
<tr>
<td>14.062</td>
<td>-0.14</td>
<td>-0.22</td>
</tr>
<tr>
<td>17.187</td>
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<td>-0.22</td>
</tr>
<tr>
<td>20.312</td>
<td>-0.14</td>
<td>-0.21</td>
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<tr>
<td>23.437</td>
<td>-0.135</td>
<td>-0.19</td>
</tr>
<tr>
<td>26.562</td>
<td>-0.135</td>
<td>-0.19</td>
</tr>
<tr>
<td>29.687</td>
<td>-0.135</td>
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<tr>
<td>32.812</td>
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<td>35.937</td>
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<tr>
<td>48.437</td>
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<tr>
<td>50</td>
<td>-0.05</td>
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</tr>
<tr>
<td>500</td>
<td>-0.035</td>
<td>-0.03</td>
</tr>
</tbody>
</table>
strength of collagen fibers are four to five times greater than articular cartilage, calculations of this study show that existence of collagen fibers in superficial zone of articular cartilage reinforces it and it could withstand under greater loading conditions. According to this study, isotropic biphasic model unlike presented model could not correctly calculate the distributing stresses of articular cartilage. But 3D biphasic-laminated composite model has the ability to be developed for studying progressive damage model of articular cartilage and for studying the effect of collagen fibers orientations in superficial zone on mechanical behavior of articular cartilage under asymmetric boundary conditions or geometrics.

**Funding**

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**References**


