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SMM Elhamian, M Alizadeh, MM Shokrieh, A Karimi and SP Madani
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The effect of collagen fiber volume fraction on the mechanical properties of articular cartilage by micromechanics models

SMM Elhamian, M Alizadeh, MM Shokrieh, A Karimi and SP Madani

Abstract

Background: Collagen fiber volume fraction of articular cartilage alters by race and age of samples and has a key role in the mechanical behavior of articular cartilage.

Methods: This study presented innovative micromechanics models to predict the mechanical properties of superficial and deep zones of any kind of articular cartilage for different races and ages as a function of cartilage component mechanical properties and collagen fiber volume fraction.

Results: Collagen fiber volume fraction as a function of normalized depth of an articular cartilage sample was calculated in this study by micromechanics models from variation measurements of its axial elastic modulus. Poisson’s ratio and aggregate modulus was also computed from this function and compared to the experimental data for verification. Since collagen fiber volume fraction was reported in a huge range for different articular cartilages with different races and ages, the effect of this variation is discussed in detail in this paper. The results showed that, as the collagen volume fraction increases, the mechanical properties of articular cartilage in all planes and directions increase as well.

Conclusion: Aging reduces all the elasticity mechanical properties of articular cartilage in all directions and planes.

Keywords

articular cartilage; micromechanics model; fiber volume fraction; mechanical properties; variations of age and race

Introduction

Articular cartilage is a hypocellular, biphasic tissue with a composite solid phase that provides a low-friction, wear-resistant, load-bearing surface in synovial joints and displays characteristic mechanical properties. Articular cartilage is considered to be structurally inhomogeneous and holds anisotropic and non-linear mechanical properties under both compressive and tensile loading. It has been reported that the mechanical properties of full-depth articular cartilage vary with degeneration and aging. Articular cartilage is considered to be structurally inhomogeneous and holds anisotropic and non-linear mechanical properties under both compressive and tensile loading. Articular cartilage is made of a solid matrix which primarily consists of proteoglycans and collagen fibers (mainly type II) and synovial as the fluid phase. Proteoglycans are mainly responsible for the compressive stiffness. Collagens also play an important role in the tensile properties of cartilage. Collagens are proteins that form the fibrillar meshwork providing cartilage with its high tensile stiffness and strength. They are the main structural elements and constitute a three-dimensional network that is adapted to the mechanical loading of the joint. The diameter, volumetric concentration and arrangement of collagen fibers also depend on cartilage depth. Based on the collagen fiber

1 School of Mechanical Engineering, Iran University of Science and Technology, Tehran 16846, Iran
2 Tissue Engineering and Biological Systems Research Laboratory, School of Mechanical Engineering, Iran University of Science and Technology, Tehran 16846, Iran
3 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
4 Department of Medical Sciences, Tehran 14198, Iran

Corresponding author:
Seyyed Mohammad Mehdi Elhamian
School of Mechanical Engineering
Iran University of Science and Technology
Tehran 16846
Iran.
Email: elhamian@iust.ac.ir
arrangement, articular cartilage falls into three different morphological zones. The superficial zone begins from the cartilage surface and covers almost 20% of the cartilage thickness. Collagen fibers in the superficial zone are parallel to surface and they are randomly oriented and distributed in the surface plane. The transition zone is in the middle of the cartilage thickness. In this zone, collagen fibers have no favored orientation. At the top of this zone, the collagen fiber angle is near to zero toward the superficial surface; this angle reaches to 90° at the end of the transition zone. The final zone of articular cartilage is the deep zone in which collagen fibers are perpendicular to the cartilage surface and tidemark. This zone covers almost 30% of the cartilage thickness and it has the largest aggregate modulus. Collagen fibers in the deep zone, and to a lesser extent in superficial zone, play leading mechanical roles in the mechanical response of cartilage under transient compression. The composition and structure of cartilage vary with depth from the articular surface. The thickness, density and alignment of collagen fibers also vary with depth. Classic experimental methods, which rely on controlling or measuring surface displacement or load, do not allow the direct measurement of spatially varying mechanical properties. Developed methods use fluorescently labeled chondrocyte nuclei and epifluorescent video microscopy to measure equilibrium deformation and estimate axial strain within isolated cartilage samples that were subjected to confined compression. Hence, for the sake of simplicity in numerical simulation, the heterogeneous, anisotropic cartilage was approximated as transversely isotropic in our model, assuming axis z to be symmetric and plane r-θ isotropic.

Materials and Methods

Micromechanics model of the deep zone of articular cartilage

Articular cartilage, by orientation of collagen fibers, is divided into three zones. In the deep zone, the collagen fibers are perpendicular to surface; in the transition zone, fiber orientation varies from 90° at the deep zone to 0° at the superficial zone. In the superficial zone, the collagen fibers are randomly oriented, but they are all parallel to surface contact (Figure 1). By the rule of mixture, the effective elastic modulus in a longitudinal fiber direction (\(E_L = E_z\)) and elastic modulus in a transverse fiber direction (\(E_T = E_r = E_θ\)) can be calculated as a function of fiber volume fraction (V_f), the mechanical properties of proteoglycan matrix and collagen fibers. However, these experimental methods could only measure the depth-dependent axial elastic modulus of the articular cartilage.

As the main function of the articular cartilage is load bearing, it is of vital importance to understand its mechanical properties. 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 cone-wise linear elasticity biphasic, hyperelastic, as well as visco-hyperelastic solutions exist to improve the accuracy of the theoretical models to predict cartilage response.

The objective of this study was to determine depth-dependent, three-dimensional mechanical properties of the superficial and deep zones of articular cartilage by micromechanics methods and study the effect of collagen fiber volume fraction on the mechanical properties of these zones.
Where $E_0$, $G_f$, 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. $V_{fr}$ represents collagen fiber volume fraction in this zone (all in z axis direction).

To calculate the Poisson's ratio, which is perpendicular to the collagen fibers' plane $\nu_{\theta \theta}$ in the deep zone, we have to consider the following issues. By knowing that total cartilage deformation perpendicular to fiber plane is equal to the sum of deformation of fibers and matrix and by using the definition of Poisson's ratio perpendicular to the fiber plane $\nu_{z \theta}$, we can write the following equations:

$$\nu_{\theta \theta} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

By knowing that the Poisson's ratio in perpendicular to the fiber plane, shear modulus in the r-θ plane would be calculated from Hashin's equation:

$$G_{r \theta} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \frac{V_{fr} \times V_f^2 + (1 - V_{fr}) \times V_m^2}{V_f \times V_f + V_m \times (1 - V_{fr})}$$

(6)

By knowing that the Poisson's ratio in perpendicular to the fiber plane, shear modulus in the r-θ plane would be calculated from Hashin's equation:

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

**Micromechanics model of superficial zone of articular cartilage**

To calculate the mechanical properties of the superficial zone of cartilage according to the orientation of the collagen fibers in this zone, it was assumed as a random-oriented, continuous strand, mat composite (CSM). The superficial zone of the articular cartilage can be modeled as a transversely isotropic material. Mechanical properties in the fiber plane (r-θ) can be obtained from:

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(3)

$$\left( G_{r \theta} \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)

$$\left( \nu_{r \theta} \right)_{\text{superficial zone}} = \nu_{z \theta} = \nu_f \times V_{fr} + \nu_m \times (1 - V_{fr})$$

(6)

$$\left( E_f \right)_{\text{superficial zone}} = \frac{E_m}{2 \times (1 + \nu_{r \theta})}$$

(7)

$$\left( G_f \right)_{\text{superficial zone}} = \frac{G_f \times G_m}{G_m \times V_{fr} + G_f \times (1 - V_{fr})}$$

(5)
ten layers of equal thickness. Since the first 10-20% of articular cartilage thickness is assumed as the superficial zone and approximately the last 30% of articular cartilage is assumed as the deep zone, therefore, the first layer of the Laasanen\textsuperscript{59} sample is in the superficial zone and last layer is in the deep zone. By equating the axial elastic modulus of these layers with the calculated axial elastic modulus from the micromechanics models of the superficial and deep zones, respectively, collagen fiber volume fractions at the top of superficial zone and at the end of deep zone have been calculated as 0.195 and 0.215, respectively. Assuming linear variation of volume fraction from the superficial zone to deep zone,\textsuperscript{61} collagen fiber volume fraction as a function of normalized depth is determined by:

\[ V_\xi = 0.195 + 0.02 \times \xi \quad (11) \]

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

### Results and Discussion

#### Calculations of different articular cartilage zone thicknesses

Since the axial elastic modulus calculated by the micromechanics model of the superficial zone with calculated collagen fiber volume fractions (equation 11) for second layer is completely different with experimental measurements (more than 50% error), this layer could not be assumed to be in the superficial zone. Thus, the thickness of the superficial zone of the Laasanen\textsuperscript{59} sample calculates 10% of its thickness. But, the axial elastic modulus calculated by the micromechanics model of the deep zone with calculated collagen fiber volume fractions for the ninth layer is approximately similar with that of experimental measurements (less than 9% error), so this layer could be assumed to be in the deep zone. However, the axial elastic modulus calculated by the micromechanics model of the deep zone with calculated collagen fiber volume fractions (equation 11) for the eighth layer is completely different to the experimental measurements (more than 50% error), so this layer could not be assumed to be in the deep zone. Thus the thickness of the deep zone of the Laasanen\textsuperscript{59} sample calculates 20% of its thickness. Figure 2 compares the axial elastic modulus \( E_z \) calculated by the micromechanics models of the superficial and deep zones with the experimental measurements for the different layers of the Laasanen\textsuperscript{59} sample in these zones. According to the results, the remaining layers of the Laasanen\textsuperscript{59} sample (layers 2 to 8) are in the transition zone, which means that 70% of the thickness of this sample between the superficial and deep zone is the region of the transition zone.

### Validations of method and calculations

To verify the micromechanics models used in this study and the calculations for predicting collagen fiber volume fraction function \( V_\xi \), the aggregate modulus in the \( z-\theta \) plane was calculated by these micromechanics models with the fiber volume fraction function \( V_\xi \) predicted from the experimental results of the Laasanen\textsuperscript{59} sample compared by the experimental measurements of the aggregate modulus \( H_{z\theta} \) of Schinagle.\textsuperscript{62} Figure 3 shows the comparison of the aggregate modulus in the superficial and deep zones layers, predicted from the micromechanics models of these zones, to experimental data. Although the mechanical properties of articular cartilage components and collagen fiber volume fraction change by alteration of the age and race of samples, this comparison shows good agreement between the micromechanics models predictions and experimental measurement \( (R^2 = 0.9998) \).

Moreover, Laasanen\textsuperscript{59} measured Poisson's ratio in the \( z-\theta \) plane of different layers of its articular cartilage sample. Figure 4 compares these measurements with the calculated Poisson's ratio in the \( z-\theta \) plane of the superficial
and deep zone layers from the micromechanics models of these zones. According to this comparison, the micromechanics models of these zones can predict other mechanical properties of the superficial and deep zones with good accuracy ($R^2 = 0.9677$).

According to the results, the micromechanics models presented for the superficial and deep zones have the ability to predict elasticity mechanical properties of these zones of articular cartilage as a function of the mechanical properties of cartilage components and collagen fiber volume fraction function $V_f(\xi)$. Besides, these results show that the collagen fiber volume fraction function of different articular cartilages with different age or race, but with similar mechanical properties, are approximately the same.

By considering the mechanical properties of collagen fiber and proteoglycan matrix from Table 1, collagen fiber volume fraction function $V_f(\xi)$ and, using the presented micromechanics models, mechanical properties of the superficial and deep zones of articular cartilage can be calculated. Table 2 shows the calculated mechanical properties of the superficial and deep zones layers of the Laasanen59 sample.

The superficial and deep zones of the articular cartilage have a main role in the mechanical behavior of cartilage,17,63,64 The presented micromechanics models, in consideration of the role of collagen fibers in their zones, could accurately predict the mechanical properties of the superficial and deep zones of articular cartilage based on the orientation of their collagen fibers. However, other models which consider collagen fibers only in tension conditions could not correctly predict the depth-dependent variations of any kind of elasticity mechanical properties of articular cartilage.65,66

**Variations of the mechanical properties of articular cartilage with changes of collagen fiber volume fraction**

Collagen fiber volume fraction varies in different races and ages and has been reported between 0.15 to 0.42.60,67 Therefore, in this section, the variation in the mechanical properties of the superficial and deep zones of articular cartilage with changes of collagen fiber volume fraction has been studied.

Figure 5 compares the elastic modulus in the axial direction $E_z$ and the elastic modulus in the radial direction $E_r$ of the superficial zone in different collagen fiber volume fractions. Figure 6 also compares the elastic modulus in the axial direction $E_z$ and the elastic modulus in the radial direction $E_r$ of the deep zone in different collagen fiber volume fractions. Due to the existence of collagen fibers in parallel with the cartilage surface plane, the elastic modulus in the radial direction $E_r$ is higher than the elastic modulus in the axial direction $E_z$ for any collagen fiber volume fraction (Figure 5). The greatest impact for the variation of collagen fiber volume fraction in the range of its variation in articular cartilage on the elastic modules of the superficial zone is on the radial modulus $E_r$ which increases 2.61 times from $V_f = 0.15$ to $V_f = 0.42$. In the deep zone, because of the collagen fiber orientation which is perpendicular to the tidemark surface, in all volume fractions of the fiber elastic modulus in the axial direction $E_z$ is higher than the elastic modulus in the radial direction $E_r$ (Figure 6). The greatest impact for variation of the collagen fiber volume fraction in the range of its variation in articular cartilage on the elastic modules of the deep zone is on the axial modulus $E_z$ which increases 2.73 times from $V_f = 0.15$ to $V_f = 0.42$.

Figures 7 and 8 compare Poisson’s ratios in different planes of the superficial and deep zones, respectively.
Table 2. Calculated mechanical properties of superficial and deep zones of articular cartilage.

<table>
<thead>
<tr>
<th>Layer number</th>
<th>zone</th>
<th>$\xi (z/H)$</th>
<th>$E_{rr}= E_{|} (\text{MPa})$</th>
<th>$E_{zz}$ (MPa)</th>
<th>$V_{rz}$</th>
<th>$V_{||}$</th>
<th>$V_{|\perp}$</th>
<th>$G_{rz}= G_{|\perp} (\text{MPa})$</th>
<th>$G_{r\theta}$ (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Layer 1</td>
<td>superficial</td>
<td>0-0.1</td>
<td>0.726</td>
<td>0.077</td>
<td>0.099</td>
<td>0.309</td>
<td>0.010</td>
<td>0.037</td>
<td>0.277</td>
</tr>
<tr>
<td>Layer 9</td>
<td>deep</td>
<td>0.8-0.9</td>
<td>0.079</td>
<td>2.158</td>
<td>0.004</td>
<td>0.204</td>
<td>0.103</td>
<td>0.037</td>
<td>0.032</td>
</tr>
<tr>
<td>Layer 10</td>
<td>deep</td>
<td>0.9-1</td>
<td>0.079</td>
<td>2.188</td>
<td>0.004</td>
<td>0.205</td>
<td>0.104</td>
<td>0.037</td>
<td>0.033</td>
</tr>
</tbody>
</table>

both zones, all Poisson’s ratios increase by increasing the collagen fiber volume fraction. If volume fraction is equal to 1, articular cartilage becomes an isotropic material and all Poisson’s ratios become equal to each other and reaches Poisson’s ratio of collagen fiber. In the comparison of Figures 7 and 8, it can be seen that, for equal fiber volume fractions, Poisson’s ratio in the r-z plane of the superficial zone equals Poisson’s ratio in the z-θ plane of the deep zone. For equal fiber volume fractions, Poisson’s ratio in the z-θ plane of the superficial zone is equal to Poisson’s ratio in the r-z plane of the deep zone.

This refers to the definition of Poisson’s ratio and reversing the orientation of collagen fibers in these zones. Poisson’s ratio in the r-z plane of the superficial zone and the z-θ plane of the deep zone increase equally 1.76 times by increasing collagen fibers from 0.15 to 0.42. However, variation of Poisson’s ratio in the z-θ plane of the superficial zone and the r-z plane of the deep zone in this range of fiber volume fractions are approximately constant.

Poisson’s ratio in the r-θ plane in the superficial zone is always higher than in the deep zone for equal collagen. As for fiber volume fractions more than 0.1, Poisson’s ratio of cartilage in the r-θ plane is approximately equal to Poisson’s ratio of collagen fibers.
Figures 9 and 10 compare shear modules in different planes of the superficial and deep zones, respectively. The shear modulus in the r-θ plane of the superficial zone due to the evenness of collagen fibers in all directions of this plane, is higher than the shear modulus in the r-z plane for equal collagen fiber volume fractions (Figure 9). The greatest impact for variation of collagen fibers volume fraction in the range of its variation in articular cartilage on shear modulus of superficial zone is G_{rθ} which increases 2.58 times from V_f = 0.15 to V_f = 0.42. The existence of collagen fibers in the deep zone did not reinforce the shear modulus in any planes, so shear modulus in the fiber plane G_{rz} and perpendicular to the fiber plane G_{rθ} have approximately the same values with the proteoglycan matrix for collagen fiber volume fraction less than 0.8 (Figure 10).

According to this study, the greatest impact of variation of collagen fiber volume fraction in the range of its variation in articular cartilage in the superficial zone are on the radial elastic modulus, Poisson’s ratio on the r-z plane and the shear modulus in the r-θ plane, but, in the deep zone, impacts are on the axial elastic modulus and Poisson’s ratio on the z-θ plane.

**Conclusions**

The micromechanics models presented in this study have the ability to predict the elasticity mechanical properties of the superficial and deep zones of any kind of articular cartilage, with different age and race as a function of elasticity mechanical properties of the collagen fiber and proteoglycan matrix and collagen fiber volume fraction. These models, considering the role of collagen fibers in their zones, could accurately predict the mechanical properties of the superficial and deep zones of articular cartilage based on the orientation of their collagen fibers. The collagen fiber volume fraction of articular cartilage varies with age and race of the samples. The result of this study, computed by micromechanics models, shows that increasing the collagen fiber volume fraction leads to an increase of the axial and radial elastic modules and also increases in Poisson’s ratio and shear modules in all planes of articular cartilage. Therefore, the aging of samples because of the reduction of chondrocyte cell activity leads to a reduction in the generation of collagen fibers, which reduces collagen fiber volume fraction and all the elasticity mechanical properties of articular cartilage. The superficial and deep zones of articular cartilage have the most important roles in mechanical behavior of articular cartilage.

Adding permeability properties of articular cartilage to the presented micromechanics models leads to a transversely isotropic biphasic model which considers the role of collagen fibers in these zones for accurately analyzing the mechanical behavior of articular cartilage.

**Declaration of Conflicting Interest**

The author declares that there is no conflict of interest.

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