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A depth dependent transversely isotropic micromechanic model of articular cartilage

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Abstract Articular cartilage owing to the variation of collagen fibers orientation through its zones has been indicated to have depth dependent mechanical properties. The aim of this study was to present an innovative micromechanics model to predict the depth dependent mechanical properties of articular cartilage as a function of collagen fibers and proteoglycan matrix mechanical properties, collagen fibers volume fraction as well as angle toward cartilage surface. The variation of collagen fibers angle toward the cartilage surface as a function of cartilage depth was computed using the micromechanics model. This function showed that the collagen fibers parallel to the cartilage surface in the superficial zone have a nonlinear angle variation in the transition zone and become perpendicular to cartilage surface in the deep zone. Depth dependent elastic modulus in perpendicular to cartilage surface plane direction was calculated using presented micromechanics model and variation function of the collagen fibers’ angle. The results revealed a suitable agreement with that of the experimental measurements in different samples at different ages and races ($R^2 = 0.944$). The results also showed that the elastic and aggregate modules perpendicular to the cartilage surface plane in the deep zone were 25.8 and 26.3 times higher than that of the superficial zone, respectively. These findings have implications not only for computing the depth dependent mechanical properties of any type of articular cartilage at different ages and races, but also of potential ability for developing a depth dependent transversely isotropic biphasic model to predict the accurate mechanical behavior of articular cartilage.

1 Introduction

Articular cartilage provides a low-friction, wear resistant bearing surface in diarthrodial joints and distributes stresses to underlying bone [1]. Cartilage is functionally highly anisotropic which results in a highly depth-dependent compressive stiffness [2]. The structure and properties of the articular cartilage are optimal for its mechanical function in the joint. Cartilage is a biphasic material consisting of an organic matrix and free interstitial fluid which is mostly water. The most important components forming the solid phase are collagen fibers (mainly type II) and proteoglycans as matrix which consist of collagen fibers [3]. Collagens are the main structural elements and constitute a three dimensional network that is adapted to the mechanical loading of the joint [4, 5].

Based on the collagen fiber arrangement, the articular cartilage can be divided into three distinct morphological zones. The superficial zone starts from cartilage surface and it covers 20 % of cartilage thickness approximately. Collagen fibers are parallel to surface in this zone and they are randomly oriented and distributed in the surface
plane [6]. The transition zone is in the middle of cartilage thickness. It almost covers 50% of cartilage thickness. In this zone, collagen fibers have no preferential orientations. At the top of this zone, collagen fibers angle are near zero toward the superficial surface, this angel reaches to 90° at the end of the transition zone [7]. The last zone of the articular cartilage is deep zone in which collagen fibers are perpendicular to the cartilage surface and tidemark. This zone covers 30% of cartilage thickness approximately [8, 9].

As the main function of articular cartilage is load bearing, it is essential to understand its mechanical behavior. Various theoretical models have been developed to explain the mechanical behavior of the articular cartilage. The first models were isotropic and elastic [10] 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 [11, 12] could also explain the time dependence of the cartilage mechanical behavior. In addition to the isotropic solution, transversely isotropic biphasic [13–15], poroviscoelastic [16], fibril reinforced poroelastic [17] solutions exist to improve the accuracy of the theoretical models to anticipate the cartilage response. However, none of these models consider the variations of mechanical properties of the articular cartilage through its depth. These variations have a direct effect on the mechanical behavior of articular cartilage. Since articular cartilage is a depth dependent transversely isotropic material and these models consider it as an isotropic or utmost transversely isotropic material, these models could not predict the depth dependent variations of stresses or exact values of stresses applied to each component of the articular cartilage which is essential for progressive damage modeling of the articular cartilage to study the osteoarthritic.

The most important factor in depth dependent variations of mechanical properties of the articular cartilages is its collagen fibers orientations. Therefore, this study is aimed to present an innovative micromechanics model to determine the continues variations of mechanical properties of articular cartilage as a function of mechanical properties of collagen fibers, proteoglycan matrix, collagen fibers volume fraction as well as orientation. The present model can successfully predict the continues variations of collagen fibers angle toward the cartilage surface or the variations of their volume fraction besides predicting the depth dependent mechanical properties of the articular cartilage from the superficial zone to the deep zone.

2 Methods

2.1 Mechanical properties of deep zone

According to Fig. 1 [36] articular cartilage was considered as depth dependent transversely isotropic material, assuming axis z to be symmetric and plane r–θ isotropic.

Modified three dimensional unidirectional micromechanics equations of rule of mixture were used to determine the effective elastic moduli in the fiber direction (E_z) and perpendicular to the fiber plane (E_r = E_h), the shear modulus (G_zr) and Poisson’s ratio (v_zr) in collagen fibers plane as a function of fiber volume frictions (V_{fz}) in this zone and the mechanical properties of proteoglycan matrix and collagen fibers.

\[
\langle E_z \rangle_{\text{deep zone}} = E_f \times V_{fz} + E_m \times (1 - V_{fz})
\]

\[
\langle E_r \rangle_{\text{deep zone}} = \langle E_0 \rangle_{\text{deep zone}} = \frac{E_f \times E_m}{E_f \times (1 - V_{fz}) + E_m \times V_{fz}}
\]

\[
\langle v_{zr} \rangle_{\text{deep zone}} = \langle v_{zθ} \rangle_{\text{deep zone}} = v_f \times V_{fz} + v_m \times (1 - V_{fz})
\]

\[
\langle G_{zr} \rangle_{\text{deep zone}} = \langle G_{zθ} \rangle_{\text{deep zone}} = \frac{G_f \times G_m}{G_m \times V_{fz} + G_f \times (1 - V_{fz})}
\]

Fig. 1 Structure of articular cartilage for both cellular and collagen fiber arrangement diagram of in the zones of articular [36]
where $E_r$, $G_r$ 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, respectively. Moreover, $V_{fz}$ represents collagen fibers volume fraction in the deep zone (all in $z$ axis direction) [18–20].

Since the total cartilage deformation in perpendicular to the fiber direction in the deep zone equals to the sum of deformation of fibers and matrix, the Poisson’s ratio in perpendicular to the collagen fibers plane $(\nu_{\theta\theta})$ can be written as follows.

$$t_e \times \nu_{\theta\theta} \times (\varepsilon_z)_c = t_f \times \nu_f \times (\varepsilon_z)_f + t_m \times \nu_m \times (\varepsilon_z)_m,$$  

where $(\varepsilon_z)_c$, $(\varepsilon_z)_f$, and $(\varepsilon_z)_m$ are cartilage, fiber, and matrix strains perpendicular to the fiber direction. In addition, $t_c$, $t_f$ and $t_m$ are standing for their thicknesses. By substituting the definition of Poisson’s ratio in fibers plane in Eq. 6, it can be written as.

$$t_e \times \nu_{\theta\theta} \times v_{zt} \times (\varepsilon_z)_c = t_f \times \nu_f \times t_f \times (\varepsilon_z)_f + t_m \times \nu_m \times (\varepsilon_z)_m,$$  

where $(\varepsilon_z)_c$, $(\varepsilon_z)_f$, and $(\varepsilon_z)_m$ are cartilage, fiber, and matrix strains in the plane perpendicular to the fiber direction. Since these strains are equal, Poisson’s ratio in the plane perpendicular to fiber direction can be driven as Eq. 8.

$$(\nu_{\theta\theta})_{\text{deep zone}} = \frac{V_{fz} \times \nu_f^2 + (1 - V_{fz}) \times \nu_m^2}{\nu_f \times V_{fz} + \nu_m \times (1 - V_{fz})}.$$  

By knowing Poisson’s ratio in the plane perpendicular to fiber direction $\nu_{\theta\theta}$, shear modulus in $r$–$\theta$ plane can be calculated from Hashin’s equation [21].

$$(G_{\theta\theta})_{\text{deep zone}} = \frac{E_r}{2 \times (1 + \nu_{\theta\theta})}.$$  

2.2 Mechanical properties of the superficial zone

To calculate the mechanical properties of the superficial zone by considering the orientation of collagen fibers in its zone, it was assumed as a random oriented continuous strand mat composite (CSM). A layer of composite with randomly oriented fibers can be considered as a laminate with a huge amount of thin unidirectional layers, each with a different orientation from $0^\circ$ to $180^\circ$ [22]. The superficial zone of articular cartilage can be modeled as a transversely isotropic material. Mechanical properties in fiber plane ($r$–$\theta$) can be obtained as

$$E_{CSM} = (E_r)_{\text{superficial zone}} = \frac{E_r^2 + 4E_rG_{LT}A + 2E_rE_T + 8v_{LT}E_TG_{LT}A - 4v_{LT}^2E_T^2 + 4E_TG_{LT}A + E_T^2}{\Delta(3E_L + 2v_{LT}E_T + 3E_T + 4G_{LT}A)}.$$  

$$(G_{CSM})_{\text{superficial zone}} = \frac{G_r + 4G_r + 2G_r + 8G_{LT} + 4G_{LT} + G_{LT}}{8A}.$$  

$$(\nu_{CSM})_{\text{superficial zone}} = \frac{E_L + 6v_{LT}E_T + E_T - 4G_{LT}A}{3E_L + 2v_{LT}E_T + 3E_T + 4G_{LT}A}.$$  

where $\Delta = 1 - v_{LT}v_{TL}$, $E_L$, and $E_T$ are longitudinal and transverse moduli of fictitious unidirectional layer having the same volume fraction as the continuous strand matrix composite and can be obtained from the Eqs. 1 and 2, respectively. In addition, $v_{LT}$ and $v_{TL}$ are obtained from the Eqs. 3 and 4. To determine the effective elastic modulus, shear modulus and Poisson’s ratio perpendicular to the fiber plane ($E_z$, $G_{zr}$ & $\nu_{zr}$), the modified three dimensional unidirectional micromechanics rule of mixture was used [18–20].

$$(E_z)_{\text{superficial zone}} = \frac{E_f \times E_m}{E_f \times (1 - V_{fr}) + E_m \times V_{fr}}.$$  

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

$$(\nu_{zr})_{\text{superficial zone}} = \frac{v_f \times V_{fr} + v_m \times (1 - V_{fr})}{(E_z)_{\text{superficial zone}}}.$$  

where $V_{fr}$ is fibers volume fraction in the superficial zone (all in $r$–$\theta$ plane). These micromechanical methods for predicting mechanical properties of superficial and deep zones of articular cartilage was validated by Elhamian et al. [23] and could be used for articular cartilages with different age and races.

2.3 Mechanical properties of the transition zone

The collagen fiber in the transition zone has a depth dependent angle $\varphi(\xi)$ between $0^\circ$ and $90^\circ$ toward the cartilage surface. We consider each collagen fiber in this zone as equivalent to horizontal and vertical fibers where volume fraction of collagen fibers are equal to sum of volume fractions of horizontal and vertical fibers ($V_f = V_{fr} + V_{fz}$).
Volume fractions of equivalent horizontal and vertical fibers could be determined as a function of fibers angle toward the cartilage surface $\phi$ and collagen fibers volume fraction $V_f$.

$$V_{fr} = \frac{V_f}{1 + \tan\phi}$$

(16)

$$V_{fz} = \frac{V_f \times \sin\phi}{\cos\phi + \sin\phi}.$$  

(17)

Thus, the transition zone of articular cartilage was assumed as a Multi Direction Composite (MDC) in which, horizontal fibers with proteoglycan matrix could be assumed as a CSM composite. The CSM composite encompasses vertical fibers as a transversely isotropic matrix. The schematic of articular cartilage structure is presented in Fig. 2.

Given the above assumption and by knowing that the displacement in the vertical fibers direction of the MDC equals to displacement of vertical fibers and displacement of the CSM composite in transverse to its fibers direction, elastic modulus of the transition zone of articular cartilage in $z$ axis can be obtained as.

$$(E_z)_{\text{transition zone}} = E_f \times V_{fz} + \left[\frac{E_f \times E_m}{V_{fr} \times E_m + (1 - V_{fr}) \times E_f}\right] \times (1 - V_{fz}).$$

(18)

Furthermore, since the displacement in $r$–$\theta$ plane of the MDC composite is equal to sum of the displacement of the CSM composite in its fibers directions and displacement of vertical fibers in transvers to the fibers direction, elastic modulus in $r$–$\theta$ plane of transition zone of the articular cartilage can be determined as following.

$$(E_r)_{\text{transition zone}} = (E_\theta)_{\text{transition zone}} = \frac{E_{CSM} \times E_f}{V_{fr} \times E_{CSM} + (1 - V_{fr}) \times E_f}.$$  

(19)

The shear stress in $r$–$z$ plane of the MDC composite also is equal to shear stresses of vertical fibers and the CSM composite. The displacement caused by this shear stress is equal to sum of the displacement of vertical fibers and displacement of the CSM composite. Shear modulus of transition zone of articular cartilage in $r$–$z$ plane, obtains as:

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

(20)

The displacement in vertical fibers direction of the MDC is equal to displacement of vertical fibers and displacement of CSM composite in transverse to its fibers direction. By using the definition of Poisson’s ratio in $z$–$\theta$ plane, Poisson’s ratio of transition zone of articular cartilage in $z$–$\theta$ plane can be written as:

$$(v_{z\theta})_{\text{transition zone}} = (v_{z\theta})_{\text{transition zone}} = \frac{v_f \times V_{fr} + (1 - V_{fr}) \times [v_f \times V_{fr} + (1 - V_{fr}) \times v_m]}{E_f \times E_m + E_m \times (1 - V_{fr}) + E_f \times V_{fr}}.$$  

(21)

By application of the Eqs. 8 and 9 of modified three dimensional unidirectional rule of mixture, Poisson’s ratio and shear modulus of transition zone of the articular cartilage in $r$–$\theta$ plane, can be obtained as:

$$(v_{r\theta})_{\text{transition zone}} = \frac{V_{fr} \times v_z^2 + (1 - V_{fr}) \times v_{z_{CSM}}^2}{V_{fr} \times v_f + (1 - V_{fr}) \times v_{CSM}}$$

(22)

$$(G_{r\theta})_{\text{transition zone}} = \frac{E_f}{2 \times (1 + v_{r\theta})}.$$  

(23)

If the fibers angle toward cartilage surface $\phi_{r\theta}$ considered to be zero, micromechanics equations presented for determining mechanical properties of transition zone (Eqs. 18–23) would be equal to the presented micromechanics equations to determine the mechanical properties of the superficial zone (Eqs. 10–15). In addition, if the fibers angle toward the cartilage surface considered to be $90^\circ$, the presented micromechanics equations for determining the mechanical properties of transition zone (Eqs. 18–23) would be equal to the micromechanics equations for determining the mechanical properties of deep zone (Eqs. 1–5, 8–9). Therefore, by knowing the mechanical properties of collagen fibers and proteoglycan matrix, depth dependent mechanical properties of each zone of the articular cartilage could be determined as a function of its fibers volume fraction $V_f$ and fibers angle toward the cartilage surface $\phi$ by using micromechanics equations of the MDC model (Eqs. 18–23).
2.4 Depth dependent collagen fiber orientation calculation $\varphi(\zeta)$

The MDC micromechanics model presented in this study enables to predict the depth dependent solid phase mechanical properties of the articular cartilage as a function of mechanical properties of collagen fibers and proteoglycan matrix, collagen fibers volume fraction, and fibers angle toward the cartilage surface. Mechanical properties of collagen fibers and proteoglycan matrix have been reported by several authors [24–27]. Table 1 shows experimental measurements of mechanical properties of collagen fibers and proteoglycan matrix used in this study [28–30].

Experimental measurements of aggregate modulus of articular cartilage in z–0 plane have been reported by Schinagle et al. [2]. To calculate the depth dependent aggregate modulus by the MDC micromechanics model, in addition to mechanical properties of collagen fibers and proteoglycan matrix, depth dependent volume fraction $V_f(\zeta)$ and angle of collagen fibers toward the cartilage surface $\varphi(\zeta)$ is required. Since the angle of collagen fibers in the superficial and deep zones are known to be 0° and 90°, respectively. By using the aggregate modulus from the experimental measurements in these zones, the collagen fibers volume fractions of the superficial and deep zones have been calculated as 0.22 and 0.20, respectively. Assuming a linear variation of the volume fraction from the superficial zone to deep zone [31], depth dependent collagen fiber volume fraction is determined by:

$$V_f = 0.22 - 0.02 \times \zeta,$$

where $\zeta$ (zeta) is a normalized depth ($\zeta = z/H$) and is 0 at the top of superficial zone and it reaches to 1 at the end of the deep zone. The depth dependent collagen fibers angular variation toward cartilage surface $\varphi(\zeta)$ has been calculated by the MDC micromechanics model by having the depth dependent collagen fiber volume fraction function and depth dependent aggregate modulus experimental values.

3 Results and discussion

The collagen fibers angle toward the cartilage surface as a function of normalized depth $\varphi(\zeta)$ was obtained by equating the aggregate modulus calculated from the MDC micromechanics model and experimental aggregate modulus of articular cartilage in each depth. Figure 3 shows variations of collagen fibers angle $\varphi$ toward the cartilage surface by normalized depth $\zeta$. According to this figure, collagen fibers are horizontal in the cartilage surface ($\varphi = 0°$) and until $\zeta < 0.3$, collagen fibers angle toward cartilage surface $\varphi(\zeta)$ is less than 5°. Therefore, by the calculated $\varphi(\zeta)$, this region could be considered as a superficial zone. Variations of collagen fibers angle $\varphi(\zeta)$ in transition zone is completely nonlinear and it reaches to 90° in the deep zone ($\zeta > 0.8$). According to the predicted collagen fibers angle toward the cartilage surface $\varphi(\zeta)$, a first 30 % of the articular thickness from its surface can be assumed as the superficial zone. Then 50 % of the thickness can be assumed as the transition zone and the remaining 20 % of thickness can be considered as the deep zone for articular cartilage sample used in this article [2].

Figure 4 compares the experimental measurements of aggregate modulus by normalized depth with aggregate modulus calculated by MDC micromechanics model.

<table>
<thead>
<tr>
<th>Table 1 Elastic properties of collagen fiber and proteoglycan matrix</th>
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<tr>
<td><strong>Elastic properties</strong></td>
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<tr>
<td>Collagen fiber</td>
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<td>Proteoglycan matrix</td>
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![Fig. 3 Variations of collagen fibers angle toward cartilage surface $\varphi(\zeta)$ by normalized depth](image-url)
In contrast, other models which consider fibers only in tension conditions could not correctly predict the depth dependent variations of aggregate modulus or other mechanical properties of articular cartilage as a function of depth [32, 33].

To validate the results of the MDC micromechanics model and the calculations for predicting collagen fibers angle toward cartilage surface as a function of normalized depth \( u(n) \), elastic modulus in z axis \( E_z \), calculated by the MDC model with \( u(n) \) predicted form the experimental results of Schinagle et al. [2] was compared to the experimental measurements of elastic modulus in z directions of another articular cartilage sample with different age and race reported by Laasanen et al. [34]. Figure 5 compares depth dependent elastic modulus values in z direction predicted from the MDC model to the experimental data. Although the mechanical properties of articular cartilage components, collagen fibers volume fraction, and angle toward cartilage surface \( \varphi(\xi) \) changed by alteration of the age and race of samples, this comparison shows good agreement between the MDC micromechanics model prediction and experiment measurement \( (R^2 = 0.9437) \). According to Fig. 5, the MDC micromechanics model has a profound ability to predict the depth dependent elastic modulus in z direction spatially in superficial and deep zones \( (R^2 = 0.9670 \text{ and } R^2 = 0.9655, \text{ respectively}) \). In addition, variation process of \( E_z \) calculated by the MDC model in the transition zone completely matches the experimental results and the difference in calculated values in this region is due to the differences of collagen fibers angles \( \varphi(\xi) \) and the sensitivity of calculations to this parameter. This comparison besides validating the MDC model indicates that the mechanical properties of articular cartilage is not so much sensitive to cartilage age or race. As in this study the mechanical properties of components of an articular cartilage was used to predict collagen fibers angular variation toward cartilage surface \( \varphi(\xi) \) of another cartilage, these calculations were used to predict the depth
dependent mechanical properties of another articular cartilage. Moreover, a sensitive analysis of articular cartilage age or race on its mechanical properties was studied by Elhamian et al. [35] and it displayed that the depth dependent mechanical properties of Schinagle et al. [2] and Laasanen et al. [34] are nearly the same despite these two studies used different articular cartilages in their experiments.

Figures 6, 7 and 8 show the depth dependent mechanical properties of the articular cartilage predicted by the MDC micromechanics model. According to Fig. 6, elastic modulus of the cartilage perpendicular to the cartilage surface increases from the cartilage surface to the deep zone, as $E_z$ in the deep zone is 25.8 times higher than the superficial zone. This is due to the alteration of collagen fiber orientation $\varphi(z)$ in the cartilage zones. Fibers are in z direction in the deep zone and this would reinforce the cartilage in this direction and increases the elastic modulus in z axis more than that of the superficial zone in which collagen fibers are parallel to the surface and would reinforce the cartilage in radial direction. According to Fig. 6, the elastic modulus calculated in the radial direction ($E_r$) decreases from the superficial to deep zone as $E_r$ in the superficial zone is 10.4 times greater than that of the deep zone.

Figure 7 shows the depth dependent Poisson’s ratio of articular cartilage in $r$–$z$, $r$–$\theta$, and $z$–$\theta$ planes. Figure 8 depicts the depth dependent variations of the shear modulus in $r$–$z$ and $r$–$\theta$ planes ($G_{rz}$ & $G_{r\theta}$). According to Fig. 8, $G_{rz}$ does not have significant changes thorough the depth but $G_{r\theta}$ decreases from the superficial zone to the deep zone by the order of 9.6 times.

According to the experimental measurements of the depth dependent mechanical properties of articular cartilage, elastic modulus of cartilage perpendicular to the cartilage surface $E_z$ in deep zone based on reinforcements of collagen fibers in this direction significantly increases, and the MDC micromechanics model presented in this study owing to considering the role of collagen fibers in all cartilage zones as a function of their volume fractions and angles has the ability to predict this process thoroughly. However, other models which consider collagen fibers only in tension conditions could not correctly

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**Fig. 6** Depth dependent a axial elastic modulus $E_z$ b radial elastic modulus $E_r$ of articular cartilage
predict the depth dependent variations of elastic or aggregate modulus or other mechanical properties of the articular cartilage as a function of depth. The depth dependent transversely isotropic biphasic model based on the MDC micromechanics model was used by Elhamian et al. [35] to predict the articular cartilage mechanical behavior under a creep indentation test and the results showed a great accuracy with that of the experimental data in calculating depth and time dependent stresses whoever other methods failed to predict the stress values correctly.
4 Conclusion

In this study, the variation of collagen fibers angle toward cartilage surface as a function of normalized depth $\varphi_{(z)}$ was calculated by using the MDC micromechanics model and experimental results of aggregate modulus of articular cartilage in various depths. The calculated $\varphi_{(z)}$ shows a nonlinear behavior in transition zone to reach collagen fibers angle from $0^\circ$ in superficial zone to $90^\circ$ in the deep zone. A comparison of elastic modulus in $z$ direction ($E_z$) calculated by this angle function $\varphi_{(z)}$ and experimental measurement of other cartilage samples with different ages or races showed that the presented $\varphi_{(z)}$ has the ability to be used as collagen fibers angle function of other articular cartilage samples with different ages or races to predict their depth dependent mechanical properties. The addition of permeability properties of articular cartilage to the MDC micromechanics model would lead to a depth dependent transversely isotropic biphasic model for analyzing the mechanical behavior of articular cartilage which is simpler than the most proposed theoretical models to analyze the mechanical behavior of articular cartilage and this reduces computational expenses. These findings have implications not only for computing the depth dependent mechanical properties of any type of articular cartilage at different ages and races, but also of potential ability for developing a depth dependent transversely isotropic biphasic model to predict the accurate mechanical behavior of articular cartilage.

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Conflict of interest The authors declared no conflicts of interest.

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