https://doi.org/10.15407/ujpe70.4.244

YU.F. ZABASHTA, M.M. LAZARENKO, L.YU. VERGUN, L.A. BULAVIN

Taras Shevchenko National University of Kyiv,
Faculty of Physics, Department of Molecular Physics
(6, Akademika Glushkova Ave., Kyiv 03022, Ukraine)

INFLUENCE OF MAGNETIC FIELD ON CARTILAGE TISSUE DEFORMABILITY: PHYSICAL MECHANISM

A structural model of cartilage tissue and the mechanism of influence of a magnetic field on the deformability of this tissue have been proposed. Proteoglycans are assumed to form layers whose surfaces are perpendicular to the direction of collagen fibers. The layers and the fibers are connected by adsorbed collagen chains. It has been shown that the magnetic field changes the arrangement of adsorbed chains. This leads to a change in the bending stiffness of the proteoglycan layers and, ultimately, a change in the deformability of the cartilage tissue as a whole. The proposed mechanism is testified making use of a model system. In the framework of this approach, the temperature dependences are obtained for the local shear compliance of a gelatin hydrogel subjected to a magnetic field. The experimental results agree with the proposed mechanism.

K e y w o r d s: cartilage tissue, deformability, magnetic field.

1. Introduction

It is known (see, e.g., [1]) that cartilage tissue performs mainly a supporting function in the processes of human life, by creating the resistance to an external load action. When describing the behavior of a physical system under the action of an external load, the system is usually considered as a continuous medium (continuum). There are various types of continua. One of them is an elastic continuum, for which (see, e.g., [2]) the rheological equation has the form

$$\boldsymbol{\varepsilon} = \boldsymbol{S}\boldsymbol{\sigma}, \tag{1}$$

where $\boldsymbol{\varepsilon}$ is the strain tensor, $\boldsymbol{\sigma}$ is the stress tensor, and $\boldsymbol{S}$ is the compliance tensor. The latter characterizes the ability of the continuum to be deformed.

In this article, we will discuss the influence of a magnetic field on the components of the $\boldsymbol{S}$ tensor of cartilage tissue and the physical mechanism that causes such an influence. As far as we know, this issue has not been covered in the literature. However, in our opinion, its challenging character is associated with the application of magnetotherapy in treating cartilage diseases (see, e.g., [3]). The article is an ideological continuation of works [4, 5] devoted to the study of the cartilage tissue deformability.

2. Ideal Framework of Cartilage Tissue

Here, hyaline cartilage tissue is considered. It is known (see, e.g., [6]) that its main components are water and polymers: collagen (10–12%) and proteo-

C i t a t i o n: Zabashta Yu.F., Lazarenko M.M., Vergun L.Yu., Bulavin L.A. Influence of magnetic field on cartilage tissue deformability: physical mechanism. *Ukr. J. Phys.* **70**, No. 4, 244 (2025). https://doi.org/10.15407/ujpe70.4.244.

glycans (7–8%). It was found [6] that collagen chains are combined into fibers that form a framework. Proteoglycans and water are located in the gaps between the fibers.

The framework model was proposed in works [6, 7]. In an idealized form, this model can be described as a cubic lattice formed by fibers directed along the main axes of inertia of the lattice (Fig. 1). Let us denote the main axes by the numbers 1, 2, 3. Since, by definition, the lattice shown in Fig. 1 has a cubic syngony, the components of the tensor $\boldsymbol{S}$ in the system of principal axes with two-subscript notations form the matrix

$$S \equiv \begin{pmatrix} S_{11} & S_{12} & S_{12} & 0 & 0 & 0 \\ S_{12} & S_{11} & S_{12} & 0 & 0 & 0 \\ S_{12} & S_{12} & S_{11} & 0 & 0 & 0 \\ 0 & 0 & 0 & S_{44} & 0 & 0 \\ 0 & 0 & 0 & 0 & S_{44} & 0 \\ 0 & 0 & 0 & 0 & 0 & S_{44} \end{pmatrix}. \tag{2}$$

The compliance of collagen fibers in the direction of their axes has an order of magnitude of 10^{-11} Pa^{-1}. Taking into account that fibers in cartilage tissue occupy approximately 10% of the volume, we have the estimate $S_{11} \sim 10^{-10}$ Pa^{-1}. Obviously, the other components must have the same order; i.e., for the examined model, the estimate

$$S_{44} \sim 10^{-10}\ \text{Pa}^{-1} \tag{3}$$

should be valid. In experiments (see, e.g., [6]), it is usually obtained that

$$S_{44} \gtrsim 10^{-6}\ \text{Pa}^{-1}. \tag{4}$$

Such a discrepancy forces us to abandon the model depicted in Fig. 1.

When constructing a structural model of cartilage tissue that would be consistent with experimental compliance data, let us turn to the mechanics of porous media (see, e.g., [8]). This domain of mechanics includes the concept of ideal porous medium. The framework of such a medium is a cubic lattice. This lattice can be formed either by rods or by plates. Accordingly, the terms "ideal rod (lamellar) framework" and "ideal rod (lamellar) porous medium" are used.

It is natural to consider rods as a continuous fiber model. From this point of view, the model shown in Fig. 1 is nothing else, but an ideal rod medium. This

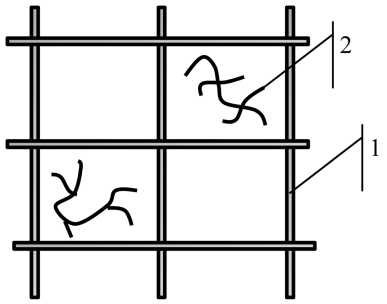

Fig. 1. Model of cartilage tissue: collagen fibers (*1*), proteoglycan chain (*2*)

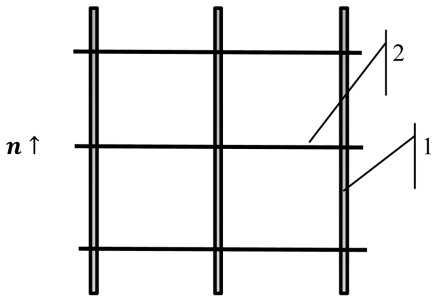

Fig. 2. Model of cartilage tissue with an ideal rod-lamellar framework: collagen fiber (*1*), proteoglycan layer (*2*)

circumstance confirms the appropriateness to use the experience of the mechanics of porous media while studying the deformation properties of cartilage tissue. In work [5], the authors proposed a model for an ideal framework consisting of rods and plates (Fig. 2). In this framework, the plates merge and form layers located at the same distance from each other. The normal $\boldsymbol{n}$ to the surface of those layers is parallel to the rods' axes. We will call such a framework "rod-plate". The unit vector $\boldsymbol{n}$ will be called the director. As was mentioned, rods are collagen fibers. Therefore, proteoglycans are located exclusively in the layers.

The lattice shown in Fig. 2 has a tetragonal syngony. Consequently, in this case, the components of the compliance tensor $\boldsymbol{S}$ form the matrix

$$S \equiv \begin{pmatrix} S_{11} & S_{12} & S_{13} & 0 & 0 & 0 \\ S_{12} & S_{11} & S_{13} & 0 & 0 & 0 \\ S_{13} & S_{13} & S_{33} & 0 & 0 & 0 \\ 0 & 0 & 0 & S_{44} & 0 & 0 \\ 0 & 0 & 0 & 0 & S_{44} & 0 \\ 0 & 0 & 0 & 0 & 0 & S_{66} \end{pmatrix}. \tag{5}$$

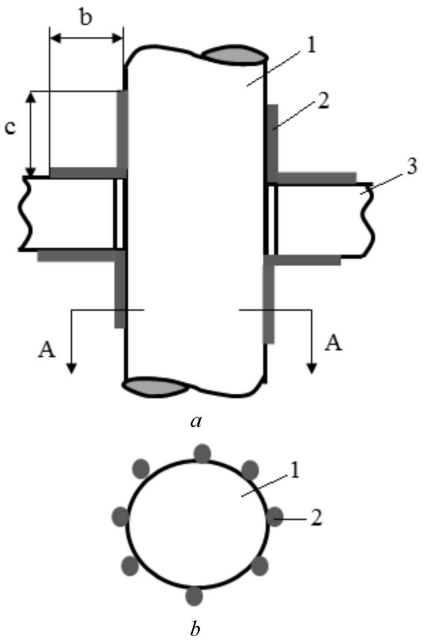

Fig. 3. Structure of a node of the ideal framework: cross-section along the fiber axis (*a*), cross-section along A-A (*b*): fiber (*1*), connecting chain (*2*), layer (*3*)

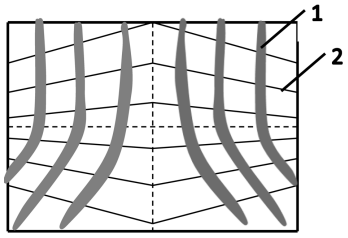

Fig. 4. Model of cartilage tissue with a real rod-lamellar forward: collagen fiber (*1*), proteoglycan layer (*2*)

According to estimate (4), the following relationships should hold for cartilage tissue:

$$S_{44} \gg S_{11},\, S_{12},\, S_{33},\, S_{13},\, S_{66}. \tag{6}$$

Hereinafter, we will call S_{44} the shear compliance.

Let h denote the layer thickness, and let D denote the fiber diameter. Obviously, relationship (6) requires the fulfillment of the inequality

$$h \ll D. \tag{7}$$

It means that the axes of the chains that form the layers should be located mainly in the plane perpendicular to the vector $\boldsymbol{n}$.

3. Nodes of Ideal Framework

The matter concerns the nodes in the lattice depicted in Fig. 2. According to this figure, each such node is a spatial region, where the fiber is connected to the layer.

Let us consider a possible variant of the node structure (Fig. 3). In this variant, the connection between the fiber and the layer is realized via adsorbed collagen chains. On an interval of length c, each such chain is connected to a fiber. Then the other part of the chain (of length b) is connected to the layer.

The connecting chains form two transition zones: upper and lower. As we can see from Fig. 3, the connecting chains are actually those chains that are adsorbed on the fiber and layer surfaces. This circumstance allows us, in this case, to use the approach used in the theory of adsorption (see, e.g., [9]). Namely, we represent the transition zone as a two-phase system. In this case, phase A is formed by the links of the adsorbed chains in contact with the layer, and phase B is formed by the links that are in contact with the fiber. Obviously, the indicated phases are two-dimensional.

In the framework of this approach, the thermodynamic potential Φ of the transition zone can be written as follows:

$$\Phi = \sigma_{\mathrm{A}}\pi\,(D+2b) + \sigma_{\mathrm{D}}\pi D + \sigma_{\mathrm{B}}\pi D + \mu_{\mathrm{A}}N_{\mathrm{A}} + \mu_{\mathrm{B}}N_{\mathrm{B}}, \tag{8}$$

where $(\mu_{\mathrm{A}}, \mu_{\mathrm{B}})$ and $(N_{\mathrm{A}},\, N_{\mathrm{B}})$ are the chemical potentials, and the number of units forming phases A and B, respectively; and σ_A, σ_{D}, and σ_{B} are the linear tensions at the outer boundary of phase A, at the boundary separating both phases, and at the outer boundary of phase B, respectively.

4. Real Framework of Cartilage Tissue

In reality, the axes of the fibers making up cartilage tissue are not parallel to one another over the whole volume occupied by this tissue. This volume can be imagined as a set of regions, where the mentioned parallelism is approximately held. The directions of the director $\boldsymbol{n}$, generally speaking, are different for different regions. In this case, the framework (we will call it real) takes the form shown in Fig. 4. The indicated regions will be called blocks. The boundaries of

the blocks in Fig. 4 are shown by dashed lines. Let l denote the block size, and let $\boldsymbol{r}^{(j)}$ be the radius vector that describes the position of the j-th block. For this block, the notation $\boldsymbol{n}^{(j)}\left(\boldsymbol{r}^{(j)}\right)$ means its director, and $\boldsymbol{S}^{(j)}\left(\boldsymbol{r}^{(j)}\right)$ and $\chi^{(j)}\left(\boldsymbol{r}^{(j)}\right)$ its compliance and magnetic susceptibility tensors.

Considering l as an infinitesimal small quantity, let us change from the discrete quantity $\boldsymbol{r}^{(j)}$ to a continuous one $\boldsymbol{r}$, by writing down the approximate equality

$$l \approx |d\boldsymbol{r}|, \tag{9}$$

thus passing to the functions of continuous argument $\boldsymbol{n}(\boldsymbol{r})$, $\boldsymbol{S}(\boldsymbol{r})$, and $\chi(\boldsymbol{r})$. The values of these functions at the point $\boldsymbol{r}$ will be called the local values of the corresponding quantities.

5. Magnetic Field and the Structure of Cartilage Tissue

Starting from a certain time moment, let the cartilage tissue be subjected to the action of an external magnetic field with the intensity $\boldsymbol{H}$. As a reaction to this action, a magnetization field $\boldsymbol{M}(\boldsymbol{r})$ should arise in the cartilage tissue; it is determined by the formula

$$\boldsymbol{M}(\boldsymbol{r}) = \boldsymbol{\chi}(\boldsymbol{r}) \cdot \boldsymbol{H}. \tag{10}$$

For the free energy density f of cartilage tissue subjected to the magnetic field action, we have the formula

$$f = f_0 - \int\limits_0^H \boldsymbol{M} \cdot d\boldsymbol{H}, \tag{11}$$

where f_0 is the f-value in the magnetic field absence.

Let us introduce the notations $\chi_\parallel$ and $\chi_\perp$ according to the equalities

$$\boldsymbol{M} = \chi_\parallel \boldsymbol{H} \quad (\boldsymbol{M} \| \boldsymbol{n}), \tag{12}$$

$$\boldsymbol{M} = \chi_\perp \boldsymbol{H} \quad (\boldsymbol{M} \perp \boldsymbol{n}). \tag{13}$$

Using the introduced notations, formula (10) takes the form [10]

$$f = f_0 - \frac{1}{2}\chi_\perp H^2 - \frac{1}{2}\left(\chi_\parallel - \chi_\perp\right)(\boldsymbol{n} \cdot \boldsymbol{H})^2. \tag{14}$$

Together with the field $\boldsymbol{M}(\boldsymbol{r})$, a field of torques should arise in the cartilage tissue. As a result, each block is subjected to the action of the torque

$$\boldsymbol{m}(\boldsymbol{r}) = \upsilon \boldsymbol{M}(\boldsymbol{r}) \times \boldsymbol{H}, \tag{15}$$

where $\upsilon \approx l^3$ is the block volume. This torque tries to rotate the block into the position $\boldsymbol{n}(\boldsymbol{r}) \| \boldsymbol{H}$, where the block free energy $F = f\upsilon$, as can be seen from formula (14), equals

$$F = \upsilon\left(f_0 - \frac{1}{2}\chi_\parallel H^2\right). \tag{16}$$

The rotation of the block as a whole is accompanied by a substantial deformation of the material surrounding the block, and there arise the considerable resistance forces that inhibit this rotation. Cartilage tissue is classified as a diamagnetic substance, for which the values of the tensor components χ are relatively small. Therefore, it is unlikely that torque (15) will overcome the aforementioned resistance forces. Therefore, we may state that the blocks remain motionless under the action of the magnetic field.

As can be seen from formulas (10), (11), and (14), besides the rotation of the blocks, there is another factor that can change the free energy of the block after the application of a magnetic field. This is the existence of the dependence

$$\chi = \chi(\boldsymbol{H}). \tag{17}$$

We assume that such a dependence is caused by the properties and the behavior of the adsorbed chains.

Let Φ_0 denote the thermodynamic potential of the transition zone in the magnetic field absence. The adsorbed chains that compose the transition zone are in contact with water. Therefore, their links, like the links of any other protein chains (see, e.g., [14]), are ionized. Obviously, owing to the action of the magnetic field on the links, the chemical potentials of those links change. For example, let those changes lead to the inequality $\mu_{\mathrm{A}} < \mu_{\mathrm{B}}$. Now, the initial state, which corresponds to the value Φ_0 of the thermodynamic potential, becomes non-equilibrium, and the transitional system tries to pass into a new equilibrium state, which corresponds to the thermodynamic potential value $\Phi_1 < \Phi_0$. The transition into this state, owing to the inequality $\mu_{\mathrm{A}} < \mu_{\mathrm{B}}$, should be accompanied by an increase in the number of links N_{A} that enter phase A. As one can see from Fig. 3, such an increase occurs due to the "creeping" of adsorbed chains from the fiber onto the layer; as a result, the size b increases, and the size c decreases. In other words, owing to the

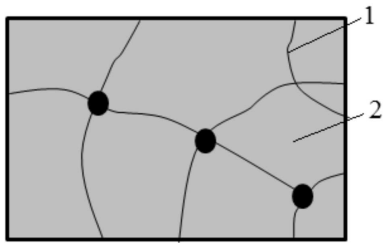

Fig. 5. Polymer network in hydrogel: chain (*1*), water (*2*)

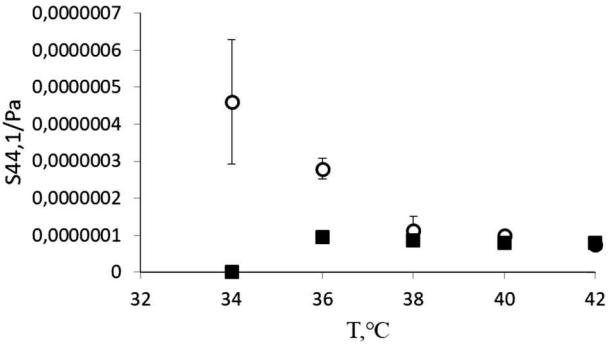

Fig. 6. Temperature dependences of the shear compliance S_{44}: under the magnetic field action (■), in the magnetic field absence (○)

mentioned creeping, the amount of polymer material increases in the direction perpendicular to the director $\boldsymbol{n}$, and decreases in the direction parallel to $\boldsymbol{n}$. In turn, such a rearrangement changes the values of $\chi_{||}$ and $\chi_{\perp}$, i.e., induces the dependence $\chi = \chi(H)$.

6. Cartilage Tissue as a Polymer Hydrogel

Currently, the study of materials called polymer hydrogels (see, e.g., [12]) is intensively developed. There are two definitions for the mentioned class of materials. According to the first definition (see, e.g., [13]), the polymer hydrogel is a "polymer–water" system, where polymer chains form a network (Fig. 5). According to the second definition (see, e.g., [14]), the polymer hydrogel is a porous medium "polymer–water" where the pores are filled with water. It is obvious that the given definitions do not contradict each other; the former concerns the molecular structural level, and the latter the continuous one.

In view of the latter definition, cartilage tissue is classified as a polymer hydrogel (see, e.g., [7]). Accordingly, following this logic, the model of the rod-lamellar framework, which was introduced for cartilage tissue (Fig. 2), should remain valid for the entire class of hydrogels. The polymer network shown in Fig. 4 also finds its place in this model. Such a network forms layers of the rod-lamellar framework. Since the thickness of such layers is relatively small, the chains in the network should be arranged in parallel to the layer surface.

7. Experimental Part

The aim of our experiment was to verify the reality of the proposed mechanism, which is responsible for the influence of a magnetic field on the cartilage tissue structure. The complexity of the cartilage tissue structure puts on the agenda the problem of using model systems in studying the properties of this tissue. When choosing a model system, we will proceed from the above-stated assumption about the applicability of the rod-lamellar framework model for both cartilage tissue itself and the corresponding model system.

It was shown above that the effect of a magnetic field on the cartilage tissue structure is practically reduced to the reorganization in the transition zone consisting of collagen chains. Therefore, the model system should be a hydrogel, where the adsorbed chains are collagen. Following this logic, gelatin hydrogel was chosen as a model system.

It was also shown above that, under the magnetic field action, the connecting chains "crawl" along the layer surface, which changes the layer stiffness. In work [5], it was found that the local value of the hydrogel shear compliance S_{44} is determined by the layer stiffness. This result leads to the following experimental setup: it is necessary to measure the local shear compliance S_{44} of gelatin gel subjected to the magnetic field action. If the magnetic field approximation changes the value of S_{44}, then the above-mentioned mechanism of the magnetic field effect is real.

In this work, we used food gelatin (bloom 200). The experiment was performed on a torsional pendulum according to the method described in work [15]. The shear modulus G was determined. Two types of samples were used. Samples of the first type were prepared as follows. Gelatin was added to distilled water at a temperature of 70 °C in an amount that pro-

vided a solution concentration of 12%. The resulting liquid system "water-gelatin" was permanently stirred to ensure homogeneity. After this liquid system was cooled down to room temperature, it was used to fill cylindrical polyethylene cuvettes, which were kept for a day at a temperature of 18–20 °C before measuring the shear modulus.

Samples of the second type were prepared as follows. A homogeneous liquid system "water-gelatin" was held at a temperature of 40 °C in a dc magnetic field with an induction of 1 mT for 30 min. After that, the liquid system was cooled down, and empty polyethylene cuvettes were filled with it (the cuvettes were also preliminarily kept for 30 min in a dc magnetic field with an induction of 1 mT). As in the first case, the cuvettes together with the examined system were kept for a day at a temperature of 18–20 °C before the experiments started.

As was shown in work [5], the local shear compliance S_{44} and the local shear modulus G of cartilage tissue are related by the formula

$$S_{44} = \frac{5}{2G}. \tag{18}$$

In Fig. 6, the temperature dependences of the shear compliance S_{44} of initial gelatin hydrogel and gelatin hydrogel after its treatment with a magnetic field are shown. The analysis of Fig. 6 demonstrates that the magnetic field changes the local shear compliance of hydrogel. This result is consistent with the proposed mechanism.

In the presented particular case, the value of S_{44} decreases. From the viewpoint of the considered mechanism, such a behavior means that the connecting chains "creep" mainly onto the layer surface under the magnetic field action and increase the layer stiffness.

8. Conclusions

From the thermodynamic viewpoint, cartilage tissue is a heterogeneous continuum consisting of regions (blocks) with different deformability values. Proteoglycans form a set of layers in each block. Fibers and layers are connected by collagen chains. Such chains consist of two sections: one is connected to the layer, and the other to the fiber. These connecting chains form two transition zones around each "layer-fiber" contact.

A magnetic field, by interacting with the electric charges in the adsorbed chain links, changes the thermodynamic potential of the transition zone. Thermodynamically more favorable becomes the configuration of adsorbed chains in which the length of the section connected to the layer changes in comparison with the value that was in the magnetic field absence. Changing the length of these sections gives rise to a change in the bending stiffness of the layers. It is this stiffness that determines the deformability of cartilage tissue.

1. O.D. Lutsyk *et al. Human Histology* (Knyga Plyus, 2010) (in Ukrainian).
2. L.A. Bulavin, Yu.F. Zabashta. *Physical Mechanics of Polymers* (Kyiv University Publishing Center, 1999) (in Ukrainian).
3. O.S. Bur'yanov, T.M. Omelchenko. *Osteoarthrosis* (LENVIT, 2009) (in Ukrainian).
4. L.A. Bulavin, K.I. Gnatiuk, Yu.F. Zabashta, O.S. Svechnikova, V.I. Tsymbalyuk. Shear modulus and elasticity of cartilage tissue. *Ukr. J. Phys.* **64**, 277 (2022).
5. Yu.F. Zabashta, V.I. Kovalchuk, O.S. Svechnikova, L.Yu. Vergun, L.A. Bulavin. Deformation and structure of cartilage tissue. *Ukr. J. Phys.* **69**, 329 (2024).
6. J. Eschweiler *et al.* The biomechanics of cartilage–An overview. *Life* **11**, 302 (2021).
7. F. Horkay, P. Basser. Composite hydrogel model of cartilage predicts its load-bearing ability. *J. Sci. Rep.* **10**, 8103 (2020).
8. N.C. Hilyand. *Mechanics of Cellular Plastic* (Applied Science Publisher LTD, 1982).
9. Y.J. Frenkel. *Kinetic Theory of Liquids* (Clarendon Press, 1946).
10. P.G. de Gennes. *The Physics of Liquid Crystals* (Clarendon Press, 1979).
11. A.R. Khokhlov, A.Yu. Grosberg, V.S. Pande. *Statistical Physics of Macromolecules* (American Institute of Physics, 1994).
12. M. Karg *et al.* Nanogels and microgels: From model colloids to applications, recent developments, and future trends. *Langmuir* **35**, 6231 (2019).
13. P.G. de Gennes. *Scaling Concepts in Polymer Physics* (Cornell University Press,1979).
14. V.C. Mow, S.C. Kuei, W.M. Lai, C.G. Armstrong. Biphasic creep and stress relaxation of articular cartilage in compression: Theory and experiments. *J. Biomech. Eng.* **102**, 73 (1980).
15. L.A. Bulavin, Yu.F. Zabashta, L.Yu. Vergun, O.S. Svechnikova, A.S. Efimenko. Boundary layers and shear elasticity of the "collagen-water" system. *Ukr. J. Phys.* **64**, 34 (2019).

Received 02.10.24.
Translated from Ukrainian by O.I. Voitenko

Ю.Ф. Забашта, М.М. Лазаренко,
Л.Ю. Вергун, Л.А. Булавін

ВПЛИВ МАГНІТНОГО ПОЛЯ
НА ДЕФОРМАТИВНІСТЬ ХРЯЩОВОЇ
ТКАНИНИ: ФІЗИЧНИЙ МЕХАНІЗМ

Пропонується структурна модель хрящової тканини і механізм впливу магнітного поля на її деформативність. Вважається, що протеоглікани утворюють шари, поверхня яких перпендикулярна напрямкові колагенових волокон. Шари і волокна з'єднуються адсорбованими колагеновими ланцюгами. Показано, що дія магнітного поля змінює розташування адсорбованих ланцюгів. Це приводить до зміни вигинної жорсткості протеоглікановых шарів і, в кінцевому підсумку, до зміни деформативності хрящової тканини в цілому. Перевірка реальності запропонованого механізму проводилася на модельній системі. В рамках такого підходу отримані температурні залежності локальної зсувної податливості для желатинового гідрогеля, який зазнав дію магнітного поля. Результати проведеного експерименту узгоджуються із запропонованим механізмом.

Ключові слова: хрящова тканина, модуль зсуву, магнітне поле.