

Material model for aortic artery bioprostheses

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Abstract

Purpose

The aim of the analysis is to develop design principles for a new material having properties similar to those of the natural aortic artery. This involves replacing the complex structure of the aortic wall with a new material with a layer-composite structure having the same strength and hemodynamic properties. The structure of the material used to construct the new aortic prosthesis consists of three layers. The fibers in inner layer are embedded in a liquid matrix, which does not degrade or change its properties in contact with the moving fiber.

Methods

The FEM was used to develop the strength properties of the new material. Constitutive equations were defined to relate the state of stress and the state of strain in the material. Based on the results of the identification process, a material specimen was prepared. Due to the orthotropic properties of the material. In the experimental studies, a specimen developed for the circumferential direction was tested.

Results

In the circumferential direction, the Young's modulus was 1090 kPa, and the fiber shape factor was 0.056. In the axial direction, the Young's modulus was 440 kPa, the fiber shape factor was 0.067.

Conclusions

The paper presents the process of optimizing the material model of a new bioprosthesis, which mechanically imitates the natural material of the aorta. A simple fiber structure was immersed in a liquid matrix and described using basic material parameters. This approach allows to obtain a material with non-linear characteristics and high compliance.

Keywords: finite element method, aortic artery, composite-layer structure

Introduction

Among the diseases of the circulatory system, one of the most prevalent is a group of diseases affecting large arteries, including the ascending aortic artery, aortic arch and abdominal aorta. In 60% of cases concerning the aorta, its dissection occurs [2]. Aortic dissection is a condition in which we observe separation of the layers of the aortic wall, which can lead to serious complications. Dissection often requires surgical intervention using an aortic prosthesis, especially when it involves the ostia of the coronary arteries and the aortic valve. Another disease resulting in the implantation of an aortic prosthesis is obliterative atherosclerosis of the arteries. It can lead to significant narrowing of the aortic orifice, which results in ischemia of organs and tissues. In such cases, it is necessary to use an aorto-femoral

prosthesis to restore normal blood flow [4][12]. It is estimated that these diseases occur in about 4.2–8.5% of the world's population. The choice of the appropriate surgical method depends on the individual characteristics of the patient and the characteristics of the disease, but a large part of it results in the implantation of an aortic artery prosthesis [13].

There are three types of aortic artery grafts, either in the form of animal tissue (heterografts) or human tissue (homografts). The material used reduces the risk of blood clots and does not require ongoing anticoagulant therapy. Their lifetime is, however, finite, and therefore they may require further replacement and more extensive diagnostics. Depending on the patient's condition as well as the nature of the disease, the most widely utilized forms of aortic prostheses are Dacron implants. Dacron is a long-lasting synthetic. Dacron is utilized to create aortic artery prosthesis sheaths. Dacron is very durable in nature, which makes aortic artery prostheses durable and resistant to wear and tear. Aortic artery prostheses do not often cause allergic reactions or rejection by the patient's system. It is less prone to infection and does not need anticoagulant therapy post-implantation [11][5].

Despite the fact that Dacron aortic prostheses possess many ideal features and characteristics, there are certain factors making this treatment stand out from the natural aortic artery whose position it takes. This prosthesis is characterized as being low compliance, which consequently leads to interference with the hemodynamic cycle of the heart. When blood is expelled, part of its volume is not occupied by the wall of the hard prosthesis and does not keep exerting pressure within the arterial system during the transient interruption of the work of the heart. This might cause the organs to function inadequately.

The very high stiffness of the Dacron prosthesis in the circumferential as well as axial direction at small deformations brings about an increase in afterload, which will be deleterious in most heart diseases, e.g. aortic valve regurgitation. Its growth in the early phase does not cause clinical symptoms, and rigid Dacron prosthesis implantation can cause a rise in regurgitation and then lead to backward flow into the left ventricle, altering the functioning of the entire heart [3].

Dacron prostheses, as an artificial organ to be implanted in the ascending aorta

location, are not ideally profiled and do not smoothly insert themselves in the aortic arch. These need to be trimmed and fashioned in the operating room such that they can be successfully fitted in the existing portion of the circulatory system. Clots form in the attachment point, which, when dislodged, can lead to strokes and other post-operative complications.

Another limitation is the possibility of this prosthesis adhering to the chest sternum, essentially closing the chest permanently to future operations. Sternum wires may damage the vascular grafts surrounding them and lead to life-threatening complications such as the formation of pseudoaneurysms. Therefore, in patients with dilated or replaced ascending aorta, prophylactic measures such as the use of sternum bands and the placement of a protective device between sternum wires and aortic grafts are recommended [1].

The second but less popular material used for the construction of aortic prostheses is expanded polytetrafluoroethylene (ePTFE). ePTFE is commonly used in aortic artery prosthesis, especially emerging techniques such as EVAR and TEVAR. PTFE, in relation to the natural aorta, has reduced elasticity but is characterized by excellent mechanical strength and resistance to degradation. In contrast to the native aortic wall, that represents living dynamic tissue and can respond to pressure changes, an ePTFE graft is idle what effects hemodynamics [8]. Its microporous nature however encourages tissue ingrowth while its low porosity decreases the risk of endoleaks in comparison with a diseased or aneurysmal aorta. This renders our atomic glow described here as a robust and stable feature of ePTFE.

The goal of the analysis is to define properties for a new material with the same characteristics as the aortic artery in its natural state. The aim of this analysis is to substitute the complex aortic wall structure with a new composite layered material of equivalent strength and hemodynamic properties. To this end the aortic artery material was identified.

Materials and Methods

The structure of the aortic artery is complex and comprises several layers and structures:

- **intima:** It is the innermost wall of the ascending aorta that directly interacts with the blood. It consists of one layer of vascular epithelial cells, which provides a smooth surface with minimum flow resistance.
- **media:** It is the middle layer of the artery, which is mainly made up of smooth muscles and elastic fibers. This arrangement allows the artery to be elastic and expandable and contractible according to the cardiac cycle. Owing to its pulsatile operation, the artery can efficiently supply blood to every part of the body. From a hemodynamic perspective, it is the most important layer. The same solutions are used in industry in the form of an accumulator of potential energy [6][15].
- **externa:** It is the external layer of the artery, which is composed of mostly connective tissue. This layer's purpose is to help retain the cylindrical form of the aorta and also hold the artery in position within the body.

The basis of this work are the results of studies on natural aortic artery material [10]. The technique applied in this paper was to cut out fragments of the aortic artery wall and place them between two glass layers. Treated preparations were immobilized in the measuring device with surgical threads in two directions tested: circumferential and axial. Stress-strain characteristics were measured and exhibited orthotropic properties of the material. Results are presented in Figure 1.

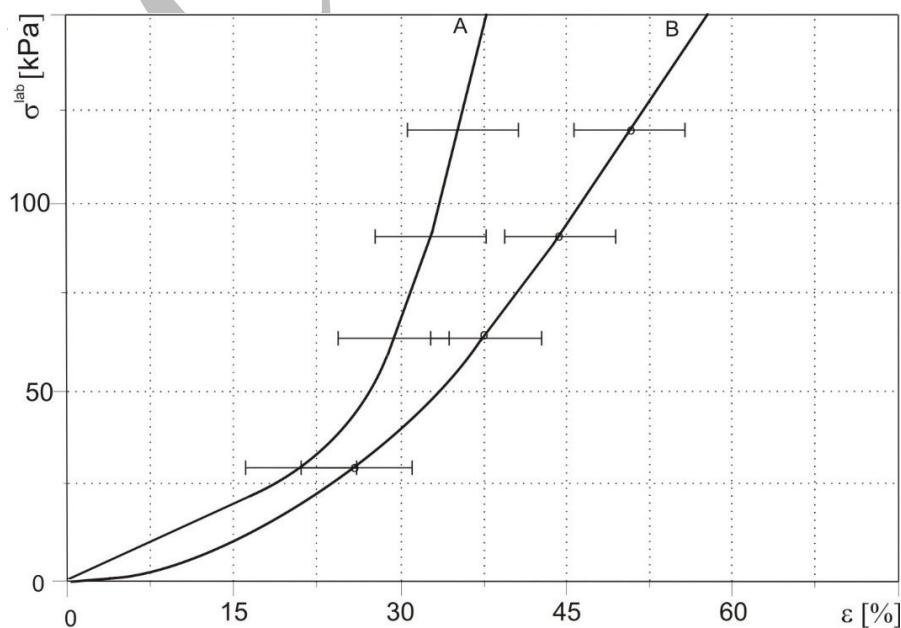


Figure 1. Results of uniaxial stretching tests of specimen of the ascending aortic artery wall. A – circumferential direction. B – axial direction [10].

The preparations were defined to loads corresponding to conditions resulting from the range of systolic and diastolic blood pressure (80 and 120 mmHg), and conditions resulting from potential diseases resulting in increased pressure, e.g. hypertension. In the study, 200 mmHg was assumed [10].

The farther part of the article focuses on the concept of a novel layered-composite material that can be used to produce an aortic artery bioprosthesis. The material is biocompatible by its nature with artificial organ-like properties, but it is a layered-composite material with natural aortic artery-like strength.

Material structure used to produce the novel aortic artery prosthesis consists of three layers.

- a) The external layers create a low stiffness flexible matrix capable of maintaining the given cylindrical shape of the organ. The layer also provides room for liquid matrix of the internal layer.
- b) The internal layer - an important part of this layer consists of fibers oriented in two directions perpendicular to one another: circumferential and axial. The role of the fibers is to introduce the right stiffness of the entire structure and to give the new material the non-linear stress-strain behavior characteristic of tissues building various elements of the vascular system and heart structures, e.g. heart valves, arterial walls. The fibers are immersed in a liquid matrix, which, due to the properties and characteristics of liquids, will not deteriorate or change its properties when it comes into contact with a moving fiber.

The geometric alignment and orientation of the fibers are the reasons why this material is characterized by strongly nonlinear elastic properties. When internal force is augmented, the contribution of the fibers in transferring the load grows and thus the stiffness of the material increases as well. The outer layers give space to the inner layer, which is functioning under load influence. Fibers are characterized by their constant geometric form and difference in density of presence circumferentially and axially.

During the work of the aortic artery, the cylindrical shell material of the artery is being subjected to enormous deformations within the elastic limits, until even several dozen percent [10]. Another particular property of this material is incompressibility due to high water

content. For the entire three-layer material, Poisson's ratio $\nu \approx 0.48$ was used. The idea of the new material is shown in Figure 2.

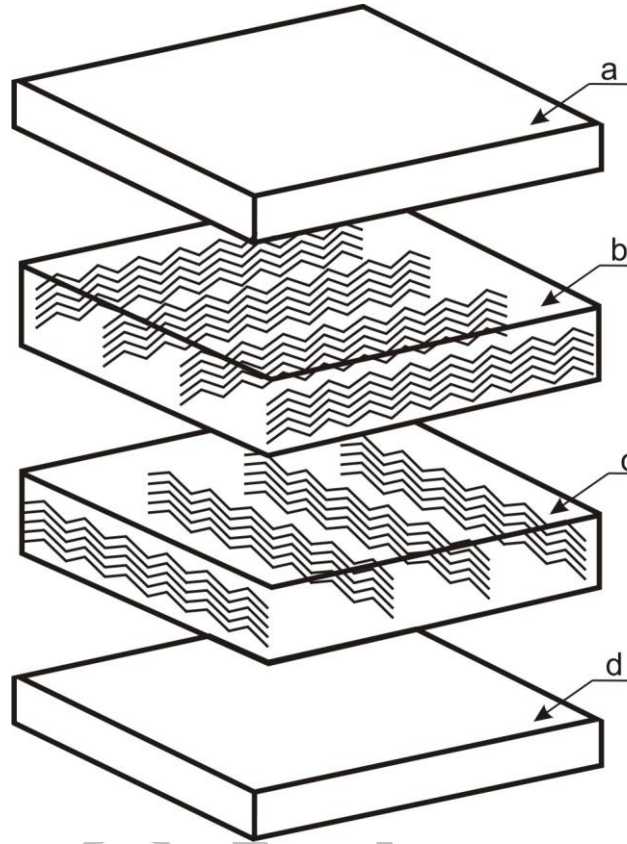


Figure 2. Structure of the new material a, d – outer layers, b – fibrous inner layer on circumferential direction, c - fibrous inner layer on radial direction

In order to create the strength properties of the new material, the finite element method was used. Constitutive relations for the relationship between the state of stress and the state of strain in the material within its operation were determined. The Hooke model was selected to represent the matrix strength and fibers, whose constitutive equation takes the following form

$$\sigma_{ij} = C_{ijkl} \varepsilon_{kl}, \quad (1)$$

where C_{ijkl} are material parameters.

The linear material model is a approximation from the very nonlinear nature of the material as indicated by laboratory tests. The nonlinearity in the model, however, came from the sinusoidal nature of the fiber, which is inherent in the natural material. The constitutive behavior of the isotropic material model is governed by only two material constants: Young's modulus E and Poisson's ratio ν . Due to orthotropy of the new material, these parameters were

provided separately for the inner layer along the circumferential and axial directions. The material's constant tensor is of the form

$$C_{ijkl} = \lambda \delta_{ij} \delta_{kl} + \mu (\delta_{ik} \delta_{jl} + \delta_{il} \delta_{jk}), \quad (2)$$

where Lamé constants are

$$\lambda = C_{1122} = \frac{Ev}{(1+\nu)(1-2\nu)},$$

$$\mu = \frac{1}{2}(C_{1111} - C_{1122}) = G = \frac{E}{2(1+\nu)},$$

The following assumptions were made for the novel aortic artery prosthesis material model:

- The material consists of two elastic outer layers and an inner layer, which comprises elastic fibers suspended within a liquid matrix.
- Geometrically, the fiber axis is a regular sinusoid in the cylindrical plane of the aortic shell. Fibers in the inner layer are oriented regularly and uniformly in two orthogonal directions aligned with the circumferential and axial directions of the aortic artery wall.
- Separate components of the layers are ideally elastic. From the point of view of the mechanics of a material body, the fibers are a homogeneous material, while the matrix is a homogeneous, continuous, incompressible medium.
- The separate layers function independently in two orthogonal directions.

The layer parameters were assumed as determined, resulting from experiments and measurements performed on samples of the natural aortic material [10]:

- Geometric parameters of the material sample: width $a = 10 \text{ mm}$, length $b = 10 \text{ mm}$, average thickness of the aorta wall $h = 2,2 \text{ mm}$, thickness of the inner layer $h_i = 1 \text{ mm}$, thickness of the outer layers $h_o = 0,6 \text{ mm}$.
- Geometric parameters of the fiber: length of one sinusoidal wave $l = 0,8 \text{ mm}$, cross-sectional area $A_w = 0,2 \text{ mm}^2$.
- Fractional share of fibers in the inner layer in both directions $\mu = 20\%$.
- The matrix of the inner layer fibers will be a fluid. Therefore, as a result of fiber deformation, we observe its movement in the fluid matrix. After the load is removed, the fiber automatically returns to the shape before the load was applied. The fluid matrix does not lose its properties under the influence of fiber movement. The strength parameters of the matrix are described by the compressibility modulus - β_o . The compressibility modulus

for water was assumed to be $\beta_o = 5 \cdot 10^{-4} [\frac{1}{\text{MPa}}]$. The influence of different types of liquid matrix on the material behavior will be demonstrated later in the article.

The fiber strength parameters and the shape factor were treated as design parameters and determined in the optimization process. These include:

- Mechanical parameters of the material:
 - E_c - Young's modulus of the fiber in the circumferential direction,
 - E_a - Young's modulus of the fiber in the axial direction.
- Geometric parameters of the fiber:
 - γ_c - fiber shape factors for the inner layer in the circumferential direction,
 - γ_a - fiber shape factors for the inner layer in the axial direction.

In order to determine the design parameters, an appropriate optimization problem was formulated, in which the minimum of the objective function was sought by the gradient method. The material parameters were determined in a series of steps according to a particular strategy. At each step, the process of stretching the material sample of the inner layer was numerically analyzed in one of the principal directions: circumferential and axial. The steps of subsequent analysis were adopted as:

1. Inner layer was stretched in the circumferential direction. E_c – Young's modulus of the fiber and γ_c – fiber shape coefficient was determined.
2. Inner layer was stretched in the axial direction. E_a – Young's modulus of the fiber, γ_a – fiber shape coefficient was determined.

The algorithm for determining the material parameters in Stage 1 is presented in detail below. The analysis of the second stage was carried out analogously. The parameters E_c, γ_c are sought by formulating the following optimization problem:

A set of design parameters was assumed

$$\mathbf{b} = (b_1, b_2)^T, \quad (3)$$

where: $b_1 = 0,1 E_c^2$, $b_2 = 8,89 \gamma_c$

The numerical coefficients in the definition of the design parameters were selected so that the design parameters in the space being analyzed took on numerical values of the same order.

For each set of the design parameters of the uniaxial tensile test, the stress-strain approximation function takes the form $\bar{\sigma}(\epsilon)$. For laboratory testing of a sample of the natural aortic artery wall in the circumferential direction, the stress-strain relationship was established

in the form of the function $\sigma^{lab}(\varepsilon)$ Figure 1. The model mismatch of stress approximation compared to the laboratory results is provided by the error functional in the form

$$F[\mathbf{b}] = \int_0^{\varepsilon_k} \frac{(\sigma^{lab} - \sigma)^2}{(\sigma^{lab})^2} d\varepsilon, \quad (4)$$

The parameters of the design are selected in a way that one obtains the minimum of the error function $F[\mathbf{b}]$. The above-given optimization problem was solved numerically using the finite element method in Galerkin form.

A flat sample of the material with size and fiber distribution as shown in Figure 3 was tested. Straight pieces were used to simulate the actual sinusoidal shape of the fibers. The cross-sectional area of the fibers was selected such that the coefficient γ was maintained (Figure 4).

According to assumptions, the fiber diameter is not relevant, but fractional proportions of individual constituents, mean distances between fibers in proportion to the amplitude d and γ are relevant.

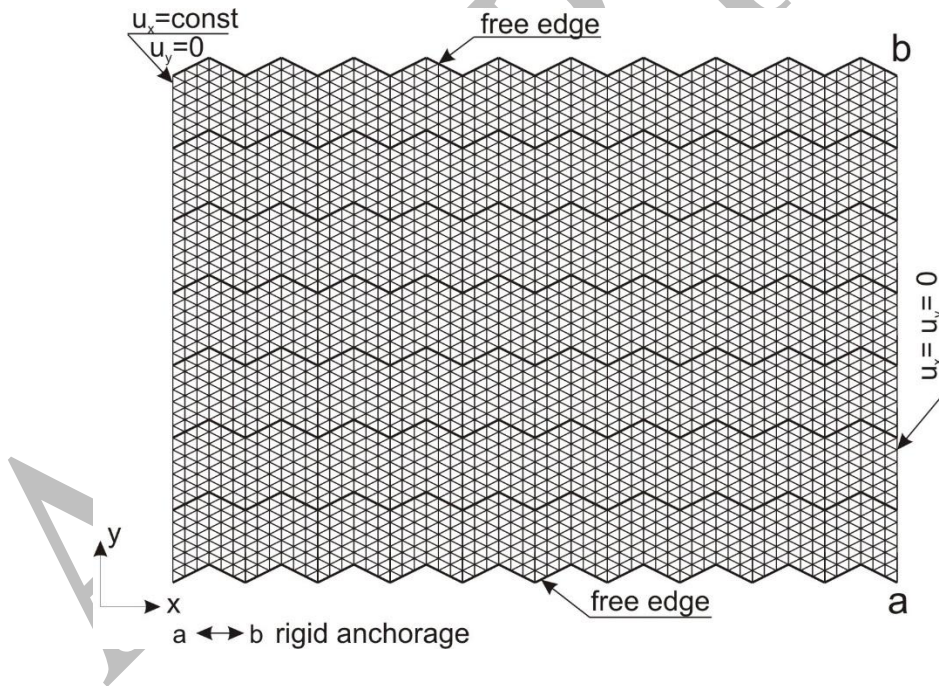


Figure 3. Definition of the gamma fiber Static model of the aortic wall's fibrous layer.

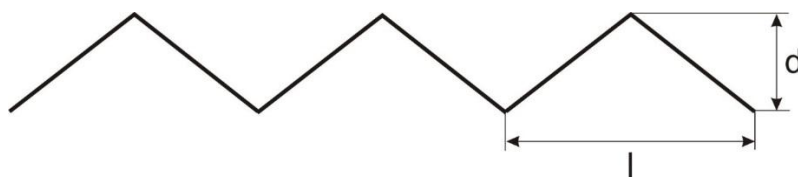


Figure 4. Definition of the gamma fiber shape coefficient.

Discrete FEM model contained 11,800 matrix elements, and 880 fiber elements, resulting in a model with 47,200 nodal parameters, (Figure 4). The fibers were spread evenly. Static calculations within geometric nonlinearity were performed with our own software. Geometric features of the model were consistent with sizes of sections in physical tests. The static scheme of the model is shown in Figure 3.

The design parameters were determined as a result of the iterative process, according to the following procedure [9]. The error function was written as a sum by dividing the interval $(0, \varepsilon_{max})$ into m segments

$$F[\mathbf{b}] = \sum_{j=1}^m \frac{(\sigma_j^{lab} - \bar{\sigma})^2}{(\sigma_j^{lab})^2}, \quad (5)$$

where $\sigma_j^{lab} = \sigma^{lab}(j\varepsilon_k/m)$, $\bar{\sigma}_j = \bar{\sigma}(j\varepsilon_k/m)$.

In each n -th iterative step, the following operations were performed. The design parameters are represented by $\mathbf{b}^{(n)}$. Then, the direction of the fastest growth of the error function is calculated.

$$\text{grad}_{(n)} F = \left(\frac{\partial F}{\partial b_1}, \frac{\partial F}{\partial b_2} \right), \quad (6)$$

where the gradient components are determined by the difference quotient. The next step in the design space is calculated by the relationship

$$\mathbf{b}^{(n+1)} = \mathbf{b}^{(n)} - \alpha \cdot \text{grad}_{(n)} F, \quad (7)$$

where the parameter α is chosen arbitrarily. The identification process is completed when the difference in the subsequent steps is less than the defined value.

$$|\mathbf{b}^{(n+1)} - \mathbf{b}^{(n)}| = \sqrt{\sum_i (\mathbf{b}_i^{(n+1)} - \mathbf{b}_i^{(n)})^2} < \tau. \quad (8)$$

The process is shown in figure 5.

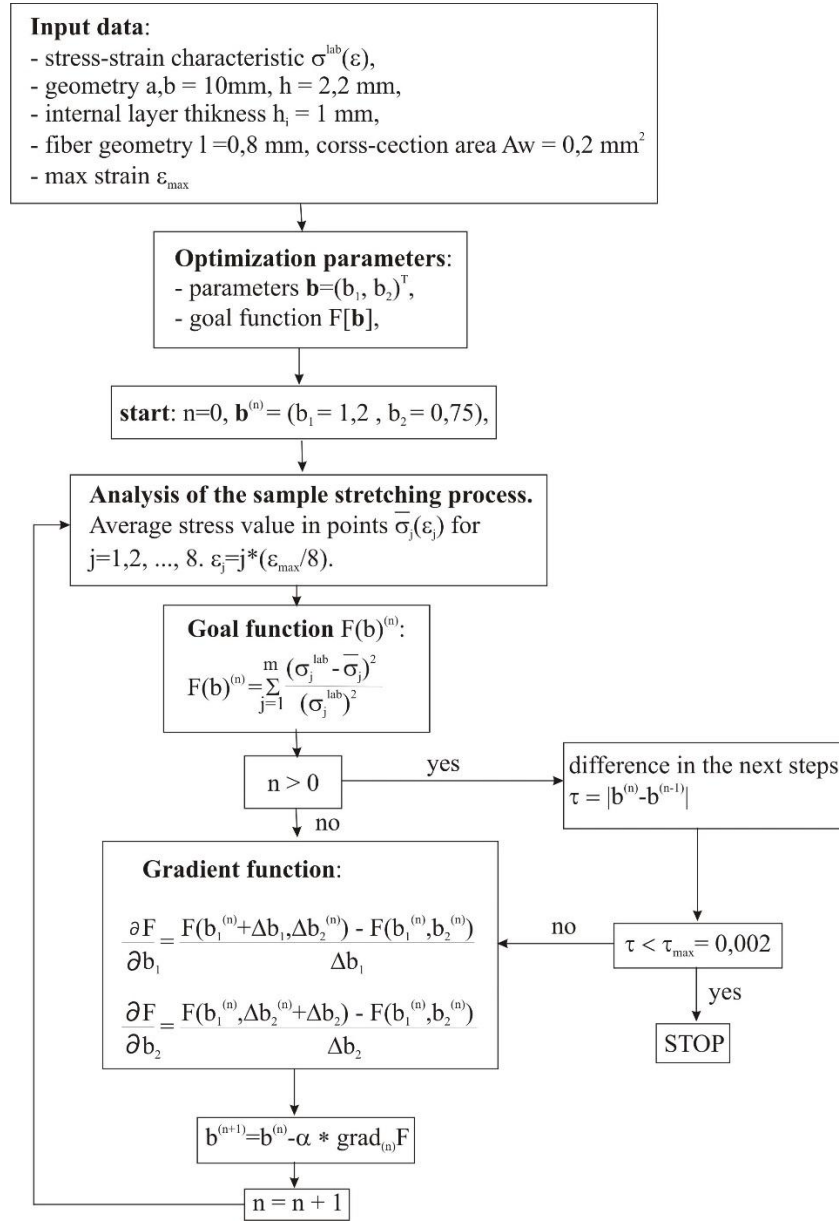


Figure 5. The optimization process algorithm.

Based on the results of the identification process (see Table 1), a material specimen was prepared. Due to the orthotropic properties of the material, its strength properties were determined separately for the circumferential and axial directions. In the experimental studies, a specimen developed only for the circumferential direction was tested due to limitations in the availability of materials and due to limitations in the method of creating the sample itself. The method chosen was 3D printing.

Table 1. Strength parameters of the inner layer of the new material.

Layer	Direction	Material parameters
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		E_c, E_a [kPa]	γ_c, γ_a
inner	circumferencial	1090	0,056
inner	axial	440	0,067

The structure of the material specimen in the circumferential direction, just like the natural material, consists of a fibrous layer surrounded by a liquid matrix. In order to maintain a constant fiber arrangement and precise immersion of the fibers in the matrix, two outer layers with very low stiffness were added. Finally, a three-layer material sample was created with the following detailed structure:

- a) The outer layers were made of a synthetic transparent polymer with a tangent Young's modulus $E_o = 150$ [kPa]. This stiffness is comparable to the matrix of the natural material [10]. These layers constitute a vessel in which the inner layer surrounded by a liquid matrix is immersed. In addition, these layers hold the fibrous layer in a given place during the uniaxial stretching process.
- b) The inner (fibrous) layer consists of a group of evenly distributed fibers with parameters determined in the identification process. The number of fibers in the layer was selected arbitrarily and distributed evenly over the entire width of the sample. The developed layer model was printed on a 3D printer using LayFomm 40 filament with a tangent Young's modulus $E_f=1000$ [kPa]. The fiber arrangement shown in Figure 6 has an additional printed anchor part to enable connection with the outer layer.



Figure 6. Fiber arrangement in the inner layer of the material sample with fixing elements.

The structure of the fibrous layer prepared in this way was secured with a material that prevents the fibers from moving or being damaged. The fiber arrangement was placed in a mold for preparation specimen of the new material, (Figure 7a), and covered with the second half of the mold, (Figure 7b). After assembling both parts of the mold (Figure 7d) of the new material using the injection method, a synthetic polymer was placed inside, which had been

previously heated to a temperature of $T_p=40$ °C. After cooling, the mold was opened, leaving the polymer outer layers for 24 hours at room temperature until full strength was achieved.

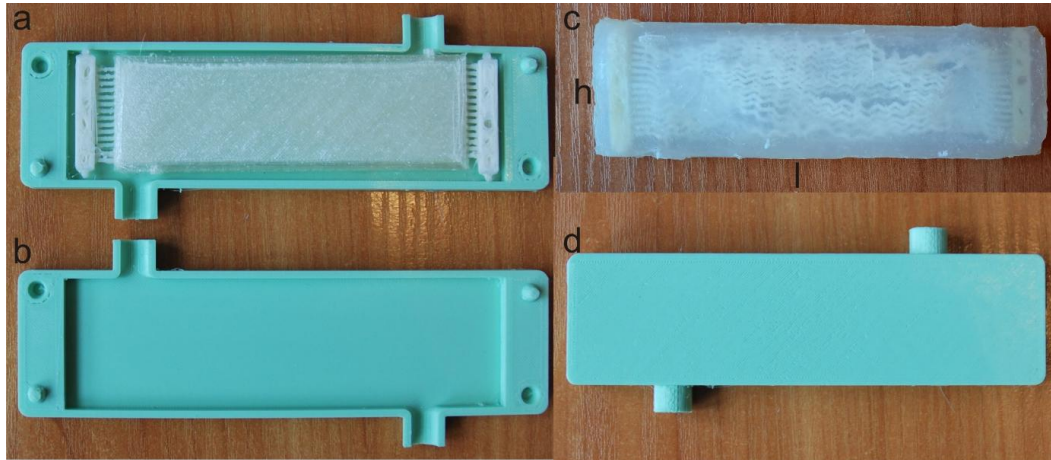


Figure 7. Sample of new material. a) Lower part of the mold with a secured fibrous layer inside. b) Upper part of the mold. c) Sample of new material with visible fibrous structure. d) Mold after assembly.

In the next stage, a small peristaltic micropump was connected to the inlet and outlet of the material specimen and generating a water flow with a flow rate of $Q=0.5$ [cm³/s] and a temperature of $T_w=24$ °C, the material was rinsed inside the material specimen, the task of which was to prevent the fibers from moving or being damaged in the polymer injection process. The rinsing process lasted 24 hours. The system with a peristaltic micropump for rinsing samples is shown in Figure 8. During the rinsing process, the internal structure of the fibers, now located between the outer polymer layers, was released.



Figure 8. The peristaltic micropump connected to specimen.

Ready to test material specimen with dimensions: length $l=80$ [mm], width $h=33$ mm and thickness $d=3$ mm are shown in Figure 7c. The sample shows internally evenly distributed fibers. After the rinsing process, the samples were filled with a glycerol solution, the task of which was to imitate the natural fiber matrix. In order to investigate the effect of the liquid matrix on the stress-strain characteristics, two glycerol solutions were prepared. The entire sample creation process is repeatable, and subsequent samples show a very similar undisturbed and regular fiber arrangement. Samples in this form in the number $n=8$ will be subjected to uniaxial stretching in order to determine the stress-strain characteristics. The sample data are summarized in Table 2.

Table 2. Material sample data.

layer	Specimen1 (n=4)	Parameter	Value	Specimen2 (n=4)	Parameter	Value
outer	synth. polimer	Young modulus	150 kPa	synthetic polimer	Young modulus	150 kPa
inner	layfoam 40	Young modulus	1000 kPa	layfoam 40	Young modulus	1000 kPa
		fiber shape coeff.	0,05		fiber shape coeff.	0,05
matrix	20% glycerol solution	viscosity	2 mPa*s	40% glycerol solution	viscosity	5 mPa*s
		density	1032 kg/m ³		density	1101 kg/m ³

Results

For the fibrous layer of the model, characteristics that identify variation in values of individual design parameters and variation in value of the error function in the iteration process were formulated. As a result of minimization of the error function, mechanical characteristics of the model approximating the actual ones were formulated. The right iterative process was executed by thorough investigations of design parameter space with the intention of gaining knowledge about the range of the parameter space. Out of the above parameter vector, parameter b_1 produced the most contribution to the value of the error function. This conclusion is in accordance with reason, as from the mechanical point of view, the fiber stiffness and geometry determine the mechanical properties of the aortic artery wall

material tested model. The analysis was carried out based on eight points of the stress-strain characteristic (Figure 9).

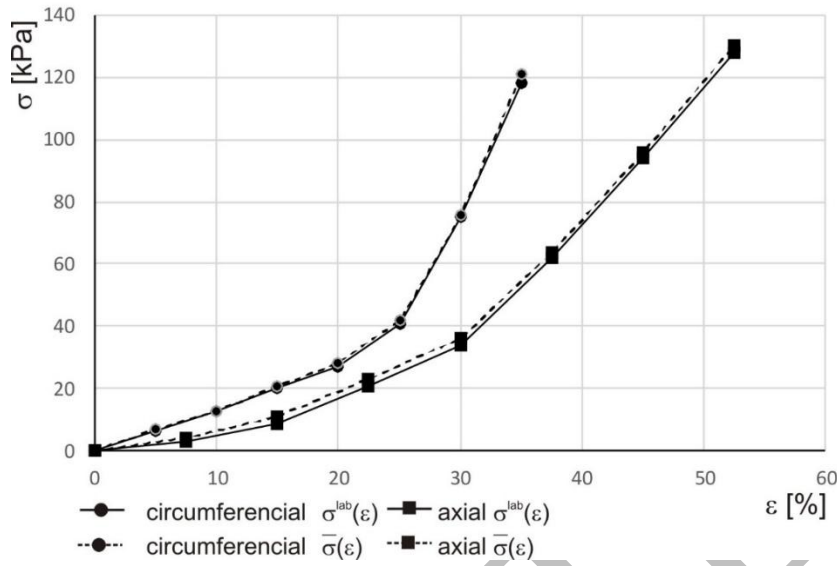


Figure 9. Comparison of mechanical characteristics obtained from experiments and the optimization process.

The results of modeling the strength parameters of the inner layer are presented in Table 1. In the circumferential direction, the Young's modulus was 1090 kPa, the fiber shape coefficient was 0.056. In the axial direction, the Young's modulus was 440 kPa, the fiber shape coefficient was 0.067.

The obtained material parameters will allow us to write constitutive equations of the inner layer and the entire material, which will be used to construct the new aortic artery bioprosthesis. The same results were obtained for tangent Young's moduli to the nonlinear behavior in paper [7].

FEM simulations were performed in our own software. Calculations were conducted in the interval of strains up to $\epsilon=53\%$ in the axis direction and $\epsilon=35\%$ in the circumferential direction. The highest stresses in the fiber up to 67 kPa ($\epsilon = 38\%$) converged in the direction of the axis when the strains were and 62 kPa ($\epsilon = 27\%$) when in the direction of the circumference. With the increase in strain, we are able to observe a rise in the highest stresses. The middle portion of the specimen tested in both directions is illustrated in Figure 10. We can observe high stress levels especially at the places where the direction of the fiber is changing. Higher value of stress can be seen in the entire fiber.

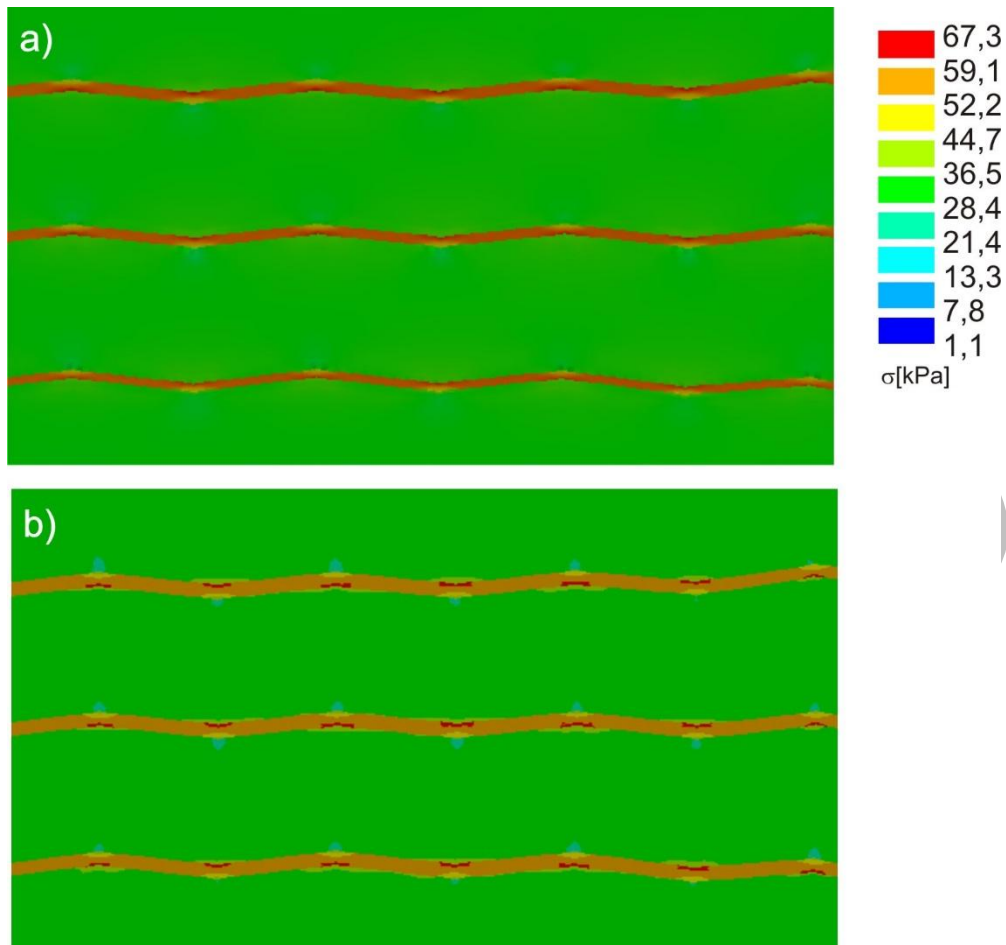


Figure 10. Map of principal stresses in the model fibers in the direction a) circumferential at strain $\epsilon=27\%$ and b) axial at strain $\epsilon=38\%$.

Because of the stretching process, we are able to witness fiber movement in the material, generating matrix flow. Velocity field obtained provides a witness to the dynamics of the material sample's stretching process. The matrix flow velocity field is shown in Figure 11. Using the model established in the work to describe the operation of the whole artificial organ under actual conditions and based on the results of this work, we will find the actual values of the matrix rate field as a function of the hemodynamic cycle of the heart. Hydrodynamic state of the matrix flow and hydrodynamic resistance values may be the other characteristic parameter that determines the new bioprosthesis. In Figure 11, we observe higher values of the velocity of the matrix flow in the fiber motion region.

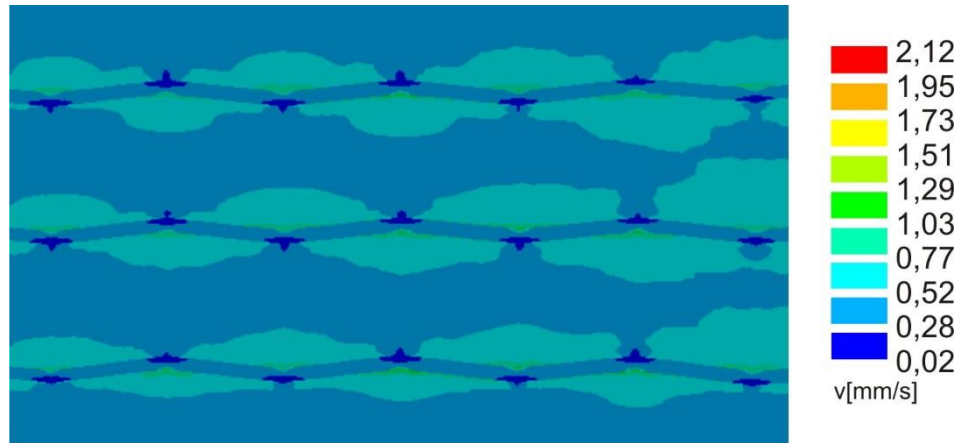


Figure 11. Matrix flow velocity field.

As a result of the uniaxial tensile test of the prepared material samples, stress-strain characteristics were obtained (Figure 12). The graph shows the nonlinear behavior of the samples characteristic of biological materials. The tangent modulus method was used to analyze the results to show that the nonlinearity of the characteristics has a direct impact on the layer structure of the tested samples.

In the initial range of the sample deformation process, we observe an increase in stress in the outer layers to the level of 45 kPa and gradual straightening of the fibers. The tangent Young's moduli have a value similar to that of the natural material and are within the range of variability.

For samples from group Specimen1 (Table 2), after exceeding the level of 25% deformation, we observe an increase in the involvement of fibers in the transfer of stresses and gradual stiffening of the material. This process lasts up to the level of 28% deformation. For samples from group Specimen2 (Table 2), the process of material stiffening is observed already at the level of 17% deformation, full fiber involvement is observed only from the level of 30% deformation. From the level of 30% strain we observe straightened fibers and a significant increase in stresses to the level of over 100 kPa.

Figure 13 shows a stretched sample with visible straightened fibers.

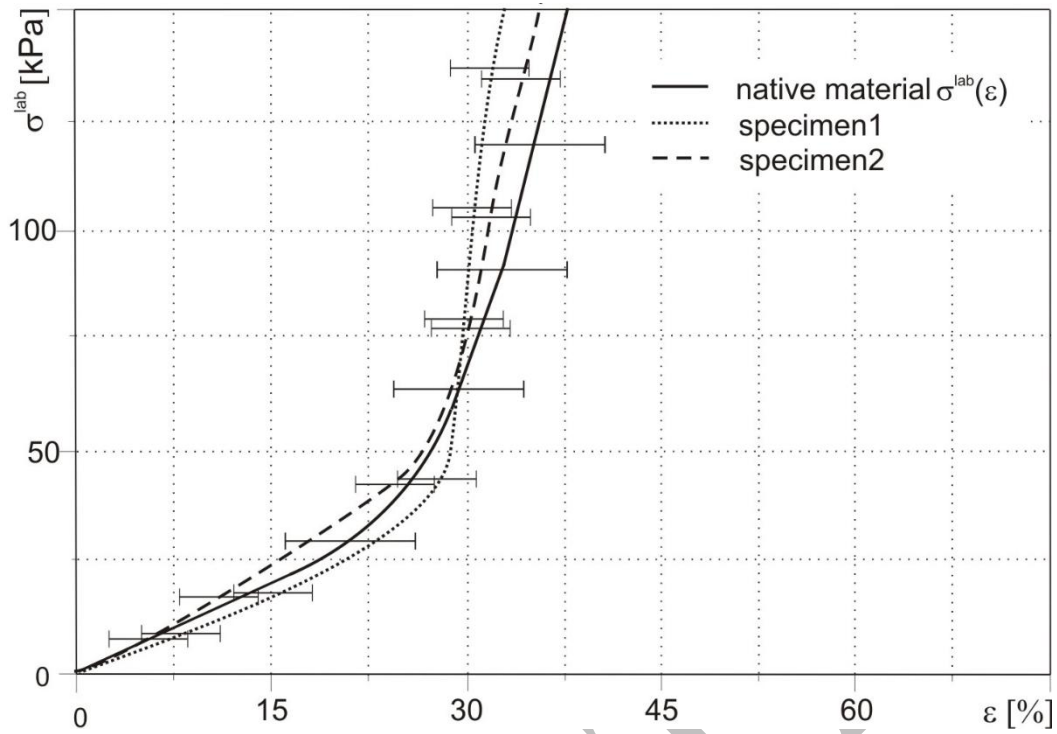


Figure 12. Comparison of stress-strain characteristics of the natural material and both types of samples.

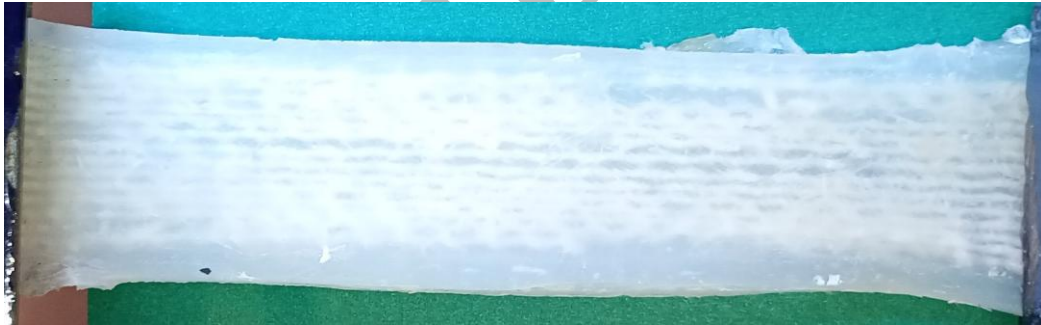


Figure 13. A sample under uniaxial test.

Discussion

The paper presents the new bioprosthesis material model optimization process, in a mechanical sense imitating the natural material of the aorta. A fiber construction of straightforward design by way of elementary material parameters through soaking into a liquid matrix allows the creation of a non-linear material of great compliance unmatched in prostheses made of Dacron. The fluid matrix used in the material model allows fiber movement that does not change the nature of the matrix. The properties of fluid about the fiber, i.e., density and viscosity, we regulate so that we can control the shape of the stress characteristic of the new material. In an unloaded state of material, the fiber is constrained and

curled. Under load action, it straightens out, transferring extra load and strengthening the material. When unloaded, the fiber returns to the restrained position. Therefore, we obtain another dynamic parameter - the relaxation time, which can be used to compute the mechanical properties of the new material. Explanation of the work of the fiber and the entire new organ synchronizes with hemodynamic cycle of the heart. Relaxation tests and fatigue analysis of the new layered-composite structure will give a chance for describing the work of the new organ for a sustained period of time. Long and long-living operation of new artificial organs is a parameter expected among new solutions in bioengineering.

The problem of changing properties of layered-composite materials in the long term should be considered as well. Based on the investigations of biomaterials with such a structure, among the key problems is delamination and changing properties of the entire composite due to changes of the properties of the composite elements. Such changes are the direct result of excessive gradients of shear stresses [14,15]. Biomaterials for cardiovascular applications experience huge deformations, which further adversely affect fiber-matrix interactions and thus shorten the working time of such a material. Use of a liquid matrix eliminates the problem of the impact of shear stresses on the strength properties of the matrix in which the fiber is aligned.

In the future, it is also necessary to investigate how the interaction process between the fibers and the matrix occurs during the hemodynamic cycle of the heart based on the given material. In particular, the interaction process between the fibers and the matrix at high pressure of the matrix fluid needs to be investigated. The new bioprosthesis can be connected to an aortic valve prosthesis and in this way the artificial organ can be used in aortic aneurysm surgeries, where the aortic valve is also damaged due to the dysfunction of the tissue building the artery wall.

The developed physical model of the aortic wall section allows the implementation of the concept of a new type of layered-composite material, in which the fibers are embedded in a liquid matrix. This concept allows the development of materials with non-linear properties based on simple materials and linear characteristics. Until now, such a concept could not be implemented because the fibers moving in the matrix exerted large tangential stresses, destroying the surrounding matrix and degrading the strength of the material. Currently, there are no materials on the market of biological and artificial materials in which we would observe nonlinear-geometric processes. The use of a liquid matrix provides such possibilities, and the subject requires further research and development.

The research results allowed us to observe a different nature of the transition of the linear characteristics of the outer layer in the initial phase of sample stretching to the linear characteristics of the fibrous layer. For specimen from group 1, this is a process that occurs suddenly (range 25-28% strain) and for group 2, this range is 17-30% strain. This difference results from the parameters of the liquid matrix, which has different density and viscosity in both groups of samples. The conclusion is that the density and viscosity of the liquid matrix have a direct impact on the course of the stress-strain characteristics, and both parameters of the fluid can be used in the process of designing a new material.

The 3D printing technology used in the process of creating the fiber system allows for the design of a fiber system dedicated in terms of strength based on numerical calculations. In the future, each element of this material, including the liquid matrix, should be created in this technology. This technology will soon allow for printing the entire organ, taking into account its geometry and orthotropic properties.

Conclusions

A layered composite material composed of fibers and a matrix with linear properties determined through the identification process allows for the construction of an aortic artery bioprosthesis with nonlinear geometric properties. A bioprosthesis with such properties corresponds to the stress-strain characteristics of the native aortic artery. The new material offers the opportunity to construct new artificial organs, including ascending artery grafts and new aortic and mitral valve prostheses with the hemodynamic properties of natural heart valves.

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