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Experimental and Numerical Study of Mechanical Properties of Artificial Blood Vessel

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Abstract. This article describes the design of equipment for the production of small diameter fiber tubular formations, which serve as carriers of tissue cells. Also their production and establishing of a simulation model, which describes their behaviour under internal pressure. For this purpose spinning device was created with a rotary collector, which allows obtaining the desired fiber orientation that is perpendicular to the axis of rotation. This orientation is necessary for the efficient proliferation of cells and future mechanical properties of artificial blood vessels. Formations created in this way were subjected to mechanical testing. Based on obtained data a suitable material model and FEM model of artificial vessels was established. This vessels were loaded with internal pressure corresponding with standard blood pressure in human body (120/80 mmHg). The results allow assessing the behaviour of the blood vessels during cultivation and proliferation of the cells in bioreactor. This is very important, because in bioreactor a medium, which pulsed under some pressure, is applied. Based on the obtained results were recommended optimization of production process, which led to suitable vessel properties.

Introduction

Currently on vascular disease dies approximately 39 % of the population. Some deaths can be prevented by replacing of the vessels. First usable vascular grafts were knitted and woven in the early fifties. Since then, their production and their use has expanded. Artificial blood vessels can be manufactured with a diameter greater than 12 mm. Blood vessels with smaller diameter have not yet been successfully created. Failure reason lies in limited proliferation of cells after vessel implantation into a patient body. One of the possibilities how to produce artificial blood vessels of smaller diameter is the creation of the multilayer nanofiber formation with different orientation of fibers in each layer. In order to establish nanofiber layers with a desired orientation was assembled machine with a modified collector electrode. This modification consists in a replacing of a stationary electrode with rotary electrode. The collector is in the form of a rod which rotates at high angular velocity perpendicularly to the created fibers, which leads to their parallelization. This enables getting of the desired shape of the pores of the nanofiber structures, in which cells proliferate easier.

Theory

One of the produced vascular grafts layer has orthotropic hyper-elastic properties. Due to uniaxial orientation the vessel has predominant strain in one direction only. This can be mathematically described by the energy-conjugated pairs, which together constitute the deformation work. For large deformation energy the pair of conjugate Green-Lagrange strain tensor $\mathbf{E}$ and 2. Piola-Kirchhoff stress tensor $\mathbf{S}$ is appropriate. This behaviour may be considered as predominantly hyper-elastic and we can use mathematical description of the strain energy of hyper-elastic anisotropic material, which is divided into two parts by an elastic part and configuration (deformed) part, which describes a change of volume (where the Jacobian of the deformation $\mathbf{J} = \mathbf{detF}$ is in the range 0 <J <1). Then the strain energy χ is given by (1) as described in the tensor dependence.

$$\chi = \chi_{\mathrm{el}}(\overline{\mathbf{C}}) + \chi_{\mathrm{vol}}(\mathbf{J}) \tag{1}$$

$$\overline{\mathbf{C}} = \overline{\mathbf{F}}^{\mathbf{T}}\overline{\mathbf{F}} \tag{2}$$

$$\mathbf{C} = \mathbf{F}^{\mathbf{T}}\mathbf{F} \tag{3}$$

$$\mathbf{F} = \mathbf{J}^{1/3}\mathbf{I}\overline{\mathbf{F}} \rightarrow \overline{\mathbf{F}} = \mathbf{J}^{-1/3}\mathbf{IF} \tag{4}$$

where $\chi_{\mathrm{el}}(\overline{\mathbf{C}})$ describes elastic part of a deformation energy part, $\chi_{\mathrm{vol}}(\mathbf{J})$ is describes deformed (volumetric) part of the deformation energy, $\overline{\mathbf{C}}$ expresses a modified Cauchy stretch tensor $\mathbf{C}$, where $\mathbf{F} = \frac{\partial \mathbf{X}}{\partial \mathbf{x}}$ describes the material deformation gradient which is a component of the Green-Lagrange strain tensor $\mathbf{E}$, $\mathbf{X}$ is the spatial coordinates, $\mathbf{x}$ is the material coordinate, $\mathbf{I}$ expresses the unit matrix and mathematical expression $\mathbf{J}^{1/3}\mathbf{I}$ is associated with part changing the volume during the deformation. The resulting stress in reinforcement is describable by 2. Piola Kirchhoff stress tensor $\mathbf{S}$ (5), will be also divided into two parts (elastic and volumetric).

$$\mathbf{S} = 2\frac{\partial \chi(\mathbf{C})}{\partial \mathbf{C}} = \mathbf{S}_{el} + \mathbf{S}_{vol} \tag{5}$$

where $\mathbf{S}_{el} = 2\frac{\partial \chi_{\mathrm{el}}(\overline{\mathbf{C}})}{\partial \mathbf{C}}$ is 2. Piola-Kirchhoff stress tensor for the elastic part of strain energy function $\mathbf{S}_{vol} = 2\frac{\partial \mathrm{A}_{\mathrm{vol}}(\mathbf{J})}{\partial \mathbf{C}}$ is 2. Piola-Kirchhoff stress tensor for the volume part of strain energy function.

Experimental Analysis of Mechanical Properties of Artificial Blood Vessel

To produce a defined orientation of the vessels is necessary to use a collector, which rotates around its own axis and allows the fibers to be arranged perpendicularly to the axis of rotation of the collector (Fig. 1). During deposition of fibers on a stationary collector, the fibers due to the residual electrical charge oriented isotropically [2], but it is for one of the layers of vascular grafts undesirable. The spinning is performed from the needle to achieve the intense electric field. The result is only a narrow strip of nanofiber layer. For uniform coverage of the fiber collector is necessary that the needle is during the spinning reversibly moved along the axis of the rotation [3]. The maximum speed is up to 600 s^{-1}. The vessels were made of polycaprolactone (PCL), which is biodegradable and after implantation to the body depredates due to metabolic processes. From the obtained tubular formations (Fig. 2) samples for tensile tests were created. Samples were prepared in radial, axial and diagonal direction from the viewpoint of the of the collector rotation. The clamping length and width of samples were 10 mm. These samples were carefully clamped in the jaws of the fibrous material. For the test dynamometer Labortech 2050 with strain gauge with a load capacity up to 5 N was used. Loading rate was 2 mm·min^{-1}. From the data orthotropic behavior of material was found. The identified properties are shown in Table 1.

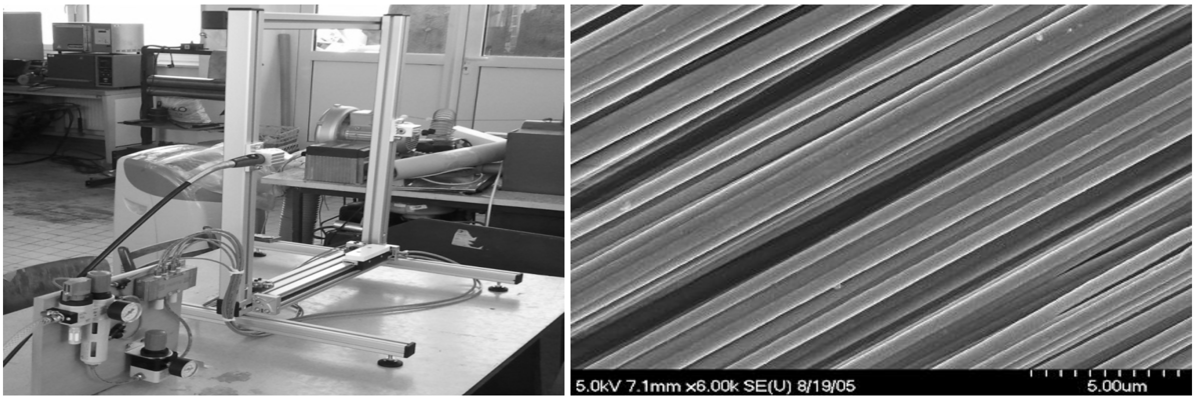

Fig. 1. The device for electrospinning (left), modified nanofiber layer (right).

Fig. 2. Vascular graft.

Table 1. Results of mechanical properties.

Young's Modulus			Poisson's Ratio			Shear Modulus		
X [MPa]	Y [MPa]	Z [MPa]	XY [-]	YZ [-]	XZ [-]	XY [MPa]	YZ [MPa]	XZ [MPa]
0.05	0.025	0.025	0.28	0.4	0.28	0.05	0.025	0.05

Numerical Analysis of Mechanical Properties of Artificial Blood Vessel

For the study of deformation and stress distribution in the construction of the experimental sample numerical model using the finite element method in software ANSYS was assembled. FEM model of the test sample with dimensions 10x10 mm and a thickness of 0.1 mm was composed. Created FEM model used the orthotropic material model based on the equations (1-5). Thereafter dimensions and hexagonal type of elements were put. Number of elements in the mesh was 1344 and the number of nodes was 2720. Into the model following boundary conditions were introduced: the one side of the model was fixed in all directions against the movement and rotations, the opposite side linear displaced with strain rate 2 $mm.min^{-1}$. Von Mises stress in the sample was observed (Fig. 3). The next step of the numerical simulation was to verify whether existing wall thickness of 0.1 mm will meet the pressure in the bloodstream. In the Ansys program the geometry of a monolayer vessel with height 100, diameter 6 mm and a wall thickness of 0.1 mm were created. The model included the properties of the orthotropic material. A structured hexagonal mesh with 8888 nodes and 4400 elements was created. Both ends of the blood vessel model were fixed in all directions against the movement and rotation. An inner wall of the vessel was loaded with a normal pressure of 120 Torr (0.0159 MPa), which corresponds to normal systolic human blood pressure [1]. Based on the obtained stress values is evident that the orthotropic material behaviour of the vessel in the X and Y axis differs due to the parallel arrangement of the fibers (to see Fig. 3). Simulated internal pressure of 1,59 $x10^{-2}$ MPa could lead to the stress in the sample is higher than the yield strength of 5478 MPa $x10^{-3}$ and a failure of the sample occurs (Fig. 4). Therefore, it is recommended to increase the thickness of the wall or a creating of additional layers with differently oriented structure.

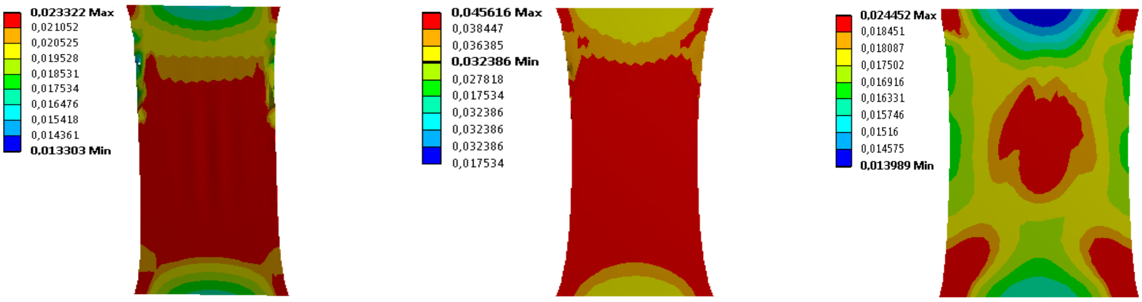

Fig. 3. Stress in principal axis: a) Y = 0.023 MPa, b) X = 0.045 MPa, c) 45° = 0.024 MPa.

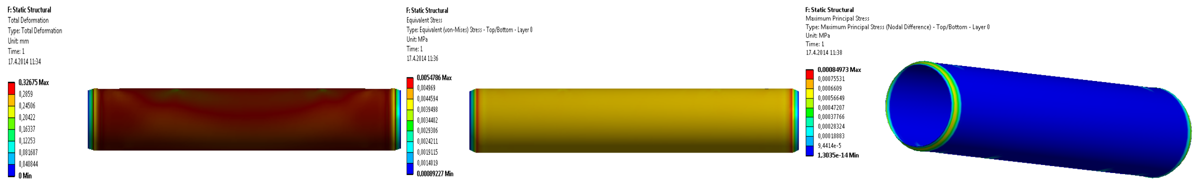

Fig. 4. Picture of scaffold (above), FEM models [total deformation = 0.32mm, von Mises stress = 5.47 kPa and maximum principal stress = 0.849 kPa] (below).

Conclusion

Created vascular graft was subjected to mechanical testing. From obtained data Young's modulus were identified and subsequently the material model in ANSYS was established. The model was verified by simulated tensile test and compared with the course of the real test. Simulated and real values are similar; therefore created model can approximated the behaviour of the real material. Further, the tubular body was loaded with internal pressure of 0.0159 MPa (120 mm Hg). The results showed that the pressure exceeds the material yield strength of 5478×10^{-3} MPa (see Fig. 4 a) and that caused failure of the sample. On the basis of this observation the modification of the vessel properties was recommended. Further aim is to prepare FEM model of a multi-layer material with different fiber orientation of the individual layers.

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