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Constitutive behavior of coronary artery bypass graft

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Abstract— **Implantation of coronary artery bypass graft (CABG) is still one of frequent revascularization methods. Biomechanics of a CABG wall may significantly affect its successful function, nevertheless, papers which concern with a constitutive modeling of the CABG are lacking in literature. The purpose of this study is to describe a constitutive behavior of the CABG tissue. A sample of the CABG underwent three years of remodeling under arterial conditions. This sample was analyzed in an inflation–extension test (vertically aligned closed-end tube was loaded by an internal overpressure and by varying axial force). Displacements of the specimen were recorded by CCD cameras and subsequently evaluated by digital image cross-correlation. The experiment comprised preconditioning cycles and measurement periods. Mechanical response of the CABG was studied using four different values of an axial force (weight). A pressure – circumferential stretch and an axial force – axial stretch data are presented. Presence of the so called inversion point in the pressure – axial stretch data was observed, manifested by an elongation of a pressurized tube under some critical value of axial stretch and by a shortening as soon as the axial stretch exceeds the critical value. This interesting phenomenon was previously reported for iliac arteries. Now we may confirm it for saphenous vein graft also. Selected data were used to fit a material model. The tissue was modeled as a one-layered composite reinforced by two families of helical fibers. The material was assumed to be locally orthotropic, nonlinear, incompressible and hyperelastic. Material parameters were estimated for the strain energy function based on a limiting fiber extensibility assumption. Model parameters are fitted by optimization based on radial and axial equilibrium equation in the thick–walled tube. Material model fits selected data successfully. Further work will be aimed at extension of material model domain on all measured data.**

Keywords— **anisotropy, constitutive model, CABG, limiting fiber extensibility, saphenous vein.**

I. INTRODUCTION

Coronary artery bypass graft (CABG) implantation is a standard treatment method for advanced coronary artery disease. Biomechanical research is mainly focused on investigation of hemodynamical conditions which are believed to play the most important role in CABG failure, e.g. [1], [2]. The venous graft implantation into an arterial system leads to its remodeling. The most obvious adaptation process in the CABG is intimal thickening related to the wall shear stress (WSS). Changes in wall tensile stresses are also studied [3]. Considerable attention is also paid to a technique of an anastomosis. Methods of computer fluid dynamics can help to decide if the end-to-side or side-to-side type of the junction is optimal with respect to the WSS [4]. But remodeling processes may be significantly affected by the stress state of wall. Finite element simulation of CABG surgery was reported in [5]. Such simulation can identify stress concentrations in the wall and may lead to improvement of a therapy.

However, papers, dealing with constitutive modeling of a CABG tissue, are scarce in biomechanical literature. Authors in [5] gained data for the CABG tissue from [6]. Hence, main goal of our study is to present constitutive behavior of the CABG.

Experimental methods, an inflation-extension test, and data post processing, are described first. Special attention is paid to *digital image correlation* (DIC), which was used for evaluation of kinematical quantities. The paper continues with description of the material behavior, especially an *inversion line* in the pressure – axial stretch data is discussed. A choice of suitable material model for anisotropic nonlinear behavior follows. Finally, estimation of material parameters based on radial and axial equilibrium in a thick-walled tube is presented.

II. INFLATION-EXTENSION TEST

The sample of CABG tissue was harvested within autopsy at the Institute of Forensic Medicine of the Faculty Hospital Kralovske Vinohrady in Prague. The sample was obtained from 66–year–old male donor who did not die in the link with cardiovascular diseases. After autopsy the sample was stored in the saline solution at temperature 4°C. The inflation-extension test was finished 65 hours after death. All measurements were performed under room temperature. For the present study, use of autopsy material from human subjects was approved by the Ethics Committee of the Faculty Hospital Kralovske Vinohrady (Prague, Czech Republic). Experiments were performed 35 months after bypass surgery.

The reference configuration of the CABG sample had approximately a tubular shape with the following dimensions: outer radius R_o = 2.31mm; thickness H = 0.25mm. Sizes were determined by an image analysis of photography. The length was not measured directly, it was determined within data post processing by DIC. Residual strains were immeasurable by our standard equipment. The sample was dusted over by pepper and coffee powder to create an artificial surface layer with a stochastic pattern what is necessary for a successful image correlation. Thereafter the sample was mounted into the experimental set up for the inflation–extension test. The experimental configuration was vertical and the tube had a closed end. The sample is shown in the Fig. 1.

The tube was pressurized manually by a syringe. After several pre–cycles (approximately 2 minutes pressurization) measurement cycles were performed. Internal pressure was measured by a pressure probe (KTS 438, Cressto, Czech Rep.) and recorded into PC by an in-house software developed in LabView (National Instruments, USA). Kinematics of the inflation and extension was recorded by digital cameras that are components of DIC system Q–450 (DANTEC Dynamics, Germany).

Within the inflation-extension test the sample is loaded by internal pressure and axial force. The force is originated by the pressure which is applied to the end of the tube (closed tube configuration) and with an additional weight. The measurement cycles were recorded for 1 minute by DIC. The pressure range within the measurement was from 0 up to 20kPa. Four recorded cycles span to 60 seconds.

Fig. 1 The CABG sample mounted in the experimental set up.

III. DIGITAL IMAGE CORRELATION

Kinematics of the experiment was recorded and evaluated by digital image correlation. The DIC is an optical method for full-field, non-contact and three-dimensional measurement of deformations and displacements. Used system Q–450 includes two high speed CCD cameras and data postprocessisng unit Istra (Istra 4D v. 4.2.1, DANTEC Dynamics, Germany). Principles of displacements and strain evaluation by DIC are presented in [7]. Important asset of the DIC method is that it allows evaluation of local strain distributions. User can monitor particular part of a sample, especially areas where assumptions of analytical solutions should be satisfied exactly. This is crucial for a consecutive regression analysis where computational models are employed. In the present work it means that only deformations in middle part of the inflated tube were incorporated. To our knowledge, the use of DIC is not so usual in blood vessel mechanics. But its successful application has already been reported in [8].

IV. MATERIAL BEHAVIOR

Mechanical response of the sample is presented in the Fig. 2 – 4. Four measurement periods were performed with different value of axial weight. They are denoted A, B, C and D and corresponding values of weight were m_A = 0g, m_B = 25g, m_C = 54g, m_D = 128g, respectively. Fig. 2 shows a typical pressure – circumferential stretch relation of the blood vessel wall characterized by a strain-stiffening. Increase in the axial weight shifts curves to the left (diameter of extended tubes decreases as expected). More complicated behavior is presented in Fig. 3 where pressure – axial stretch responses are shown. Two different kinds of material response to internal overpressure were observed. When no axial weight was added, increasing pressure lead to inflation and elongation of the tube (curve A depicted by black diamonds). In this case the axial force is given only by an internal pressure which acts on the end of the tube. However, when an axial weight was added (in this case the total axial force is given as a sum of added weight and the force rising from internal pressure), the tube was shortened with the increasing pressure. This phenomenon was observed for measurement periods B, C and D. The shortening of the pressurized blood vessel was reported by several authors, e.g. [9] or [10]. The value of axial stretch when the dependence of pressure on axial stretch changes from increasing to decreasing is called the *inversion line* [9]. The value of axial weight (and corresponding additional force) seems to be decisive for the axial behavior, even if it represents only about 30% of the total axial load.

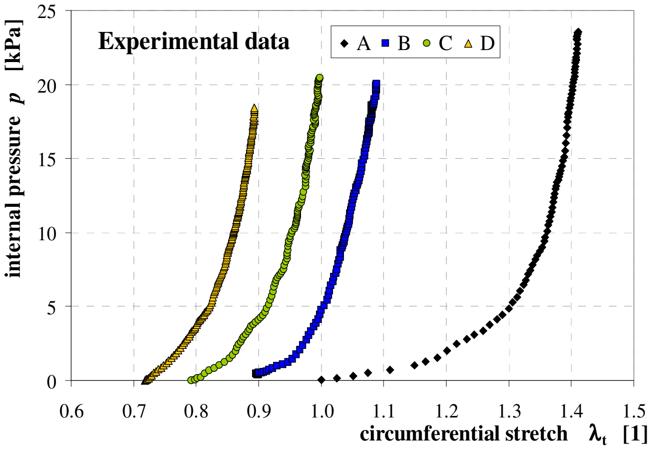

Fig. 2 Pressure – circumferential stretch response of CABG tissue for different values of axial weight (m_A = 0g, m_B = 25g, m_C = 54g, m_D = 128g).

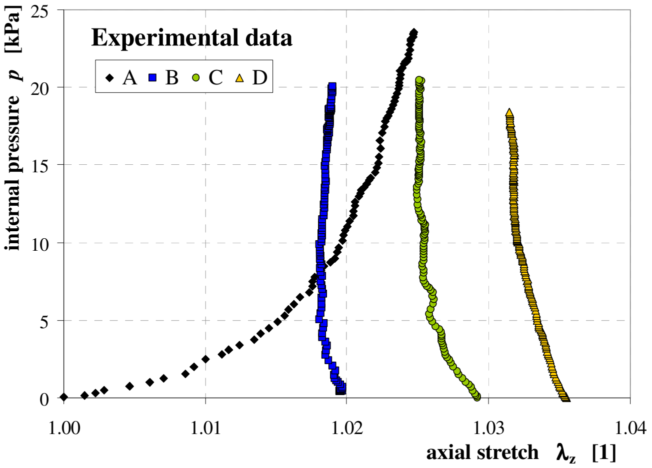

Fig. 3 Pressure – axial stretch response of CABG tissue for different values of axial weight (m_A = 0g, m_B = 25g, m_C = 54g, m_D = 128g). Axial behavior of the tube depends on the value of axial weight. Under no axial weight the tube inflates and elongates, but non-zero weight leads to inflation and axial shortening (with respect to axially loaded sample). It is important to note that the intersection of A and B curves is caused by projection (values of circumferential stretch are different). That is apparent in the Fig. 4.

The literature relates the value of axial stretch, where inversion occurs, to the in vivo axial pre-strain [9], [10]. If we demonstrate it in the strains' domain, it should be a line of constant axial strain. Fig. 3 and 4 shows that while the A, C and D curves monotonically increase or decrease, the B curve behaves non-monotonically. Possible explanation of this behavior may be in the sample twisting. The problem of superimposed twisting on an axially extended incompressible tube from a strain-stiffening material has been recently discussed in [11].

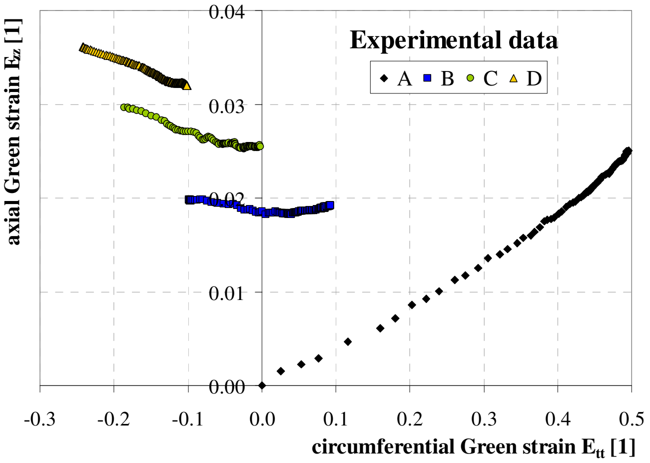

Fig. 4 Strains' domain of inflation – extension test. Plotted symbols correspond to different values of axial weight (m_A = 0g, m_B = 25g, m_C = 54g, m_D = 128g). Distinction between elongation and shortening is clear. Figure proves that A and B curve do not cross, they only overlap.

V. CONSTITUTIVE MODELING

A computational model for the CABG tissue under inflation and extension was based on the one-layered thick-walled tube assuming incompressible, nonlinear and anisotropic hyperelastic material. Residual and shear strains were not included. A presence of collagenous fibers is expected, thus the material is modeled as a fiber reinforced composite. Supposed structure contains two families of collagenous fibers arranged in two symmetrical helical coils with the same mechanical response. Thus the wall anisotropy could be regarded as a local orthotropy. Details of this approach can be found in [12]. We supposed the strain energy dependence upon the two invariants I_4 and I_6 of the right Cauchy – Green strain tensor having the same form

$$I_4 = I_6 = \lambda_t^2 \cos^2\beta + \lambda_z^2 \sin^2\beta \,. \tag{1}$$

Here λ_t and λ_z denotes circumferential and axial stretch, respectively. β is helix angle (material parameter). Two particular forms of the strain energy density function were used for the material parameter estimation. The first was the exponential model proposed in [12].

$$\psi = c\left(I_1 - 3\right) + \frac{k_1}{k_2}\left(\mathbf{e}^{k_2\left(\lambda_t^2\cos^2\beta + \lambda_z^2\sin^2\beta - 1\right)^2} - 1\right) \,, \tag{2}$$

where ψ denotes the strain energy density function, c and k_1 have meaning of stress-like material parameters and k_2 is a dimensionless parameter. I_1 denotes the first invariant of the right Cauchy–Green strain tensor. The second model is the so-called *limiting fiber extensibility* model originally proposed in [13]. Slightly modified expression is used in (3).

$$\psi = c\left(I_1 - 3\right) - \mu J_m \ln\left(1 - \frac{\left(\lambda_t^2 \cos^2\beta + \lambda_z^2 \sin^2\beta - 1\right)^2}{J_m^2}\right) \qquad (3)$$

Meaning of symbols is the same as in the previous exponential model except μ and J_m representing the mean shear modulus and the limiting fiber extensibility parameter.

Table 1 Material parameters

Model	c [kPa]	k_1 [kPa]	k_2 [1]	β [°]
(2)	0	15.17	6.248	43.8
(2)	2.858	15.17	6.962	45.11
Model	c [kPa]	μ [kPa]	J_m [1]	β [°]
(3)	0	24.41	0.7493	41.82
(3)	2.5	24.41	0.7498	41.93

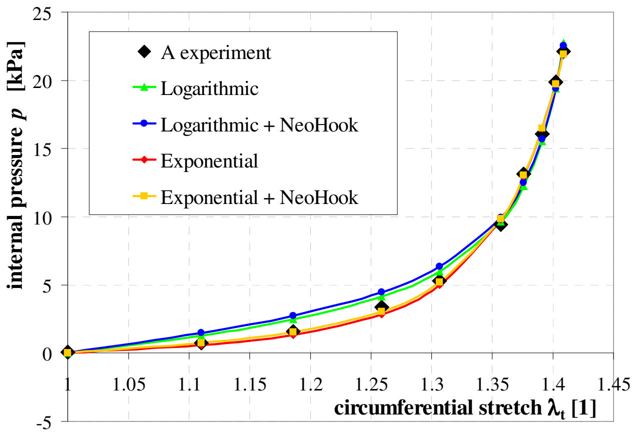

Fig. 5 Results of regression. Model predictions and experimentally measured values of internal pressure for cycle A (no axial weight).

VI. RESULTS AND CONCLUSION

Least square optimization based on radial and axial equilibrium of the cylindrical thick-walled tube with closed end yields estimations of the material parameters in the models (2) and (3). The regression was performed for the A cycle only (zero axial load), results are shown in Fig. 5 and estimated parameters are presented in Tab. 1. They slightly differ from data published previously in [14] because here a more accurate evaluation of the reference configuration was used. Both models fit selected cycle successfully. Model (2) fits slightly better than (3). The neo-Hook term is not important from the regression point of view. But it is included due to compatibility with linear theory, see paper [15]. Relative error of the axial force prediction never exceeded 7.5%. The results suggest hypothesis that there exists no unique relationship between the deformation and the stress state.

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