Kronek, J., Horny, L., Adamek, T., Chlup, H., & Zitny, R. (2014). Site-specific mechanical properties of aortic bifurcation. *IFMBE Proceedings, 41*, 939-942. DOI: 10.1007/978-3-319-00846-2_232 MANUSCRIPT VERSION
Publisher link: http://link.springer.com/chapter/10.1007%2F978-3-319-00846-2_232

Site-Specific Mechanical Properties of Aortic Bifurcation

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Abstract— **This study describes stiffness of arterial wall near aortic bifurcation. Uni-axial tensile tests with specimens cut out from human abdominal aorta, aortic bifurcation and common iliac arteries were carried out. Material anisotropy was verified as well as site-specific stiffness. The stiffness was described by initial secant modulus of elasticity (E_{INIT}). This parameter was gained from stress/strain data at 12% of deformation. The tissue samples in various regions and directions were concluded as described as follows: region of aorto-iliac bifurcation is significantly stiffer in longitudinal direction than in the circumferential direction in all of three examined parts of arterial tree. The comparison between tissue samples cut and loaded in longitudinal direction indicates that arterial tree stiffens in caudal direction. No significant change in material properties was found in the apex of bifurcation in comparison with adjacent tissue. Stiffening of arterial tree during ageing was verified.**

Keywords— **aortic bifurcation, mechanical properties, atherosclerosis, anisotropy**

I. Introduction

The relationship between atherosclerosis and the mechanical state of arterial wall have been studied extensively. For instance [1] reported that regions of arterial wall near ostia of branches, which represent concentrators of the wall stress, are saturated by low-density lipoproteins (LDL) more than straight parts of arteries. Similar results were obtained by Bratzleret in [2]. It suggests that mechano-biological interaction could be a key factor for atherosclerosis development. The question is which one is the main factor and which factors do not play a role. Authors of [3] wonder, why intramyocardial coronary arteries (inside the heart) remain free of atherosclerosis, whereas the epicardial arteries (on the heart surface) do not. The result probably is, that surrounding tissue (myocardium) acts as a support, which decrease the transmural pressure and thus it decrease circumferential stress and strain of these arteries. Previous complex study [4] examined, among others, the distribution of LDL in the vertebral artery. Amount of LDL in the parts of the vertebral artery passing through bone canal is lower than in the parts of the artery between the two bone canals. Surrounding stiff bones also play a role of support protecting the artery from systolic stretch. Both of these studies suggest that the transmural pressure, which causes circular (and partly axial) stress, is a main atherosclerosis generator / accelerator instead of flow conditions. Another evidence of relationship between transmural pressure and atherosclerosis is a well known risk factor of high blood pressure. According to Fuster et al. in [5] the probability of coronary artery disease and stroke increases exponentially with blood pressure.

Stress and strain distribution within arterial wall is next to boundary conditions (pressure, surrounding) given by both geometry (thickness, branch and non-planarity angles) and material properties (nonlinear stiffness, viscoelasticity, anisotropy). For instance the geometry of a FEM model of the artery bifurcation [6] is based on precise medical imaging, material properties are not site-specific. This study and the later publication [7] deal with bifurcations and branches of arteries as stress concentrators.

Physiological principle of homeostasis governs mechanobiological interaction of living tissues with mechanical environment. A response of a tissue to changes in mechanical environment can be mediated by changes in geometry (radius, thickness) but also with a change in internal structure which is then observed as a change in macroscopical mechanical properties. That is why an understanding of site-specific stress-strain relationships is essential for etiology or tissue engineering.

Aorto-iliac bifurcation (AB) is an area dividing high pressure blood flow from descending abdominal aorta into pelvis and lower limbs.

AB seems to be a suitable part of arterial tree for studying phenomena of material non-uniformities because of several reasons: 1) it is large enough to prepare samples for mechanical tests from it, 2) it is geometrically fairly uniform (more than the other bifurcations of large arteries) and 3) obstructed arteries in the area of AB is also clinically serious (decreased blood supply of lower extremities often causes pain and even may lead to amputation).

There are only a few of articles [8-10] dealing with geometrical parameters of AB systematically. Moreover, to the best of our knowledge no study investigating material properties at AB linked to pathological conditions is available in the literature.

II. Material and Methods

Five cadaveric tissue donors were involved in present study (age 22, 50, 51, 58, 64 years, all male). Abdominal aorta (AA), aortic bifurcation (AB) and common iliac arteries (CIA) were resected from each cadaver. The tissue specimens were cut out and prepared for mechanical testing immediately after removing. Rectangular strips were prepared and grouped according to their location and orientation as schematically described in Fig.1.

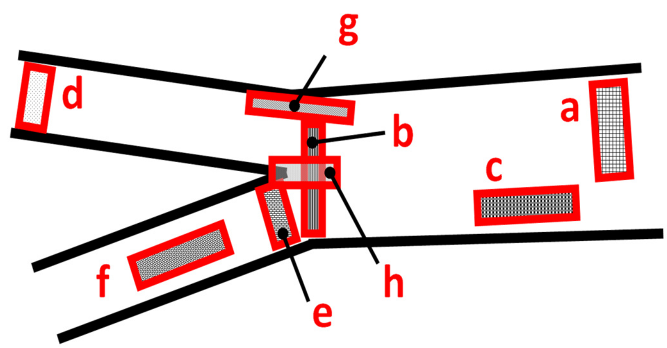

Fig 1 Position and orientation of samples groups: samples cut out from AA in the circumferential direction a) distant from AB (n=6) and b) near the AB (n=3), c) samples cut out from AA in the longitudinal direction (n=6), samples cut out from CIA in the circumferential direction d) distant from the AB (n=3) and e) near AB (n=5), f) samples cut out from CIA in the longitudinal direction (n=4), g) samples cut out from lateral side of AB (n=4) and h) samples cut out from apex (flow divider) of AB (n=4).

A pair of lengthwise and a pair of crosswise lines were painted on intimal surface of the samples to allow us optical recording of deformation during loading. Each sample was loaded in the direction of its longer side in customer-specific tensile testing machine (Zwick/Roell). Four precycles (up to the stretch λ=1.1) and stress relaxation for 100s were applied prior to final testing (up to λ=1.4 or a failure; Fig.2). Loading speed was 0.1mm/s (crosshead velocity). It means that stretch rate was between $0.003s^{-1}$ and $0.01s^{-1}$ according to the sample length. Acting force was recorded by force transducers (U9B HBM). The geometry of samples was measured before testing. Nominal engineering stress-strain curves were evaluated from the final loading cycle.

Stress-strain data were fitted by two-parameter Mooney-Rivlin hyperelastic model (not presented in this paper). In order to simplify interpretation of mechanical properties of the tissue, initial secant modulus of elasticity (E_{INIT}) was also evaluated considering it as a slope of the line from the origin to a point [ε=0.12, $\sigma(\varepsilon=0.12)$] or a point of failure initiation. The strain was calculated as a ratio between the elongation (actual length minus reference length) and the reference length. Reference length is the length of a sample in the beginning of a final loading cycle. Also width and thickness of the samples were measured in reference configuration. An example of stress-strain curve is shown in the Fig. 3.

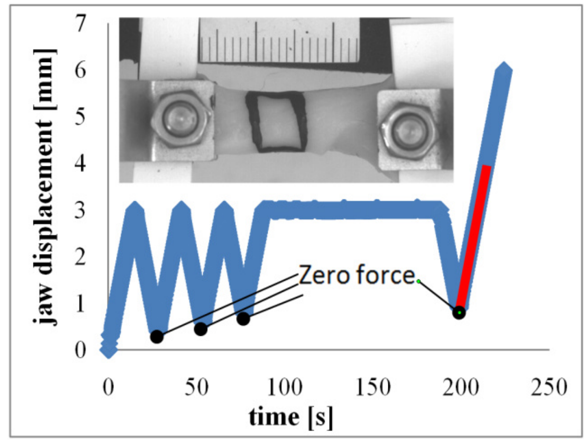

Fig. 2 Loading regime consists in four precycles (up to λ=1.1), 100s stress relaxation and final loading cycle up to a failure. Red line denotes evaluated zone. The picture of clamped sample is inserted.

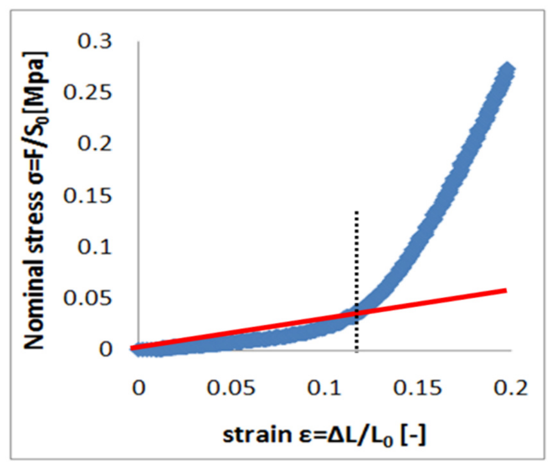

Fig. 3 Example of stress-strain relationship. The slope of red line represents initial secant modulus of elasticity.

Computed E_{INITs} were normalized with respect to mean E_{INITs} calculated for each donor. Subsequently normalized E_{INIT} were grouped with respect to the location and orientation (Fig. 1) and averaged (Fig. 4). To better understand the procedure, there is a sequence of calculation steps: 1) parameters E_{INITs} were assigned to each measured sample, 2) mean E_{INITs} were calculated for each donor separately, 3) normalized E_{INITs} were calculated as E_{INIT} divided by relevant mean E_{INIT}, 4) normalized E_{INIT} from the same sample group were averaged.

III. RESULTS AND DISCUSSION

Considerable differences between parameters E_{INIT} donor by donor were found. Therefore the parameters were normalized. Results (see Fig. 4) suggest that in the region of AB arterial wall is significantly stiffer in longitudinal direction than in the circumferential direction (groups c vs. a, g vs. e or b and f vs. d). Comparison between f, g and c indicates that arterial tree stiffens in caudal direction. This is in accordance with well-known decrease in the number of elastic lamellae in the wall. When a variance of the results is considered, the data does not seem to suggest significant trend or differences in the circumferential direction. It should however be noted that results were obtained only for five donors as they are a part of the feasibility study.

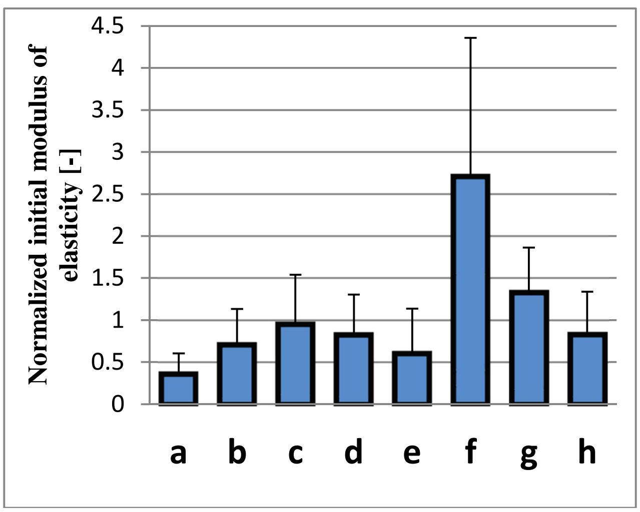

Fig. 4 Normalized initial moduli of elasticity. The figure shows how many times the one group of samples is stiffer than the average. The horizontal tick above each bar represents standard deviation.

Authors are aware of inaccuracies associated with the fact, that naturally anisotropic material was cut and tested in each case only in one direction. This procedure was chosen for two reasons. First geometrical complexity of arterial wall in bifurcations does not allow us to carry out a biaxial tensile test. Second the aim of the study was to illustrate site-specific mechanical properties as clear as possible based on the only one parameter. Normalized E_{INIT} was chosen as an appropriate parameter.

Referring to [7], the stress concentration factors (K_σ) differ only by 7% if an effect of anisotropy is taken into account or is not (evaluated from the FEM model of arterial branch). K_σ is a ratio between max stress (at the branch ostium) and nominal stress (at uniform tube-like part of a main artery). It could justify the performance of uni-axial tensile test with anisotropic material.

The relaxation zone prior to final testing was included in order to evaluate viscoelastic properties of the tissue. Unfortunately these stress-on-time data showed no significant trend and do not provide any additional information.

Finally it should be noted that the basic hypothesis of this approach is that material properties of arterial tree affect (initialize or accelerate) atherosclerosis, but also reverse causality applies. Thus, it is possible, that atherosclerosis consequence was measured rather than its reason. An influence of a plaque occurrence on wall stiffness was discussed in a large number of studies. During first stage of atherosclerosis, LDL enter the intima layer. The intima layer begins to be foam-like and soften. However, at a later stage of atherosclerosis calcification deposits make an arterial wall stiffer.

The positive correlation between the age and probability of arteriosclerosis is known. Our investigation confirmed this correlation. Generally the tissue samples of "older" donors (age 51, 58 and 64) were less ductile and stiffer than the tissue samples from "younger" donors (age 22 and 50). Comparison of E_{INIT} averaged over all samples from each donor can be seen in Tab.1. It is clear at first sight, why it was not possible to compare groups of samples without normalizing parameters E_{INIT} by the averages.

age of the donor	mean E_{INIT}
22	0.28
50	0.66
51	2.63
58	2.57
64	2.40

Tab. 1 Mean E_{INIT} for each donor. It is calculated from all samples for each donor separately. The tab. Illustrate, that arterial wall obtained from youngest donor, was generally weaker than the rest of samples.

IV. CONCLUSIONS

Mechanical properties of arterial wall were reduced to only one parameter – initial moduli of elasticity in corresponding direction.

This study showed that the region of AB does not differ sharply in mechanical properties from the arteries proximal and distal. It is implicated from above that this area represents a transition region, where mechanical properties change continuously from relatively compliant aorta to relatively stiff arteries of lower extremities. The tissue of arterial wall is stiffer in the longitudinal direction compared to the circumferential direction. No significant trend was found in circumferential direction.

ACKNOWLEDGMENT

The project was supported by the Grant Agency of the Czech Technical University in Prague, grant SGS13/176/OHK2/3T/12, the Czech Ministry of Health, grant NT 13302 and the project TACR TA0101185.

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