Acta of Bioengineering and Biomechanics
Vol. 7, No. 1, 2005

Analysis of the mechanical parameters of human brain aneurysm

BRIGITTA K. TÓTH

Budapest University of Technology and Economics,
1111 Budapest, Műegyetem r. 3., Hungary, brigitta.toth@biomech.bme.hu

GÁBOR RAFFAI

Clinical Research Department, Second Institute of Physiology, Semmelweis University of Medicine,
1082 Budapest, Üllői út 78/a., Hungary, raffai@elet2.sote.hu

IMRE BOJTÁR

Budapest University of Technology and Economics,
1111 Budapest, Műegyetem r. 3., Hungary, ibojtar@mail.bme.hu

Cardiovascular disease is one of the most frequent reasons of mortality in the western word. Nowadays the mechanical properties of biological soft tissues were treated from a continuum mechanical perspective. The aim of this article is to investigate the mechanical response of arterial tissue. We present some three-dimensional finite element model to study the mechanical effects. The arterial wall is composed mainly of an isotropic matrix material (elastin) and collagen fibers from two families which are arranged in symmetrical spirals. These fibers induce the anisotropy in the material response. So the constitutive law of an artery is orthotropic. We want to develop a new constitutive law for arterial wall mechanics. In addition we make a comparative study of some material model used in the literature to describe the mechanical response of arteries. These are the following models: 1. Linearly elastic model. 2. Neo-Hookean model for incompressible materials. 3. Mooney–Rivlin model for incompressible materials. For this reason we make uniaxial and biaxial measurements to have appropriate parameters for the underlying material models. We investigate the biomechanical properties of strips from human cerebral aneurysms from surgery and cadavers. (An aneurysm is a bulge along a blood vessel.) Meridional and circumferential, thick and thin parts were distinguished respectively. This paper focuses on the analysis of the haemodynamic pattern and biophysical properties of cerebral aneurysms, diagnosed and delineated in living human individuals. The aim of this research is to estimate stresses at critical points of the aneurysm wall and its parent artery, and to estimate the likelihood of a later aneurysm rupture.

Key words: arterial tissue, cerebral aneurysm, finite element model, material model, vascular biomechanical properties, viscoelasticity, in vitro tests

1. Introduction

Brain arterial aneurysms are common forms of arterial deformation occurring in about 5% of the adult population. The aneurysm is a bulge along the artery hanging there embedded in the surrounding tissue. In most situations, it usually appears around a joining of two arteries. This bifurcation is the part of the supplier of the brain vascular bed system so if its blow-out (rupture) causes incalculable chain reaction, there is no safe solution without any side-effect to protect the patient against unpleasant consequences, see for instance [3], [28], [40]. In the majority of the cases, the patient does not notice any symptoms of the aneurysm, in some cases, however, the aneurysm bursts leading to stroke and immediate death. Figure 1a presents a photo of an aneurysm obtained by planar angiography, while figure 1b illustrates the characteristic places of aneurysms in the human brain.

At present, the therapeutic decision on unruptured aneurysms is made purely on the basis of the size and location of the lesion in the belief that those are the only factors influencing the likelihood of rupture.

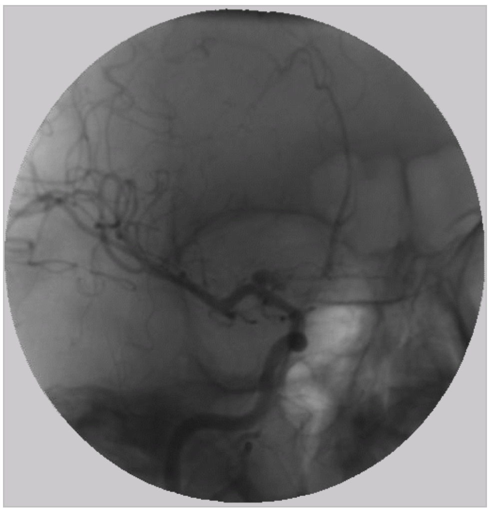


Fig. 1a. Photo of the aneurysm

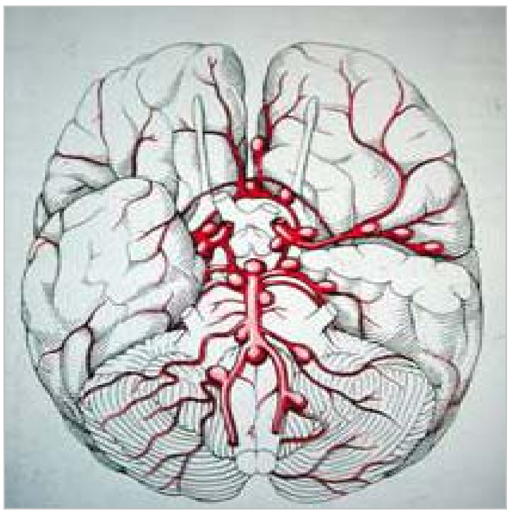


Fig. 1b. The usual places where brain aneurysms occur

The literature results providing the basis for this practice have been seriously criticized by many researchers and clinical decision is frequently based on the personal experience and judgment of the doctor. Our work will provide the physicians – and the patients – with a much more accurate prognosis of the disease that will allow

for
a more appropriate decision about treatment. We note that besides its scientific merit, the potential for providing information about the prognosis of the disease and about the optimal technique for its treatment would greatly enhance the value of the modern angiography systems.

Nevertheless the importance of this area cannot be overestimated – the most important causes of death in developed countries are arterial deceases. Research budgets and a public interest in this subject have been grown continuously.

2. Methods

2.1. Research strategy

The geometrical and morphological data as well as physiological parameters of the patients were collected at the National Institute of Neurosurgery and combined with physiological information about the vessel wall and aneurysm wall provided by the Department of Human Physiology (Semmelweis Medical University, Budapest). The determination of the material parameters of the aneurysm wall is the first step of our research. Based on this information, the researchers from the Research Centre for Biomechanics (TU Budapest) prepared 3D coupled (flow and solid) finite element models of the aneurysms. Strength calculations were done on these models in order to predict their mechanical strength (allowable blood pressure, etc.) and to compare the effects of different possible medical treatments. This is the second step of the research activity.

The final goal of this research program was to construct non-intrusive diagnostic tools detecting the presence of aneurysms and to assess the necessity of medical intervention.

Since the process is highly unsteady, and the elastic deformation of the wall and the flow in the arteries are strongly coupled, the solution is far from trivial. After performing simulations on rigid models for fluid mechanics on the one hand and on wall only models for the elasticity studies on the other, the next step will be to couple both phenomena and the data will be transferred in each time step to provide time-dependent boundary conditions for both simulations.

Specific aim of this study was to characterize quantitatively the behaviour of strips from human cerebral aneurysm and to simulate the mechanical behaviour of the vascular wall using three-dimensional finite element models

2.2. Material parameters

One of the problems is that different constitutive models in the available literature are based on data from different types of arteries [27], [2], [4]. Moreover, cardiovascular disease like human cerebral aneurysm can only be studied in detail if a reliable constitutive model of the arterial wall is available. In order to understand the sterically inhomogeneous behaviour of cerebral aneurysms, we measured the mechanical properties of the aneurysm tissue as a function of strain in different regions (thin and thick) and in different directions (meridional and circumferential). The strips from aneurysms showed typical hyperelastic-plastic behaviour in the stress–relaxation tests. Meridional thin strips exhibited larger tensile strenghts than the meridional thick ones, see [32].

First we summarize the theoretical framework of the description of the arterial wall mechanics. We begin by giving a brief description of the histological structure of arterial walls and we outline the general characteristic of the mechanical response of arteries. An artery can practically be treated as a thick-walled circular cylinder which is appropriate for the analysis of bending, extension, inflation and torsion of the tube. In the literature, some models are able to provide a full three-dimensional description of the state of stress in the artery, but a large number of material constants may lead to parameter identification problems. Several models use geometrical simplifications, too.

Then we present the uniaxial clinical studies which constitute the basis for quantifying material properties such as Young's modulus.

2.3. Human arterial histology

In general, arteries are subdivided into two types: elastic and muscular arteries [1], [5]. Elastic arteries have relatively large diameters and are located close to the heart, while muscular arteries are located on the periphery. We focus our attention on the microscopic structure of muscular arterial walls composed of three distinct layers. These are the tunica intima, the tunica media and the tunica adventitia.

The tunica intima is the innermost layer of the artery. The arterial wall is lined with a single layer of endothelial cells. In healthy young arteries, the intima is very thin and does not contribute significantly to the solid mechanical properties of the human arterial wall. But it is known that pathological changes of the intimal components are associated with significant alterations in the mechanical properties.

The tunica media is the middle layer of the artery and from mechanical perspective it is the most significant layer. It consists of a complex three-dimensional network of smooth muscle cells, elastin and collagen fibrils. The so-called fenestrated elastic laminae separate the media which form concentrically fiber-reinforced layers, and this middle layer is separated from the intima and the adventitia by the internal elastic laminae and the external elastic laminae, respectively. The smooth muscle cells, the elastin and collagen fibrils and the fenestrated elastic laminae constitute a continuous, almost circumferentially oriented fibrous helix. This arrangement gives high strength and resilience.

The tunica adventitia is the outermost layer of the artery. The thickness of the adventitia depends on the type of the artery and its topographical site. Apart from a histological ground substance it also consists of fibroblasts, fibrocytes and thick

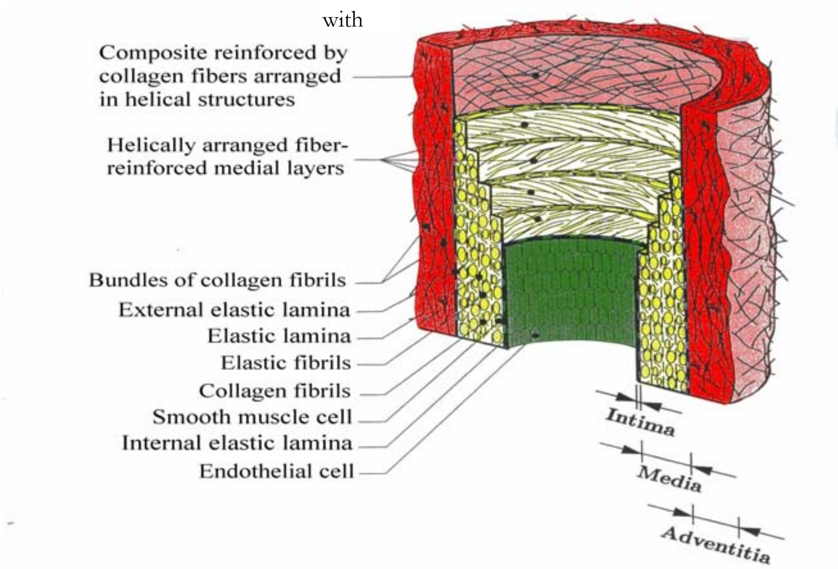


Fig. 2. Model of the maior components of a healthy artery composed of three layers: intima, media and adventitia

bundles of wavy collagen fibrils, which are arranged in helical structures. They contribute to the strength and stability of the arterial wall. The adventitia is less stiff at low pressure than the media but at higher pressures the collagen fibrils straighten out and the adventitia turns into a stiff tube.

2.4. Typical mechanical behaviour of arterial walls

The reliability of material parameters is related to the quality of the experimental data [13], [18], [20], [22]. It may come from *in vivo* tests or from *in vitro* tests. By *in vivo* tests the artery is observed under real life conditions, while *in vitro* tests mimic real loading conditions in a physiological environment. The complex anisotropic material response can only be measured in an *in vitro* experiment, though the exact physiological circumstances can be rather difficult to simulate.

Arteries do not change their volume in the physiological range of the deformation, for this reason they can be regarded as incompressible – rubber like – materials. Therefore we have set the task to determine the mechanical properties from biaxial tests: uniaxial extension tests are certainly insufficient to quantify completely the mechanical behaviour of arterial walls.

The mechanical behaviour of arteries depends on physiological and chemical environmental factors, therefore they were tested in appropriately oxygenated, temperature-controlled salt solutions.

Whereas the composition of arterial walls varies along the arterial tree and thus the shape of the stress–strain curve for blood vessels depends on the anatomical site, the general mechanical characteristics are the same.

The artery is a heterogeneous system and it can be regarded as a fiber-reinforced composite biomaterial. The layers of the arterial walls are composed mainly of an isotropic matrix material (associated with the elastin) and the fibers (associated with the collagen) from two families, the latter being arranged in symmetrical spirals. We note that we have made a simple independent finite element simulation of the biomechanical behaviour of arterial wall to check the effect of the different parameters in different constitutive equations.

2.5. Continuum-mechanical framework

Fundamental equations are essential to characterize kinematics, stresses and balance principles and hold for any continuum body [14]. Generally we use a functional relationship as a constitutive equation, which determines the state of stress at any point x of a continuum body. Our main goal is to study various constitutive equations, in the field of solid mechanics, appropriate for approximation techniques. We follow the so-called phenomenological approach which describes the macroscopic behaviour of living tissues as continua.

Numerous materials can sustain finite strains without any noticeable volume changes. Such materials can be regarded as incompressible, according to a common idealization in continuum mechanics. Materials which maintain the constant volume throughout a motion are characterized by the incompressibility constraint $J = 1$, where J means the determinant of the gradient tensor. In general, these materials are referred to as constrained materials.

The stress response of hyperelastic materials is derived from a given strain energy function Ψ:

$$\sigma_{ij} = \frac{\partial \Psi(\boldsymbol{\varepsilon})}{\partial \varepsilon_{ij}} .$$

In the next section, we summarize the most important energy functions frequently used in biomechanics.

2.5.1. The Ogden model for incompressible rubber-like materials

The postulated strain energy function Ψ describes the principal stretches λ_a, $a = 1, 2, 3$, that change from the reference configuration to the current configuration:

$$\Psi = \Psi(\lambda_1, \lambda_2, \lambda_3) = \sum_{p=1}^{N} \frac{\mu_p}{\alpha_p} (\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3),$$

where N is a positive integer which determines the number of terms in the strain energy function, μ_p are constant shear moduli and α_p are dimensionless constants, $p = 1, \ldots, N$. Only three pairs of constants are required to give an excellent correlation with experimental stress–deformation data.

After differentiation we find that the three principal values σ_a of the Cauchy stresses assume the form:

$$\sigma_a = -p + \sum_{p=1}^{N} \mu_p \lambda_a^{\alpha_p}, \quad a = 1, 2, 3,$$

where p is a scalar not specified by a constitutive equation. It is determined from a boundary condition of the problem examined.

2.5.2. The Mooney–Rivlin model for incompressible rubber-like materials

In the Mooney–Rivlin model, we assume the following values: $N = 2$, $\alpha_1 = 2$, $\alpha_2 = -2$. Using the strain invariants I_1, I_2 and the constraint condition $I_3 = \lambda_1^2 \lambda_2^2 \lambda_3^2 = 1$ we arrive at:

$$\Psi = c_1(\lambda_1^2 + \lambda_2^2 + \lambda_3^2 - 3) + c_2(\lambda_1^{-2} + \lambda_2^{-2} + \lambda_3^{-2} - 3) = c_1(I_1 - 3) + c_2(I_2 - 3)$$

with the constants $c_1 = \mu_1/2$ and $c_2 = -\mu_2/2$.

The derivatives of the strain energy function of the Mooney–Rivlin model with respect to the invariants I_1 and I_2 give the simple associated stress relations:

$$\boldsymbol{\sigma} = -p\mathbf{I} + 2c_1\mathbf{b} - 2c_2\mathbf{b}^{-1},$$

where the strain tensor $\mathbf{b}^{-1}$ is the inverse of the left Cauchy–Green tensor $\mathbf{b}$, which is defined based on the gradient tensor $\mathbf{F}$: $\mathbf{b} = \mathbf{F}\mathbf{F}^T$ ($\mathbf{F}$ is on the left). It is an important strain measure in terms of spatial coordinates. $\mathbf{I}$ denotes the second-order unit tensor.

2.5.3. Neo-Hookean model for incompressible rubber-like materials

In the neo-Hookean model, $N = 1,\ \alpha_1 = 2$ are assumed. Using the first principal strain invariant I_1 we find that:

$$\Psi = c_1(\lambda_1^2 + \lambda_2^2 + \lambda_3^2 - 3) = c_1(I_1 - 3)$$

with the constant $c_1 = \mu_1 / 2$. The strain energy function involves a single parameter only and relies on phenomenological considerations.

Derivatives of the strain energy function of neo-Hookean model with respect to the invariants I_1 give the simple associated stress relations:

$$\sigma = -p\mathbf{I} + 2c_1\mathbf{b}\,,$$

where the strain tensor **b** is the left Cauchy–Green tensor, and **I** is the unit tensor.

2.5.4. Some other constitutive models for arterial walls

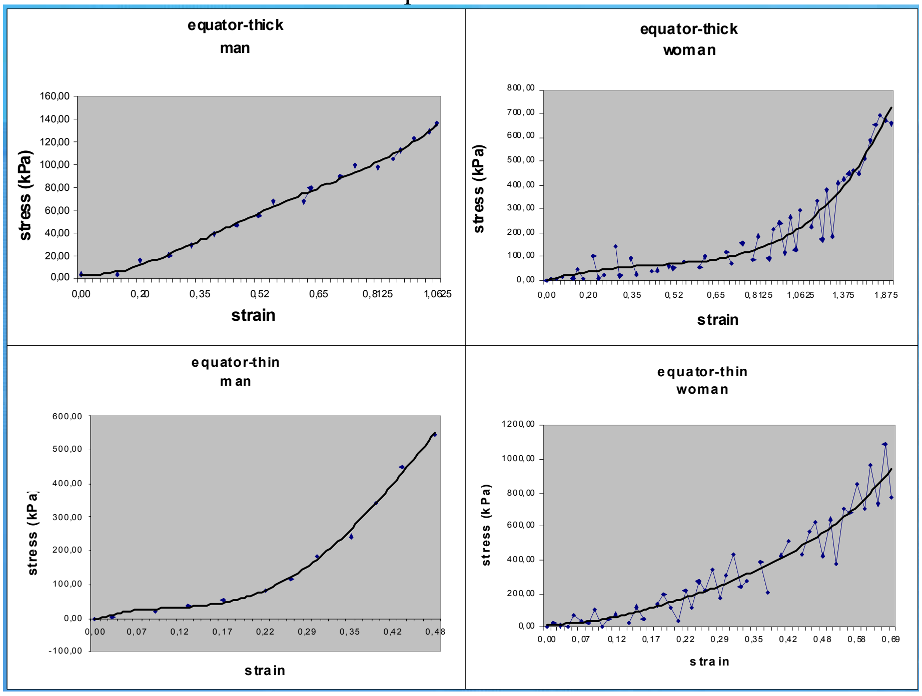
We note that other versions of the constitutive equations for human artery walls can be found in the literature, see for instance the works of Delfino, Vaishnav, Fung [52] and especially Holzapfel. In these models, an active mechanical behaviour of arterial walls is governed mainly by the intrinsic properties of elastin and collagen fibers and by the degree of activation of smooth muscles; however, a passive mechanical behaviour is quite different and is governed mainly by the elastin and the collagen fibers. Most constitutive models describe the artery as a macroscopic system and capture the response near the physiological state.

In our program – based on our laboratory tests – we applied the most common strain energy functions. In figure 3, our uniaxial and biaxial test machines can be seen, both connected to the computer. In our recent numerical simulations, the test results of the biaxial specimens were not applied yet because of the insufficient number of experimental results.

Fig. 3. The uniaxial and biaxial laboratory test machines

In figure 4, we show some characteristic experimental diagrams. We had 53 different specimens from 30 persons. All tests were carried out immediately after the operations, within 24 hours. Meridian and circumferential strips were cut and measured from the aneurysma sack.

Based on these experiments we calculated the material parameters of the Mooney–Rivlin and neo-Hooke an nonlinear hyperelastic models (see the table). Based on simple finite element tests all material parameters were checked.



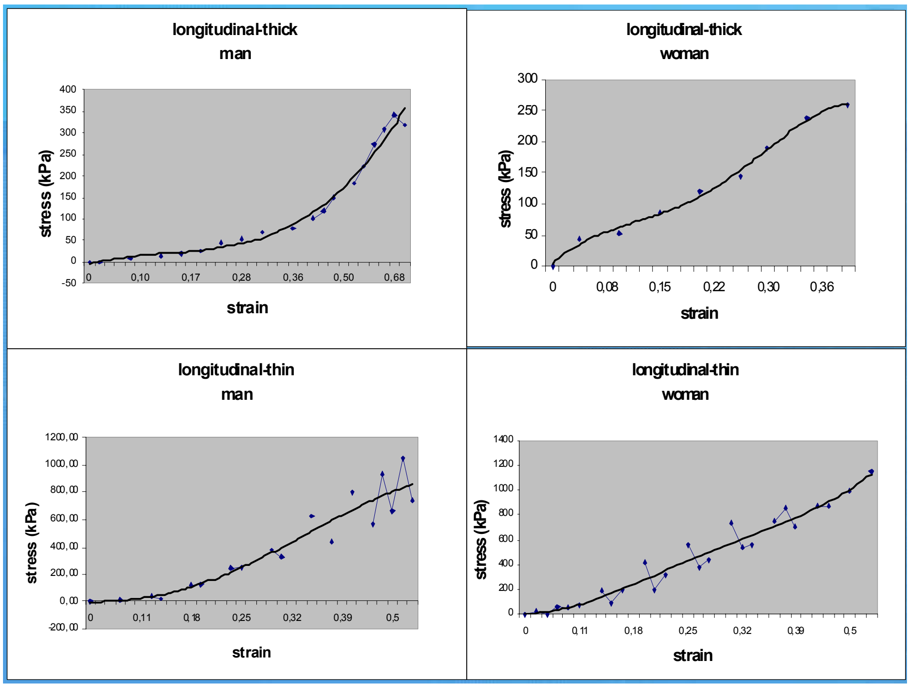


Fig. 4. Uniaxial stress–strain curves for meridian and circumferential aneurysma strips in passive condition (based on the tests performed in the authors' laboratory). A thick solid line represents an approximate engineering response of the biomaterial

Table. The Mooney–Rivlin and neo-Hookean material parameters calculated from experiments

	ε		σ [N/cm²]		**E** [N/cm²]	
	Feminine	Masculine	Feminine	Masculine	Feminine	Masculine
Equator-thick	1.80	1.10	70	14	39	12
Equator-thin	0.90	0.60	93	55	134	108
Longitudinal-thick	0.40	0.70	40	34	65	49
Longitudinal-thin	0.60	0.55	100	84	167	152

	Mooney–Rivlin				neo-Hookean	
	C10 [N/cm²]		C01 [N/cm²]		C10 [N/cm²]	
	Feminine	Masculine	Feminine	Masculine	Feminine	Masculine
Equator-thick	5.2	1.6	1.3	0.4	6.5	2.0
Equator-thin	17.9	14.4	4.5	3.6	22.4	18.0
Longitudinal-thick	8.7	6.5	2.2	1.6	10.9	8.1
Longitudinal-thin	22.3	20.3	5.6	5.0	2.79	25.3

2.6. Finite element analysis

The finite element simulations were performed using MSC Marc 2001, a commercially available finite element software. The cycle of finite element analysis involves six distinct steps: Step 1: mesh generation. Step 2: definition of the boundary conditions, initial conditions and links. Step 3: material and geometric properties. Step 4: contact. Step 5: load cases and jobs. Step 6: interpretation of the results.

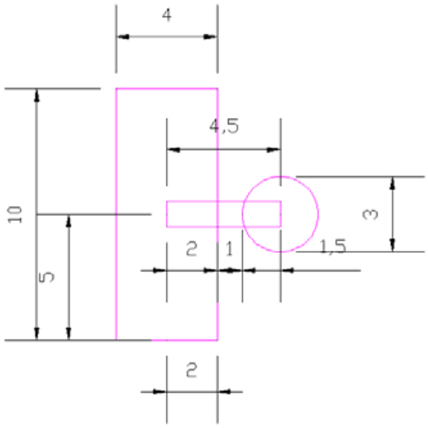


Fig. 5. Basic geometry of a typical aneurysm model. The dimensions of the structure are in [mm]

The radius to thickness ratio of the structure of aneurysm warrants the use of shell theory instead of a full three-dimensional analysis based on hexahedral elements. The aneurysm is modelled as a cylinder–cylinder-sphere intersection. Because of the symmetry, only a part of the aneurysm needs to be modelled. The thickness to radius ratio is small enough to allow us to use the shell approximation. The boundary conditions can be applied to the shell edge without affecting the stresses at the intersections. During the phase of mesh generation we used the four-noded shell element. The artery was considered to be a cylindrical tube with incompressible wall subjected to combined inflation, extension and torsion that mimic real conditions in a physiological environment.

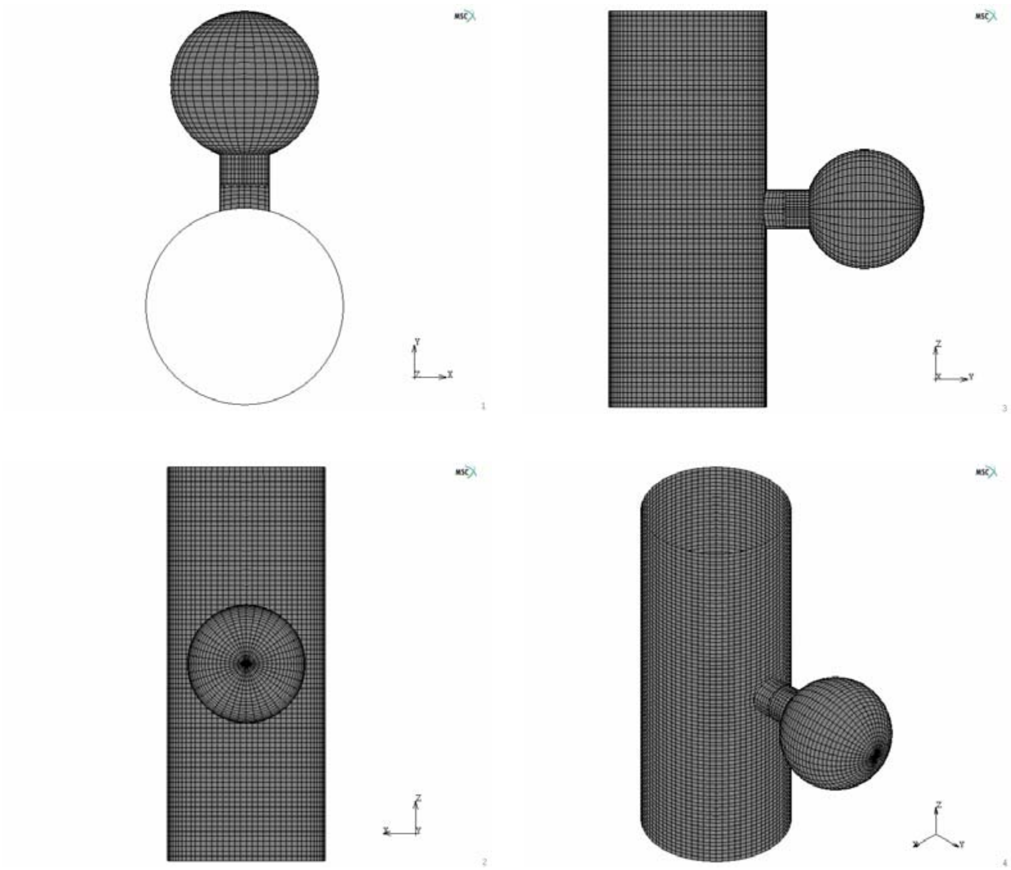


Fig. 6. A simplified model of aneurysm and its mesh generation

There are two types of boundary conditions: expressed in global and in local coordinates. The artery was considered to be a cylindrical tube with an incompressible wall subjected to combined inflation, extension and torsion that mimic real conditions in a physiological environment. Figure 5 shows the distribution of the Cauchy stress through the deformed arterial wall and the aneurysm.

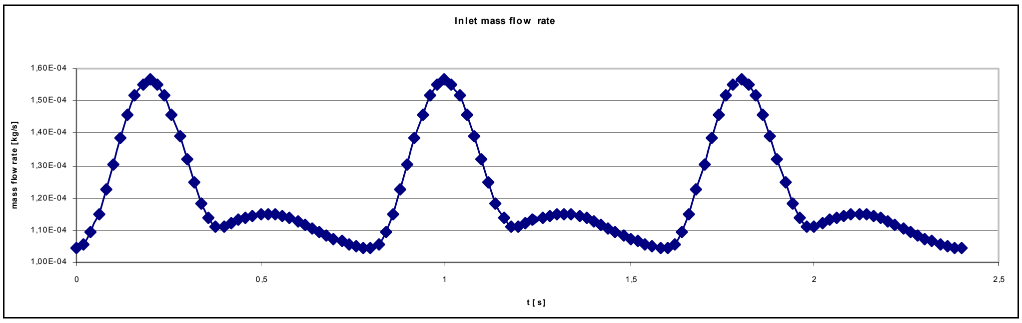


Fig. 7. Inlet boundary condition. Pulsing pressure in the artery; in domain: 80–120 Hgmm

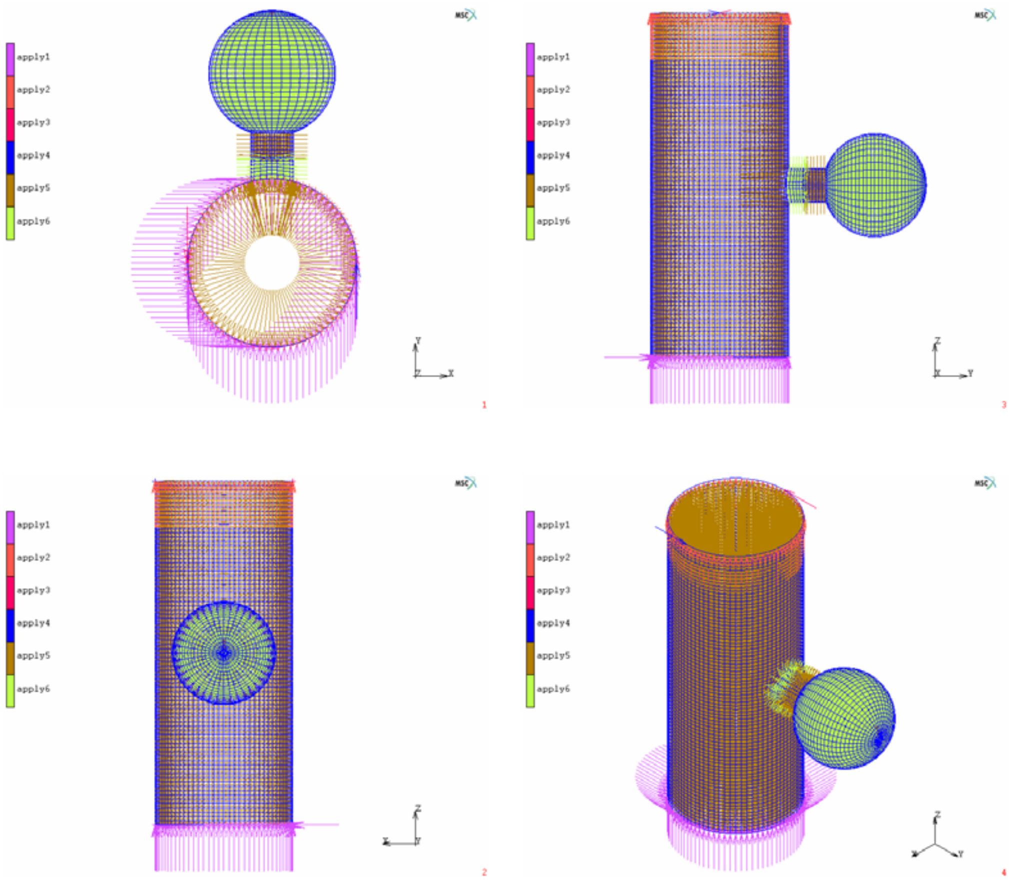


Fig. 8. Arterial tube and the aneurysm enlarged under combined inflation, extension and torsion. The loading conditions mimic real loading conditions in a physiological environment

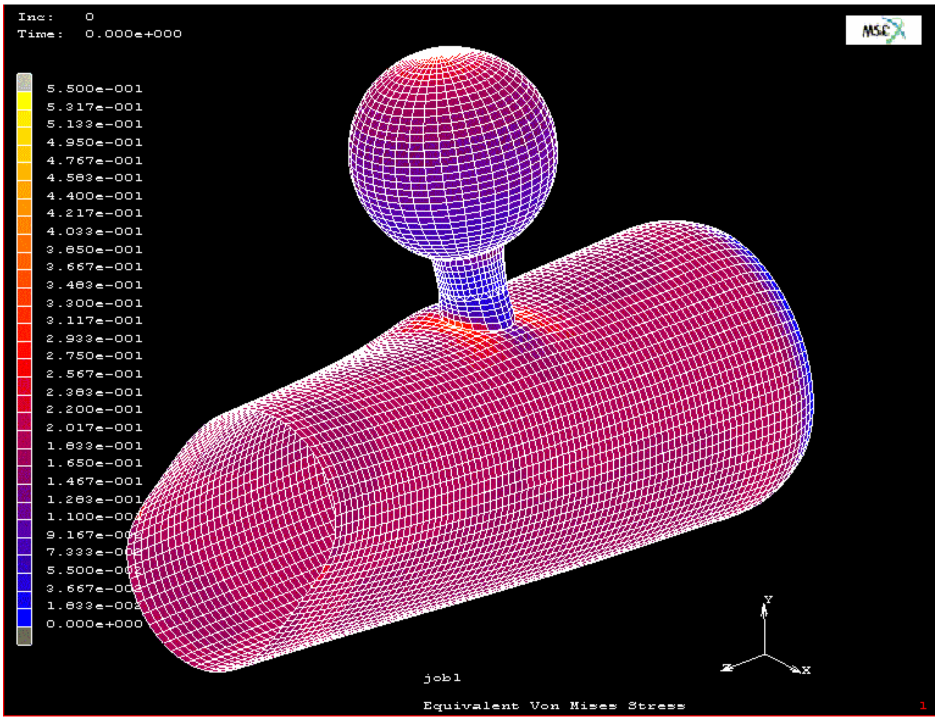


Fig. 9. Deformed states of the artery and the distribution of the Cauchy stress through the deformed arterial wall

Some preliminary results demonstrate how the behaviour of aneurysm can be simulated. This idealization may be far from the real aneurysm, but it is simple enough to test the model behaviour. The idealized model is made up of two cylinders and a sphere.

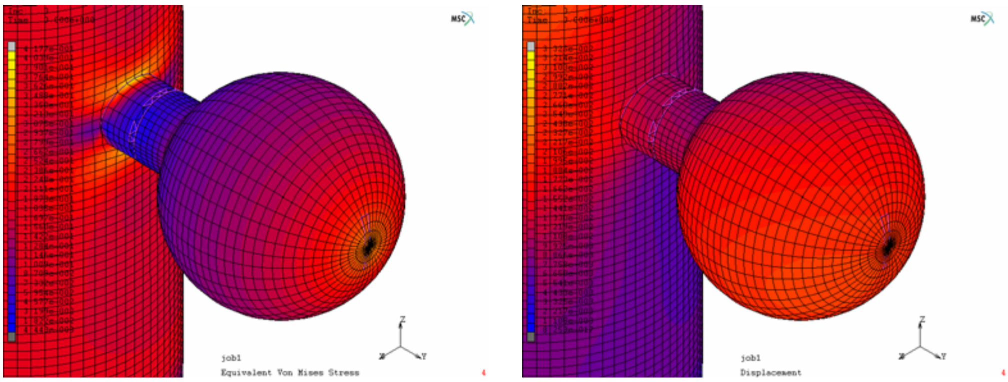


Fig. 10. The equivalent von Mises stress (left) and displacement (right) along the aneurysm walls

Figure 9a shows that the equivalent von Mises stress in the artery wall on the top of the aneurysm "bulge" is significant. This is in harmony with the medical observations. Note that higher stress values can only be detected at the "shoulders" of the aneurysm. This, however, cannot be the cause of the rupture since experience shows that it usually happens along the "equator" of the aneurysm.

We have made another finite element mesh too, which reproduces three layers of artery, although the number of layers can be changed. During the phase of mesh generation we used the hexahedron type elements with 8 nodes. Eight finite elements were applied along the wall thickness (two for the intima, four for the media and two for the adventitia). It was assumed that the media occupied 2/3 of the arterial wall thickness. In the circumferential and axial directions, an arterial layer was discretized by 40 and 40 finite elements, respectively. The configuration of one arterial layer was taken to correspond to a circular cylindrical tube with wall thickness, inner radius and length. The boundary conditions were the same. Figure 5 shows the distribution of the Cauchy stress through the deformed arterial wall. It can be seen that the stress distribution is mainly determined by the media, which is in agreement with the experimental findings. This result is an approach, since the adventitia is very soft in the associated strain domain and the intima is not of mechanical interest.

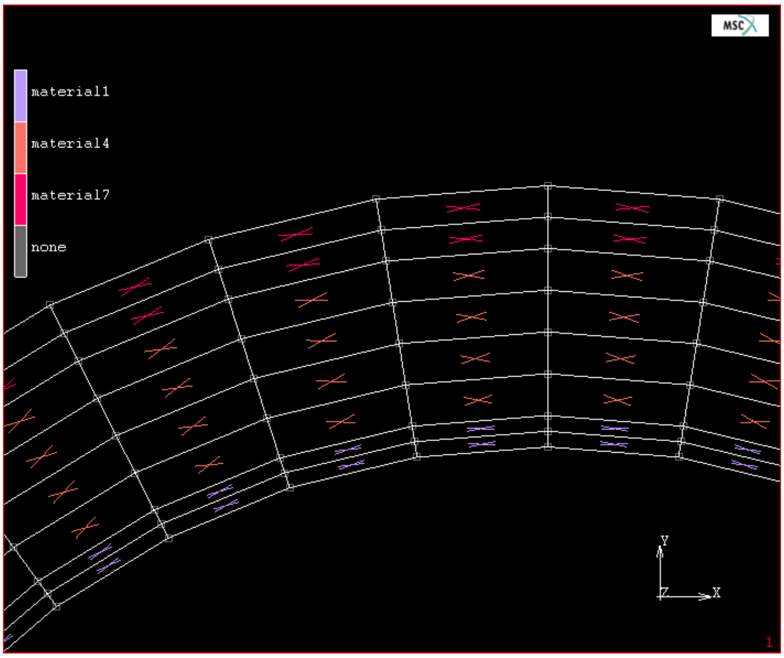


Fig. 11. Finite element model of a healthy elastic artery composed of three layers: intima (the innermost 2 layers), media (the 3rd–6th layers), adventitia (the 7th–8th layers)

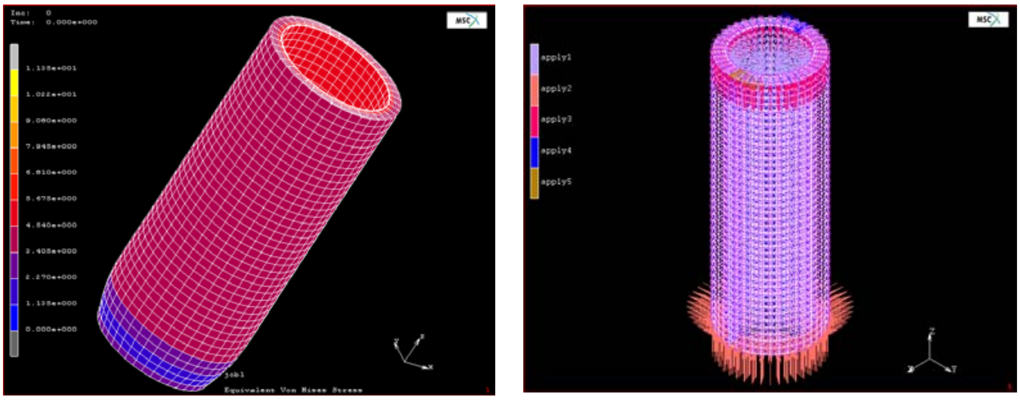


Fig. 12. Left: arterial tube under combined inflation, extension and torsion.
The loading conditions mimic the real loading conditions in a physiological environment.
Right: states of the artery deformation and the distribution of
the Cauchy stress through the deformed arterial wall

In the immediate future, we will embark on biaxial measurements. For this reason we have made a comparative study to decide which type of the grip is more favourable and which type has a smaller domain with disturbations. The two possible arrangements are shown in figure 13.

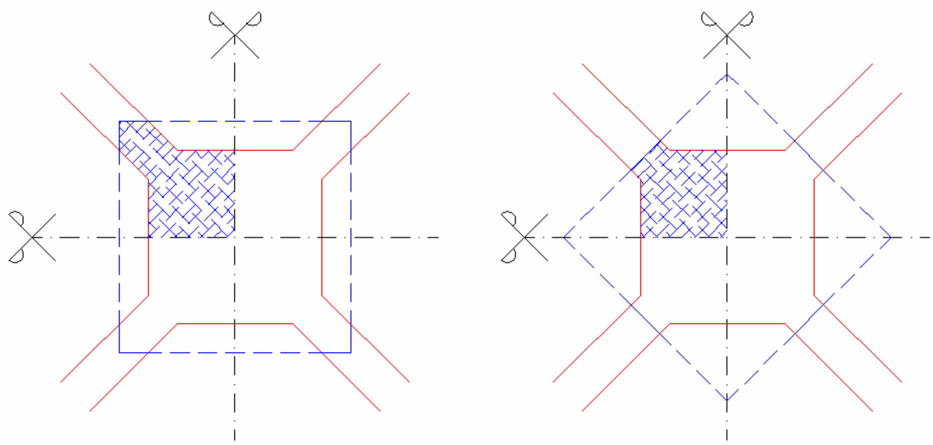


Fig. 13. Two possible arrangement of arterial strips

Due to symmetry it is sufficient to analyze the quarter of the vessels shown in figure 13 (see the sections marked by lines).

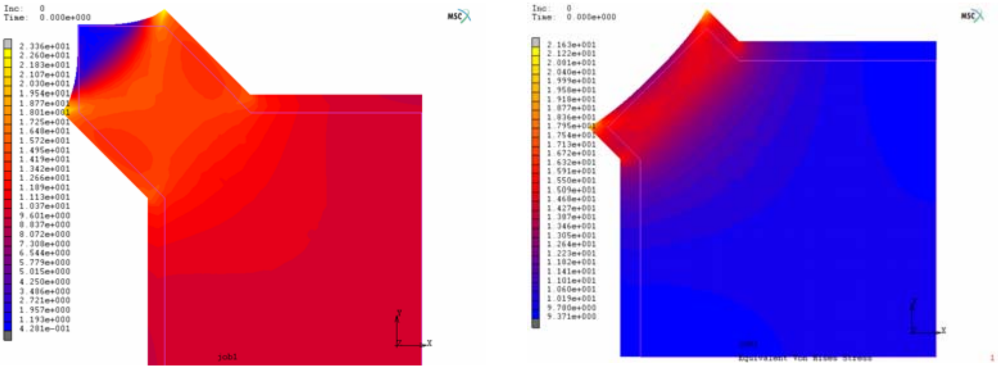


Fig. 14. Displacement and von Mises stresses by 5% extension
in two mentioned types, respectively

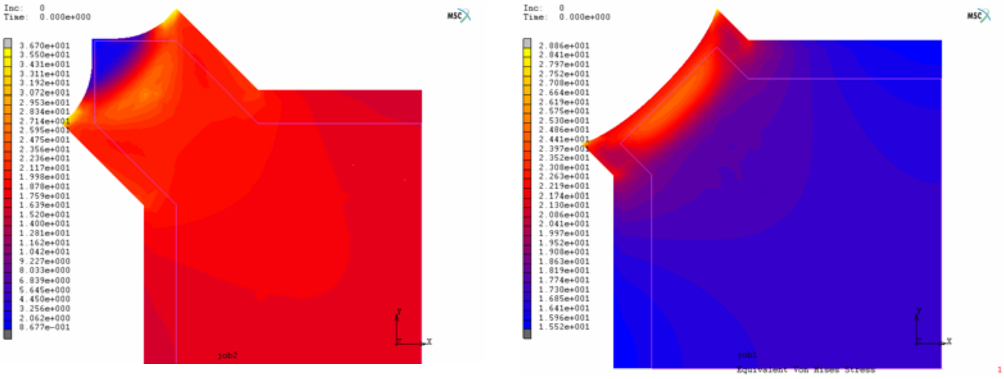


Fig. 15. Displacement and von Mises stresses by 10% extension
in two mentioned types, respectively

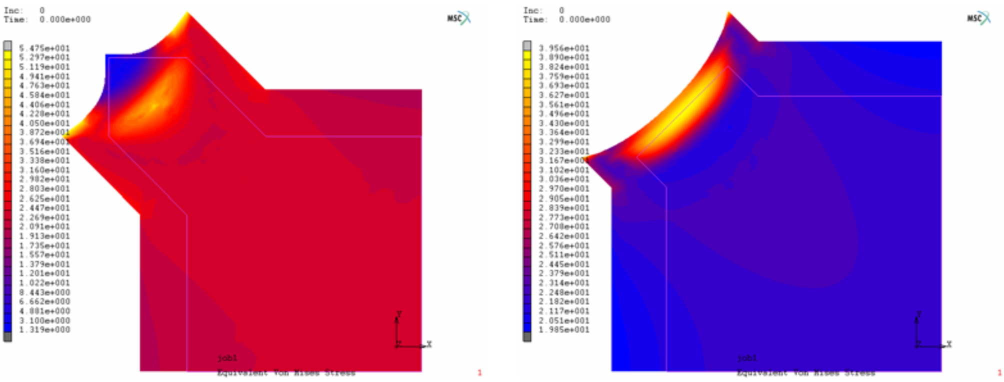


Fig. 16. Displacement and von Mises stresses by 15% extension
in two mentioned types, respectively

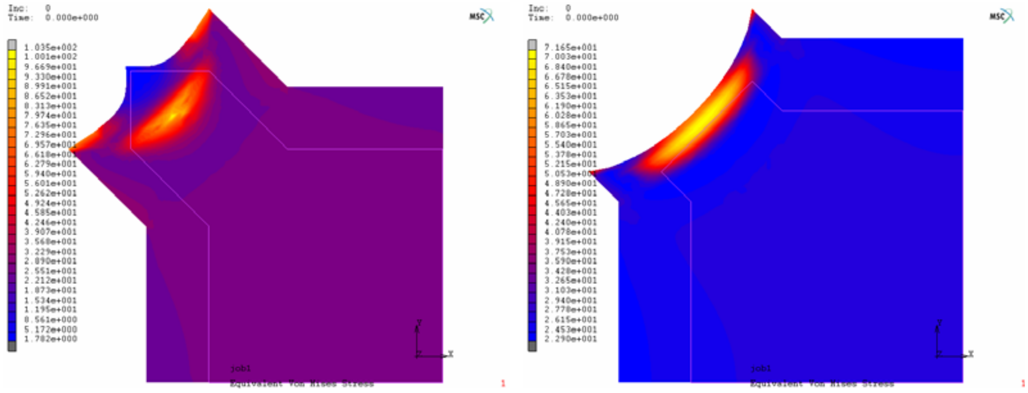


Fig. 17. Displacement and von Mises stresses by 20% extension
in two mentioned types, respectively

These results show that we cannot make a significant distinction between the two types. The difference between the colours in the middle of the artery segment has no physical sense. The stress values that we can read in contour plots are the same. Only the maximum and minimum values differ from each other.

Due to angiography we can build a model of the existent aneurysm, and this model can be used for Finite Element Analysis calculation, too. The angiography allows us to design a real three-dimensional model with an original geometry. When the aneurysm material parameters are used the system could help the doctors to analyze the case and to decide whether or not a patient needs an urgent operation. This way of geometrical modelling is much more complicated than the previous one. Many different tests are needed to declare that the system works reliably. Moreover, the analysis of different material models and boundary conditions need more and more calculations. With this part of the task we have just started to work. The system is almost ready for the final testing. Thanks to the engineers of General Electric, we are able to gain three-dimensional geometrical data from angiography.

3. Summary

All the models discussed are based on phenomenological approach in which the macroscopic nature of blood vessels is modelled. From this study it may be concluded that there is a need for a constitutive model which describes the viscoelastic behaviour of the human arterial wall. We have proposed an approach in which the arterial wall is approximated as a three-layer thick-walled tube, with each layer modelled as a fiber-reinforced composite in the domain of large deformations and we have examined a human arterial aneurysm.

In cooperation with the co-workers of the National Institute of Neurosurgery and Human Sciences, we made the first step in the complex numerical simulations of brain aneurysms.

This work was done with the help of an OTKA grant (principal investigator Doctor István Szikora, Nat. Inst. of Neurosurgery).

References

[1] FONYO A., *Book of Medical Physiology* (in Hungarian), Medicina, Budapest, 1999.
[2] CHOUNG C.J., FUNG Y.C., *Three-dimensional stress distribution in arteries*, J. Biomech. Engr., 1983, 105, 268–274.
[3] STEINMAN D.A., *Image-based computational simulation of flow dynamics in a giant intracranial aneurysm*, American J. of Neuroradiology, 2003, 24, 559–566.
[4] VORP D.A., RAJAGOPAL K.R., SMOLINSKY P.J., BOROVETZ H.S., *Identification of elastic properties of homogeneous orthotropic vascular segments in distension*, J. Biomech., 1995, 28, 501–512.
[5] MONOS E., *Hemodinamika: a vérkeringés dinamikája. SE egyetemi jegyzet*, Budapest, 2001.
[6] HUDETZ G., MONOS E., *A viscoelastic model of mechanically induced and spontaneous contractions of vascular smooth muscle*, Acta. Physiol. Hung., 1985, 65(2), 109–123.
[7] VAN DIJK M., WIERINGA P., VAN DER MEER A.M., LAIRD D.J., *Mechanics of resting isolated single vascular smooth muscle cells from bovine coronary artery*, Am. J. Physiol., 1984, 246(3 Pt 1), C277–C287.
[8] VIIDIK C., DANIELSEN C., OXLUND H., *On fundamental and phenomenological models, structure and mechanical properties of collagen, elastin and glycosaminoglycan complexes*, Biorheology, 1982, 19, 437–451.
[9] STEFANDALIS C., STRATOS C., VLACHOPOULOS S., MARAKAS H., BOUDOULAS S., TOUTOUZAS L., SIOROS, TOUTOUZAS P., *Pressure-diameter function and a new method of determination*, Circulation, 1995, 92(8), 2210–9.
[10] MONOS E., *Biomechanics of vascular wall*, Medicina, Budapest, 1986.
[11] HOLZAPFEL G.A., WEIZSÄCKER H.W., *Biomechanical behavior of the arterial wall and its numerical characterization*, Comp. Biol. Med., 1998, 28, 377–392.
[12] HOLZAPFEL G.A., GASSER T.C., *A viscoelastic model for fiber-reinforced composites at finite strains: continuum basis, computational aspects and applications*, Comput. Methods Appl. Mech. Engr., 2000.
[13] HOLZAPFEL G.A., GASSER T.C., OGDEN R.W., *A new constitutive framework for arterial wall mechanics and a comparative study of material modes*, Journal of Elasticity, című folyóirat, 2000, 61, 1–48.
[14] HOLZAPFEL G.A., *Nonlinear Continuum Mechanics*, Hyperelastic Materials, 2002.
[15] HOLZAPFEL G.A., *Nonlinear Solid Mechanics. A Continuum Approach for Engineering*, John Wiley & Sons, Chichester, 2000.
[16] LANGEWOUTERS G.J., WESSELING K.H., GOEDHARD W.J.A., *The pressure dependent dynamic elasticity of 35 thoracic and 16 abdominal human aorta in vitro described by a five element model*, J. Biomech., 1985, 18(8), 613–620.
[17] NÁDASY G.L., MONOS E., MOHÁCSI E., KOVÁCS A.G.B., *The background of hysteretic properties of human umbilical arterial wall*, Acta Physiol. Hung., 1988, 71(3), 347–361.
[18] FINLAY H.M., MCCULLOUGH L., CANHAM P.B., *Three-dimensional collagen organization of human brain arteries at different transmural pressures*, J. Vasc. Res., 1995, 32, 301–312.
[19] TAKAMIZAWA H., HAYASHI K., MATSUDA T., *Isometric biaxial tension of smooth muscle in isolated cylindrical segments of rabbit arteries*, Am. J. Physiol., 1992, 236(32), H30–H34.

[20] WEIZSÄCKER H.W., PINTO J.G., *Isotropy and anisotropy of the arterial wall*, J. Biomech., 1988, 21, 477–487.

[21] WEIZSÄCKER H.W., LAMBERT H., PASCALE K., *Analysis of the pound carotid arteries*, J. Biomech., 1983, 16, 703–715.

[22] RHODIN J.A.G., *Architecture of vessel wall*, [in:] H.V. Sparks Jr., D.F. Bohr, A.D. Somlyo and S.R. Geiger (Editors), *Handbook of Physiology. The Cardiovascular System*, 2. kötet, 1–31 oldal. American Physiologial Society, Bethesda, Maryland, 1980.

[23] HUMPHREY J.D., KANG T., SAKARDA P., ANJANAPPA M., *Computation with a new electro-mechanical test system*, Ann. Biomed. Eng., 1993, 21(1), 33–4.

[24] APTER J.T., RABINOWITZ M., CUMMINGS D.H., *Correlation of viscoelastic properties of large arteries with microscopic structure*, Circ. Res., 1966, 19, 104–121.

[25] TAKAMIZAWA K., HAYASHI K., *Strain energy density function and uniform strain hypothesis for arterial mechanics*, J. Biomech., 1987, 20, 7–17.

[26] HANSEN M.E., YUCEL E.K., MEGERMAN J., L´ITALIEN G.J., ABOTT W.M., *Determination of human arterial compliace*, Cardiovasc. Intervent. Radiol., 1994, 17(1), 22–6.

[27] O'ROURKE F.M., *Vascular mechanics in the clinic*, J. of Biomechanics, 2003, 36, 623–630.

[28] THUBRIKAR M.J., *Wall stress studies of abdominal aortic aneurysm in a clinical model*, Annals of vascular surgery, 2001, 15, 355–366.

[29] OROSZ M., MOLNÁRKA G., MONOS E., *Curve fitting methods and mechanical models for identification of viscoelastic parameters of vascular wall – a comparative study*, Mon. Sci. Monit., 1997, 3(4), 599–604.

[30] OROSZ M., MOLNÁRKA G., NÁDASY G.L., RAFFAI G., KOZMANN G., MONOS E., *Validity of viscoelastic models of vessel wall*, 64th An. Meeting of the Hung. Physiol. Soc., Budapest, 1997.

[31] OROSZ M., MOLNÁRKA G., GILÁNYI M., MONOS E., *Computerized identification of viscoelastic parameters of biological tissues using viscoelastic models*, Proceedings of COMBIO´96, KFKI-1996-06/M-H, K. Tarnay and Z. Fazekas (Editors), 85–90, 1996.

[32] OROSZ M., MOLNÁRKA G., TÓTH M., NÁDASY G.L., MONOS E., *Viscoelastic behavior of vascular wall simulated by generalized Maxwell models – a comparative study*, Med. Sci. Monit., 1999, 5(3), 549–555.

[33] TÓTH M., G NÁDASY. L., NYÁRY I., KERÉNYI T., OROSZ M., MOLNÁRKA G., MONOS E., *Sterically inhomogeneous viscoelastic behavior of human saccular cerebral aneurysms*, Érbetegségek, IV/2, J. Vasc. Res., 1997, 35, 345–355.

[34] ZATZMAN M., STACY R.W., RANDALL J., EBERSTEIN A., *Time course of stress relaxation in isolated arterial segments*, Am. J. Physiol., 1954, 177, 299–307.

[35] DOBRIN P., ANIDJAR S., *Pathophysiology of arterial aneurisms*, Arch. Mal. Coeur., 1991, 84(3), 57–62.

[36] DOBRIN P., CANFIELD T., *Identification of smooth muscle series elastic component in inact carotid artery*, Am. J. Physiol., 1977, 232(2), H122–H130.

[37] DOBRIN P., *Biaxial anisotropy of dog carotid artery*, J. Biomech., 1986, 19(5), 351–358.

[38] DOBRIN P., *Vascular muscle series elastic element stiffness during isometric contraction*, Circ. Res., 1974, 34, 242–250.

[39] LERMUSIAUX P., LEROUX C., *Aortic aneurysm: construction of a life-size model by rapid prototyping*, Annals of vascular surgery, 2001, 15, 131–135.

[40] VAN LOON P., *Length force and volume–pressure relationship*, 1977.

[41] COX R.H., *Influence on muscle length on series elasticity on arterial smooth muscle*, Am. J. Physiol., 1978, 234(5), C146–C154.

[42] COX R.H., *Viscoelastic properties of canine pulmonary arteries*, Am. J. Physiol., 1984, 246, H90–H96.

[43] ASPDEN R.M., *Fibre reinforcing by collagen in cartilage and soft connective tissues*, Proc. R. Soc. Lond. B., 1994, 258, 195–200.

[44] OGDEN R.W., SCHULZE-BAUER C.A.J., *Phenomenological and structural aspects of the mechanical response of arteries*, Mechanics in Biology, J. Casey, G. Bao (Editors), New York, AMD-Vol. 242/BED-Vol. 46, ASME 2000, The American Society of Mechanical Engineers, 125–140.
[45] OGDEN R.W., *Nearly isochoric elastic deformations: Application to rubberlike solids*, J. Mech. Phys. Solids., 1987, 26, 37–57.
[46] RÖHLICH P., *Szövettan egyetemi tankönyv I. kötet*, Folpress Nyomdaipari Kft., 1999.
[47] KANG S.M., HA J.W., CHUNG N., JANG K.J., SHIN M.S., RIM S.J., *The elastic properties of the descending thoracic aorta in patient with a stroke*, Echocardiographe, 2000.
[48] DENG S.X., TOMIOKA J., DEBES J.C., FUNG Y.C., *New experiments on shear modulus of elasticity of arteries*, Am. J. Physiol., 1994, 266, H1–H10.
[49] AZUMA T., HASEGAWA M., *A rheological approach to the architecture of arterial walls*, Jpn. Physiol., 1971, 21, 27–47.
[50] IMURA T., YAMAMOTO K., KANAMORI K., MIKAMI T., YASUDA H., *Measurement of the elastic properties of the human abdominal arteries*, 1986.
[51] FUNG Y.C., FRONEK K., PATITUCCI P., *Pseudoelasticity of arteries and the choice of its mathematical expression*, Am. J. Physiol., 1979, 237, H620–H631.
[52] FUNG Y.C., *Biomechanics: Mechanical Properties of Living Tissue*, Springer-Verlag, New York, 2, Kiadás, 1993.
[53] FUNG Y.C., *Biomechanics: Motion, Flow, Stress, and Growth*, Springer-Verlag, New York, 1990, 11.3 (fejezet).
[54] FUNG Y.C., *Mechanical Properties and Active Remodeling of Blood Vessels Biomechanics*, Springer-Verlag, New York, 1990, 8 (fejezet).
[55] LUO Y., COOKE R., PATE E., *A modelof stress relaxation in cross-bridge systems: effect of a series elastic element*, Am. J. Physiol., 1993, 256, C279–C288.
[56] LUO Y., COOKE R., PATE E., *Effect of series elasticity on delay in development on tension relative to stiffness during muscle activation*, Am. J. Physiol., 1994, 267, C1598–C1606.