Viscoelasticity of Biological Materials – Measurement and Practical Impact on Biomedicine

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Running title:

Viscoelasticity of Biological Materials

Summary

Mechanical behavior of biological structures under dynamic loading generally depends on elastic as well as viscous properties of biological materials. The significance of “viscous” parameters in real situations remains to be elucidated. Behavior of rheological models consisting of a combination of inertial body and two Voigt’s bodies were described mathematically with respect to inverse problem solution, and behavior in impulse and harmonic loadings was analyzed. Samples of walls of porcine and human *aorta thoracica* in transverse direction and samples of human bone (*caput femoris, substantia compacta)* were measured. Deformation responses of human skin *in vivo* were also measured . Values of elastic moduli of porcine aorta walls were in the interval from 10^2 kPa to 10^3 kPa, values of viscous coefficients were in the interval from 10^2 Pa.s to 10^3 Pa.s. The value of shear stress moduli of human *caput femoris, substantia compacta* range from 52.7 to161.1 MPa, and viscous coefficients were in the interval from 27.3 to 98.9 kPa.s. The role of viscous coefficients is significant for relatively high loading frequencies – in our materials above 8 Hz in aorta walls and 5 Hz for bones. In bones, the viscosity reduced maximum deformation corresponding to short rectangular stress.

Keywords

Viscoelasticity • Bones • Arterial walls • Skin

Introduction

Mechanical behavior of biological structures under dynamic loading generally depends on elastic as well as viscous properties of biological materials (Saito and Werff 1975, Katsamanis and Raftopoulos 1990, Stump 1998, Bia 2005).Consequently, classical "elastic" approach should be replaced with viscoelastic models (Linde 1994, Doubal 2000A, Doubal 2000B, Doubal and Klemera 2002, Rabkin 2002, Li 2004, Mezerova 2004). On the other hand, the significance of "viscous" parameters in real situations for realistic values of elastic and viscose parameters remain to be elucidated (ZHU 2003, BIA 2005).

The theoretical aim of this paper is to derive the methodology of quantification of the significance of viscous parameters to demonstrate the applicability of the methodology by practical measurements.

For this purpose, we measured mechanical responses of selected biological structures, identified their rheological models and derived theparameters of the models.

Viscoelastic models

Viscoelastic properties of biological materials are often considered to be sufficiently described by the model in Fig. 1 (Voigt‘s model). As this model doesn’t take into account inertial forces, we will deal also with Voigt‘s model with inertial body (Fig.2), as a more general representation of reality.

Equilibrium of forces for Voigt‘s model with inertial body (Fig.2) in tensile, pressure and bending stress:

$$F_{EXT} = F_H + F_N + F_M \ , \tag{1}$$

where F_{EXT} is the external force (stress), F_H the elastic force, F_N the dumping (viscous) force, F_M is the inertial force.

For the forces it holds that

$$F_H = H \ \Delta L \ , \ F_N = N \ \frac{dL}{dt} \ , \ F_M = M \frac{d^2 L}{dt^2} \ ,$$

where H is the Hooke coefficient, N is the Newton coefficient, M is the effective mass, ΔL is the change of length or bending.

Analogously, equilibrium of momenta for Voigt‘s model with inertial body (Fig. 2) in torsion stress is defined as

$$J_{EXT} = J_H + J_N + J_M \ , \tag{2}$$

where J_{EXT} is the external torque, J_H the elastic torque, J_N the viscous torque, J_M the inertia torque,

For the momenta it holds that

$$J_H = H^* \Delta\varphi \ , J_N = N^* \frac{d\varphi}{dt} \ , J_M = J^* \frac{d^2\varphi}{dt^2}$$

where H^*, N^*, J^* are coefficients of Voigt’s model for torsion loading, φ is the angle of torsion, t is time.

The equation of motion for Voigt‘s model with inertial body in tensile, bending stress:

$$0 = M \frac{d^2 L}{dt^2} + N \frac{dL}{dt} + H \Delta L \quad . \tag{3}$$

The equation of motion for Voigt‘s model in torsion stress is then

$$0 = J^* \frac{d^2 \varphi}{dt^2} + N^* \frac{dL}{dt} + H^* \Delta \varphi \quad . \tag{4}$$

Supposing unit harmonic deformations ($\Delta L = \sin \omega t$) we have for Voigt‘s model with inertial body:

$$F_H = H \sin \omega t \quad , \tag{5}$$

$$F_N = \omega N \cos \omega t \quad , \tag{6}$$

$$F_M = -\omega^2 M \sin \omega t \, , \tag{7}$$

Similar relationships hold for torsion loading.

Dynamic responses of Voigt‘s model

Impulse characteristic

In this case, the impulse characteristic is a deformation response to an infinitely short impulse of force.

$$\Delta L(t) = \frac{I}{N} \cdot \exp\left(\frac{H}{N} \cdot t \right) \tag{8}$$

where *I* is the impulse of the force, *t* is time.

Transient characteristic

In this case, the transient characteristic is a deformation response to a “jump” ΔF of the force between two constant levels:

$$\Delta L(t) = \frac{\Delta F}{H} \cdot \left[1 - \exp\left(-\frac{H}{N} \cdot t \right) \right] \tag{9}$$

Frequency characteristic

In this case, the frequency characteristic is the relationship between Fourier transform of the force and deformation.

$$\Delta L(i\omega) = \frac{F(i\omega)}{i\omega N + H} \tag{10}$$

where *i* is the complex unit,

For amplitude frequency characteristic it holds that

$$|\Delta L(\omega)| = \frac{|F(\omega)|}{\sqrt{(\omega N)^2 + H^2}} \tag{11}$$

The phasor of the deformation with respect to the phasor of the force of angle φ is

$$\mathrm{arctg}\varphi = -\omega N/H \tag{12}$$

The above formulae hold naturally for simple Voigt‘s model if the mass approaches to zero.

From our previous experiments on vein walls, bones and skin it follows that mechanical behavior of many biological structures corresponds to the behavior of the model according to Fig 3. Deformations in static loading corresponding to elements 2 were in the interval from 10 to 30 % of total deformation. This ratio decreases with frequency in dynamic loading. Results in this paper deal with the data for rapid element 1.

Methods

Measuring appliance

The applied version of the apparatus (Fig. 4) permits measurement of samples of material under tensile pressure and torsion loadings (Doubal 2000A, 2000B).

The samples are fixed and connected with a gauge. Changes in deformation are detected by means of an inductive transducer. The deformation force is produced by inserting weights on a pan on the top of the gauge (electronically or manually). The signal from transducer is electronically processed, amplified, transformed to digital form by A/D transducer and processed in a computer. The time constant of electronics is 1.25 ms. Minimum detectable change of deformation is about 2 μm

Material

Aorta walls

Samples of walls of porcine *aorta thoracica* obtained from 18 piglets (6 months, male, *Sus scrofa* f. *domestica*) in transverse (circumferential) direction were measured in two successive series. The aim of the first series (10 aortas) was to verify the methodology. In the second series, 8 aortas were examined using a standard procedure. In the following text, the results of the second series are described. Further, samples of walls of human aorta (*pars thoracica*, 5 men, 8 women, age from 37 to 78 years. Measurements were performed under tensile stress.

Bones

Samples of bone (*caput femoris, substantia compacta)* taken from 10 human subjects (mean age 75 ± 1.5 years, 4 men, 6 women) were measured. Bone specimens were withdrawn through hip joint surgery. They were kept in physiological solution and measurements were performed within two weeks of the operation.

The availability of research material in our experiments was optimal, we utilized the remnants of bones after surgery on hip joints. Replacement of the hip joint is not uncommon after low impact fractures of the femoral neck. The head of the femur was extracted during the replacement of the hip joint and our samples were cut from this remaining part of the femur. Measurements were performed under bending and torsion stress.

Human skin *in vivo*

Two groups consisting of 15 men and 27 women (age range from 20 to 58 years) were used in the experiment. The deformation response was measured on the left palm (above the middle of abductor *pollicis brevis*) under pressure stress.

Results

Aorta walls

The values of elastic moduli E lay in theinterval from 10^2 kPa to 10^3 kPa, values of viscous coefficients η were in the interval from 10^2 Pa.s to 10^3 Pa.s.

The elastic moduli were calculated from Hooke's coefficients ($H_1 = a.E$), where a depends on the shape of the sample. Similarly, the viscous coefficients η were calculated from Newton's coefficients ($N_1 = a\ \eta$.).

Bones

Young's moduli E of cortical bones were calculated from the measurement of bending stress. Values of these moduli ranged from 139.4 to 453.2 MPa.

Shear stress moduli G and viscous coefficients were calculated from measurements during twisting stress. The computed values of G ranged from 52.7 to161.1 MPa, values of viscous coefficients η were in the interval from 27.3 to 98.9 kPa.s.

Young's moduli were calculated from Hooke's coefficients ($H_1 = b.E$), where b depends on the shape of the sample. Similarly, the viscous coefficients η were calculated from Newton's coefficients ($N_1 = b\ \eta$.).

Human skin in vivo

Results of experiments were published previously (Doubal and Klemera 2002). Typical values of viscoelastic parameters are in the following table (for symbols see Fig. 3:

Woman, 58 years	N_1 (kg.s^{-1})	H_1 (kN.m^{-1})	N_2 (kg.s^{-1})	H_2 (kN.m^{-1})
	3.33	190.1	651.24	223.4

Discussion

From our previous experiments on vein walls, bones and skin itfollows that mechanical behavior of many biological structures corresponds to the behavior of the model according to Fig 3.

In all cases, the time constant (see Eq. 9) in "rapid" element 1 was of the order of 1 ms, and about two orders of magnitude lower in comparison with the time constant of "slow" element 2.

Deformations in static loading corresponding to elements 2 were in the interval from 10 % to 30 % of total deformation. This ratio will decrease with the frequency in dynamic loading. Results in this paper deal with data for element 1.

In the case of aorta walls, the satisfactory description of mechanical properties is crucial for analysis of mechanical matching of arteries and surrogates (stents, replacements etc.) and for understanding the mechanical situation at the boundary between healthy and pathological parts of the walls. In addition, viscous coefficients potentially influence pulse wave propagation.

Taking into account our results for aorta walls and applying formulae 5, 6 and 7 dependences of inertial, viscous and elastic forces on frequency may be calculated. The significance of the viscous parameter increases with frequency and the significance of inertial parameter increases with the square of frequency. According to our results the influence of viscous parameters is significant for frequencies above 8 Hz. For lower frequencies the viscous force is less than 10 % of the elastic force. This limiting frequency increases with *E* and decreases with η. The mass of the sample must be taken into account for frequencies above 5 Hz for materials with $E < 10^5$ Pa and samples longer than 2 cm. This limiting frequency increases with E and decreases with the length of the sample. The lower the limiting frequency, the more pronounced the effect of viscous properties of the material.

In bones, viscosity reduced maximum deformation corresponding to short rectangular stress. The reduction is significant for rectangular impulses of force shorter than 10 ms., i.e. the viscosity will significantly dump the impact of impulses shorter than 10 ms.

Consequently, the viscous properties will affect the "shape" of the response to stresses containing harmonic components of frequencies above limiting frequency. In other words, the responses to rapid and "sharp" courses of stress will be more influenced by the viscosity of materials.

References

BIA D, ZOCALO Y, PESSANA F, ARMENTANO R, PEREZ-CAMPOS H, SALDIAS M, ALVAREZ I: Femoral arteries energy dissipation and filtering function remain unchanged after cryopreservation procedure. *Transpl Int* **18**:1346–1355, 2005.

BIA D: In vitro model to study arterial wall dynamics. *Latin Amer Appl Res* **35**: 217–224, 2005.

DOUBAL S: Visco-elastic parameters of biological material - measurement and analysis of creep curves. *Folia Pharmaceut Univ Carol* **25**: 35–43, 2000.

DOUBAL S: Apparatus for measurement of mechanical properties of solids and visco-elastic bodies. *Folia Pharmaceut Univ Carol* **25:** 45–48, 2000.

DOUBAL S, KLEMERA P: Visco-elastic response of human skin and aging. *J Amer Aging Assoc* **3**:115–117, 2002.

KATSAMANIS F, RAFTOPOULOS D: Determination of mechanical properties of human femoral cortical bone by the Hopkinson bar stress technique. *J Biomech* **23**: 1173–1184, 1990.

Li LP, HERZOG W: The role of viscoelasticity of collagen fibers in articular cartilage: theory and numerical formulation. *Biorheology* **41**: 181–194, 2004.

LINDE F: Elastic and viscoelastic properties of trabecular bone by a compression testing approach. *Dan Med Bull* **41:**119–138, 1994.

MEZEROVA L: Experimental verification of viscoelastic properties of spongy bone tissue. *Eur Cells Materials* **7**: 45–46, 2004.

RABKIN E: Cardiovascular tissue engineering. *Cardiovasc Pathol* **11:** 305–317, 2002.

SAITO GE.,WERFF TV: The importance of viscoelasticity in arterial blood flow models. *J Biomech* **8:** 237–245, 1975.

STUMP DM: Circumferentially flexible vascular grafts1. *J Biomech* **31**: 705–711, 1998.

ZHU H., WARNER JJ, GEHRIG T.R, FRIEDMAN M. H: Comparison of coronary artery dynamics pre- and post-stenting. *J Biomech* **36:** 689–97, 2003.

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