Visco-elastic mechanical behaviour of human abdominal fascia

L. Hadjikov, M. Kirilova, S. Stoytchev, D. Pashkouleva

Bulgarian Academy of Sciences, Institute of Mechanics,
Acad. G.Bonchev Str. Bl.4, 1113 Sofia, Bulgaria

Abstract: Time dependant mechanical behaviour of human abdominal fascia was studied. Mathematical modeling of visco-elastic properties was accomplished applying the non-linear and linear theories proposed by Maxwell-Gurevich- Rabinovich (MGR). The results show that the non-linear theory describes better the experimental data. A method for determining the material constants was developed and their values were compared to data found in the existing literature.

Keywords: human fascia, stress relaxation, mathematical model

1. Introduction

Fascia is a dense connective tissue membrane, which surrounds all organs, muscles, bones and nerve fibers [1]. As part of the abdominal wall the abdominal fascia forms a sheath around the abdomen. The abdominal fascia is formed of two layers of undulating collagen and elastic fibers which are separated by adipose tissue. In each layer the bundles of collagen fibers are parallel to each other but the direction of fibers varies according to the layer [2].

There is an increasing interest in the role that mechanical properties of abdominal fascia play in hernia development. The mechanical properties of fascia transversalis have been studied using tensile tests, punching tests, stress relaxation experiments and elasticity measurements. The elasticity measurements were performed by Junge et al using a suction device [3]. Punching tests were applied by T. Wolloscheck et al. [4] and tensile tests were performed by Wolloscheck et al. [4], Kureshi et al. [5], Minns et al. [6], Kirilova et al. [7]. Stress relaxation experiments of human abdominal fascia were done by Minns et al. [6] and Kirilova et al. [8].

Like most soft biological tissues the abdominal fascia has to be considered as a visco-elastic solid because its stress state is not uniquely determined by strain, but depends also on the history of deformation. So the full description of the mechanical behavior of the fascia requires applying a theory, which can account for the nonlinear, time dependant stress-strain characteristics of the fascia.
Very few investigators however dealt with the theoretical description of the mechanical behavior of fascia. Zeng et al., used power function to fit the stress-strain curves of the nasal fascia and compare the coefficients found [9]. The quasi-linear visco-elastic theory of Fung was applied to relaxation process of abdominal fascia [8]. It was shown that the relaxation parameters were of the same order as parameters of human small intestine.

Another theory, which can be used for mathematical modeling of visco-elastic properties of soft biological tissues, is the theory of Maxwell-Gurevich- Rabinovich (MGR) [10]. So far it is applied for describing the relaxation behavior of the human small intestine [11].

The aim of this work is to develop a method for finding the model parameters based on linear and nonlinear variants of the MGR theory. In the next Section 2 we describe the experimental methods and the material samples used. Later in Section 3, we explain the MGR theory and a method for the determination of material parameters from experimental recordings. Finally, we provide examples of visco-elastic parameter identification in Section 4. .

2. Experimental methods

The fascia specimens were extracted from posterior wall of inguinal canal and umbilical region. They were taken during autopsy of cadavers within 12 hours of death, freed of fat and immersed in physiological solution. Specimens with dimensions between (10x40) - (10x65) mm were tested. The samples were oriented parallel to the collagen fibers and perpendicular to it. Tests were performed immediately after cutting and realized at ambient temperature (21 ± 2) °C [8].

Uniaxial relaxation tests of 6 flat specimens taken from umbilical fascia and 8 samples taken from the transversalis fascia in longitudinal and transversal direction were performed. An investigation was done using computer equipped testing machine FU1000/E. The specimens were subjected to a few cycles of loading and unloading until repeatable mechanical performance was obtained. Thus we reached a steady state called a “preconditioned state” [12].

During relaxation tests the extension of the specimen was kept constant while the load was recorded with time. The experiments were performed at 1.26 mm/sec rate of elongation. The initial length of the specimens was measured after preconditioning of the specimens. Strips were loaded from 2% to 15% of their initial length.

The following mechanical characteristics were calculated from the experimentally found load-elongation curves: Lagrangian stress σ - tensile force per unit undeformed cross-sectional area and stretch ratio $\lambda = L / L_0$, where L_0 is the undeformed length and L is the deformed length of the specimen. Strain ε was calculated as $\varepsilon = \lambda - 1$. The experimental data were presented as stress-time relationships.

A normalized relaxation function G [%], which gives the reduction of the stress during relaxation process, was defined as:

$$G = \frac{(\sigma_{init.} - \sigma_{\infty})}{\sigma_{init.}} 100 \ [\%] \tag{1}$$

where $\sigma_{init} = \sigma(t = 0)$ and $\sigma_{\infty} = \sigma(t = 600\,\text{sec})$. Function G [%] was used as a parameter for the estimation of the total stress reduction.

3. A method for the determination of visco-elastic parameters based on MGR theory

A theory was developed to describe the deformation and relaxation process in polymers, but it was successfully applied for soft biological tissues, too [11]. In theory, the total strain ε is a sum of an elastic strain e and viscous strain $\varepsilon*$.

$$\varepsilon = e + \varepsilon^{*} \tag{2}$$

It is assumed that elastic strain e results from intermolecular distances, while viscous strain ε^{*} is conditioned by conformational changes in long polymer molecules. Based on Gibbs thermodynamics, the relaxation time and stress are derived from the activation energy applying external forces. Stress relaxation is governed by the following differential equation

$$\frac{d\sigma}{dt} = -\frac{E}{\eta_0^{*}} f^{*} \exp\left\{ \frac{\frac{1}{3}\gamma^{*}\sigma + \left| f^{*} \right|}{m^{*}} \right\} \tag{3}$$

where η_0^* is the initial coefficient of viscosity, m^* is the parameter, which determines strain rate influence on stress and γ^* is volumetric strain (compressibility) coefficient. Function f^* is defined as:

$$f = \sigma - E_\infty \varepsilon^* = (1 + \frac{E_\infty}{E})\sigma - E_\infty \varepsilon_0 \qquad (4)$$

Here ε_0 is the initial step-wise deformation of the sample, σ_0 is the initial stress at $t = 0$, $\sigma_{\min}$ is the values of stress when relaxation is fully completed, $\varepsilon_{\max}^*$ is the viscous strain maximal values, E is the initial elastic modulus defined as:

$$E = \frac{\sigma_0}{\varepsilon_0} \qquad (5)$$

and E_∞ is the visco-elastic modulus when the relaxation process finishes

$$E_\infty = \frac{\sigma_{\min}}{\varepsilon_{\max}^*} \qquad (6)$$

The following assumptions are made: at the beginning of relaxation initial strain $\varepsilon = const = \varepsilon_0$; and there is no volumetric strain so its coefficient is $\gamma^* = 0$. The main assumption of this model is that the instantaneous nonlinear elastic and time-dependant viscous responses are independent.

The solution of the differential equation (3) is

$$t = \frac{\eta_0^*}{E + E_\infty}\left[- E_i(-\xi^*) + E_i(-\xi_0^*)\right] \qquad (7)$$

where E_i is the exponential integral function and the values of the argument ξ^* are calculated as

$$\xi^* = \frac{f^*}{m^*} = \left|(1 + \frac{E_\infty}{E})\sigma - E_\infty \varepsilon_0\right| / m^* \qquad (8)$$

The linearized equation of Maxwell-Gurevich-Rabinovich [10] assumes that the parameter $m^* \to \infty$. Thus, stress variation in time is:

$$\frac{d\sigma}{dt} = \frac{E}{\eta_0^*} f^* \qquad (9)$$

Taking into account that stress and viscous strains are connected by the following equation

$$\sigma = \sigma_0 - E\varepsilon^*, \qquad (10)$$

and substituting (10) in equation (9), we find

$$\frac{d\varepsilon^*}{dt} = \frac{f^*}{\eta_0^*} \quad , \tag{11}$$

where function f^* now takes the form:

$$f^* = E\varepsilon_0 - \varepsilon^*(E + E_\infty) = \sigma_0 - \varepsilon^*(E + E_\infty) \tag{12}$$

After substituting the expression for f^* in equation (11) we obtain

$$\frac{d\varepsilon^*}{dt} + \frac{E + E_\infty}{\eta_0^*}\varepsilon^* = \frac{\sigma_0}{\eta_0^*} \tag{13}$$

Whit initial values $t = 0$ and $\varepsilon^* = 0$, the solution of equation (13) reads:

$$\varepsilon^* = \frac{\sigma_0}{E + E_\infty}(1 - \exp(-\frac{E + E\infty}{\eta_0^*}t) \tag{14}$$

So far we determined the expression of the time dependence of the visco-elastic strain ε^*. Having in mind equations (2) and (10) and substituting Eq. (14) into Eq. (10), after some transformations we finally obtain

$$\sigma(t) = \sigma_0(1 - \frac{1 - \exp(-(\frac{E + E_\infty}{\eta_0^*}t)}{1 + \frac{E_\infty}{E}}) \tag{15}$$

Summarizing the obtained formulas, we note that:

- Equations (7) and (8) represent stress relaxation behavior according to the non-linear theory of MGB;
- Eq. (15) represents stress relaxation behavior according to the linear theory of MGB;
- The linear model includes five parameters, namely:

a) experimentally recorded initial stress σ_0 and initial strain ε_0;

b) local elasticity modulus at the beginning of relaxation process E and visco-elastic modulus when the relaxation process is completed E_∞. They can be calculated through Eqs. (5) and (6);

c) Initial viscosity coefficient η_0^*, which is a constant not depending on the time.

The non-linear theory includes the parameters mentioned above and also additional dependence of the relaxation process on the strain rate, the so-called strain rate modulus m^* (See Eq.3). Modulus m^* influences tissue viscosity through the following relationship:

$$\eta^*(t) = \eta_0^* \exp(\frac{E_\infty \varepsilon^* - \sigma(t)}{m^*}) \tag{16}$$

Thus we come to the conclusion that the non-linear MGR theory supposes that tissue viscosity changes during relaxation and changes are exponentially proportional to the differences between actual and equilibrium (complete relaxation) stresses.

Therefore, only the two material parameters η_0^* and m^* are to be found. They can be calculated using suitable numerical procedure so that the following objective function F could reach its minimum:

$$F = [(\sum_{i=1}^{N} \{\sigma_i^{teor}(\eta_0^*, m^*) - \sigma_i^{\exp}\}^2)/N]^{1/2} \tag{17}$$

σ_i^{theor} (η_0^*, m^*) is the predicted stress computed from the model, σ_i^{exper} is the experimental stress relaxation data.

4. Results and discussion

Typical experimental stress-time curves are presented in Fig.1 by filled symbols. The three samples were investigated at initial deformation approximately 6%. They were approximated analytically using the non-linear MGR theory. The model parameters were found by the method described in the previous Section and they are listed in Table 1. The results can be summerized as follows: a) stress relaxation exhibits a decaying exponential form like other soft biological tissues; b) the relaxation process could be conventionally divided in two parts – fast relaxation, during which the stress rapidly falls down, and slow relaxation, during which the stress decreases slightly. Fast relaxation usually lasts up to 60 sec; c) fast relaxation strongly depends on the initial stress level σ_0. Nevertheless, the stress relaxation parameter G slightly varies between 30% and 42%; d) the non-linear MGR theory describes very well the experimental curves. Indeed, the objective function F, defined by Eq.17, has a value below 1.3%. As a result, the theoretical and experimental dots in Figure 1 are almost indistinguishable; e) the initial viscosity parameter η_0^* varies in a wide range where its values change tenfold. Our preliminary analysis shows that the values of η_0^* well correlate with the values of the initial stress σ_0.

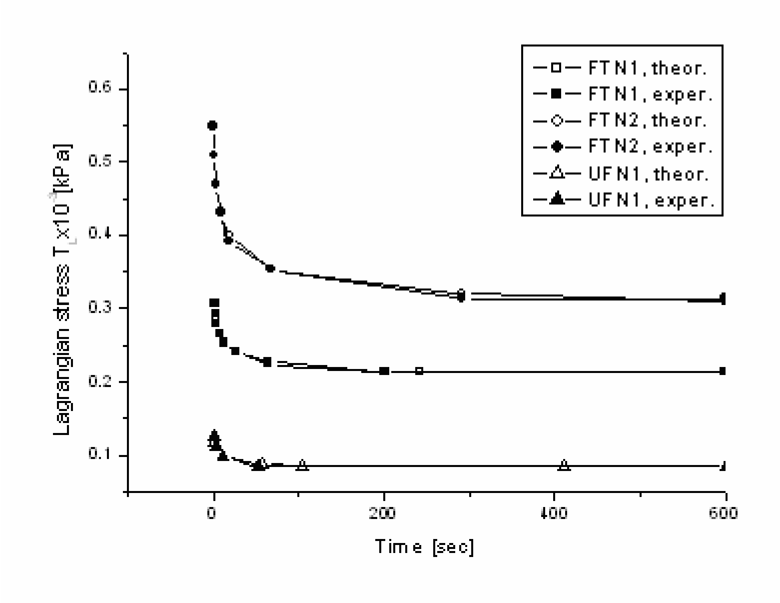

Fig.1. Lagrangian stress as a function of time.

Table 1.
Values of the parameters of nonlinear MGR theory.

Type	ε_0	ε^*_{max}	E [kPa]	E_∞ [kPa]	η_0^* [kPa.sec]	m^* [kPa]	G [%]	F
FTN1	0.06	0.018	5110	11 677	956 920	165.5	30.45	0.01311
FTN2	0.0588	0.025	9334	12 447	3 546 000	143.0	42.85	0.01148
UFN1	0.063	0.021	2000	4000	285 816	45.0	33.33	0.01314

The application of the linear MGR theory for a sample of human umbilical fascia is shown in Figure 2. It is evident that the coincidence with the experimental data is very poor especially in the first part of the experimental curve. For this reason we decided to apply the nonlinear theory of MGR in the future investigations of the visco-elastic mechanical behaviour of human abdominal fascia.

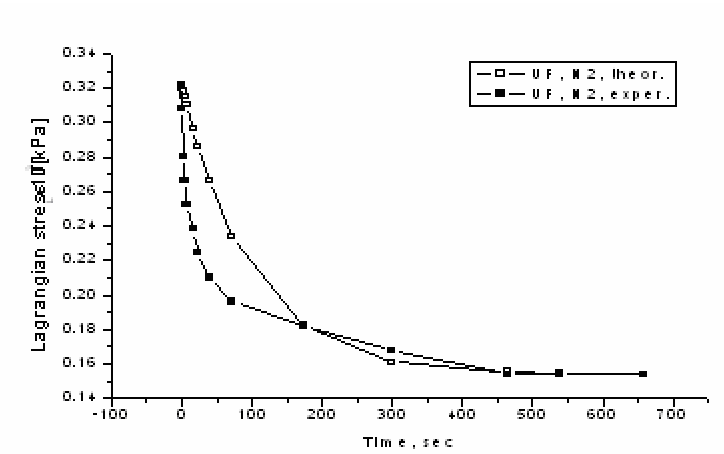

Fig. 2 Experimental (filled squares) stress-relaxation curves and analytical approximation based on linear MGR theory.

It is interesting to compare predictions of the non-linear theory of MGR concerning other soft biological tissue. To our knowledge only Hadjikov et al. [11] have applied it to stress relaxation experiments in human intestines. The numerical values of the coefficients are represented in Table 2.

The differences in elastic modulae E for the intestines and fascia can be explained with differences in the initial step strain because they are the local elastic modulae which depend on the strain level. Values of E_∞ for the compared tissues are also quite different because this parameter depends inversely on the values of the initial strain ε_0.

The discrepancies in the values of m^* could be explained turning back to the physical meaning of the parameter. According to the MGR theory it has dimension of stress and depends inversely on the strain rate. Thus the smaller values of m^*, as for the human intestines, listed in Table 2, correspond to a pronounced dependence on the strain rate while the higher values of m^* (See Table 1) indicate less strain rate sensitivity. On the other hand, it is well known that the elastic or pseudo-elastic materials display negligible strain rate dependences and viscous effects [12]. In this sense, since the abdominal fascia consists mainly of collagen fibers while the gastro-intestinal tract is composed of muscle cells which exhibit considerable viscous behaviour, the values of the coefficient m^* reflect the differences in tissue composition and individual fiber visco-elastic properties of the investigated materials. According to the MGR theory the higher values of m^* mean also that relaxation duration will be shorter. This fact is evident from the experimental results – the end of the relaxation process for the fascia is between 150-300 sec (See Fig.1) , while the relaxation process for the human intestines lasts about 600 sec [11].

Comparing the values of apparent viscosity η_0^* of both materials it is seen that the range of viscosity η_0^* for human intestines is broader than the range of parameter η_0^* for fascia. According to the results it can be supposed that the viscous properties of fascia are closer to those of intestines in circumferential direction.

The most popular visco-elastic theory for biological tissues is that proposed by Fung [12]. Fung noted that since the stress-strain relationship for soft biological tissue is insensitive to strain rate, the relaxation function must be broad. Therefore the theory includes two time constants which are connected with the beginning and the end of the relaxation process. The third constant of the theory determines the degree to which viscous effects are presented.

The visco-elastic theory of MGR comprises larger amount of effects: material compressibility γ^*, initial, or more correctly apparent viscosity η_0^* of the material, time course of the apparent viscosity $\eta^*(t)$, influence of strain rate on the apparent viscosity m^* . The goal of our future investigations will be the study of the detailed time course of the apparent viscosity and comparison of the results about the stress relaxation process based on Fung's and MGR theory.

Table 2.
Values of the relaxation parameters for human small intestines as reported in [11].

Tissue/direction	ε_0	ε^*_{max}	E [kPa]	E_∞ [kPa]	η_0^* [kPa.sec]	m^* [kPa]
Human small intestine						
Circumferential	0.372	0.342	2430	442.48	3 095 700	50.203
Longitudinal	0.217	0.165	7870	2246.8	90 670.2	36.166

Acknowledgments

The authors thank Prof. V. Kavardzhikov for his help during the experimental part of this study. The study was supported by the National Science Fund of the Ministry of Education and Science; project TH-1502/2005**.**

References

[1] Snell, R.S., 1984. Clinical and Functional Histology for medical students, Little, Brown and Company**.**
[2] Stecco, C., Porzionato, A., Macchi, V., Tiengo, C., Delmas, V., De Caro, R., 2006. A histological study of the deep fascia of the upper limb. It.J.Anat.Embryol 111, n2, 125-132.
[3] Junge, K., Klinge, U, Prescher, A., Giboni, P., Niewiera, M., Schupelick, V., 2001. Elasticity of the anterior abdominal wall and impact for reparation of the incisional hernias using mesh implants. Hernia 5, 113-118.
[4] Wolloscheck, T., Gaumann, Terzic, A., Heintz, A., Jungiger, T., Konerding, M., 2004. Inguinal hernia: Measurement of the biomechanics of the lower abdominal wall and the inguinal canal. Hernia 8, 233-241.
[5] Kureshi, A., Vaiude, P., Nazhat, S., Barker, S., Brown, R., 2006. Mechanical properties of Transversalis Fascia and Hernia formation. European Cells and Materials 11, suppl.3, 33.
[6] Minns, R.J., Tinckler, L.F., 1976. Structural and mechanical aspects of prosthetic Herniorrhaphy. J.Biomechanics 9**,** 435-438.
[7] Kirilova, M., Stoytchev, S., Pashkouleva, D., Kavardzhikov, V., Tsenova, V., 2006. Mechanical properties of human fascia transversalis, Comptes rendus de l'Academie bulgare des Sciences 59, No. 7, 731–736.
[8] Kirilova, M., 2006. Visco-elastic properties of human abdominal fascia: Experiments and mathematical model, In Proceedings of 2nd Euro Summer School on Biorheology&Symposium on Micro Mechanobiology of Cells, Tissues and Systems. Varna, Bulgaria, 62-66.
[9] Zeng, Y., Sun, X., Yang, J., Wu, W., Xu, X., Yan, Y., 2003. Mechanical properties of nasal fascia and periosteum. Clinical Biomechanics 18, 760-764.
[10] Rabinovich, A.A., 1970. Introduction to mechanics of reinforced polymers, Moscow.
[11] Hadjikov, L., Stoytchev, S., Alexandrova, D., Egorov , V.I., Baranov, A.O., Shtastlivtsev, I.V., Turusov, R.A., 2005. Non-linear mechanical model of small human intestine at stress relaxation experiments. In Proceedings of 10 National Congress of Theoretical and Applied Mechanics. Varna, Bulgaria, 124-127.
[12] Fung, Y., 1972. Stress-strain history relations of Soft tissues in simple elongations. In: Biomechanics. Its Foundations and Objectives, Prentice-Hall, Englewood Cliffs, New Jersey, 181-208.