EXPERIMENTAL STUDY OF LOCAL SHEAR STRAIN REMOTELY INDUCED BY PULSED ULTRASOUND IN TISSUE-MIMICKING PHANTOMS

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Abstract

The present study revealed the significant dependence of displacement magnitude and strain relaxation on phantom elasticity and viscosity. It has been shown that simultaneous analysis of temporal behavior and magnitude of shear strain induced by the radiation force of focused ultrasonic beam gives the necessary data for quantitative estimation of tissue viscosity and shear modulus due to the known functional dependencies of displacements on these parameters. As a result the simplest calibration procedure of acoustic radiation force based methods is performed for used excitation beam.

These findings were tested, in particular, using data obtained for specially prepared phantoms containing calf liver and muscle tissue *in vitro*. The observed complex character of shear strain relaxation in some tissue phantoms and tissues *in vitro* reduces the preciseness of viscoelastic properties estimation.

Introduction

Shear Wave Elasticity Imaging (SWEI) [1-3] and Acoustic Remote Palpation (ARP) [4] methods, that are based on the remote generation of shear strain in tissue by radiation force of focused ultrasonic beam, suggest considerable clinical potential for diagnostic of soft tissue elasticity as well as some other methods. However not all of them are capable to provide the actually local measuring of tissue parameters that results in application of rather complicated mathematical procedures for spatial reconstruction of tissue elasticity. Besides that not all of these methods are potentially capable to submit the identical volume of diagnostically important information to estimate a physiological state of examined tissues and organs.

The previously performed experiments [5] show a great importance of viscosity in the process of excitation and relaxation of shear strains induced by focused pulse of radiation force of the ultrasonic wave beam. Therefore the main aim of the present work was the experimental testing of the algorithm for simultaneous quantitative assessment both of the tissue elasticity and viscosity within the limits of ARP and SWEI methods, which provide such possibility.

Methods

The present experimental research of viscoelastic properties of tissue mimicking phantoms and soft tissues *in vitro* was performed using ultrasound Doppler technique. The experimental schematic and the peculiarities of the measurement procedure were the same as in [3,5]. The set of used gelatin phantoms and the phantoms containing soft tissues is given in Table 1. The quantity of gelatin in phantom 1 and phantoms 2-5 were 40 g/l and 50 g/l, respectively. In order to change the viscosity and elasticity characteristics of some phantoms with the same ratio C_W/C_G of the volume content of water and glycerol the temperature was varied and the proper values of

Table 1 : Tissue phantoms and soft tissues *in vitro*

#	C_W/C_G	T, ºC	c_t , m/s	c_l , m/s	α , cm^{-1}
1	1/0	24	0.32	1480	0.0228
2	1/0	23.5	0.37	1480	0.030
3	1/0	21	0.87	1480	0.0347
4	4/1	23	0.75	1510	0.0344
5	2/1	22	0.98	1660	0.0366
6	liver	22	0.76	1540	0.050
7	beef	22	4.0	1540	0.040

shear wave speed c_t, the speed of ultrasound c_l, and the ultrasound attenuation coefficient α at used carrier frequency $1MHz$ are given in Table 1. The value of c_t was evaluated by SWEI method [1,3], as is shown in Fig. 1, using the measured delay time caused by passing of induced shear waves to the observation points with the different radial coordinates in the focal plane. On particular, the upper curves show the dependence of delay time on radial coordinate in

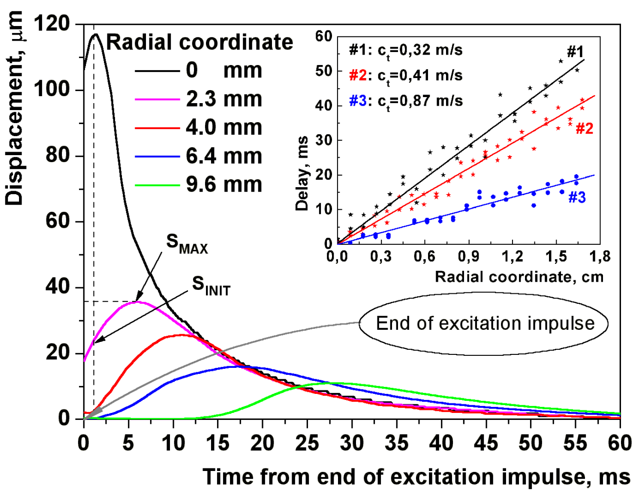

Figure 1 : Evaluation of c_t by SWEI method

phantoms 1, 2 and 3. Note that the displacement in the focal point reaches its maximum value at the time moment approximately equal to $t_0 = \alpha\gamma^{-1}/c_t - \tau$, where $\alpha\gamma^{-1}$ is the effective radius of the beamwidth in the focal plane, $\tau = 2.18\,ms$ is the duration of the excitation pulse. At that the delay time $t_0 + \tau$ is caused by passing shear waves across the beamwidth [6].

Results and discussion

Lateral distribution of shear displacement

The typical displacement fields induced by acoustic radiation force are shown in Fig.2 together with the intensity distribution in the focal plane of focused excitation beam (dotted line). Curves 1 and 2, respectively, show the distributions of maximum displacement S_{MAX} in the induced packet of shear waves for the points of observation with different radial coordinate r and "initial" displacements S_{INIT} of phantom points, corresponding to the moment of time t_0. For the most of used phantoms this time point approximately contemporizes with the end of the excitation pulse (see Fig.1).

It is clear that the maximum displacements in the propagating shear waves for all observation points are greater than initial displacements. It is not difficult to see that both distribution of displacement 1 and 2 are wider than the distribution 3 of acoustic intensity and, consequently, of the distribution of the radiation pressure force. The same result for the distribution of maximum displacements was obtained by Andreev *et al.* [2]. The largest difference of curves 1 and 2 from curve 3 is observed for the points with radial coordinate $r \geq \alpha\gamma^{-1}/\sqrt{2}$. The latter testifies particularly to the fact, that the forces of viscous

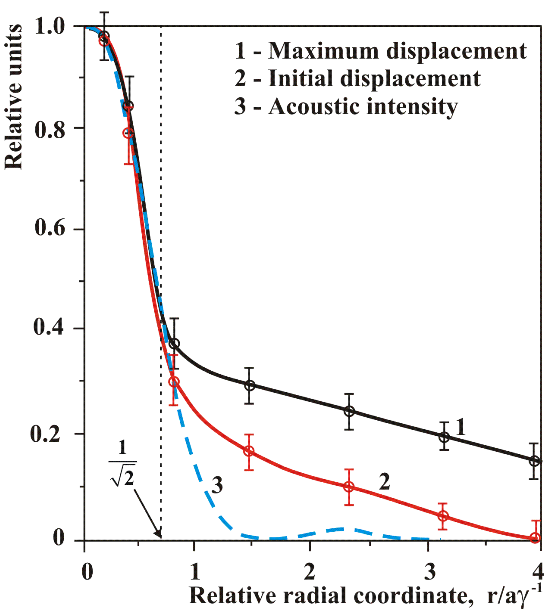

Figure 2 : Displacement fields in phantom 1

friction for points aside the focal area are mainly accelerating, as it is shown in [6], and results in evident displacements outside the region of maximum action of radiation pressure force.

Maximum displacement and strain relaxation

Figure 3 shows the difference of maximum displacement value and the relaxation curves in phantoms 1-5, which are different in shear module and viscosity. At first sight these curves differ

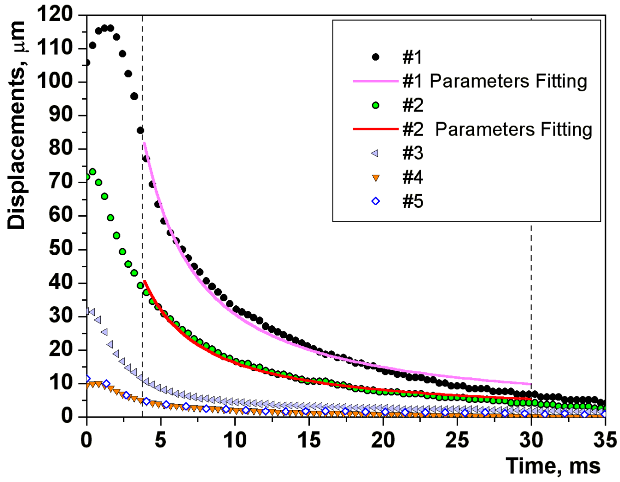

Figure 3 : Displacement through the time in the focal point in phantoms 1-5

rather insignificantly and mainly only in scale. At the same time the difference of maximum displacements in phantoms 1 and 2 cannot be explained only by the difference in shear elastic modules of the material of these phantoms. As it is seen from Fig. 1 the shear wave velocities and, correspondingly, the shear modules in these phantoms differ rather insignificantly. Through analysis of curves using functional dependencies for induced displacements makes it possible to determine a great contribution of viscosity in their value.

According to [6] the dependence of displacement in the focal point on time in the process of shear strain relaxation was approximated by the formula

$$S(t) = P_0 \frac{P_1(c_t)}{t+\tau/2}\left(1+\frac{P_2(c_t,\eta)}{t+\tau/2}\right)+P_3, \qquad (1)$$

where dimensional parameters P_1 and P_2 are determined by both the parameters of excitation wave beams (the ultrasound intensity, the excitation pulse duration, the beamwidth in the focal region) and the physical parameters of the soft tissue under study. In the theoretical analysis the dimensionless parameter P_0 depends only on the model assumptions made [1,6], for example as to the excitation pulse envelope and the profile of the beam, and therefore must be determined experimentally for every particular ultrasonic focused wave beam. Finally, the dimensional parameter P_3 has been introduced for taking into account the possible noise term.

Table 2 shows the results of estimating parameters $\hat{P}_0, \hat{P}_2$ and $\hat{P}_3$ for phantoms 1-5 with the usage of marked regions of relaxation curves within the time interval approximately from 3.9 to 30 ms, as it is shown in Fig.3. The numerical analysis of the form of the obtained relaxation curves and the parameters fitting in accordance with Eq. (1) was made using the standard packet of data processing programs Origin 6.0. Fig. 3 gives the result of approximation of relaxation curves by means of Eq. (1). At that for each phantom the calculated value of parameter P_1 and measured value of c_t were used (Fig.1).

Table 2 : Parameters estimated under calibration

#	$\hat{P}_0$	P_1	$\hat{P}_2$	$\hat{P}_3$	$\hat{\eta}$
1	15.3	18.7	2.11	0	0.057
2	10.3	15.8	1.24	0	0.070
3	9.46	3.11	2.89	0.91	0.161
4	6.23	4.30	0.357	0	0.177
5	7.44	1.81	1.08	0.102	0.263

It is easy to see that the average value of the parameter $\overline{P}_0 = 9.74$ is one of the adjusting parameters of the theory [6]. So the described above procedure of the estimation of parameter $\overline{P}_0$ represents the easiest variant of calibration of ARP and SWEI methods of soft tissue viscosity and elasticity evaluation. Knowing parameter $\overline{P}_0$ it is possible to determine values $\hat{P}_1$ and $\hat{P}_2$ using the relaxation curve in the observation point of soft tissue on the analogy with the procedure of calibration. The dependence of values P_1 and P_2 on the viscoelastic parameters of the soft tissue under study is known [6], that gives a principal possibility to determine the shear modulus and viscosity by solving the combined equations $P_1(c_t) = \hat{P}_1$ and $P_2(c_t, \eta) = \hat{P}_2$.

At the same time the assessed value of parameter $\hat{P}_2$ given in Table 2 was characterized by a large spread in values that results in great error in the estimation of viscosity. That is why at the estimation of viscosity $\hat{\eta}$, the assessed values of which are given in the Table 2, another method has proved to be more reliable. This method uses additionally the information about the dependence of maximum displacement in the focal point on the excitation beam parameters and the viscoelastic properties of the examined tissue. The value of maximum displacement in the focal point can be written in the form:

$$S_{MAX} = \overline{P}_0 P_1'(c_t)(1 - P_2'(c_t, \eta)), \quad (2)$$

where dimensional parameters P_1' and P_2' on the analogy with P_1 and P_2 are determined by physical characteristics of the examined soft tissue and the parameters of the excitation pulse. Therefore the shear wave velocity and tissue viscosity can be found by solving the combined equations:

$$P_1(c_t) = \hat{P}_1 \quad (3)$$

$$P_2'(c_t, \eta) = 1 - S_{MAX} / \overline{P}_0 P_1'(c_t) \quad (4)$$

Figure 4 shows the relaxation curves in the gelatin phantom 5 and phantoms 6 and 7 containing raw calf beef (phantom 7) and perfused by physiologic solution liver (phantom 6). At that it was analyzed the sample of relaxation curve in phantom 5 differing from the calibration one. These phantoms have been chosen for verifying of proposed method together with the gelatin phantoms 1 and 4 having the assessed values of $\hat{P}_0$ differing most significantly from the mean value $\overline{P}_0 = 9.74$ (see Table 2). According to the described algorithm the values of $\hat{c}_t$ and $\hat{\eta}$ were reconstructed for every phantom. These values are given in Table 3 together with the values of velocity c_t measured by SWEI method.

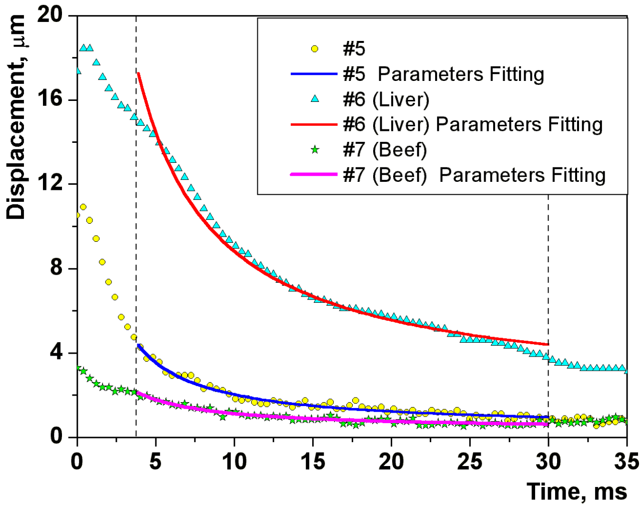

Figure 4 : Displacement through the time in the focal point in phantoms 5-7

Table 3 : Estimated parameters of testing phantoms

#	$S_{MAX}, \mu m$	$\hat{P}_1, \mu m \cdot ms$	$\hat{\eta}, Pa \cdot s$	$\hat{c}_t, m \cdot s^{-1}$	$c_t, m \cdot s^{-1}$
1	117	29.4	0.040	0.26	0.32
4	10.50	2.75	0.228	0.94	0.75
5	10.94	1.70	0.278	1.10	0.98
6	18.8	7.84	0.156	0.57	0.75
7	2.85	0.91	0.674	2.54	4.0

Note that the shear viscosity of the phantoms used for calibration and testing was not measured by independent method, but the obtained values fairly agree with the known typical value for water solutions of gelatin [7].

It is seen from Table 3 that from three gelatin phantoms the greatest error in the estimation of the shear wave velocity, as it was expected, took place for phantoms 1 and 4. However, even for these phantoms it did not exceed 20-25% while for phantom 5 the error is about 9%. Relatively low preciseness of the velocity estimation, which is equal to 24 and 36%, respectively, was observed also for phantoms 6 and 7. This is explained by rather complicated nature of shear strain relaxation, which is often observed in soft tissues. Possible reasons of it are discussed, in particular, in Refs. [3,5].

Conclusion

It is generally accepted that the tissue displacement induced by pulsed acoustic radiation force is inversely proportional to the local stiffness of the tissue. The reasons for this statement are the known calculations [8] for the displacement value under the effect of the localized static loading, which are not applicable, generally speaking, in the case of pulse loading. The experimental data given in the present paper and the results of quantitative estimation of viscoelastic parameters testify that the displacement magnitude of induced shear strain is inversely proportional to the shear wave velocity $c_t = \sqrt{\mu/\rho} \approx \sqrt{E/3\rho}$, i.e. to the square root of the Young modulus E .

Moreover, a great contribution into displacement magnitude is made by the tissue shear viscosity. That is why the one- and two-dimensional images of displacement distribution obtained, for example, by ARP method, are not actually a map of space distribution of shear hardness but some integral images of hardness and viscosity of the tissue. It can be easily seen if we compare the results for phantom 3 and phantoms 4 which have nearly equal values of the shear modulus. At the same time they are by several times different as to observed displacement magnitude (see Fig.3) because of different composition and considerably different viscosity.

The present research suggests the easiest procedure of calibration of ARP and SWEI methods necessary for simultaneous estimation of tissue elasticity and viscosity by experimental data of the shear strain excitation and relaxation. It is clear that a more precise calibration procedure should use a set of phantoms with the known and measured by independent method shear modulus and viscosity. It will make possible for every particular ultrasonic device to specify some adjusting constants, which in theoretical analysis turn out to be dependent on the introduced model assumptions. The experimental research of the calibration procedure of such kind is the necessary stage for developing a real technology suitable for immediate application in medical ultrasonic scanners.

However, the preciseness of estimation of shear velocity using the test phantoms after the proposed simple calibration procedure is quite satisfactory in our opinion. As it is known the shear elasticity modulus in the tissues with pathological changes is often an order of magnitude greater than that of normal tissue, as it has place for example for breast lesion tissue [9]. Therefore the obtained preciseness of tissue elasticity estimation, which is not worse than 30%, is suitable for the medical applications already now.

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