Elasticity Moduli Estimation Using Different Elastography Methods: A Phantom Based Approach

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Received: 20.11.2015

Abstract. Recently elastography has become a rapidly developing set of non-invasive techniques for estimation of soft tissues mechanical characteristics. The article represents the results of direct comparison of static (strain), vibration (based on mechanically induced vibrations) and shear wave elastography methods. For the first two methods the possibility of quantitative elastic moduli estimation has been proved previously. The results of the study on gelatin phantoms show satisfactory correspondence between all three methods. Besides of that, the results of comparison of static and vibration methods were very good.

Keywords: static, transient, shear wave elastography, elastometry, ultrasonic diagnostics, soft tissues strain, gelatine phantoms, ultrasound

INTRODUCTION

Elastography is a non-invasive ultrasonic visualization technique that allows examination the mechanical properties of soft tissues and organs and overcomes certain limitations of palpation, the technique used for thousands of years in the past and widely used nowadays, such as limited accessibility to internal organs and disease detectability, subjectiveness. The European Federation of Societies for Ultrasound in Medicine and Biology (EFSUMB) attempted to classify elastography methods [1] and proposed recommendations on their clinical use. All elastography methods can be divided in static (or strain) and dynamic ones. In static methods mechanical forces or pressure are considered constant (or changing slowly enough to get a set of images for comparative analysis to get a field of displacements in quasi-static methods). Dynamic methods pre-suppose the exposure of the soft tissue to variable forces like impulses of acoustic waves (focused or not focused at depth) or external pressure applied to the tissue surface.

The elasticity imaging emerged in the late 1980s from static and quasi-static methods, but still these methods are qualified as qualitative and not quantitative [1, 2]. The idea behind the methods is quite simple: being under compression, a softer tissue deforms greater than a stiffer tissue due to the difference of elasticity. In medical diagnostics, the inverse problem is

to be solved: to get elasticity characteristics of the tissue (shear modulus), it is necessary to know displacement and stress data in a zone of interest. While the first can be obtained using conventional ultrasonic scanners (at selected points) or by using equipment realizing the quasi-static method (as B-mode image), the second depends on many factors including pressure of surrounding tissues, pose of patient, tissue and inhomogeneity geometry, pressure under compression by external device or free-hand. However, as noted in [2], model-based approaches could decrease the shear modulus estimation error.

A static model of soft tissue inhomogeneity deformation proposed in [3] allows quantitative evaluation of the shear modulus for round-shaped inhomogeneity located at specified depth in homogeneous isotropic tissue. According to [4], the static method potentially can be up to 10 times more sensitive to small changes in shear modulus than vibration sonoelastography dynamic method.

Vibration elastography is based on applying the dynamic pressure to a soft tissue to induce periodical longitudinal and transversal displacements, which can be detected as vibration speed in specified tissue region by using the Doppler system of conventional ultrasonic scanners. Additional setup is required to create the dynamic pressure. The shear module can be evaluated by measuring the vibration speed in transversal direction and setting the amplitude and frequency of vibrations at zone of interest.

Shear Wave Elastography (SWE) is based on imaging the shear wave velocity [1]. The shear wave is exited by short radiation force impulses focused at various depths (or sweeped over depth fast enough to create a Mach cone of shear wave). No additional devices such as vibration generators or pressure sensors are needed for SWE methods, which allows so-called "free-hand" scanning technique, which is convenient in practice. On the other hand, specially designed acoustic sensors and ultrasonic scanners are needed for SWE option, so there is no compatibility with convenient ultrasonic equipment. Also, the "free-hand" method implies the absence of pressure requirement. Practically, shear vibrations are hard to excite from surface, but due to viscosity and complexity of tissues, focusing ultrasonic excitation techniques allow to produce longitudinal waves propagating in forward direction together with very fast fading shear vibrations propagating with very slow speeds around 3–10 m/s in transverse direction. There are several techniques to detect shear waves based on phased array sensors. Despite the fact that SWE technology is quite complex and obtained images are subjected to artifacts, large local variations and a little depth dependency in measured values, it allows to construct 2D elasticity map of a selected tissue region.

As stated in [5], due to different physical principles and computational algorithms used, static methods are suitable for local diseases, while shear wave methods are better for diffuse diseases.

Because of the widespread application of elastography in the medicine, it is necessary to make a direct comparison of different elastography methods and numeric values of soft tissue mechanical parameters given by them with general aim to increase the accuracy and reliability of elastography methods being developed. In this paper the research is based on comparison the static, vibration and shear wave elastography methods when measuring elasticity moduli of the same gelatin phantoms with inhomogeneities.

METHODS AND EQUIPMENT

To perform the study we created three gelatin phantoms with one cylindrical inhomogeneity in each of them as shown in Fig. 1. The main volume of phantoms was made of water solution with gelatin of 0.1 g/ml concentration to imitate the soft biological tissue. The first inhomogeneity in phantom 1 imitating the stiff tumor consists of 0.3 g/ml gelatin with small amount of talc powder. The second inhomogeneity in phantom 2 contains the same

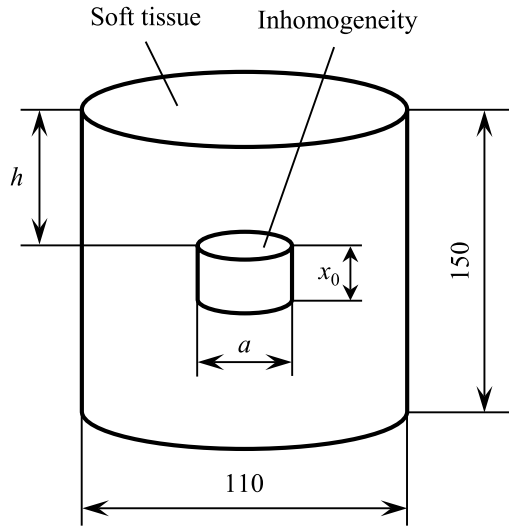


Figure 1. Scheme of soft tissue with inhomogeneity gelatin phantoms

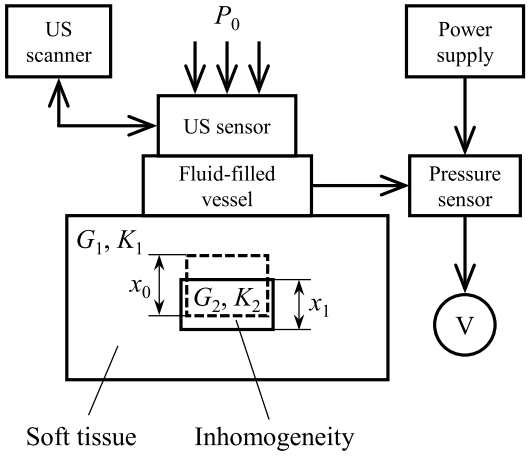


Figure 2. Functional block-diagram of the experimental setup for the static elastography

amount of gelatin but without talc. The third inhomogeneity in phantom 3 was made of 0.05 g/ml solution of gelatin in water to imitate softer (benign) tumor. Inhomogeneity parameters are presented in Table 1.

Table 1. Parameters of inhomogeneities

No.	Depth h, mm	Height x_0, mm	Diameter a, mm	Mass m, g	Density ρ, kg/m^3	Longitudinal waves velocity C_l, m/s
1	75	15	36	16	1048	1503
2	75	14	36	15	1053	1500
3	75	12	36	10	819	1375

For static elastography method, we used physical model of compression with constant pressure P_0 applied to the soft tissue [4] based on linear elasticity theory for isotropic media, according to which all the elastic moduli are connected with the following equations:

$$\nu = \frac{C_l^2 - 2C_t^2}{2C_l^2 - 2C_t^2}, \tag{1}$$

$$E = \frac{C_l^2 \rho(1+\nu)(1-2\nu)}{1-\nu}, \tag{2}$$

$$G = C_t^2 \rho = \frac{E}{2(1+\nu)}, \tag{3}$$

$$K = \frac{E}{3(1-2\nu)}, \tag{4}$$

where ν – Poisson's ratio; C_l and C_t – longitudinal and transverse wave velocities, m/s, correspondingly; E – Young's modulus, Pa; ρ – density, kg/m^3; G – shear modulus, Pa; K – bulk elasticity (modulus of dilatation), Pa.

The block diagram of experimental setup is presented in Figure 2. The convenient ultrasonic scanner (in our case, the Edan U50 portable system by Edan Instruments, Inc. with C352UB convex array sensor) works in standard B-mode and provides graphical interface for

distance measurements (assuming $C_l = 1540$ m/s) necessary for longitudinal strain ε_{xx} evaluation. Pressure P_0 is controlled by hand of researcher holding the ultrasonic sensor and is measured by specially designed device consisting of fixed soft silicone vessel filled with water, an infusion system and a pressure sensor with linear characteristic curve. Two measurements, before ($P_0 = 0$) and during compression ($P_0 > 0$), are necessary to evaluate the deformation ε_{xx}.

The inhomogeneity shear modulus G_2 evaluation is based on numerical dependencies [3] of deformation ε_{xx} on pressure P_0 and shear modulus G_2 obtained with the use of finite element method modeling and taking into account geometrical and physical peculiarities of phantoms.

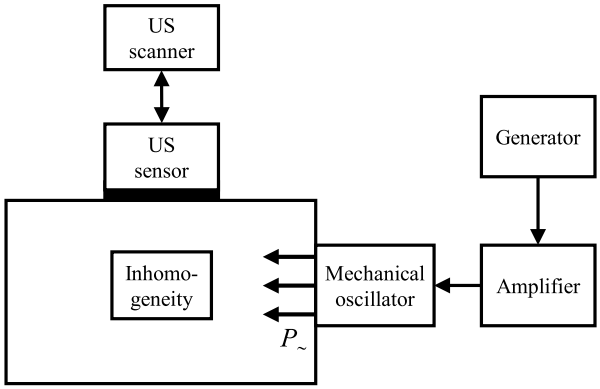


Figure 4. Block diagram for vibration elastography method

The block diagram for vibration elastography method is presented in Fig. 4. Mechanical oscillator supplied by amplified harmonic signal of 60–150 Hz frequency from a function generator creates the variable pressure $P_\sim$ in volume near the inhomogeneity from a lateral phantom wall. Ultrasonic scanner Edan U50 operates in power Doppler mode and displays vibration speed V measured at zone of inhomogeneity as a color map or in spectrogram-like form (vibration speeds versus time) at maximum of the first harmonic. The Young's modulus E was then calculated from the equation [4]:

$$V = 2.13\pi f_{\mathrm{US}} \frac{\sigma_{xx}\lambda}{E},$$

where f_{US} – ultrasound frequency, σ_{xx} – mechanical harmonic stress, λ – wavelength.

Shear wave elastography was performed using Aixplorer® ultrasonic system (by SuperSonic Imagine) with convex SC6-1 probe with following technical specifications: 192 elements, spatial resolution 2.9 mm, precision on elasticity displayed ±15 %, elasticity range 0–300 kPa, penetration depth >75 mm. The system measures the shear (transversal) wave propagation velocity C_t in m/s and then converts it to a Young's modulus E, using the approximate equation $E = 3\rho C_t^2$, where the tissue density ρ is assumed to be close to that of water and equal to 1000 kg/m^2 [6]. The output on-screen Young's modulus values $E_{\min}$ (minimum), $E_{\max}$ (maximum) and E_{mean} (mean) are computed for selected round shaped area of 2 cm in diameter by corresponding equations:

$$E_{\min} = 3000 \cdot (V_{\min})^2, \quad E_{\max} = 3000 \cdot (V_{\max})^2,$$

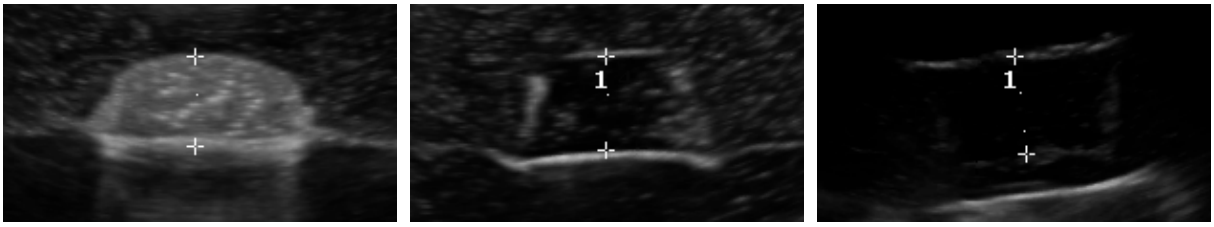
$$E_{\mathrm{mean}} = 3000 \cdot \left(V^2\right)_{\mathrm{mean}} = \frac{3000 \cdot \sum_{i=1}^{n} V_i^2}{n},$$

where $n \in \mathrm{N}$ – amount of values measured inside the area.

RESULTS AND DISCUSSION

Static elastography requires performing the linear dimensions measurement in B-mode. Because it is necessary to do at least two measurements – before and during compression, the

measurement accuracy highly depends on clearness of the region of interest borders. B-mode images of inhomogeneities in gelatin phantoms obtained by Edan U50 ultrasonic scanner are presented in Fig. 5. Borders can be clearly seen although to get an accurate result the probe should not be taken off the surface or revolved about its axis between measurements. The first inhomogeneity looks grey because of the talc powder presence, and its strain ε did not exceed 8.8 % (1.4 mm) under pressure P_0 of 4.2 kPa. Considering model dependencies of strain ε on elastic and geometric parameters of the phantom described in [3], the estimated Young’s modulus E value is 60 kPa. The strain of the second inhomogeneity ε under the same pressure is 12.5 % (2.2 mm), and the Young’s modulus E is 31 kPa. The third inhomogeneity is softer than surrounding media, its borders are not that contrast, but the strain under the same pressure P_0 = 4.2 kPa is larger: ε = 28.8 % (3.2 mm), so the Young’s modulus E = 18 kPa.



Inhomogeneity 1 (stiff) Inhomogeneity 2 Inhomogeneity 3 (soft)

Figure 5. Measurements of inhomogeneity height in B-mode (static elastography)

Vibration speed measurement accuracy may vary depending on the Power Doppler mode implementation, scaling to low values under 10 cm/s and pixel resolution in the ultrasonic scanner. An example of vibration elastography measurement on Edan U50 is shown in Fig. 6. Vibration speed spectrogram at frequencies under 20 Hz starts to be discontinuous in the time domain, and under 10 Hz it becomes hard to distinguish the peak value. On the other hand, waves of lower frequencies are able to propagate deeper. So, the appropriate rate to expose inner organs, tissues or inhomogeneities in gelatin phantoms, as in our case, to sonic vibrations excited from the surface should belong to the range of 10–200 Hz. The resulting values of Young’s modulus E for the first, the second and the third inhomogeneities, are 62, 36, 23 kPa, correspondingly.

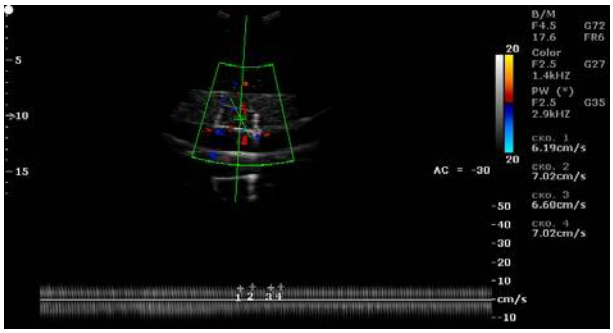
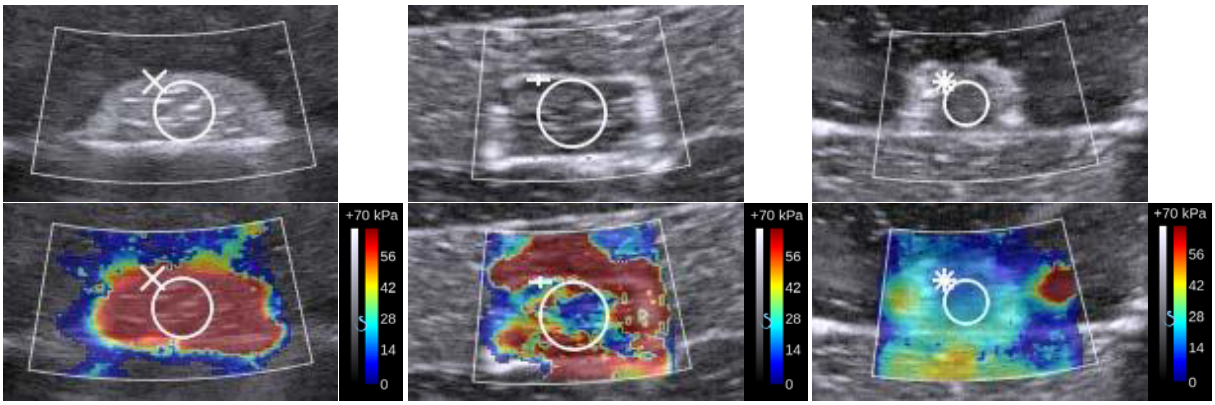


Figure 6. Vibration speed measurement (vibration elastography)

2D elasticity images of the same inhomogeneities obtained from Aixplorer ultrasonic system, implementing shear wave elastography method, and presented in Fig. 7 shows that stiffness of inhomogeneities varies from hard to soft. Apart from the colormap, the min, mean and max values over the round area are displayed. The first inhomogeneity is clearly mapped red and the mean E = 84 kPa. The second inhomogeneity is not very explicit, and the colormap changes a lot during measurements. As a result, the image contains extremely small (E_{min} = 1 kPa) and extremely large (E_{max} = 160 kPa) values. From Fig. 7, b it can be seen that the border area is more stiff than the central area. The reasons of this may be specific phantom properties or imaging artefacts, which need more study. Nevertheless, the mean Young’s modulus value E = 41 kPa. The third inhomogeneity in Fig. 7, c looks like it is stiffer than surrounding media, which is wrong (it may be a border effect), and the red spot on the right seems to be an artefact. Inside the circle, the mean Young’s module value is equal to reasonable 22 kPa.



Inhomogeneity 1 (stiff) Inhomogeneity 2 Inhomogeneity 3 (soft)

Figure 7. B-mode and Young's modulus distribution images (shear wave elastography)

Table 2. Elastic moduli of inhomogeneities in gelatin phantoms estimated using different elastography methods

No.	Young's modulus *E*, kPa					Shear modulus *G*, kPa	
	Shear Wave Elastography (Aixplorer®)			Vibration elastography	Static elastography	Vibration elastography	Static elastography
	Min	Mean	Max				
1	61	84	103	62	60	20	19
2	1	41	160	36	31	12	10
3	15	22	29	23	18	8	6

The results of comparative study are accumulated in the Table 2. We can see satisfactory correspondence between all examined elastography methods and better correspondence between static and vibration methods.

CONCLUSION

Static and vibration elastography methods are applicable for quantitative measurements of elastic moduli in viscoelastic media, for example, gelatin phantoms, soft tissues or rubber-like materials, using conventional ultrasonic scanners. For soft tissues diagnostics more specific model is required considering geometric and elastic properties of the tissue itself and its surroundings (other tissues, blood vessels). Whereas static and vibration methods allows to estimate integral shear moduli values over all of the inhomogeneity depth, the Shear Wave Elastography methods allows to get local elastic moduli values at the region of interest. Large local deviations in measured values of the Young's modulus *E* could be caused by inhomogeneous phantoms structure and imaging artefacts. There is a need in less stiff and more homogeneous phantoms for soft tissues imitation.

The work was supported within the project No. 3.751.2014/K as a project part of the state task of the Ministry of Education and Science of the Russian Federation to FSBEI HPE "Kalashnikov Izhevsk State Technical University" for 2014–2016 years.

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