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Shear-Wave Sources for Soft Tissues in Ultrasound Elastography

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Abstract

Elastography is an imaging tool for soft-tissue characterization in the human body, and it is based on the propagation of shear waves. Since its first use in the ultrasound landscape, the generation of shear waves has had practical and theoretical issues. In this review, the use of four methods of shear-wave generation are considered: an external vibrating source; a radiation pressure source; a natural source; and an electromagnetic source. The pros and cons of these methods are detailed and discussed.
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Keywords: Ultrasound; Elastography; Shear waves

1. Introduction

The ultrasound community long considered the soft tissues of the human body as a fluid where only pressure waves can propagate, as because of the strong attenuation, ultrasound shear waves can barely propagate. The discovery of low-frequency shear waves in dynamic elastography redefined human soft tissues in terms of their solid properties. As in any solid, in soft tissues the elasticity can be characterized by two parameters: the Lamé coefficient λ which is related to compression waves; and the Lamé coefficient μ for shear waves (also known as the shear modulus). Soft tissues are considered as quasi-incompressible tissues ($\lambda \gg \mu$). This implies that the elasticity measured by palpation is related to the shear modulus. Measurement of the speed of shear waves in this medium is thus quantitatively equivalent to palpation. Therefore, studies have usually been based on the different approaches that can be used to observe the shear waves. In the present paper, we review the methods used in elastography for shear-wave generation, as an external vibrating source, a radiation pressure source, a natural source, and an electromagnetic source. Other studies have been carried out in elastography, such as that of Ophir et al. that was based on quasi-static elastography [34]. However, this is not the main purpose of the present review.

2. External vibrating source

In 1983, Eisenscher et al. presented an ultrasound method for use in soft tissues that was called 'echosonography' [15]. This method used a mechanical piston placed on the surface of the skin that produced low-frequency vibrations. Eisenscher et al. were thus the first to mix ultrasound with low-frequency mechanical vibration. However, this only provided qualitative characterization of the tissues.

In 1987, Krouskop et al. proposed a method for quantitative measurements of wave propagation in tissue through the measurement of the shear-wave speed [28]. This experimental method consisted of the use of an external vibrating source in contact with the limb tissue, and the measurement of the induced displacement by the ultrasonic Doppler method (see Fig. 1). These authors showed that this method can detect the local displacement field and further quantify the global elasticity properties of the soft tissues.

In 1988, Lerner et al. also used an external vibrating source, but with a color-flow map imaging system that was designed for real-time display of blood flow [29]. In this method, the frequency of the vibrations might have been limited to a few Hz because of the low frame rate of the imaging device. Contrary

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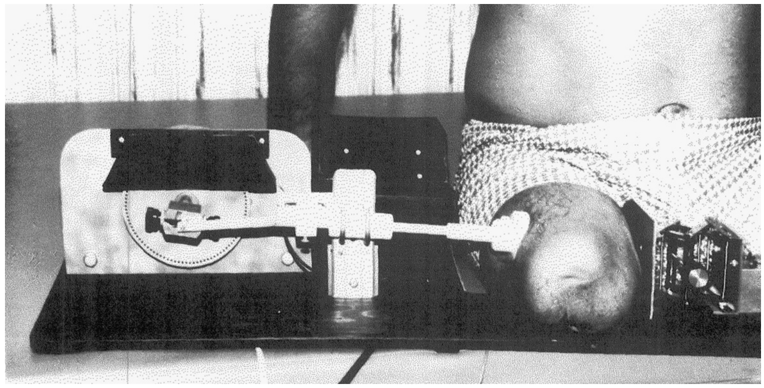

Fig. 1. Ultrasonic sensor used to characterize the modulus of soft tissue by Krouskop et al. [28].

to that of Krouskop et al. [28], this method was only qualitative, although it allowed the creation of an image. This method is now known as 'sonoelasticity'.

The measurement method of Lerner et al. [29] was modified by Yamakoshi et al. in 1990 [48]. They still used an external vibrating source, but they replaced the measuring device by an ultrasound phased-array transducer. This was used to obtain amplitude and phase maps from the Doppler frequency. This method allowed them to use higher frequencies (i.e., several hundred Hz), because of the higher frame rates. Increasing the frame rate thus allowed them to create quantitative images.

Previous methods for sonoelasticity were based on harmonic imaging. However, there was a problem here because the shear-wave speed did not agree with the theoretical models, as observed by Fujii et al. in 1995 [16]. These authors tried to use a different model that took into account the source. However, it was only in 1999 that this issue was solved, with the development of transient elastography by Catheline et al. [12,11]. These authors used a similar set-up, but changed the emitted vibration to a pulse. This was made possible by the development of ultrafast ultrasound imaging by Sandrin et al. [39]. The difference between the shear-wave propagation models and the experiments were due to a longitudinal component of the near field. The correct description was then made possible in soft solids by the use of pulse emission and the elastic Green's function. This harmonic Green's function is defined as the angular frequency ω between two points at a distance r apart and between the direction of polarization m and n, according to [1]:

$$G_{mn}(\omega,r)=\frac{1}{4\pi\rho\alpha^2}\frac{\gamma_m\gamma_n}{r}e^{iqr}+\frac{1}{4\pi\rho\beta^2}\frac{\delta_{mn}-\gamma_m\gamma_n}{r}e^{ikr}+\frac{1}{4\pi\rho}\frac{3\gamma_m\gamma_n-\delta_{mn}}{r^3}\times\left[\frac{e^{ikr}}{i\omega}\left(\frac{r}{\beta}-\frac{1}{i\omega}\right)-\frac{e^{iqr}}{i\omega}\left(\frac{r}{\alpha}-\frac{1}{i\omega}\right)\right] \quad (1)$$

where α and β are the compression and shear-wave velocities, $q=\omega/\alpha$ and $k=\omega/\beta$ are the corresponding wave numbers, and $\gamma_i=x_i/r$ is the cosine director. Transient elastography was then developed in 1-D for hepatic fibrosis [41,40], as a study of hepatic fibrosis and in 2-D for breast tumors [4]. These techniques were based on the measurement of the longitudinal component of the shear waves, as described in Equation (1) [12,8].

3. Acoustic radiation force

The previous section dealt with external sources for shear-wave production. However, the use of an external vibrating source has practical limitations, such as not being 'user-friendly' for clinical application by a physician, because it can be too heavy and/or too large. Another method was proposed by Sugimoto et al. in 1990 [43]. This consisted of the use of an acoustic radiation force. An acoustic radiation force is a unidirectional force that can be applied to absorbing and reflecting targets [45]. This corresponds to a transfer of momentum between the acoustic wave and the target. Therefore, in a medium, displacements can be created due to this force. Thus, this corresponds to shearing created by a compression wave. In their study, the acoustic radiation force was created by a geometrically focused transducer, and the shear strain was measured by an ultrasonic transducer. Sugimoto et al. [43] wanted to investigate the different parameters that they could use to measure the tissue hardness.

One of the first ideas was to use the acoustic radiation force in order to mimic the palpation by a physician, shear wave propagation was not considered yet. So, one of the possibilities was to measure the elastic modulus from the relation between the force and the deformation of the medium, which is also known as the spring ratio. They also observed the relaxation time of the medium. In 2001, Nightingale et al. applied this idea to the use of an ultrasonic transducer array [32]. This allowed the generation of multiple pushes and track beams, to create a two-dimensional map of the stiffness. This method was based on the measurement of the displacement. They showed the feasibility of this method in a phantom, and later they showed the clinical feasibility *in vivo* [33]. However, this method has some limitations, as it is a qualitative method and it requires many pushes to image the medium. Using too many ultrasonic pushes can also exceed the US Food and Drugs Administration recommendations.

Previous studies have used an acoustic radiation force to create inner tissue palpation through a quasi-static method. However, some studies began to consider the case of the shear

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