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● *Review Article*

ACOUSTIC WAVES IN MEDICAL IMAGING AND DIAGNOSTICS

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Abstract—Up until about two decades ago acoustic imaging and ultrasound imaging were synonymous. The term *ultrasonography*, or its abbreviated version *sonography*, meant an imaging modality based on the use of ultrasonic compressional bulk waves. Beginning in the 1990s, there started to emerge numerous acoustic imaging modalities based on the use of a different mode of acoustic wave: shear waves. Imaging with these waves was shown to provide very useful and very different information about the biological tissue being examined. We discuss the physical basis for the differences between these two basic modes of acoustic waves used in medical imaging and analyze the advantages associated with shear acoustic imaging. A comprehensive analysis of the range of acoustic wavelengths, velocities and frequencies that have been used in different imaging applications is presented. We discuss the potential for future shear wave imaging applications. (E-mail: urban.matthew@mayo.edu) © **2013 World Federation for Ultrasound in Medicine & Biology.**

Key Words: **Compressional wave, Shear wave, Elasticity, Viscoelasticity, Acoustic imaging, Dispersion, Anisotropy.**

INTRODUCTION

Two decades ago, in the field of medical imaging, the terms *acoustic imaging* and *ultrasonic imaging* were synonymous. The only acoustic waves used for imaging biological structures were ultrasonic compressional (or longitudinal) waves. In the 1990s, a new acoustic imaging technology started to emerge that was based on shear (or transverse) acoustic waves. In the remainder of this article, we use the term *shear wave* to denote the entire family of transverse waves.

The wave speeds of these different kinds of waves are governed by two different types of moduli. Compressional wave speed is related to the bulk modulus of the tissue, whereas shear wave speed is related to the shear modulus. Compressional wave speed does not vary significantly for biological tissues compared with the variation of the shear wave velocity in the same tissues. For this reason, elasticity imaging, which is targeted at imaging the shear modulus of tissue, has a wide dynamic range that can be exploited (Sarvazyan et al. 1998).

The purpose of this article is to explore the differences between imaging with compressional and shear waves. We explore the ranges of relevant frequencies used in each modality and the ranges of acoustic wave speeds. We explore how different imaging techniques exploit parameters obtained with the use of shear waves and discuss regions of these parameter spaces that have yet to be explored.

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MECHANISMS OF CONTRAST IN ACOUSTIC IMAGING

The wave motion in a medium is governed by the wave equation. For simplicity we assume a linear, elastic, isotropic and homogeneous medium (Manduca et al. 2001)

$$\rho \frac{\partial^2}{\partial t^2} u(\mathbf{x}, t) = (\lambda+\mu)\nabla(\nabla \cdot u(x,t)) + \mu \nabla u(\mathbf{x}, t), \qquad (1)$$

where ρ = mass density; λ_L and μ_L = Lamé parameters; t = time; and $\mathbf{x}$ = spatial vector defined as $\mathbf{x} = [x, y, z]$. A solution of the wave equation is given by

$$u(\mathbf{x}, t) = u_0 e^{i(\omega t - k\mathbf{x})}, \qquad (2)$$

where u_0 = displacement amplitude; ω = angular frequency; and k = wavenumber: $k = \omega/c$, where c is the speed of the acoustic wave. As stated above, compressional waves have been used for more than 60 years to

image the internal structures of the body. Conventional B-mode imaging is based on differences in acoustic impedance of tissue, Z, which is given by

$$Z = \rho c_c, \tag{3}$$

where c_c = compressional wave velocity. Compressional wave velocity is related to both Lamé parameters by

$$c_c = \sqrt{\frac{\lambda_L + 2\mu_L}{\rho}} = \sqrt{\frac{E(1-v)}{\rho(1+v)(1-2v)}}. \tag{4}$$

The above relationship can also be written in terms of the bulk modulus, K, and the shear modulus, μ, where $K = \lambda_L + 2\mu_L/3$ and $\mu = \mu_L$.

$$c_c = \sqrt{\frac{K + 4\mu/3}{\rho}}. \tag{5}$$

It is shown later that the bulk modulus is typically several orders of magnitude larger than the shear modulus in tissues, so the compressional wave velocity is almost solely determined by the bulk modulus of the tissue. The bulk and shear moduli can also be written as (Sarvazyan et al. 2011)

$$K = \frac{E}{(1+v)(1-2v)}, \tag{6}$$

$$\mu = \frac{E}{2(1+v)}, \tag{7}$$

where E = compressional Young's modulus, assuming a Hookean elastic solid; and v = Poisson's ratio. In most soft tissues, Poisson's ratio is assumed to be very close to 0.5, which is the condition for incompressibility.

Both fundamental components of bulk acoustic impedance of a material, density and bulk modulus, are dependent on the molecular composition of that material. Water is the main molecular component of soft tissue; thus, the speed of compressional waves in all soft tissues lies within the range ±10% that of water (Duck 1990; Goss et al. 1978, 1980; Sarvazyan and Hill 2004). It is well known that the speed and attenuation of compressional waves in soft tissue are defined mainly by its molecular content rather than structure: disintegration, that is, mechanical homogenization of tissue, generally does not lead to substantial, immediate change in these acoustical parameters (Pauly and Schwan 1971; Sarvazyan et al. 1987). The speed of compressional waves in liver tissue samples of different levels of structural integrity (intact, ground and highly homogenized) differs less that 0.5% (Sarvazyan et al. 1987). The wave speed in the ground tissue is slightly higher than that in the intact tissue, and this small increase is explained by an increase in the level of hydration of some biopolymers released in the ground tissue. Both compressibility and density, which define compressional wave speed in tissue, are determined by short-range intermolecular interactions, and water, as the major component of soft tissue, contributes the most to these bulk properties of tissue. Therefore, the images obtained using compressional waves represent mainly the pattern of water hydrating the molecules composing the tissue (Sarvazyan and Hill 2004). This pertains to imaging of most soft tissues such as liver, kidney, heart and skeletal muscle, which contain 70%–75% water and which are the main target of ultrasonic imaging, but this is not the case for tissues like lung, cartilage and fat, which do not have a high water content.

This strong association of acoustic tissue properties with those of water becomes evident from the comparison of temperature dependences of compressional wave speed in pure water and soft tissues. The most characteristic acoustical feature of water is a unique non-linear temperature dependence of sound speed resulting from the temperature-induced changes in the dynamics of hydrogen-bonded clusters. No other fluid or substance has similar dependence. The temperature dependence of compressional wave speed in tissue closely resembles that of pure water (Fig. 1) (Sarvazyan et al. 2005).

Although conventional sonography is based mainly on visualizing the spatial distribution of acoustic impedance, which is defined by short-range molecular interactions in the tissue, the imaging based on the use of shear waves shows quite a different set of macroscopic structural features determined by long-range molecular and cellular interactions (Sarvazyan 2001; Sarvazyan and Hill 2004).

Shear waves have been used over the last two decades to explore the underlying material properties of soft tissues (Sarvazyan et al. 2011). Shear wave speed is dependent on the shear modulus of the material as given by

$$c_s = \sqrt{\frac{\mu}{\rho}} = \sqrt{\frac{E}{2\rho(1+v)}}. \tag{8}$$

The shear modulus of a material is highly dependent on tissue architecture and structural makeup. This tissue architecture varies greatly depending on the organ and its state. For example, the liver is largely isotropic and homogenous with an intricate series of blood vessels running through it so that the liver can filter the blood supply. Alternatively, other organs are very specifically arranged. Skeletal muscle can be assumed to be transversely anisotropic. That is, it is arranged into bundles of fibers in a semi-crystalline architecture. In this case,