A Review of Theoretical and Experimental Aspects of Imaging the Elastic Attributes of Tissue In Vivo

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The literature describing the gross mechanical properties of tissues is primarily concerned with muscle (skeletal and cardiac) as well as with blood vessel walls [1]. Relatively little has been written about the mechanical properties of other normal and pathological tissues. Pathological changes are generally correlated with changes in tissue elastic modulus; in fact, the standard medical practice of soft tissue palpation is based on qualitative, low-resolution assessment of the static elastic modulus of tissue. Many cancers, such as scirrhous carcinoma of the breast, appear as extremely hard nodules [2]. In quite a few cases, despite the difference in elastic modulus, the small size of a pathological lesion and/or its location deep in the body precludes its detection and evaluation by palpation. In general, the lesion may or may not possess echogenic properties that could make it ultrasonically detectable. For example, tumors of the prostate or the breast could be invisible in standard ultrasound examinations, yet be much harder than the embedding tissue.

Diffuse diseases such as cirrhosis of the liver are known to significantly reduce the elastic modulus of the liver tissue as a whole [2], yet they may appear normal in conventional ultrasound examination. Because the ultrasonic echogenicity and the elasticity of tissue are generally uncorrelated, it is expected that imaging tissue elastic modulus will provide new information that is related to tissue structure and/or pathology (Fig. 1).

Biological tissues can be considered as approximating homogeneous gels [3]. Different modes of elastic wave propagation in such media are determined primarily by their bulk (K) and shear (G) elastic moduli. In biological soft tissues, the value of K far exceeds that of G. The bulk properties (and hence the ultrasonic properties) are determined by the molecular composition of the tissue [1], whereas the shear properties are determined by the higher level of tissue organization [3]. Because deformable soft tissues are essentially volume incompressible (i.e., Poisson ratio, ~0.5), their shear moduli are proportional to their longitudinal (Young's) moduli [4]. It follows that estimation and imaging of the Young's moduli of tissue should in principle convey information about their shear properties, and hence about the higher level of tissue organization.

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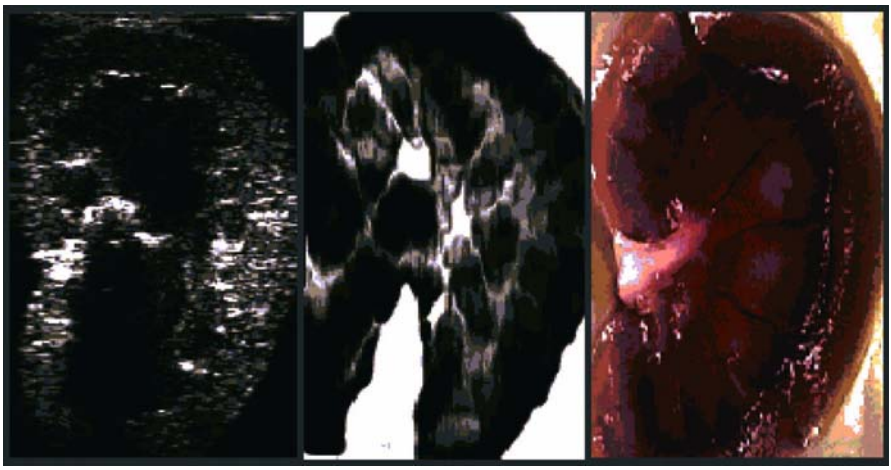

FIG. 1. A sonogram, elastogram, and gross pathology slide showing the longitudinal appearance of an ovine kidney in vitro. Note the distinct appearance of the pyramids in the elastogram and the contrast in stiffness between the cortex and the medulla on the elastogram (*black*, stiff; *white*, soft)

No known modality is capable of imaging these important elastic properties of tissue directly. It is therefore necessary to apply an external mechanical stimulus to the tissue system and to observe the tissue response in terms of local internal deformations. In principle, any high-resolution imaging modality may be used for such observations. The use of ultrasound for this purpose, however, has several important advantages that include real-time imaging capabilities, very high resolution in motion estimation (~1 μm), simplicity, noninvasiveness, and relative low cost.

Ultrasonic methods for deriving information related to the elastic properties of soft tissues have been described in the literature in the past 10 years. These techniques rely on one of the following procedures: Doppler tissue velocity measurements [5–7], decorrelation techniques to quantify motions in tissues [8–11], and visual inspection of M-mode and B-mode image. Internal mechanical excitation (motion of cardiac structures, arterial pulsation) or external vibrational sources of motion are used to produce displacement of the tissues under investigation. The displacements of different tissues are then analyzed by one of these techniques.

Elastography is performed by obtaining a set of ultrasonic echo signals from a target, subjecting the target to a small axial deformation, and obtaining a second set of echo signals. Time-delay estimations along the direction of the applied load are computed by performing piecewise cross-correlation analysis on congruent pairs of signal segments. Using the gradient operator, the time-shift estimates are then converted to longitudinal strain information, which could be displayed in the form of a two-dimensional strain image named elastograms. We have demonstrated that the

strain distributions in tissues are correlated to the distribution of tissue elastic moduli under certain conditions. Furthermore, general inverse methods for calculating elastic modulus images from elastograms have been described by Skovoroda et al. and by Sumi et al. [12]. In addition to axial strain elastograms, it is also possible to obtain lateral strain elastograms and Poisson's ratio elastograms, which convey additional tissue mechanical information.

We have constructed an elastography device based on an Philips/ATL HDI-1000 imaging system, which was designed to produce sonograms and corresponding elastograms of the breast in vivo. This device is able to acquire elastograms at a rate of up to 15/s. We have also been working on elastographic imaging of the prostate. Elastograms can produce good-quality images of tissue strain that easily demonstrate stiff yet ultrasonically occult, isoechoic lesions. The images in Fig. 1 demonstrates the different sonographic and elastographic appearance of an ovine kidney in vitro.

We have also developed a theoretical framework (called a strain filter or strain transfer function) that can predict the upper bound of imaging performance of elastography given the engineering system parameters. This tool is based on the Cramer–Rao lower bound for the variance of time-delay estimations and on the analysis of the decorrelation of shifted signals. Once the system bandwidth, center frequency, sonographic signal-to-noise ratio (SNR), and window size are specified, and the estimation algorithm known, the elastographic parameters of elasticity SNR, dynamic range, sensitivity, contrast-to-noise ratio, and resolution may be calculated. The strain filter may be derated by nonstationary effects such as transducer beam profiles and attenuation in the tissues.

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