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● *Original Contribution*

FREEHAND ULTRASOUND ELASTOGRAPHY WITH A 3-D PROBE

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Abstract—This paper presents the first near-real-time freehand ultrasound elastography system using a (3-D) mechanical probe. Acquisition is complete within two sec, and only an additional 20 sec are required for generation of a full 3-D strain volume. The strain is axial, with estimates of lateral and elevational tissue movement used to increase the accuracy of the axial strain measurement. This is the first time all system components have been extended to 3-D, *i.e.,* 3-D windows are used to track displacement, which is tracked in all directions, and 3-D kernels are used for least-squares gradient estimates. Normalization of the freehand 3-D strain data is also applied across the whole volume. The system is tested using a novel research 3-D radiofrequency (RF) system with real-time control over the stepper motor driving the ultrasound probe, and real-time streaming of RF ultrasound data. The paper proves the concept, rather than making significant comments on the achievable accuracy in 3-D, although we demonstrate that the high performance of the 2-D techniques that we extend appears to carry through to *in-vitro* and *in-vivo* 3-D data. The result is a fast and high-resolution 3-D image of normalized axial strain. (E-mail: gmt11@eng.cam.ac.uk) **© 2008 World Federation for Ultrasound in Medicine & Biology.**

Key Words: **Strain imaging, Elastography, Three-dimensional, Freehand.**

INTRODUCTION

Ultrasound strain imaging, or elastography, is a way of visualizing tissue stiffness: it can be seen as the imaging equivalent of manual palpation. It is currently receiving much attention from within both the research and commercial sectors. Although its clinical usefulness has yet to be proven beyond doubt, it seems very likely that this will become an important technique. Applications have already been indicated in imaging certain tumours (Garra et al. 1997), atherosclerosis (de Korte et al. 1998) and any other masses that are expected to be stiffer than the surrounding tissue. It may also have a role in studying the physiological and pathological mechanical properties of soft tissue (Gennisson et al. 2004; Vogt and Ermert 2005).

High-quality quasistatic freehand 2-D ultrasound strain imaging is currently possible in real time (Lindop et al. 2007a) by comparing radiofrequency (RF) data from sequential ultrasound images during a slight deformation of the tissue because of contact pressure from the ultrasound probe. It has recently been shown that it is possible to produce stable, high-quality strain data over a wide range of freehand motions of the probe (Lindop et al. 2007b). The approach has considerable benefits over other techniques in that no additional equipment is required to induce the required stress field in the tissue: the clinician simply moves a normal ultrasound probe over the anatomy. This potentially makes the technique simpler to use as well as easier to implement.

In just the same way as for 3-D ultrasound, the availability of 3-D strain data would be beneficial for more detailed geometry and more accurate measurements, in particular of the volume of small or irregular masses. Because the real strain during applied pressure from an ultrasound probe has a pronounced 3-D variation, it may also be beneficial to show images of true strain, rather than axial strain, which is the current convention. However, the lateral and elevational components of true strain can generally be estimated with much less precision. Hence, in common with most work in this area, we focus on 3-D volumes of axial strain, making use of estimates of lateral and elevational tissue movement only to increase the accuracy of such axial strain measurements.

Previous work on 3-D strain imaging has generally followed one of two directions. In one approach, 2-D data is acquired as the probe is gradually moved in the elevational direction, forming a sequential 3-D dataset. Neighboring frames are used to estimate 2-D strain images, which when stacked together create a 3-D dataset of 2-D strain

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estimates. This technique has been successfully implemented for freehand 3-D ultrasound (Lindop et al. 2006), and for intravascular ultrasound (Schaar et al. 2005; Li et al. 2006). The alternative approach involves the use of mechanical swept 2-D probes or 2-D phased array probes. Here the probe is held still, a volume of data is acquired and either a controlled compression is applied (Krueger et al. 1998; Lorenz et al. 1999; Fisher et al. 2006) or, in the case of strain estimation as a predictor of temperature change, no compression at all is applied (Anand et al. 2007). An exception is the approach of Insana et al. (1997), where a 3-D precompression volume is compared with a 2-D post-compression scan, resulting in a 2-D strain image with displacements tracked in 3-D. Three-dimensional acquisition has also been investigated for the related techniques of sonoelastography (Taylor et al. 2000) and prostate mechanical imaging using a pressure pad (Egorov et al. 2006).

The system presented in this paper is novel in several respects. First, to the authors knowledge it is the first time that fully 3-D windows have been used for displacement and gradient estimation. This is an important step because if we make the windows more representative of the fundamental resolution of the ultrasound data, we can maximize the quality of the strain estimates for a given strain resolution. This is also the first attempt to acquire 3-D strain data with a freehand quasistatic approach on a mechanical 3-D probe, with subsequent normalization of the data also applied to the entire 3-D dataset (Treece et al. 2007). We achieve this in a system that matches the fastest reported processing time of about 20 sec (Anand et al. 2007), although this technique only used 1-D windows.

We start by outlining the important stages in this system, before demonstrating results over a range of *in-vitro* and *in-vivo* data and drawing conclusions.

METHOD

Three-dimensional RF data is acquired in real time from a 3-D ultrasound probe, which has a geared internal stepper motor controlling a 2-D linear phased-array probe head. Each volume of data is acquired with the probe held steady and the motor sweeping in the same direction. A very slight additional pressure (or relaxation) is then applied to the probe in an approximately axial direction before a further volume is acquired. This can be achieved conveniently during the slight pause of just less than 1 sec while the motor returns to the starting position and the RF data is transferred to PC memory. The whole process, including the gap between each frame, takes just less than 2 sec. Good results can be achieved with just two volumes of data; however, it is also possible to acquire several volumes with slight pressure changes between each, and then combine them in subsequent processing. The results presented hereafter make use of only two volumes.

Having acquired the RF data, strain estimation follows the procedure shown in Fig. 1. The axial displacement estimator we use requires base-band analytic ultrasound signals. Often, such in-phase and quadrature (IQ) signals are available directly from the probe interface, in which case the displacement estimation can begin directly. However, in our research platform the ultrasound signals are digitized as RF passband signals —hence these must first be converted to baseband analytic form. This can be done easily by the use of a pair of passband Hilbert filters, followed by demodulation at the nominal center frequency of the ultrasound probe. The baseband signal r is therefore:

$$r = (p \otimes h_r + p \otimes h_i j) e^{-j\omega_c t} \quad (1)$$

where p is the original RF passband signal, ω_c is the approximate probe center frequency and h_r; h_i are the filters. Best results are achieved when h_i is a Hilbert (antisymmetric coefficients) filter with a passband covering only the expected ultrasound probe frequency range and h_r is a symmetric filter with a frequency response carefully matched to h_i. The convolutions must be calculated at the sampling frequency of 66.67 MHz (synchronous to the ultrasound machine clock) to retain high accuracy phase information. All other operations can be performed after down-sampling, usually by a factor of 5, with substantial benefit in processing speed and negligible cost to the strain accuracy; a suitable choice of factor depends on the ratio of sampling frequency to signal bandwidth.

Calculating axial, lateral and elevational offsets

We estimate axial displacement using the Efficient Phase Zero Search (EPZS) of Lindop et al. (2006), which is a variant of phase root seeking. This is a fast and accurate method, providing that the tracking of displacements is handled appropriately; we explain how this is achieved in the following section. EPZS gives axial displacements that are inherently subsample resolution. It can be performed very efficiently, because only a handful of iterations are required to converge on a given displacement estimate. Indeed, because we do not require normalized correlations (EPZS relies on the phase of the cross-correlation, and this remains unaffected by normalization), it is generally only necessary to perform one correlation per displacement estimate. Assuming our initial displacement estimate is d, on the first iteration we require the complex cross-correlation c of the pre- and postdeformation signals, r_a and r_b at relative shift d, in samples:

$$c = \sum_{i,j,k} r_{a(\mathrm{i,j,k})}\, r_b{}^*_{(\mathrm{i+d,j,k})} \quad (2)$$

where (i, j, k) is the extent of the 3-D window over which