

● *Original Contribution*

VOLUMETRIC ELASTICITY IMAGING WITH A 2-D CMUT ARRAY

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Abstract—This article reports the use of a two-dimensional (2-D) capacitive micro-machined ultrasound transducer (CMUT) to acquire radio-frequency (RF) echo data from relatively large volumes of a simple ultrasound phantom to compare three-dimensional (3-D) elasticity imaging methods. Typical 2-D motion tracking for elasticity image formation was compared with three different methods of 3-D motion tracking, with sum-squared difference (SSD) used as the similarity measure. Differences among the algorithms were the degree to which they tracked elevational motion: not at all (2-D search), planar search, combination of multiple planes and plane independent guided search. The cross-correlation between the predeformation and motion-compensated postdeformation RF echo fields was used to quantify motion tracking accuracy. The lesion contrast-to-noise ratio was used to quantify image quality. Tracking accuracy and strain image quality generally improved with increased tracking sophistication. When used as input for a 3-D modulus reconstruction, high quality 3-D displacement estimates yielded accurate and low noise modulus reconstruction. (E-mail: tgfisher@wisc.edu) Published by Elsevier Inc. on behalf of World Federation for Ultrasound in Medicine & Biology.

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INTRODUCTION

Strain imaging has shown promise in differentiating breast tumors (Garra et al. 1997). A multi-institution study on breast elastography concluded that viewing axial strain images along with conventional ultrasound images aids in differentiation between malignant and benign solid breast masses (Burnside et al. 2007). One-dimensional (1-D) and 1.25D array transducers (Wildes et al. 1997) are the standard for most ultrasound imaging applications. Two-dimensional (2-D) strain images can be readily formed from radio-frequency (RF) data acquired with these transducers by obtaining an RF echo frame, deforming the tissue, obtaining another RF frame and then tracking the motion that occurred among those two frames of RF data. Echo signal decorrelation limits the ability to track motion among frames of RF echo data. A major

source of signal decorrelation is out-of-plane motion (motion that is perpendicular to the imaging plane) (Insana et al. 1997; Chen et al. 2005; Rao and Varghese 2008). Thus, it is highly desirable to acquire volumetric data to better track tissue motion, as well as to view the lesion in multiple parallel planes and/or perpendicular planes to further aid with diagnosis. When trying to include a third spatial dimension in strain imaging, two major problems arise: obtaining a 3-D data set and tracking motion between 3-D data sets.

Various methods have been used to obtain a volume of 3-D RF data with varying amounts of success. Translating a 1-D array in the elevation direction may be the simplest to do but creating volumes without plane-to-plane rotation or shear motion is difficult. Insana et al. (1997), Deprez et al. (2009) and Richards et al. (2009), used laboratory fixtures to scan a rectangular phantom for 3-D tracking but this approach has limited clinical applicability. Lindop et al. (2006) used an optical position sensor to track the position of the 1-D ultrasound transducer and place the acquired 2-D data within a 3-D volume. The transducer was moved in the elevation

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direction acquiring about 3 cm of data in 10 s (Lindop *et al.* 2006). Significant artifacts were apparent when viewing the strain volume in the perpendicular direction. Sweeping a 1-D array in the elevation direction with transducers such as the Siemens C7F2 provides data points in a curvilinear grid with nonuniform spacing. This curvilinear sampling also adds an element of complexity that ideally would be avoided. There is a tradeoff with these devices between spatial and temporal sampling that results (potentially) in spatial over-sampling in some areas and under-sampling (for motion tracking) in others, though recent advances with rocked arrays have been promising (Fisher *et al.* 2007; Bharat *et al.* 2008; Treece *et al.* 2008). The GE system used in Treece's work has the advantage of relatively fast acquisition times (Treece *et al.* 2008) while the Siemens system used in Bharat's work has the advantage of using a clinical transducer with the stepper motor as an integral part of the transducer handle that is controlled directly by the ultrasound machine (Bharat *et al.* 2008). Both systems suffer from a fixed elevational focus and mechanical steering of the beam in the elevation direction. That approach is inherently slower than a system that uses electronic steering since mechanical steering cannot utilize parallel receive beams in the elevation direction.

Other methods of acquiring a volume include using a 1.5-D array with stepping of the aperture in elevation or using a 1.75-D array (Wildes *et al.* 1997) with steering the beam in elevation. With these techniques, a few scan planes could be acquired but an appreciable volume cannot be obtained (Konofagou and Ophir 2000). Two-dimensional arrays show promise for acquiring a 3-D volume. Simulations of 1.5-D (Konofagou and Ophir 2000) and 2-D transducers (Konofagou and Ophir 2000; Chen *et al.* 2005) have been performed for various applications. Two-dimensional piezoelectric transducers have been manufactured for cardiac applications (von Ramm *et al.* 1991) but the relatively low operating frequency and small aperture of these transducers is not well suited for breast imaging. The advent of capacitive micro-machined ultrasound transducers (CMUTs) has created an opportunity to manufacture reliable, relatively low cost 2-D array systems with the ability to obtain high frequency rectilinear 3-D data quickly and easily (Oralkan 2003; Daft 2005).

Block-matching algorithms (BMA) are an efficient way to track relative motion due to the fact that the heuristic search is performed only in limited regions, with *a priori* knowledge and not the entire image (Zhu and Hall 2002; Chen and Varghese 2009). A normalized cross-correlation (NCC) metric has proven popular for motion tracking but the high computational cost makes surrogate metrics desirable. Sum-squared difference (SSD), sum-absolute difference (SAD) and other metrics

(Viola and Walker 2003) have been implemented for motion tracking with lower computational cost and higher speed image formation compared with NCC.

When presented with 3-D RF data, 1-D, 2-D or 3-D tracking can be performed to estimate motion. One-dimensional (Dotti *et al.* 1976), 2-D (Morrison *et al.* 1983; Trahey *et al.* 1987) and 3-D (Bashford and von Ramm 1996) tracking were first used in ultrasound to track blood flow as an alternate method to Doppler or to compensate for unwanted physiologic motion. One-dimensional tracking (Wilson and Robinson 1982; Bonnefous and Pesque 1986; Ophir *et al.* 1991) is an application of the simple 1-D time delay estimation problem (Carter 1987) to motion tracking. With axial deformation, 1-D motion tracking estimates the dominant motion along the acoustic beam direction. This approach ignores the fact that a stress applied to any surface of a 3-D object generally results in 3-D motion. One-dimensional tracking is accurate provided the off-axis motion is well within the width of the given acoustic beam but it suffers from significant decorrelation error (due to 3-D motion). Two-dimensional tracking (Morrison *et al.* 1983; Zhu and Hall 2002; Chen *et al.* 2009) recognizes the limitation of 1-D tracking and tracks motion within the restrictions of the data available from a 1-D array transducer. Two-dimensional tracking has demonstrated the ability to track relatively large deformations ($>2\%$ strain) in simple media (Chaturvedi *et al.* 1998). For a review of the early history of 1-D and 2-D motion tracking in elastography see Hein and O'Brien (1993).

Three-dimensional tracking has been used to create a 2-D strain image (Insana *et al.* 1997; Konofagou and Ophir 2000; Chen *et al.* 2005) that is higher in quality than can be obtained with 2-D tracking. Three-dimensional strain volumes have also been formed for deformations of approximately 0.1% (Bharat *et al.* 2008), 0.5% (Lindop *et al.* 2006) and 1% (Treece *et al.* 2008).

We present here 2-D and 3-D strain images acquired with a 2-D CMUT array and compare the performance of several different motion tracking strategies. The accuracy of the displacement estimates (*i.e.*, registration of the pre- and postdeformation RF signals) is measured by comparing cross-correlation between the predeformation and motion-compensated postdeformation RF echo signal. The quality of the resulting strain image is measured by the contrast-to-noise ratio metric. As a particularly demanding test of displacement field accuracy, the modulus distribution within the phantom was reconstructed from the measured displacement fields. Modulus inversion is sensitive to the noise and therefore is a good test for the data quality (Zhu *et al.* 2003; Oberai *et al.* 2004). The inverse elasticity problem is solved to recover the shear modulus distribution within the

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