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● *Original Contribution*

MODEL-BASED CORRECTION OF TISSUE COMPRESSION FOR TRACKED ULTRASOUND IN SOFT TISSUE IMAGE-GUIDED SURGERY

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Abstract—Acquisition of ultrasound data negatively affects image registration accuracy during image-guided therapy because of tissue compression by the probe. We present a novel compression correction method that models sub-surface tissue displacement resulting from application of a tracked probe to the tissue surface. Patient landmarks are first used to register the probe pose to pre-operative imaging. The ultrasound probe geometry is used to provide boundary conditions to a biomechanical model of the tissue. The deformation field solution of the model is inverted to non-rigidly transform the ultrasound images to an estimation of the tissue geometry before compression. Experimental results with gel phantoms indicated that the proposed method reduced the tumor margin modified Hausdorff distance (MHD) from 5.0 ± 1.6 to 1.9 ± 0.6 mm, and reduced tumor centroid alignment error from 7.6 ± 2.6 to 2.0 ± 0.9 mm. The method was applied to a clinical case and reduced the tumor margin MHD error from 5.4 ± 0.1 to 2.6 ± 0.1 mm and the centroid alignment error from 7.2 ± 0.2 to 3.5 ± 0.4 mm. (E-mail: thomas.s.pheiffer@vanderbilt.edu) **© 2014 World Federation for Ultrasound in Medicine & Biology.**

Key Words: **Image-guided surgery, Tissue deformation, Finite-element model, Biomechanical model, Boundary conditions, Image registration.**

INTRODUCTION

Ultrasound is commonly used as an intra-operative imaging modality to monitor surgical targets such as tumors. The need to maintain acoustic coupling between the probe and tissue often results in significant compression of the target by the user. This is especially a concern when using ultrasound strain imaging, in which a certain level of pre-compression of the tissue may be necessary. However, this tissue deformation affects the geometry of the scanned objects and the resulting images. Soft tissue can undergo surface compression on the order of 1 cm during routine freehand imaging (Artignan et al. 2004; Xiao et al. 2002). This leads to incorrect estimates of the size and location of landmarks within the ultrasound images.

Compressional effects from the probe are especially apparent in image-guided procedures, which align intra-operative data with pre-operative tomographic images. In these procedures, it is important that data collected during the surgery are accurately registered to high-resolution computed tomography (CT) or magnetic resonance (MR) image volumes for optimal guidance. Typically this is done by digitizing physical landmarks on the patient with a tracked instrument, selecting the corresponding landmarks in the tomograms and computing a rigid transformation that best aligns the two coordinate spaces. Although there are a variety of methods to track and calibrate an ultrasound probe such that each image slice is recorded with a known pose in physical space (Blackall et al. 2000; Boctor et al. 2006; Hsu et al. 2008a, 2008b; Mercier et al. 2005; Muratore and Galloway 2001), the usefulness of tracked ultrasound relies on an accurate registration. Registration accuracy is compromised by non-rigid tissue deformation such as that which occurs with manipulation of the ultrasound probe. The goal of this work was to improve the usefulness of tracked ultrasound in image-guided procedures by improving this registration.

There are several approaches in the literature that have sought to address the problem of tissue deformation

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exerted by an ultrasound probe. One method is to create a digital representation of the surface and then use a combination of Bayesian theory and prior knowledge of the surgical scene to create a deformation that matches the observed ultrasound data (King et al. 2000), but this approach did not incorporate a physical model of tissue which could be used to provide more realistic priors. Another approach is to acquire B-mode or raw radiofrequency data from the ultrasound and use non-rigid image-based registration and positional tracking to correct for deformation (Treece et al. 2005; Xiao et al. 2002), but this approach requires a series of ultrasound images to provide sequential estimates of compression correction. There has also been work done to model tissue compression using data from a force transducer attached to the ultrasound probe along with a position sensor to drive a tissue model (Burcher et al. 2001; Sun et al. 2010). Our proposed method is similar to this method, but eliminates the need for a force measurement apparatus on the probe by using measured 3-D surface displacements, rather than force, to drive the model. Our method uses just the tracking system which is routinely used in surgical procedures such as image-guided neurosurgery. To our knowledge, there has not been an attempt to model the tissue deformation from the physical probe surface itself in the correction. This work presents a compression correction method that measures and compensates for this effect using a biomechanical tissue model with validation in simulations, phantoms and a preliminary clinical case.

METHODS

We present our compression correction method as one component in the context of a patient-specific data pipeline for image-guided therapy. Before correction, we perform several data acquisition and processing steps. The procedures described below were used in all phantom experiments and were similar for the acquisition and analysis of clinical data.

Phantom construction

Two compliant phantoms were each constructed by mixing 7% by mass polyvinyl alcohol (PVA) in water, 10% by volume glycerol and heating to 80°C to ensure saturation (Fromageau et al. 2007; Surry et al. 2004). For each phantom, a smaller amount of PVA was treated with barium sulfate powder for CT contrast and poured into a separate mold to act as the tumor target. The tumor was subjected to five freeze-thaw cycles in which it was frozen at –40°C for 12 h and then thawed for an additional 12 h, to produce a stiffer material. The tumor was then suspended in the bulk phantom mixture, and the phantom underwent one freeze-thaw cycle to produce a tissue-like phantom containing a stiff tumor. The volumes of the tumor and bulk phantom mixtures were 3.2 and 720 cm^3, respectively. The stiffness properties for the bulk tissue and tumor were tested using small samples with an ElectroForce 3100 instrument (Bose, Eden Prairie, MN, USA). One of the phantoms was constructed in a small cup-like container covered in fiducial markers and was used for the baseline accuracy test described under Phantom Experiments. The second phantom was fixed to a rigid base, which contained eight evenly distributed fiducial markers used in the image-to-physical registration, and was used to test the compression correction method.

Patient model from pre-operative image volume

Computed tomography image volumes of the phantoms were acquired using a clinical CT machine. These data simulated a typical pre-operative tomogram acquisition, and were defined in the experiment as the baseline un-deformed state against which our corrected ultrasound data would be compared. All volumes were 512 × 512 × 422 with 0.6-mm isotropic voxels. The phantom structures were segmented using intensity thresholding tools within Analyze 9.0 (Mayo Clinic, Rochester, MN, USA). Isosurfaces were generated from the bulk phantom and tumor segmentations *via* the marching cubes algorithm, and were smoothed using a Laplacian filter. A tetrahedral mesh was generated from the segmentation surfaces using custom-built mesh generation methods (Sullivan et al. 1997). One phantom and mesh are illustrated in Figure 1.

Intra-operative data collection

All ultrasound images were acquired with an Acuson Antares ultrasound machine (Siemens, Munich, Germany), using a VFX13-5 linear array probe with a 6 cm depth setting at 10 MHz. The ultrasound unit was also capable of producing strain images *via* the eSie Touch elasticity software. For the compression correction experiment, B-mode images were collected, as were strain images, and both types of images were analyzed to evaluate the effect of correction on target locations in ultrasound images having different contrast mechanisms. Ultrasound data were tracked in 3-D space by synchronizing the ultrasound video and tracking data using software based on the Visualization Toolkit on a host PC (Boisvert et al. 2008; Pace et al. 2009). The video was captured by a Matrox Morphis Dual card (Matrox Imaging, Dorval, QC, Canada), which recorded the analogue video output of the ultrasound machine in real time. A passive optical tracking rigid body (Northern Digital, Waterloo, ON, Canada) was fixed to the ultrasound probe, as illustrated in Figure 2. The pose of the rigid body was measured with a Polaris Spectra (Northern