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## **2pBAb5. Validation of three-dimensional strain tracking by volumetric ultrasound image correlation in a pubovisceral muscle model**

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Little is understood about the biomechanical changes leading to pelvic floor disorders such as stress urinary incontinence. In order to measure regional biomechanical properties of the pelvic floor muscles *in vivo*, a three dimensional (3D) strain tracking technique employing correlation of volumetric ultrasound images has been implemented. In this technique, local 3D displacements are determined as a function of applied stress and then converted to strain maps. To validate this approach, an *in vitro* model of the pubovisceral muscle, with a hemispherical indenter emulating the downward stress caused by intra-abdominal pressure, was constructed. Volumetric B-scan images were recorded as a function of indenter displacement while muscle strain was measured independently by a sonomicrometry system (Sonometrics). Local strains were computed by ultrasound image correlation and compared with sonomicrometry-measured strains to assess strain tracking accuracy. Image correlation by maximizing an exponential likelihood function was found more reliable than the Pearson correlation coefficient. Strain accuracy was dependent on sizes of the subvolumes used for image correlation, relative to characteristic speckle length scales of the images. Decorrelation of echo signals was mapped as a function of indenter displacement and local tissue orientation. Strain measurement accuracy was weakly related to local echo decorrelation.

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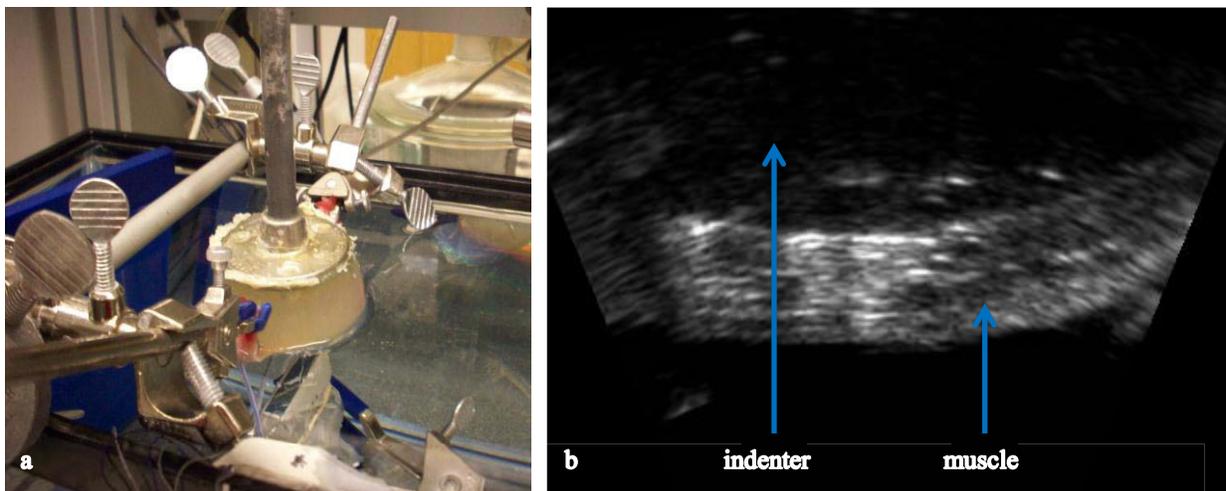
## Introduction

The pubovisceral muscle (PVM) is essential to the support of pelvic organs and maintaining continence, because it forms a sling around the urethra and vagina.<sup>1</sup> It can be damaged by childbirth and other biomechanical changes that can occur due to the aging process and excess intra-abdominal pressure.<sup>2</sup> Methods to measure and monitor these biomechanical changes are needed for early detection of pelvic floor disorders and determination of biomechanical parameters for finite element models of the pelvic floor.<sup>3</sup> One promising approach employs Bayesian texture correlation to track regional deformation and strain in three dimensional (3D) ultrasound images of the pubovisceral muscle. Preliminary efforts to validate and optimize this algorithm are described here.

Echo decorrelation imaging is a pulse-echo ultrasound approach to detect changes in ultrasound signals to monitor and guide radiofrequency ablation (RFA)<sup>4</sup> and other thermal ablation methods used in the treatment of liver cancer and other tumors. While the goal of such monitoring is to detect thermal ablation of tissue,<sup>5,6</sup> motion by the transducer or tissue can cause artifactual decorrelation between images. Here, the same measurements used to validate our strain mapping algorithm are also employed to characterize echo decorrelation measurements associated with out-of-plane transducer displacement and in-plane tissue displacement.

## Materials and Methods

The experimental setup employed is shown in Fig. 1 a. Within a water tank, a model of the PVM was created using bovine skeletal muscle wrapped in a sling around a gelatin indenter, with size and shape constructed to mimic the human pelvic floor geometry measured by Dietz.<sup>7</sup> The gelatin material was designed with proportional n-propanol and formaldehyde content to achieve sound speed and stiffness similar to human soft tissue as described by Hall.<sup>8</sup> Stress was applied to the muscle tissue by controlled displacement of the indenter, simulating intra-abdominal pressure on the PVM. After each indentation was made, a group of parallel B-scan images, comprising a single volumetric image, were recorded at regular spatial intervals. An ultrasound imaging transducer was held 2-3 cm below the muscle and displaced in the horizontal plane by a stepping motor system (Velmex, Bloomfield, NY, USA). Volumetric ultrasound images were obtained by recording parallel two dimensional (2D) ultrasound images while stepping the transducer in set intervals. An example ultrasound image is shown in Fig. 1 b.



**FIGURE 1.** (a) Study setup showing muscle slung around the gelatin indenter, with the imaging array held below. (b) Ultrasound image of the muscle and indenter, obtained using the Voyager imaging system with a 10 MHz probe. The area shown is 50 mm × 36 mm.

### *Texture correlation*

Sonomicrometry sensors (Sonometrics, London, ON, Canada) were sutured onto the muscle to provide independent strain measurements for validation of the strain tracking algorithm. These sensors use 1.5 MHz ultrasonic signals to record the distance between sensors with a precision of 0.013 mm. The muscle, with the sonomicrometry sensors attached, was slung around the indenter and held in place by clamps. Images were obtained using a portable Voyager imaging system (Arden Sound, Mesa, AZ, USA) with a 10-MHz, mechanically scanned two-element

transducer. B-scans were obtained at a spacing of 0.08 mm between parallel planes. Muscle was displaced by the indenter in 7 steps of 0.2 mm each. B-scan images were exported in DICOM format and cropped for analysis by the texture correlation algorithm.

The strain tracking algorithm used here is a 3D extension of the 2D Bayesian mapping method described by Clocksin.<sup>9</sup> A subvolume of voxels in an undeformed image centered on a chosen point is mapped to a location in a deformed image by maximizing the sum of a prior probability and a likelihood expression:

$$\hat{\theta} = \operatorname{argmax}(\beta \cos \frac{\sqrt{u^2+v^2+w^2}}{\pi s} + \sum \exp(-\frac{d_i^2}{\alpha^2})), \quad (1)$$

where  $\hat{\theta}$  is the position of the displaced subvolume,  $\beta$  is a user-defined scale factor to adjust the effect of the prior probability,  $u$ ,  $v$ , and  $w$  are displacements in the  $x$ ,  $y$ , and  $z$  directions respectively,  $s$  is the radius of the search region,  $d_i$  is the difference between the intensity values of the two images for voxel index  $i$ , and  $\alpha$  is a user-defined parameter that adjusts steepness of the likelihood map. The cosine prior probability term weights smaller displacements as more likely than larger displacements. The exponential-sum likelihood term quantifies similarity in texture between the two image subvolumes. This process is repeated for a sequence of 3D ultrasound images obtained under progressively increasing, externally applied stress. Large 3D deformations are tracked as the sum of a series of smaller displacements. This algorithm has been implemented by custom software written in MATLAB (Mathworks, Natick, MA, USA). Images were analyzed using this method with cubic subvolumes with side lengths ranging from 4 to 6 mm and  $\alpha$  values equal to 0.1, 1, and 100. For comparison, images were also analyzed using subvolume displacements estimated from the Pearson correlation coefficient.

#### *Echo decorrelation*

Using the same PVM model, images were taken with a 7 MHz linear array with a z.one imaging system (Zonare Medical Systems, Mountain View, CA, USA), with an interval of 0.1875 mm between B-scan image planes. Muscle was displaced by the indenter in nine steps of 1 mm each, followed by five more steps of 0.5 mm each for a total displacement of 12 mm.

Complex, beamformed echo signals were recorded as IQ data by the z.one system and used to compute echo decorrelation images<sup>5,6</sup> in MATLAB. Echo decorrelation was computed within a region of interest (ROI) in the muscle tissue, using a Gaussian spatial window with width parameter 1 mm. To measure echo decorrelation caused by out-of-plane transducer motion, position-dependent decorrelation was computed between images separated in the out-of-plane direction by 0.1875 mm (one step) to 2.25 mm (twelve steps). Average decorrelation within the ROI was then computed as a function of image separation distance. To measure echo decorrelation caused by in-plane tissue motion, average decorrelation within the ROI was computed between images from the middle of the sample at each indenter displacement step and averaged for displacements of 0.5 and 1.0 mm.

## **Results**

For validation of the strain tracking algorithm, the engineering strain (relative change in length) measurement between sonomicrometry sensors was compared to the engineering strain between two image points corresponding to muscle in the vicinity of each sensor. Fig. 2 a illustrates the effect of the subvolume side length, the likelihood parameter  $\alpha$  from Eq. (1), and the correlation type on the relative strain error.

The effect of out-of-plane displacement on echo decorrelation is illustrated in Fig. 2 b. Decorrelation increases monotonically up to a separation distance of about 1.8 mm, comparable to the image slice thickness for the 7 MHz linear array employed. For in-plane displacements of 0.5 mm and 1.0 mm, echo decorrelation was also substantial, with an average decorrelation of 0.83 for 0.5 mm displacement and 0.87 for 1.0 mm displacement.

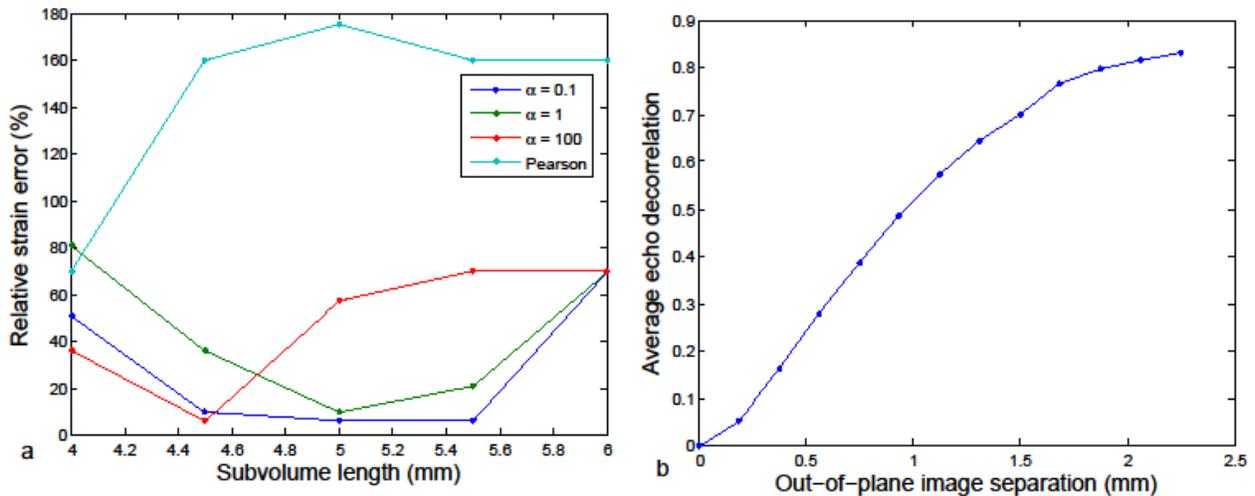


FIGURE 2. (a) Relative strain error for texture correlation analysis vs. sonomicrometry measurements as a function of subvolume side length. (b) Average echo decorrelation within ROI as a function of out-of-plane image separation.

## Discussion

The results shown in Fig. 1 suggest that more accurate strain tracking is obtained using the Bayesian estimator from Eq. (1) with a small likelihood parameter  $\alpha$ . Larger  $\alpha$  values effectively smooth the likelihood map to reduce the possibility of erroneously selecting local maxima. However, smoothing of the likelihood map also results in less precise mapping. Strain tracking based on the Pearson correlation coefficient consistently performed worse than the exponential likelihood function of Eq. (1). Inspection of image data indicated that strain tracking using the Pearson correlation coefficient sometimes resulted in matching of subvolumes with similar speckle texture but different overall brightness, while the exponential likelihood function matched only subvolumes with both similar texture and brightness. Strain tracking accuracy varied with subvolume side length, with a length of 4.5-5.5 mm providing best results for the data shown in Fig. (1). Subvolume size may be further optimized based on the correlation length of the speckle pattern in each direction.

Echo decorrelation values increased with both in-plane and out-of-plane displacement, as expected. For the results shown in Fig. 2, echo decorrelation on the order of 0.1-0.2, comparable to that observed for thermal ablation *in vivo*,<sup>6</sup> was observed for out-of-plane displacements on the order of 0.3-0.5 mm. Thus, when thermal ablation is monitored using B-scans recorded at 50 frames per second, echo decorrelation artifacts will be comparable to ablation-induced decorrelation when the tissue velocity exceeds about 15 mm/s in the out-of-plane direction. Similar comparisons for in-plane motion will require computation of echo decorrelation for smaller indenter displacements.

## ACKNOWLEDGMENTS

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