

Deformable model for 3D intramodal nonrigid breast image registration with fiducial skin markers

Mehmet Z. Unlu¹, Andrzej Krol^{2,1,3}, Ioana L. Coman^{4,1,2}, James A. Mandel⁵, Karl G. Baum⁶, Wei Lee¹, Edward D. Lipson^{3,1,2}, David H. Feiglin²

¹Department of Electrical Engineering and Computer Science, Syracuse University

²Department of Radiology, SUNY Upstate Medical University

³Department of Physics, Syracuse University

⁴Department of Mathematics and Computer Science, Ithaca College

⁵Department of Civil and Environmental Engineering, Syracuse University

⁶Department of Imaging Science, Rochester Institute of Technology

ABSTRACT

We implemented a new approach to intramodal non-rigid 3D breast image registration. Our method uses fiducial skin markers (FSM) placed on the breast surface. After determining the displacements of FSM, finite element method (FEM) is used to distribute the markers' displacements linearly over the entire breast volume using the analogy between the orthogonal components of the displacement field and a steady state heat transfer (SSHT). It is valid because the displacement field in x, y and z direction and a SSHT problem can both be modeled using Laplace's equation and the displacements are analogous to temperature differences in SSHT. It can be solved via standard heat conduction FEM software with arbitrary conductivity of surface elements significantly higher than that of volume elements. After determining the displacements of the mesh nodes over the entire breast volume, moving breast volume is registered to target breast volume using an image warping algorithm. Very good quality of the registration was obtained. Following similarity measurements were estimated: Normalized Mutual Information (NMI), Normalized Correlation Coefficient (NCC) and Sum of Absolute Valued Differences (SAVD). We also compared our method with rigid registration technique.

Keywords: Intramodal image registration, finite element method, deformable breast model, fiducial skin markers.

1. INTRODUCTION

Some medical breast imaging applications such as image guided surgery, detection of the breast cancer using contrast agents and progress observation of the breast cancer requires more than one MRI scans. Any motion such as patient movement, respiratory or cardiac changes between the scans or using different scanners may cause non-rigid deformation on the breast volume. This makes difficult diagnosing, observing and treatment of the breast cancer. Therefore, we developed a finite element method (FEM) deformable breast model to correct motion artifacts between the scans and extract the corrected uptake/washout information. Finite-element method (FEM) deformable tissue models have been applied to predict mechanical deformations of tissues or organs based on biomechanical tissue properties including brain shift modeling [1], heart kinetics modeling [2], breast compression simulation, such in x-ray mammography [3] and breast image registration [4-6]. However, physically-based deformable breast models are very difficult to implement due to complex and patient-specific breast morphology and highly nonlinear (hyperplastic) and difficult-to-measure elastic properties of different types of tissues in the breast, as well as explicitly unknown boundary conditions [5]. Since physically based models are difficult to implement, we developed a method, which doesn't require any patient-specific elastic breast data for nonrigid intramodal breast-image registration. Our model is applicable only for small changes in stress conditions in the imaged breast during dynamic scans. Large breast displacements are prevented by careful patient prone positioning and by use of a breast-support device.

2. MATERIALS AND METHODS

2.1. MRI

Gradient echo MRI dynamic sequences (TE/TR=5.4/2.1, field of view: 360mm x 360mm, reconstruction in 512x249 matrix) with patients positioned prone with both breasts suspended into a single well housing the receiver coil were acquired on a few patients and a healthy volunteer using Philips 1.5 T Intera MR system. Nine fiducial skin markers made of polyethylene tubes; 2 mm I.D. and 4 mm in length filled with diluted Magnevist were used for each breast. I/V line was placed in antecubital vein (22 or 20 gage needle) in the contra lateral side to the breast with suspicious lesion. Gd-DTPA (Magnevist; Schering AG) was delivered (0.15 mmol/kg) at constant flow 1.5 ml/s directly followed by 20 ml physiologic saline solution after Pre-Gd scan was acquired. Five scans at 90s intervals were acquired. Our model estimates the displacement field for any location in the breast volume from observed displacements of the external fiducial markers. The patient-specific geometry of the breast is obtained from moving MRI scan segmenting out the breast surface using ImageJ¹. Meshing is performed using ANSYS, ver. 5.7, software package.

2.2. Finite Element Method (FEM)

Finite element method (FEM) [7, 8] is an approximate method to solve partial differential equations by dividing a continuous domain into elements and constructing basis functions across the elements. The finite elements are connected to each other at discrete locations called nodes on the surface of the finite elements. In this manner, a piecewise approximation of the real problem is defined. Theorems have been proven that define the necessary conditions to insure convergence of finite element solutions to the solution of the real problem, as the density of the finite elements is increased.

The basic steps in the finite element method are:

First, the domain of the problem is divided into finite elements. Second, Interpolation functions are selected to define the variation of the field variable over the domain of the finite element in terms of the nodal values of the field variable. Polynomials are normally chosen for interpolation functions.

There exist a number of commercial packages dedicated to solving FEM problem. In our work we employed ANSYS, version 5.7².

3. DEFORMABLE FEM BREAST MODEL

We have developed and implemented an imaging strategy and a suitable FEM model for nonrigid registration of dynamic MR breast scans. This approach does not require information on patient-specific breast morphology and elastic tissue properties. However, it can be applied only if the stress conditions in the imaged breast are virtually the same between dynamic MRI scans. This is accomplished by use of identical patient support and positioning systems on all scans. Under these conditions, the observed intramodal displacements, after rigid alignment of MRI scans, are predominantly due to underlying biological and physical differences in the imaging process and in the reconstruction algorithm used, including differences in the scanners spatial distortion and resolution, and signal-to-noise ratio [9, 10]. Our model compensates for these dissimilarities. In addition, it compensates for small discrepancies in patient positioning and for minor displacements resulting from physiological and other motion. The model can be classified as a point-based registration method and requires a small number of non-invasive fiducial skin markers visible in all MRI scans placed on the surface of the examined breast.

3.1. Fiducial skin markers and their localization

The fiducial markers visible in MR are made of polyethylene tubes, 2 mm I.D. and 4 mm in length, filled with diluted Magnevist and sealed. Nine markers per breast were used. They were attached using medical adhesive tape to patient's breasts prior to MRI scans forming two rings (4 markers each); one at the breast base, one at the midplane, and one on the nipple. The MRI image acquired prior to Gd-DTPA administration is considered a fixed image, while the series of images acquired on 90 seconds interval post Gd-DTPA administration are considered target images. The corresponding pairs of markers in the fixed and target images were manually defined. The localization of markers was performed by calculating an intensity-based centroids using the method described by Wang et al [11]. In this iterative knowledge-

¹ <http://rsb.info.nih.gov/ij/>

² <http://www.ansys.com/>

based method, a user provides information on the spatial extent of a marker and identifies a voxel belonging to a selected marker and then the algorithm finds a set of voxels connected in 3D to a selected voxel that form an object with proper geometrical properties to be a marker.

3.2. FEM model construction

In the framework of 3D FEM, the breast volume and its surface are discretized (meshed) by a set of finite elements (tetrahedrons for the volume and triangles for the surface) connected through nodes located on element boundaries. First, the discrete values of displacement vectors are estimated for each corresponding pair of fiducial skin markers located on target and on moving images. A dense displacement field is then obtained using FEM, by first distributing linearly the Cartesian components of the fiducial displacement vectors over the breast surface and then throughout its volume. An analogy between Cartesian components of the displacement field and the temperature differences in steady-state heat transfer (SSHT) in solids can be used and a unique FEM solution for displacement vectors at each node can be obtained via standard heat conduction FEM software. Such displacement interpolant can be described by the Laplace-Poisson equation as follows

$$\nabla \cdot (\kappa \nabla T) = -q \tag{3.1}$$

where κ is the thermal conductivity, T is the temperature, and q is the internal heat generation rate. In this approach, the Cartesian components of the observed displacements of fiducial skin markers are considered to be equal to the SSHT temperature loads and consequently the displacement field components u_x, u_y, u_z are mathematically equivalent to temperature differences in the SSHT problem and are all distributed piecewise linearly over the breast spatial domain. Accordingly, for steady-state conditions 3.1 can be rewritten as follows

$$\nabla \cdot (\kappa \nabla u_j) = 0 \quad j=x, y, z \tag{3.2}$$

where κ is the pseudoconductivity and $\kappa_{surface} = 1000\kappa_{volume}$. To perform calculations the ANSYS (ver. 5.7) is used.

3.3. Finite element mesh generation

In the first step, patient-specific geometry of the breast is obtained from target image segmentation via thresholding the breast surface using ImageJ software. In the second step, the breast meshing is performed using the ANSYS package (Figures 3.2.a and b). The following elements were chosen from the ANSYS library: SHELL57 (2-D Thermal Shell) for the breast surface and SOLID70 (Tetrahedral Thermal Solid) for the bulk of the breast volume (Figures 3.1.a and b).

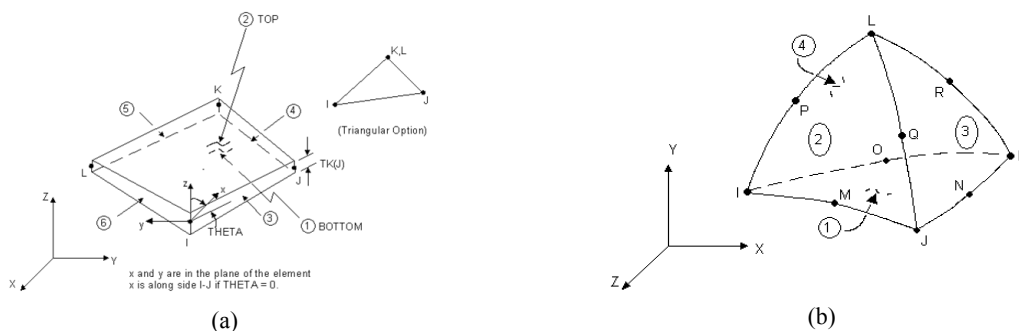


Figure 3.1. Thermal mesh elements (a) SHELL57 2-D Thermal shell mesh element. (b) SOLID70 3-D Tetrahedral thermal solid mesh element.³

³ Reprinted from ANSYS ver. 5.7. manual with permission.

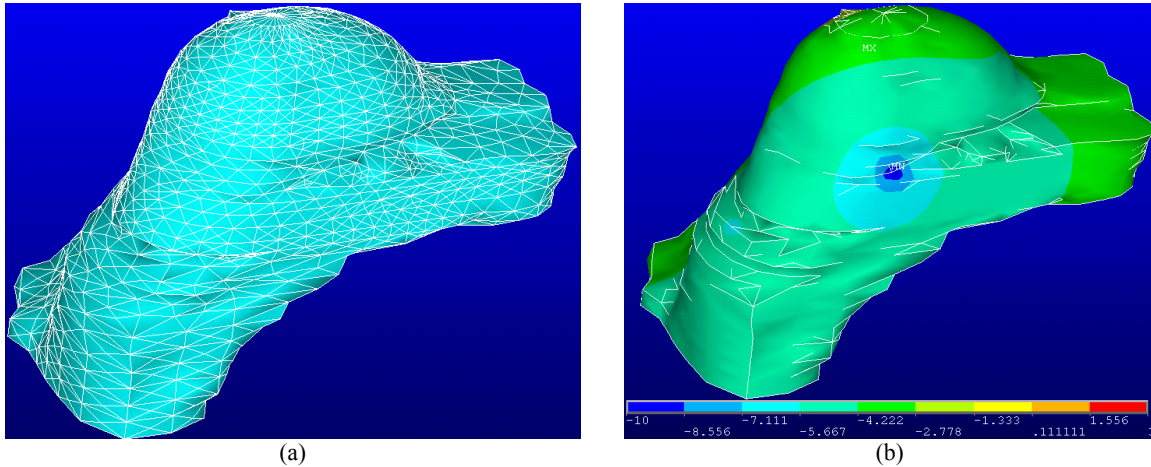


Figure 3.2. (a) Meshed breast surface and volume using ANSYS package. (b) Distribution of the marker displacements over the entire breast surface and volume.

3.4. FEM displacements and interpolation of FEM solutions

The SSHT FEM produces a displacement vector at each FEM node. Displacement vectors for each location (voxel or subvoxel) within an FEM element can be interpolated using a weighted sum of the element’s nodal displacements with the weights equal to the element’s node shape function [7],

$$\bar{u} = \sum_{i=1}^{N_{nodes}} N_i^{el} \bar{u}_i^{el} \quad (3.3)$$

where N_{nodes} is the number of nodes in the element, N_i^{el} is the element’s node shape function, and u_i^{el} is the nodal displacement vector. The exact FEM interpolation given by 3.3 was used to obtain dense displacement field within each FEM element, which in turn was used to interpolate the image gray values via a truncated sinc interpolation kernel. We call this process warping of the floating (moving) image to the fixed (target) image. We emphasize that only the voxels inside the FEM mesh are affected by this kind of processing.

4. RESULTS

We have acquired MRI dynamic clinical sequences with fiducial skin markers on a few patients and a healthy volunteer using the protocols and the processing methodology described above. We have applied our SSHT FEM registration technique to perform intramodal MRI/MRI image registration in 3D. We first segmented out the surface of the moving volume via simple thresholding with good definition of the breast surface. Segmented breast surface was used to create FEM mesh with the ANSYS package. The geometric centroids of fiducial markers visible in both MRI scans were estimated using the iterative method described above, and the discrete displacement field was obtained. These data allowed our FEM model to obtain a dense displacement field (i.e. a displacement vector for each mesh node), which in turn was used to warp the post-contrast MRI image to the pre-contrast MRI image, to obtain MRI differential dynamic series images

We registered dynamic MRI scans and obtained differential dynamic image series (preGd-postGd1). The examples of obtained images are shown in Fig. 4.1. Our method of registration yielded excellent differential image series that clearly revealed lesions (Figure 4.1.a,b and c) that were not visible or difficult to discern in the unregistered differential image series (Figure 4.1.d,e and f). Further, it produced clinically useful maximum intensity projection (MIP) 3D images, in contrast to MIPs provided by the unregistered dynamic series (Figures 4.2.a and b).

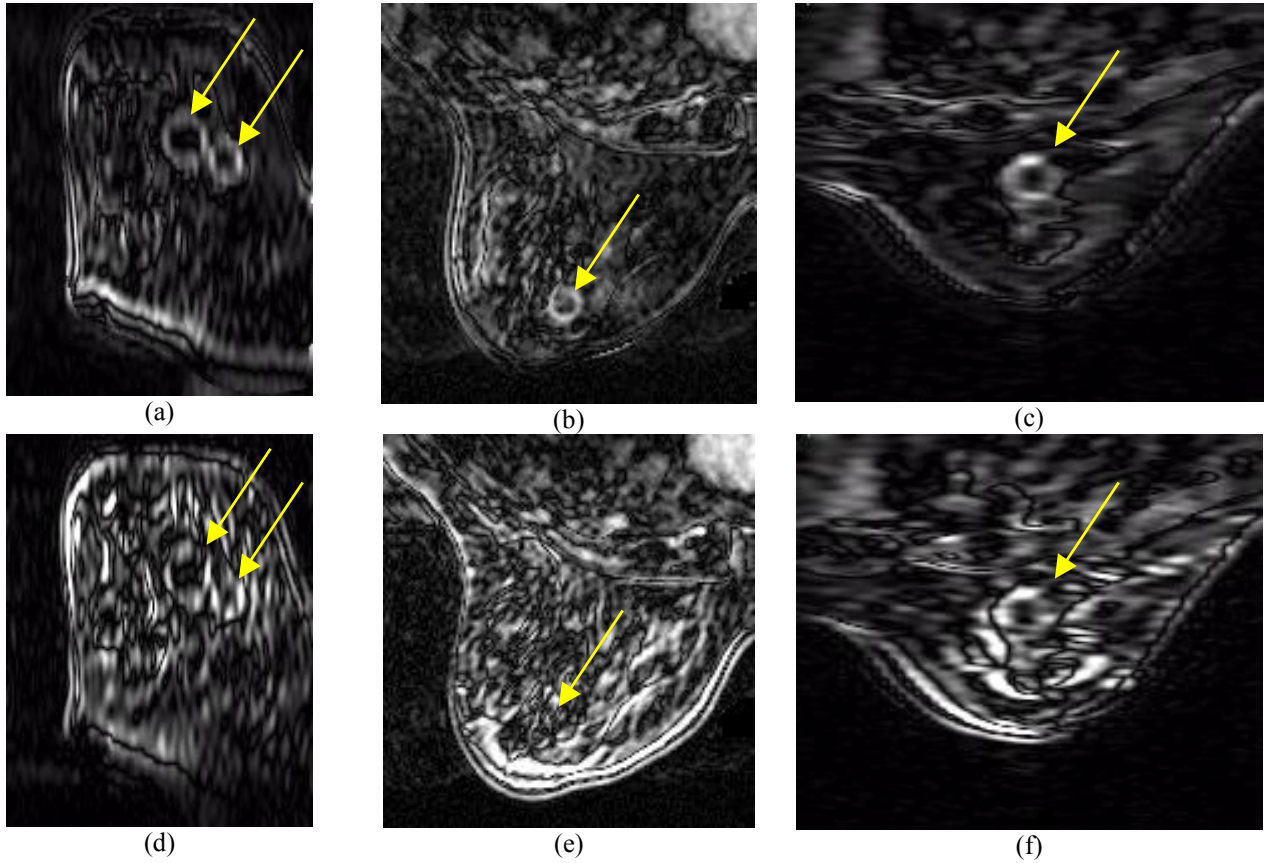


Figure 4.1. MRI differential dynamic image series (preGd-Gd1) using gradient echo (GRE) sequence. (a) coronal; (b) transaxial, (c) sagittal views of the left breast after SSHT FEM image registration and warping. (d) coronal; (e) transaxial; (f) sagittal views of left breast before SSHT FEM registration and warping. Note that the lesions are invisible or difficult to discern before SSHT FEM registration. Arrows indicate the lesions locations.

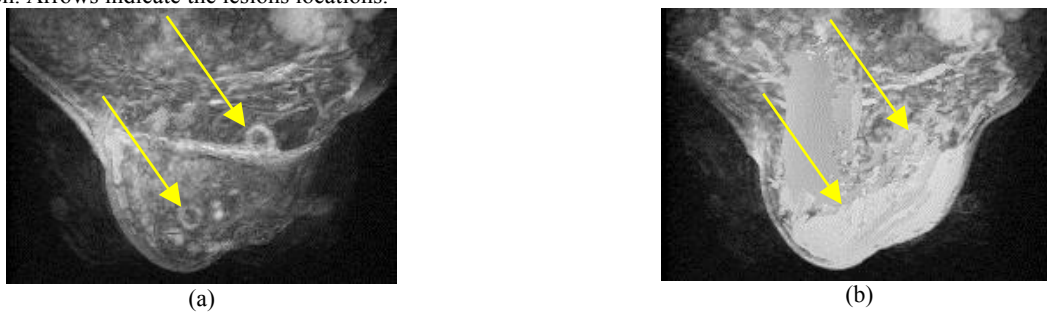


Figure 4.2. Maximum intensity projections (MIP) of MRI differential dynamic image series (preGd-Gd1) of the left breast using GRE sequence: (a) MIP at 190° after SSHT FEM, (b) MIP at 190° before SSHT FEM.

In addition, for the same patient as shown in Figs. 4.1. and 4.2., we estimated various similarity measures, which demonstrate improvement after SSHT FEM registration, (Tables 4.1 and 4.2).

Table 4.1. Image similarity measures used [12, 13]

Similarity Measure	Formula
Mutual Information	$MI(T) = H(I_1) + H(I_2, T) - H(I_1, I_2, T)$
Normalized Mutual Information	$NMI(T) = \frac{H(I_1) + H(I_2, T)}{H(I_1, I_2, T)}$
Normalized Correlation Coefficient	$NCC(T) = \frac{\sum_{x \in I_1} (I_1(x) - I_1^*) \cdot (I_2(T(x)) - I_2^*)}{\left[\sum_{x \in I_1} (I_1(x) - I_1^*)^2 \cdot \sum_{x \in I_2} (I_2(T(x)) - I_2^*)^2 \right]^{\frac{1}{2}}}$
Sum of Absolute Valued Differences	$SAVD(T) = \sum_{x \in I_1} I_1(x) - I_2(T(x)) $
<i>I₁</i> : Target image, <i>I₂</i> : Moving image, <i>I₁[*]</i> : Mean intensity value of target image, <i>I₂[*]</i> : Mean intensity value of moving image <i>H(I_i)</i> : Shannon entropy of the image <i>I</i>	

Table 4.2. Calculated image similarity measures

MI before	MI after	NMI before	NMI after	NCC before	NCC after	SAVD before	SAVD after
1.0945	1.6580	1.1529	1.3868	0.9231	0.9528	1.61xE7	1.27xE7
Best cases: NMI: 2.0, NCC: 1.0, SAVD: 0							

5. CONCLUSIONS

We have developed a deformable breast model for nonrigid registration of intramodal breast images. This model requires fiducial skin markers placed on the breast surface. It also requires careful patient prone positioning to make sure that stress conditions are not changed between the scans. The experimental results show that this model corrects most of the motion artifacts between the scans and thus would allow extracting the corrected uptake/washout information. The similarity measurements also demonstrate improvement, as compared to rigid registration only. However, in some of the differential images small misregistration artifacts are still visible even after application of our method. This could be caused by errors in segmentation of the breast surface. Presently, we are implementing improved segmentation and meshing process and applying our method to intermodal MRI/PET breast image registration [14, 15].

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