Finite-element modeling of compression and gravity on a population of breast phantoms for multimodality imaging simulation

Gregory M. Sturgeon^{a)}

Carl E. Ravin Advanced Imaging Laboratories, Department of Radiology, Duke University Medical Center, Durham, North Carolina 27705

Nooshin Kiarashi

Carl E. Ravin Advanced Imaging Laboratories, Department of Radiology, Duke University Medical Center, Durham, North Carolina 27705 and Department of Electrical and Computer Engineering, Duke University, Durham, North Carolina 27708

Joseph Y. Lo

Carl E. Ravin Advanced Imaging Laboratories, Department of Radiology, Duke University Medical Center, Durham, North Carolina 27705; Department of Electrical and Computer Engineering, Duke University, Durham, North Carolina 27708; Medical Physics Graduate Program, Duke University, Durham, North Carolina 27705; and Department of Biomedical Engineering, Duke University, Durham, North Carolina 27708

E. Samei

Carl E. Ravin Advanced Imaging Laboratories, Department of Radiology, Duke University Medical Center, Durham, North Carolina 27705; Department of Electrical and Computer Engineering, Duke University, Durham, North Carolina 27708; Medical Physics Graduate Program, Duke University, Durham, North Carolina 27705; Department of Biomedical Engineering, Duke University, Durham, North Carolina 27708; and Department of Physics, Duke University, Durham, North Carolina 27708

W. P. Segars

Carl E. Ravin Advanced Imaging Laboratories, Department of Radiology, Duke University Medical Center, Durham, North Carolina 27705; Department of Electrical and Computer Engineering, Duke University, Durham, North Carolina 27708; and Medical Physics Graduate Program, Duke University, Durham, North Carolina 27705

(Received 22 July 2015; revised 10 March 2016; accepted for publication 21 March 2016; published 5 April 2016)

Purpose: The authors are developing a series of computational breast phantoms based on breast CT data for imaging research. In this work, the authors develop a program that will allow a user to alter the phantoms to simulate the effect of gravity and compression of the breast (craniocaudal or mediolateral oblique) making the phantoms applicable to multimodality imaging.

Methods: This application utilizes a template finite-element (FE) breast model that can be applied to their presegmented voxelized breast phantoms. The FE model is automatically fit to the geometry of a given breast phantom, and the material properties of each element are set based on the segmented voxels contained within the element. The loading and boundary conditions, which include gravity, are then assigned based on a user-defined position and compression. The effect of applying these loads to the breast is computed using a multistage contact analysis in FEBio, a freely available and well-validated FE software package specifically designed for biomedical applications. The resulting deformation of the breast is then applied to a boundary mesh representation of the phantom that can be used for simulating medical images. An efficient script performs the above actions seamlessly. The user only needs to specify which voxelized breast phantom to use, the compressed thickness, and orientation of the breast.

Results: The authors utilized their FE application to simulate compressed states of the breast indicative of mammography and tomosynthesis. Gravity and compression were simulated on example phantoms and used to generate mammograms in the craniocaudal or mediolateral oblique views. The simulated mammograms show a high degree of realism illustrating the utility of the FE method in simulating imaging data of repositioned and compressed breasts.

Conclusions: The breast phantoms and the compression software can become a useful resource to the breast imaging research community. These phantoms can then be used to evaluate and compare imaging modalities that involve different positioning and compression of the breast. © 2016 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4945275>]

Key words: finite-element, compression, breast, phantom, mammography

1. INTRODUCTION

1.A. Breast phantoms

Computational breast phantoms are finding a wide use in breast imaging research to evaluate and improve new imaging systems and techniques.^{1–4} They provide a practical means to perform studies that would otherwise be difficult or impossible with human subjects due to the high cost or risks to patient safety. For these phantoms to be truly useful in optimizing and comparing different modalities, it is essential to be able to simulate the breast in the various positions and states of compression that the modalities require. For example, in dedicated breast CT (bCT), the breast is imaged in the prone position without any compression, while in mammography and tomosynthesis, the patient is standing and the breast is compressed.

We are developing a series of over two hundred realistic and detailed 3D computational breast phantoms based on high-resolution bCT datasets (250 μm pixel width and 500 μm slice thickness) from normal breasts obtained from the work of Dr. John Boone of UC Davis.^{1,2,4} The breast volumes underwent noise reduction via 3D bilateral filtering, voxel intensity normalization, and 3D nonuniformity correction before conversion to isotropic voxels. Using a bias-corrected fuzzy C-means algorithm,⁵ the bCT datasets were segmented into six materials representing skin, adipose, and varying levels of fibroglandular tissue as seen in Fig. 1. Since they are based upon patient imaging data, the phantoms provide a realistic representation of the breast anatomy. However, the breasts in our population were imaged in an uncompressed pendant state. In order for them to be applicable to imaging modalities other than bCT, a method is needed to reposition them and simulate varying levels of compression.

In contrast to the patient-based phantoms, other breast phantoms^{3,6} utilize rule-based region growing algorithms to fill a hemiellipsoidal volume with compartments of adipose and fibrous materials. The mathematical nature of these phantoms allows for the generation of large ensembles of phantoms. Other investigators have extended these rule-based phantoms to contain the power law noise characteristics of mammograms.⁷ However, they lack the realistic 3D geometry inherent to patient based phantoms.

1.B. Finite-element (FE) breast models

The FE method provides a framework to apply and solve differential equations to a complex domain by discretizing it into a system of smaller simpler elements. The FE method

has been used extensively to model breast deformations and compression, including use for breast augmentation,^{8,9} multi-modality registration,^{10–18} image-guided surgery,^{19–21} and tumor tracking.^{22,23} Several studies have tracked landmarks to within 5 mm.^{15,24} Other studies have also included the effects of gravity.^{9,21,22,25}

Current state-of-the-art methods often optimize the initial breast position¹⁶ and/or material parameters^{16–18,20} through a FE-registration framework. Registrations, between breast MR images at different levels of compression, have also been used to provide boundary constraints to define the displacement of the region posterior to the compression plates in a corresponding FE model.²⁴ Some of these FE methods also model the stress-free reference state to account for the unknown distribution of the initially loaded state.^{26,27}

Our methodology is more aligned with earlier generation breast finite-element models, with a particular emphasis on ease of use and model robustness. Corresponding mammograms or compressed image data were not available for the bCT dataset used in our study. So, we were unable to optimize the material parameters or use ground truth displacements as boundary constraints and thus used material properties¹⁸ and boundary constraints from the literature. We also choose to neglect the initial (unknown) stress present in the prone breast. These assumptions were similar to other breast FE models.^{28,29}

1.C. Finite-element modeling for deformable breast phantoms

There are a multitude of papers dealing with breast FE models. However, there are relatively few applying these methods to create a library or database of anthropomorphic breast phantoms. This limited adoption to breast phantoms may be, in part, due to a perceived learning curve to creating FE breast models.

O'Connor *et al.* imaged mastectomy specimens in compressed and uncompressed states, using bCT, to avoid the need to model this deformation computationally.^{30,31} Kellner *et al.* used a simplified FE scheme which essentially used the voxels of down-sampled versions of the bCT as the finite elements.²⁸ In the Bakic breast phantoms, FE compression was applied using a rectangular slice approximation⁶ or as a homogeneous isotropic material.³

Our application extends a previous nonlinear contact analysis FE model,²⁹ out of our lab, to include gravity and automate the FE mesh creation. The previous method involved an interactive FE mesh generation process including mesh decimation and multiple remeshing steps to improve element quality. This

FIG. 1. Example breast phantoms with six segmentation classes shown as different grayscale levels.

was done, however, using the proprietary mesh generation tools and FE modeling solvers, Hypermesh (Hypermesh 10, Altair Engineering, Inc., Troy, MI), and LS-DYNA (Livermore Software Technology Corp., Livermore, CA). Such tools are very expensive and cannot be distributed freely to the research community.

Here we provide a user-friendly, scriptable application with a simplified FE model to ease/eliminate this learning curve and provide useful deformable breast phantoms to the imaging community. This will enable researchers to deform our library of voxelized breast phantoms (or their own breast phantoms) to create compressed versions at a desired thickness in a CC or MLO view. The results can be represented as either boundary surface representations of the phantoms or in a voxelized format.

Our FE application uses the freely available (for noncommercial use) software FEBio (Musculoskeletal Research Laboratories, University of Utah, Salt Lake City, UT).³² The application and our breast phantom library will be available for distribution to researchers and will enable simulation, optimization, and comparative studies for mammography and tomosynthesis.

2. METHODS

This work used the FE method to deform breast phantom models originally created from an uncompressed pendant breast geometry to account for the effect of gravity with changed patient position and compression. Figure 2 provides an overview of our FE application and how it interacts with our phantoms. Detailed descriptions of the various steps are provided below.

2.A. Creation of the FE model for a given breast phantom

The internal fibroglandular structure of the breast, and hence our phantoms, is geometrically complex and varied from patient to patient. In our method, we have chosen to fit elements to the outer skin surface only. A geometric smoothing scheme³³ was used to optimize the quality of the internal elements. Then the material properties of each element were set based on the median composition of the segmented voxels contained in each element. This allowed for a known hexahedral element scheme to be automatically fit to a wide range of geometries.

FIG. 2. Overview of the steps to set up and solve for the compressed breast geometry using finite-element analysis. Once the FE analysis is complete, the deformation is applied to the phantom to reposition and compress the breast.

The alternative approach of fitting tetrahedral finite elements to the complex fibroglandular structure of the breast is also a reasonable approach that has been used by a number of researchers.^{9,12,22,24,29} However, since the intended audience of our FE application may not have a background in biomechanics, we chose to not burden the end users with the need to create or potentially troubleshoot these meshes.

2.A.1. Model fitting

We created a generic/template hexahedral FE model capable of fitting the different phantoms. Hexahedral (brick shaped) elements are better suited to large deformation modeling, as experienced during mammographic compression, than the tetrahedral (pyramid shaped) elements used in our previous study.²⁹ The template consisted of 59 904 hexahedral elements with additional 32 elements being used to fit the nipple region when indicated by the user. A relatively large number of elements were used in order to achieve a mean edge length of 2.0 mm.

Given a user-specified breast phantom, the elements were automatically fit to the outer skin surface of the breast. The fitting began by aligning the nipple and then fitting the chest wall edge based on a radial sampling. The exterior nodes of the FE model were then projected to the skin surface mesh of the phantom using a nearest neighbor projection. Next, the internal elements were updated based on a thin plate splines deformation. Finally, the element quality was improved following the geometric smoothing scheme of Vartziotis and Wipper, which regularizes the dual octahedron of the hexahedral mesh.³³

The nearest neighbor projection and element quality improvement are applied iteratively. Final iterations of the fitting process replace the nearest neighbor projection with a slower and more accurate projection to the implicit surface (defined from a Euclidean distance map of the segmented images) utilizing a Newton–Raphson method.

2.A.2. Material properties

To represent the internal structures of a given phantom, the material properties of each element were set based on the classification of the voxels contained in the element. The glandular and adipose material properties were interpolated into five levels corresponding to the adipose and fibroglandular levels in the segmented phantom. The skin was its own class in the segmentation, with its own material properties, which was sometimes also used to represent the densest glandular tissue depending on the results from the fuzzy-*C*-means classification. The material of each element was set based on the median fibroglandular level for the voxels contained in the element.

The material properties were assigned using moduli of elasticity (E_{skin} 88 kPa, $E_{\text{glandular}}$ 10 kPa, E_{adipose} 1 kPa) which have been previously used for breast FE simulations^{28,29,34,35} and a nearly incompressible Poisson's ratio of 0.49.

These properties were used to define a hyperelastic Neo-Hookean material with a nonlinear stress–strain relationship

as

$$W = \frac{\mu}{2}(I_1 - 3) - \mu \ln J + \frac{\lambda}{2}(\ln J)^2, \quad (1)$$

where I_1 is the first invariant of the right Cauchy–Green deformation tensor and J is the determinant of the deformation gradient tensor.

2.B. Finite-element simulation

The repositioning and compression of a breast phantom were simulated using FEBio. FEBio has been extensively validated versus other established and verified FE applications.³⁶ The following methods were used to simulate gravity and compression.

2.B.1. Boundary conditions

Boundary conditions were set up within FEBio to constrain the rigid body motion and model the attachment of the breast to the rest of the body. These constraints were designed to essentially model a sliding contact between the breast and underlying ribcage, similar to the constraints used by other researchers.^{28,29,37} The chest wall surface of the breast was constrained in the anterior/posterior direction. To simulate the attachment of the breast to the rest of the body and to provide reaction forces that limit the overall rotation and translation, the skin surface was extended 48 mm in the posterior direction and the nodes on the posterior surface were constrained to have zero displacement. These constraints allow the breast to have the freedom to conform to the paddle/detector surface while limiting rotation and translation.

2.B.2. Gravity

The effects of gravity were simulated by a body force comprised of the acceleration of gravity and the mass of each element. Reversing the effects of gravity provides an approximation to the unloaded reference state without the additional complexity of solving for the unknown stress that is initially acting on the breast in the prone position. Others have modeled the stress-free reference state by modeling the buoyance of the breast (physically or with a hydrostatic pressure)²⁰ or by solving an iterative²⁷ or reverse problem.^{26,27}

One G was applied in both the posterior and vertical directions to model the patient in a standing position, with the breast in contact with the detector. For an MLO view, the detector was set to a 45° angle and the breast was modeled to slide along this surface.

2.B.3. Contact

A contact analysis was used to simulate the interaction between the breast and the rigid paddles. The contact analysis was a nonlinear problem where the gap function and contact traction (pressure) were solved by using a Newton–Raphson method. The contact tractions were then applied with a Lagrangian multiplier in order to achieve a specified

tolerance. Decreasing the contact tolerance can adversely affect the convergence and runtime of the analysis and, conversely, increasing the tolerance can lead to increased penetration (where the objects in contact begin to intersect each other).

2.B.4. Compression

The amount of breast compression was a user-defined parameter specified as either as a desired thickness in millimeter or a percent compression. Breast compression is a nonlinear large deformation problem. So, rather than hindering the computation further by attempting to prevent any penetration, a second stage of compression was applied to reduce the penetrating regions. An additional flattening stage was applied, since the flatness of the skin surface was a primary concern for the purposes of simulated mammography. This final stage displaced any penetrating nodes to be in contact with the paddle/detector and used thin plate splines to deform the internal nodes.

2.C. Application of the FE-based deformation to the phantom

Once the FE analysis was completed within FEBio, the results were used to deform the given breast phantom. Boundary surface meshes were used as the primary data structure for our breast phantom rather than the original segmented images (see the Appendix for details). To apply the deformation, the vertices of the surface meshes defining the breast phantom were located in the (u,v,w) parametric space of the undeformed finite elements. Then the deformed location of the vertices was found corresponding to their (u,v,w) location in the deformed elements. Figure 3 shows surface mesh vertices located in a sample element in the undeformed and deformed states. The location of the sample element within the breast provides a sense of scale.

The mapping within each element was diffeomorphic, since the Jacobian of the elements was checked during the FE analysis. Thus the entire mapping from the uncompressed to compressed states was diffeomorphic, provided that none of the finite elements intersected another element.

Note that the use of a diffeomorphic deformation field does not necessarily guarantee that the resulting deformation applied to a piecewise linear solid geometry will be free from folding. It would be necessary to introduce additional nodes and/or degrees of freedom to the deformed mesh to guarantee diffeomorphism. This would greatly increase the memory required to define the breast phantoms. In practice, given the short edge length of the nested meshes, intersections are unlikely to occur as a result of the deformation.

2.D. Packaging the FE method into a user-friendly application

Use of the template finite-element model, which was fit to the geometry and material composition of a given breast phantom, allows for the FE-based deformation of the breast

phantom to be automated. A MATLAB script was developed toward that end. The script sets up the FE analysis, calls FEBio to perform the analysis, reads the results, and creates a deformed surface mesh representation of the breast phantom. The user only needs to specify which voxelized breast phantom to use, the compressed thickness, and orientation of the breast.

The deformed breast phantom was designed to be used with medical imaging simulators in one of two ways. The mesh representation can be input directly into a simulator that can work with surface meshes such as the x-ray projector we developed,³⁸ or a deformed voxelized version of the phantom can be created.

2.E. Program evaluation

The finite-element compression was compared to a simplistic compression² previously developed in our lab (which was an incremental approach that did not account for gravity or material properties of the breast). Simulated CC and MLO mammographic views were created using a ray-triangle intersection algorithm to project through the deformed mesh representation of the phantom. These models were also compared to a FE model of a homogeneous breast, to demonstrate the effect of modeling the material properties of the adipose and fibroglandular tissues.

Additionally, the robustness of the FE compression was investigated by determining how many test cases would be required to achieve 40 successful cases at 55% compression in a CC orientation.

3. RESULTS

3.A. Comparison to simplistic compression/homogenous breast model

As could be expected, there were subtle differences in compressed shape and local structures between the finite-element model and our previous simplistic compression algorithm. The stretch λ , where $\lambda = l/L$ with undeformed length L and deformed length l (maximum lengths were used, which provide a global measure of stretch), was defined to be equal for the medial–lateral and anterior–posterior directions in the simplistic compression. Thus, the ratio $\lambda_{AP}/\lambda_{ML} = 1$ in the simplistic compression. In the finite-element approach, the stretch was influenced by the stiffness and shape of the breast. There was characteristically greater stretch in the medial–lateral direction compared to anterior–posterior. The mean $\lambda_{AP}/\lambda_{ML}$ was 0.89 with a standard deviation of 0.05 for the finite-element compression. A similar trend was observed in the experimental setup of Tanner *et al.* utilizing MRI of breasts in the compressed and uncompressed states.³⁹

Furthermore, the finite-element compression modeled gravity to account for the change in patient positioning from prone to standing with the breast in contact with the detector. In the MLO view, the detector was modeled at a 45° angle and the breast was modeled to slide along this surface giving a characteristically different appearance than that without modeling gravity. The variation in the orientation and

FIG. 3. Deformation of the internal structure based on the deformation of the finite elements. A boundary surface mesh representation of the phantom was used to avoid the regriding and partial volume effect associated with deforming the voxelized phantoms. Two of the six surface meshes are shown in the top row. The portion of the surface mesh contained in a given element deforms with that element. A sample element and the contained surface meshes are shown in the undeformed state (middle row) and compressed state (bottom row), respectively. The location of the elements within the breast is shown to provide a sense of scale.

appearance of the fibroglandular structure was also influenced by the modeling of local variation in material properties between adipose and glandular regions. These differences can be qualitatively observed in Fig. 4.

To demonstrate the effect of modeling the breast with a multimaterial model, a finite-element model was created for five test cases with the interior of the breast modeled as a homogenous adipose material for comparison. The difference in displacement between the multimaterial and homogenous models is shown in Table I. Maximum displacements ranged from 15.3 to 2.7 mm. In general, modeling the breast with a homogenous adipose material would have a greater effect for larger dense breasts.

The distribution of the displacement magnitude can be seen in Fig. 5 to have a general agreement with the distribution of fibroglandular tissue in the corresponding phantom. Local

variation due to the multimaterial model provides additional fidelity with little additional computational cost.

3.B. Robustness of the FE compression

In testing our FE application on 40 of our breast phantoms, we did not experience any cases that failed to converge when undergoing 55% compression. The average calculation time, including model fitting and deformation of the surface mesh representation, was 2 h 13 min per phantom with an Intel 3.4 GHz processor and 32 GB of RAM. The FE application was capable of deforming a diverse range of breast geometries as seen in Fig. 6 showing the compressed CC views for 40 variable breast phantoms (breast size ranging from 74 to 1455 ml and breast compositions ranging from 8% to 65% glandularity).

FIG. 4. FE compressions shown in the left column and simplistic compression in the right column with CC views in the top row and MLO views in the bottom row. The breast phantom had a volume of 1037 ml with a density of 27% and was compressed by 55% in each orientation.

The deformed surface mesh representation of the phantoms (described in the Appendix) was used along with a ray-triangle intersection algorithm to produce the simulated mammograms shown in Fig. 6. For these proof-of-concept images, the integer values of the fibroglandular classes were used as a correlate for attenuation and the x-ray source was modeled with parallel beams.

4. DISCUSSION

It should be noted that the modeling of gravity and the compression, in our application, was not necessarily intended to predict the actual deformation of a given breast. The deformable breast phantoms in this work were based on patient bCT data where corresponding mammograms were not available,

which inherently limits the ability to validate the FE compression. Such validations are also limited by the substantial uncertainties caused by positioning and variations in compression force/thickness, even for exams of the same breast on the same

TABLE I. Comparison of displacement between multimaterial and homogeneous models.

	Test cases				
	1	2	3	4	5
Fibroglandular density (%)	62	44	32	12	8
Volume (ml)	243	537	678	1200	970
Max displacement (mm)	5.4	11.1	15.3	6.8	2.7
Mean displacement (mm)	2.8	2.3	7.2	1.9	0.8

FIG. 5. Magnitude of displacement (in mm) between the homogenous and multimaterial finite-element models (test case 4) shown on right with a corresponding slice of the deformed voxelized phantom.

day. For this study, the use of well validated FE algorithms and parameters should lead to a reasonable image appearance, which lies within the limits of typical clinical uncertainties.

If attempting to match corresponding mammograms, the initial positioning of the breast prior to compression would have an impact on the model performance.¹⁶ Had corresponding mammograms been available, an alternative method would have been to use an iterative surface based registration algorithm similar to Krüger *et al.* to provide a mapping between the bCT data and a compressed shell developed from the mammogram.⁴⁰ It should also be noted that the desired thickness specified by the user may intentionally differ from the compressed thickness that may actually occur in a corresponding mammogram. This would allow a researcher to either include or exclude variability of breast thickness in their study design. For the purposes of multimodality deformable breast phantoms, we were seeking reasonable deformations to simulate realistic compressed geometries.

Simplifications were introduced to make the FE model a tractable solution. The boundary conditions at the chest wall were one of the larger simplifications. Tanner *et al.* were able to set the boundary conditions to model the actual deformation of the region beyond the paddle based on registration between pendant and compressed breast MRs of the patient.²⁴ We only had the pendant geometry for the breast CT data used to build our phantom population. The simplification of the first slice of the CT being fixed in the anterior–posterior direction was consistent with what other researchers have done.^{28,29,37} We have also chosen to neglect the initial stress in the prone breast and approximate the unloaded state by reversing the load due to gravity. Other researchers have modeled the stress-free reference state by solving an iterative²⁷ or reverse problem,^{26,27} which increases the complexity of the model. Another simplification was not to attempt to model any selfcontact between the breast and itself. Folding could potentially occur as the breast was compressed, which would result in a selfintersection in our finite-element model. If this were to occur, these cases

would have to be excluded. In practice, the technologist helps to guide the breast and helps to prevent pinching or folding of the breast. The manual positioning of the breast by the technologist during compression and its potential variability would have been difficult to model.

The breast tissue is typically modeled as being incompressible ($\nu_{\text{incompressible}} \approx 0.5$), which means that the volume of the breast is conserved as it is deformed. However, material properties of breast tissue are based on measurements from small *ex vivo* samples^{41,42} and do not take into account the change in blood volume experienced under the load of mammographic compression. Since the pressures induced during mammographic compression exceed systolic blood pressures,⁴³ it is likely to result in decreased blood flow to the breast. Studies of hemodynamic response to breast compression have shown approximately 25% decrease in total hemoglobin under fractional mammographic compression (approximately 20% of clinical mammographic compressive force).⁴⁴ The potential impact of the hemodynamic changes on the compressibility of breast tissue may be an area for future studies.

Unique to our approach was the use of a template FE model composed of hexahedral elements that were fit to each specific breast automatically. The material properties were then set based on the fibroglandular classes of the voxels contained in each element. This approach, along with the use of the freely available (for noncommercial use) FEBio software, enabled us to package the deformable breast phantoms into a robust user-friendly application. This approach was also robust enough to allow for modeling of the effect of gravity in combination with compression. Additionally, we utilize a boundary mesh representation of the breast phantom to avoid the regriding and partial volume effects that would be associated with deforming a voxelized model.

We produced simulated mammograms with realistic appearance, but note that the resolution of these simulated mammograms was limited by the original resolution of the bCT data set. As more bCT data sets become available, we

FIG. 6. Variation in breast size, shape, and glandularity accounted for in this breast phantom library. Projections are shown in a compressed CC orientation.

intend to compare simulated mammograms, compressed through our FE application, to the clinical mammograms from the same patient.

5. CONCLUSION

We developed a FE application that will allow researchers to reposition and apply varying levels of compression to our series of realistic breast phantoms. The FE application will give our phantoms the flexibility needed to investigate the variety of techniques used in breast imaging. Different

imaging modalities such as breast CT, MRI, mammography, tomosynthesis, and nuclear medicine require different positioning of the breast and may or may not require compression. With the FE application we developed, our breast models have the potential to evaluate and compare these modalities.

ACKNOWLEDGMENT

This project was sponsored by NIH Grant No. R01-CA-134658.

FIG. 7. Comparison of fibroglandular isosurfaces created at the original resolution (left) and using down-sampled float images (right). Even the fine fibrous structures were well preserved by down-sampling with the added benefits of decreased face count and increased smoothness. The region of interest shown represents a $21 \times 21 \times 44$ mm volume.

APPENDIX: PHANTOM MESH REPRESENTATION

Surface meshes were used to represent the boundaries between fibroglandular classes. These meshes are capable of representing complicated geometry of arbitrary topology and deforming without the rebinning and associated partial volume effects that would be associated with deforming a voxelized image.

Creating meshes from a segmented image has become a fairly common task in medical imaging. However, the mesh creation will still be discussed at some length. There are a variety of options when creating meshes, some of which were better suited to our particular application.

Typically, in a segmented image, the intensities correspond to an organ designation such as liver, lungs, and heart. To create a mesh corresponding to a specific organ, it is first necessary to create a binary image containing only that object. Then the isosurface half-way between the object and the background can be determined by methods such as the Marching Cubes.⁴⁵ However, the segmented breast image included neighboring complementary regions, each with a certain percent glandularity. Using the mesh creation scheme described above would result in duplicated surfaces with numerous small intersections due to the ambiguities in the Marching Cubes.

In our application, the intensities of the segmentation roughly correspond to the percent glandularity. For example, a mesh at the $n_k + 0.5$ isosurface creates a boundary between the glandularity classes g_k and g_{k+1} . This correspondence to percent glandularity also allows the integer segmented image to be down-sampled into a float image with $1/8$, the number of voxels with little loss of featural information and some beneficial smoothing. Note that prior to down-sampling, the background was set to the next integer value higher than the 100% glandular tissue. This down-sampling resulted in mesh

files that were roughly $1/5$ the size. The down-sampling also allowed for greater variability when interpolating the edges of the Marching Cubes, resulting in smoother meshes as seen in Fig. 7.

Meshes were created to have a nested structure, such that the n_{k-1} mesh lies within the n_k . It is possible for multiple meshes to be contained within the original voxel spacing, especially in areas with a sharp transition between intensities of the segmented image.

^{a)}Author to whom correspondence should be addressed. Electronic mail: greg.sturgeon@duke.edu

¹W. P. Segars, A. I. Veress, J. R. Wells, G. M. Sturgeon, N. Kiarashi, J. Y. Lo, E. Samei, and J. T. Dobbins, "Population of 100 realistic, patient-based computerized breast phantoms for multi-modality imaging research," *Proc. SPIE* **9033**, 90331X (2014).

²C. M. Li, W. P. Segars, G. D. Tourassi, J. M. Boone, and J. T. Dobbins III, "Methodology for generating a 3D computerized breast phantom from empirical data," *Med. Phys.* **36**, 3122–3131 (2009).

³P. R. Bakic, C. Zhang, and A. D. Maidment, "Development and characterization of an anthropomorphic breast software phantom based upon region-growing algorithm," *Med. Phys.* **38**, 3165–3176 (2011).

⁴C. M. Hsu, M. L. Palmeri, W. P. Segars, A. I. Veress, and J. T. Dobbins III, "Generation of a suite of 3D computer-generated breast phantoms from a limited set of human subject data," *Med. Phys.* **40**, 043703 (11 pp.) (2013).

⁵M. N. Ahmed, S. M. Yamany, N. Mohamed, A. A. Farag, and T. Moriarty, "A modified fuzzy c-means algorithm for bias field estimation and segmentation of MRI data," *IEEE Trans. Med. Imaging* **21**, 193–199 (2002).

⁶P. R. Bakic, M. Albert, D. Brzakovic, and A. D. Maidment, "Mammogram synthesis using a 3D simulation. I. Breast tissue model and image acquisition simulation," *Med. Phys.* **29**, 2131–2139 (2002).

⁷B. A. Lau, I. Reiser, R. M. Nishikawa, and P. R. Bakic, "A statistically defined anthropomorphic software breast phantom," *Med. Phys.* **39**, 3375–3385 (2012).

⁸L. Roose, W. De Maerteleire, W. Mollemans, and P. Suetens, "Validation of different soft tissue simulation methods for breast augmentation," in *International Congress Series* (Elsevier, New York, NY, 2005), Vol. 1281, pp. 485–490.

- ⁹A. P. Del Palomar, B. Calvo, J. Herrero, J. Lopez, and M. Doblaré, "A finite element model to accurately predict real deformations of the breast," *Med. Eng. Phys.* **30**, 1089–1097 (2008).
- ¹⁰L. Roose, D. Loeckx, W. Mollemans, F. Maes, and P. Suetens, "Adaptive boundary conditions for physically based follow-up breast MR image registration," in *Medical Image Computing and Computer-Assisted Intervention—MICCAI* (Springer, New York, NY, 2008), pp. 839–846.
- ¹¹C. Tanner, J. A. Schnabel, D. L. G. Hill, D. J. Hawkes, M. O. Leach, and D. R. Hose, "Factors influencing the accuracy of biomechanical breast models," *Med. Phys.* **33**, 1758–1769 (2006).
- ¹²L. Han, J. Hipwell, Z. Taylor, C. Tanner, S. Ourselin, and D. J. Hawkes, "Fast deformation simulation of breasts using GPU-based dynamic explicit finite element method," in *Digital Mammography* (Springer, New York, NY, 2010), pp. 728–735.
- ¹³H. M. Reynolds, J. Puthran, A. Doyle, W. Jones, P. M. F. Nielsen, M. P. Nash, and V. Rajagopal, "Mapping breast cancer between clinical x-ray and MR images," in *Computational Biomechanics for Medicine* (Springer, New York, NY, 2010), pp. 81–90.
- ¹⁴N. V. Ruiter, R. Stotzka, T. O. Muller, H. Gemmeke, J. R. Reichenbach, and W. A. Kaiser, "Model-based registration of x-ray mammograms and MR images of the female breast," *IEEE Trans. Nucl. Sci.* **53**, 204–211 (2006).
- ¹⁵N. V. Ruiter, T. O. Müller, R. Stotzka, H. Gemmeke, J. R. Reichenbach, and W. A. Kaiser, "Finite element simulation of the breast's deformation during mammography to generate a deformation model for registration," in *Bildverarbeitung für die Medizin* (Springer, New York, NY, 2003), pp. 86–90.
- ¹⁶T. Mertzaniidou, J. Hipwell, S. Johnsen, L. Han, B. Eiben, Z. Taylor, S. Ourselin, H. Huisman, R. Mann, and U. Bick, "MRI to x-ray mammography intensity-based registration with simultaneous optimisation of pose and biomechanical transformation parameters," *Med. Image Anal.* **18**, 674–683 (2014).
- ¹⁷T. Hopp, M. Dietzel, P. A. Baltzer, P. Kreisel, W. A. Kaiser, H. Gemmeke, and N. V. Ruiter, "Automatic multimodal 2D/3D breast image registration using biomechanical FEM models and intensity-based optimization," *Med. Image Anal.* **17**, 209–218 (2013).
- ¹⁸T. Hopp, N. Duric, and N. Ruiter, "Image fusion of ultrasound computer tomography volumes with x-ray mammograms using a biomechanical model based 2D/3D registration," *Comput. Med. Imaging Graphics* **40**, 170–181 (2015).
- ¹⁹T. Carter, C. Tanner, N. Beechey-Newman, D. Barratt, and D. Hawkes, "MR navigated breast surgery: Method and initial clinical experience," in *Medical Image Computing and Computer-Assisted Intervention—MICCAI* (Springer, New York, NY, 2008), pp. 356–363.
- ²⁰T. J. Carter, C. Tanner, and D. J. Hawkes, "Determining material properties of the breast for image-guided surgery," *Proc. SPIE* **7261**, 726124 (2009).
- ²¹L. Han, J. Hipwell, T. Mertzaniidou, T. Carter, M. Modat, S. Ourselin, and D. Hawkes, "A hybrid fem-based method for aligning prone and supine images for image guided breast surgery," in *IEEE International Symposium on Biomedical Imaging: From Nano to Macro* (IEEE, New York, NY, 2011), pp. 1239–1242.
- ²²L. Roose, W. Mollemans, D. Loeckx, F. Maes, and P. Suetens, "Biomechanically based elastic breast registration using mass tensor simulation," in *Medical Image Computing and Computer-Assisted Intervention—MICCAI* (Springer, New York, NY, 2006), pp. 718–725.
- ²³F. S. Azar, D. N. Metaxas, and M. D. Schnall, "A finite element model of the breast for predicting mechanical deformations during biopsy procedures," in *Proceedings of IEEE Workshop on Mathematical Methods in Biomedical Image Analysis* (IEEE, New York, NY, 2000), pp. 38–45.
- ²⁴C. Tanner, M. White, S. Guarino, M. A. Hall-Craggs, M. Douek, and D. J. Hawkes, "Large breast compressions: Observations and evaluation of simulations," *Med. Phys.* **38**, 682–690 (2011).
- ²⁵J. Chung, V. Rajagopal, P. Nielsen, and M. Nash, "A biomechanical model of mammographic compressions," *Biomech. Model. Mechanobiol.* **7**, 43–52 (2008).
- ²⁶V. Rajagopal, J. H. Chung, D. Bullivant, P. M. Nielsen, and M. P. Nash, "Determining the finite elasticity reference state from a loaded configuration," *Int. J. Numer. Methods Eng.* **72**, 1434–1451 (2007).
- ²⁷B. Eiben, V. Vavourakis, J. H. Hipwell, S. Kabus, C. Lorenz, T. Buelow, and D. J. Hawkes, "Breast deformation modelling: Comparison of methods to obtain a patient specific unloaded configuration," *Proc. SPIE* **9036**, 903615 (2014).
- ²⁸A. L. Kellner, T. R. Nelson, L. I. Cerviño, and J. M. Boone, "Simulation of mechanical compression of breast tissue," *IEEE Trans. Biomed. Eng.* **54**, 1885–1891 (2007).
- ²⁹C. M. Hsu, M. L. Palmeri, W. P. Segars, A. I. Veress, and J. T. Dobbins III, "An analysis of the mechanical parameters used for finite element compression of a high-resolution 3D breast phantom," *Med. Phys.* **38**, 5756–5770 (2011).
- ³⁰J. M. O'Connor, M. Das, C. Didier, M. Mah'd, and S. J. Glick, "Development of an ensemble of digital breast object models," in *Digital Mammography* (Springer, New York, NY, 2010), pp. 54–61.
- ³¹J. M. O'Connor, M. Das, C. Didier, M. D. Mufeed, and S. J. Glick, "Using mastectomy specimens to develop breast models for breast tomosynthesis and CT breast imaging," *Proc. SPIE* **6913**, 691315 (2008).
- ³²S. Maas and J. Weiss, FEBio, Musculoskeletal Research Laboratories, University of Utah, available at <http://mrl.sci.utah.edu/software/febio>.
- ³³D. Vartziotis and J. Wipperf, "A dual element based geometric element transformation method for all-hexahedral mesh smoothing," *Comput. Methods Appl. Mech. Eng.* **200**, 1186–1203 (2011).
- ³⁴J. A. Schnabel, C. Tanner, A. D. Castellano-Smith, A. Degenhard, M. O. Leach, D. R. Hose, D. L. Hill, and D. J. Hawkes, "Validation of nonrigid image registration using finite-element methods: Application to breast MR images," *IEEE Trans. Med. Imaging* **22**, 238–247 (2003).
- ³⁵C. Tanner, A. Degenhard, J. Schnabel, A. C. Smith, C. Hayes, L. Sonoda, M. Leach, D. Hose, D. Hill, and D. Hawkes, "A method for the comparison of biomechanical breast models," in *IEEE Workshop on Mathematical Methods in Biomedical Image Analysis, MMBIA* (IEEE, New York, NY, 2001), pp. 11–18.
- ³⁶S. A. Maas, B. J. Ellis, D. S. Rawlins, and J. A. Weiss, "A comparison of FEBio ABAQUS and NIKE3D results for a suite of verification problems," SCI Technical Report No. UUSCI-2009 (University of Utah, 2009).
- ³⁷T.-C. Shih, J.-H. Chen, D. Liu, K. Nie, L. Sun, M. Lin, D. Chang, O. Nalcioglu, and M.-Y. Su, "Computational simulation of breast compression based on segmented breast and fibroglandular tissues on magnetic resonance images," *Phys. Med. Biol.* **55**, 4153–4168 (2010).
- ³⁸W. Segars, M. Mahesh, T. Beck, E. Frey, and B. Tsui, "Realistic CT simulation using the 4D XCAT phantom," *Med. Phys.* **35**, 3800–3808 (2008).
- ³⁹C. Tanner, M. White, S. Guarino, M. A. Hall-Craggs, M. Douek, and D. J. Hawkes, "Anisotropic behaviour of breast tissue for large compressions," in *IEEE International Symposium on Biomedical Imaging: From Nano to Macro, ISBI'09* (IEEE, New York, NY, 2009), pp. 1223–1226.
- ⁴⁰J. Krüger, J. Ehrhardt, A. Bischof, and H. Handels, "Breast compression simulation using ICP-based B-spline deformation for correspondence analysis in mammography and MRI datasets," *Proc. SPIE* **8669**, 86691D (2013).
- ⁴¹T. A. Krouskop, T. M. Wheeler, F. Kallel, B. S. Garra, and T. Hall, "Elastic moduli of breast and prostate tissues under compression," *Ultrason. Imaging* **20**, 260–274 (1998).
- ⁴²A. Samani and D. Plewes, "A method to measure the hyperelastic parameters of *ex vivo* breast tissue samples," *Phys. Med. Biol.* **49**, 4395–4405 (2004).
- ⁴³J. De Groot, M. Broeders, W. Branderhorst, G. den Heeten, and C. Grimbergen, "A novel approach to mammographic breast compression: Improved standardization and reduced discomfort by controlling pressure instead of force," *Med. Phys.* **40**, 081901 (11pp.) (2013).
- ⁴⁴S. Carp, J. Selb, Q. Fang, R. Moore, D. Kopans, E. Rafferty, and D. Boas, "Dynamic functional and mechanical response of breast tissue to compression," *Opt. Express* **16**, 16064–16078 (2008).
- ⁴⁵W. E. Lorensen and H. E. Cline, "Marching cubes: A high resolution 3D surface construction algorithm," in *ACM Siggraph Computer Graphics* (ACM, New York, NY, 1987), Vol. 21, pp. 163–169.