Finite Element Simulation of the Breast's Deformation during Mammography to Generate a Deformation Model for Registration

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Abstract. For registration of X–ray mammograms and MR volumes, the deformation of the female breast during mammography has to be considered. A Finite Element simulation of this deformation is presented. Different material models for breast tissue are examined, to see if they provide a sufficiently accurate simulation. A neo–hookean model results in a simulation with an average displacement smaller than two voxels. It was found to be homogeneous and has boundary conditions which imitate the deformation between two plates. This enables the model to predict the position of the smallest visible lesion within a MRI.

1 Introduction

X–ray mammograms and Magnet Resonance Images (MRI) of the female breast provide complementary information for breast cancer diagnosis. To use this information in a combined manner, the position of a lesion detected in a X–ray has to be determined in the MRI and vice versa. The images cannot be compared directly. To acquire a mammogram the breast is deformed between two plates as far as 50 % of its former diameter before the X–ray projection is performed (see fig. 1). MRI displays the undeformed breast in a three–dimensional (3D) volume. Our goal is to estimate the location of a lesion detected in a X–ray mammogram in the MRI, based on automatic registration. Hence the spatial correlation between the deformed projection and the undeformed volume has to be defined. A model of the deformable behavior of the female breast is build, to cope with the problems arising from the huge deformations during mammography. The deformation is simulated using the Finite Element Method (FEM) based on the volume of the breast, as given in the MRI.

Recently some new approaches for FEM simulations of the female breast have been proposed. Samani (e.g. [1]) depicts mere qualitative results. Azar [2] simulates a mild deformation of the breast as applied in MRI–guided biopsy. He gives the accuracy of his simulations only for the displacement of lesions within

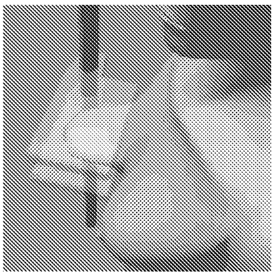

Fig. 1. Deformation during mammography.

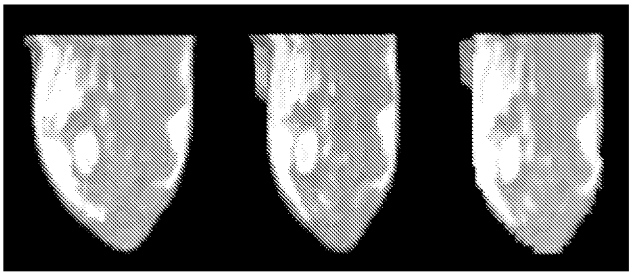

Fig. 2. Cut through (left) undeformed, (middle) deformed and (right) MRI after simulation.

the breast. Tanner [3] compares different material models using the displacement of the whole surface of the breast as boundary conditions for the deformation.

Our simulation model is designed to meet the problems arising from X–ray mammogram and MRI registration. Firstly; the spatial boundary conditions which drive the deformation process are not known in detail. E.g. the displacement of the breast's surface can not be recovered from a 2D projection as given by the X–ray. Then only the resulting thickness of the breast has to be utilized as an user–defined condition of the simulation [4]. Secondly; to locate the smallest visible tumor in a MRI, the required accuracy based on the resolution of a MRI is 3 to 5 mm. Thirdly; the material model, describing the mechanical properties of the breast tissue, should provide the needed accuracy and be as simple as possible. In the following sections different models of breast tissue are examined, to determine how far they satisfy this specifications.

2 Material Models for Breast Tissues

Recent publications describe the behavior of the breast tissue and are presented below. Azar, Wellman [5] and Krouskop [6] assume exponential, Samani hyperelastic and Krouskop and Bakic [7] linear elastic stress–strain properties of the material models and use different material parameters. Samani uses a hyperelastic material model to approximate Wellman's stress–strain properties. Azar applies the same material model as Wellman, but uses a corrected stress–strain relationship for fat. He assumes that the elastic moduli for fat, embedded in a grid of connective tissue, stiffens and becomes similar to glandular tissue above strains of 15.5 %. Because the strains of mammography are considerably higher, a simple additional material model (Azar, homogeneous) was considered, using only the properties of glandular tissue. All authors imply nearly incompressible materials with a Poisson ratio of $\nu \approx 0.5$.

The described models are quite different. They differ in the general definition of the material model, in the material parameters assigned to specific tissues and in the ratio of stiffness of gland and fat. Hence an initial assumption might be that simulation results, based on the different models should be quite different.

On the other hand the deformation of the breast is a tightly conditioned problem. E.g. nearly all surface nodes are moved by the plates and are displaced to approximately the same points for the same deformation configuration. With two–sided plate deformation the displacement of nodes in the middle of the FEM model is expected to be small too. Hence, only the displacements of intermediate points display the differences of the models and might be marginal.

To examine these effects quantitatively two simple phantom experiments have been carried out as described in the following section. Furthermore different simulations based on real MRI data are compared and the results discussed accordingly. All simulations are carried out using ANSYS [8] and the simplest available hyperelastic model (neo–hookean) is used. Because the deformation displayed in the real data is 21 % the phantoms are deformed by this amount too.

3 Performance of Different Material Models

First the variances caused by different stiffnesses of gland in respect to fat were examined, by variation of the ratio of the elastic moduli E_g/E_f. The elastic moduli of fat and gland are approximated by linear elastic stress–strain formulations at strains of 21 %. In Wellman's and Samani's model gland is 6.7 times harder than fat; in Krouskop's models 4.5; in Bakic's model 1.2; and in Azar's model 1.0. The phantom is composed as a stack of three equally sized finite elements. To approximate the thickness of the breast in the real data, the overall length is set to 8 cm. The top and bottom elements are assigned to material parameters of fat and the middle element is assigned to be glandular tissue. The displacement of the nodes connecting the elements are expected to show the maximal differences between two simulations. These nodes are allowed to move freely, because the breast tissue can move in almost all perpendicular directions to the deformation. The differences of the displacement of an arbitrary node, used as the phantom is symmetrical, is quite small. The maximal variance is 0.1 mm for the simulations with the ratios 1.0 and 6.7. Only simulations with ratios greater than 100 give a significant variance above 1.0 mm. Thus we expect the variance, due to different ratios of elastic moduli, to be small by means of the needed accuracy.

The variance due to different stress–strain relationships of the material models, have been examined using a hemispherical phantom made of homogeneous tissue. It has been subjected to mammographic deformation by using exponential, neo–hookean and linear elastic stress–strain properties. The variances are calculated using the displacement of all nodes of the FEM mesh. For 21 % deformation the mean euclidian distance and the maximal distance of the nodes were calculated. The results of this comparison are displayed in tab. 1. A neo–hookean model is, in this application, considered to be a good approximation of an exponential model, as the maximal displacement due to different material models is only 0.5 mm. Whereas the linear elastic approximation results in 1.9 mm maximal displacement. Thus the variations caused by exponential and neo–hookean material models are expected to be small.

Table 1. Differences of simulation results due to different material models. In $[mm]$. (Average euclidian distance μ, max. euclidian distance $\max_\mu$, max. distance normal to direction of deformation $\max_x$, $\max_z$ and in direction of deformation $\max_y$).

Comparison of	μ	$\max_\mu$	$\max_x$	$\max_y$	$\max_z$
Exponential/neo–hookean	0.2	0.5	0.4	0.4	0.4
Exponential/linear elastic	0.5	1.9	1.3	1.2	1.5
Neo–hookean/linear elastic	0.5	1.4	1.3	1.2	1.5

4 Simulation of Real Data

To evaluate the accuracy of simulations, two MR volumes of a healthy volunteer with and without applied deformation were used. The deformation in the deformed MRI is based on a medio–lateral mammographic deformation (left–to–right) with an amount of 21 %. Sixteen point landmarks were defined on the borders between fat and gland, corresponding in both images. The average distance, standard deviation and maximum distance of the landmarks are used for quantitative description of a simulation's accuracy. The average distance between the landmarks of the original data is 18.3 mm (± 6.5 mm). A finite element model of the breast is build based on the uncompressed MRI, by changing only the tissue properties. The resulting displacement field serves to generate a MRI of a deformed breast and to calculate the deformed positions of the point landmarks. Fig. 2 shows the comparison of the original data with a generated MRI from a simulation. (Azar, homogeneous).

In tab. 2 the results of the simulations with different material models, are displayed. In general the results of the neo–hookean and exponential models fulfill the requirements stated in the first section. The average distances are approximately 3 mm, and the maximal distances are near to 5 mm, as demanded for simulation accuracy in the first section. The linear elastic approaches have very high maximal deviations in respect to the needed accuracy (e.g. 6.8 mm for the linear elastic approximation of Krouskop's model).

These results confirm what we expected based on the phantom experiments. Even with quite different stiffness ratios of gland and fat, the results do not vary within a significant range in regards to the required simulation accuracy. The exponential and the neo–hookean models can be used as approximations, whereas the linear elastic approaches do not perform that well. The simplest tissue model, which performs within the accuracy limits, is a neo–hookean model ignoring the differences between the material properties of gland and fat for the breast simulation.

5 Discussion and Conclusion

A simulation model of the deformable behavior of the female breast was built based on a clinical MRI, which imitates the deformation of the breast as applied during mammography. The average deviation of the simulations is smaller than

Table 2. Average landmark distances (μ), standard deviations (σ) and maximal distance (max) of simulations with different material models. In $[mm]$.

Model description	μ_{exp}	σ_{exp}	$\max_{exp}$	μ_{neo}	σ_{neo}	$\max_{neo}$	μ_{lin}	σ_{lin}	$\max_{lin}$
Wellman [Samani]	3.0	1.4	4.8	[3.1]	[1.4]	[4.7]	(3.5)	(1.6)	(6.4)
Azar (inhomogeneous)	3.1	1.3	5.0	-	-	-	-	-	-
Azar (homogeneous)	3.1	1.3	5.0	(3.3)	(1.2)	(5.1)	(3.3)	(1.9)	(6.0)
Krouskop	3.1	1.3	5.1	(3.1)	(1.3)	(4.8)	(3.5)	(1.7)	(6.8)
Bakic	-	-	-	(3.3)	(1.2)	(5.1)	3.3	1.5	5.8

two voxels and hence enables the estimation of the location of the smallest visible tumors in the MRI. The proposed tissue model has two major advantages. The neo–hookean modeling allows Poisson ratios very near to 0.5 and have all the same good convergence properties with ANSYS, and it is not necessary to segment the different breast tissues.

Coopers ligaments are ignored in the model, as they are not displayed in the MRI. They give structural support to the breast and should be considered for more advanced simulations. The spatial contortions due to the imaging methods have been neglected so far. The model was tested on one individual dataset. In future more patient data will be simulated to evaluate the results with a larger data pool. The breast in the clinical data was subjected to 21 % deformation, due to difficulties in obtaining higher deformation within the mamma coil of the MRI. The result was within the lower range of the usually applied deformation during mammography. A first application of the model to register patient data with approximately 50 % deformation is described in [4]. It could be shown, that the deviation of the central point of a lesion is 3.8 mm, well inside the required accuracy limits.

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