Towards Navigated Breast Surgery Using Efficient Breast Deformation Simulation

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Abstract. Localization of target structures in open surgical breast procedures relies on metal wire guides to give orientation hints, together with radiological images. Patient positioning, however, is different for image acquisition and surgery. We propose to simulate the breast deformation between these positions to track and visualize the target position, while acquiring the interventional breast shape to guide the deformation. In the proposed setup, a structured light scanning system allows the scan of the surface, and the display of information onto the patient skin. An interactive update of the simulated deformation requires a fast scanning procedure and a real-time capable, robust deformation simulation. In this paper, besides the general system we propose specific extensions of a highly efficient dynamic corotated finite element method (FEM) to incorporate non-linear material properties while maintaining stability and speed of the original simulation. We assess the prone-supine surface distance after deformation of the prone data on a set of five volunteer images.

Fig. 1. The nipple position and a landmark position (cross) visualized in renderings of the prone (a) and supine (b) breast.

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1 Introduction

Motivation and state of the art. Breast conserving breast cancer surgery, i.e. lumpectomy and open biopsies, suffer a high recurrence rate (8% to 14%) and high number of positive margins (17% to 59%, [3]) and often poor cosmetic results, which is due to the way of indicating the target area with metal wire guides most often inserted under mammographic image guidance.

Several attempts have been made to improve surgical outcomes, relying on ultrasound imaging, on mammography, and on MRI. Methods employing robotic approaches for the placement of biopsy needles may eventually become available [10], but solve only a very constrained problem of placing a needle in a fixated breast, while approaches that help the surgeon to navigate more safely are more challenging and less developed. The study of Alderliesten et al. using MRI-based navigation tackles the problem of tracking the breast surface from a supine MRI scan to the surgery position [1], but neither support navigation in open breast surgery nor account for deformations in the prone positioning. Support of open surgery is attempted in the ultrasonography-based approach presented by Sato et al. [15]. Their approach, however, requires a tracking equipment to be positioned in the operating room, and will display the superimposed target area only in a computer monitor.

Previous work tried to match prone and supine breast shapes by employing finite element analysis using non-linear material laws [13]. However, the computational complexity of this approach is high and thus difficult to be performed in clinical routine. The fastest available implementations of dynamic non-linear models [7] are based on explicit finite element approaches, which limit the magnitude of the largest possible time step in a dynamic simulation. Furthermore, the deformation simulation has to be followed by a non-rigid registration step to achieve the final result. The contribution of Carter et al. addresses intra-interventional visualization of target structures with a technical setup that is not feasible practically [2]. Most closely resembling the setup in our contribution is the work of del Palomar [11], who matched the deformed mesh to a body scan of the women standing upright. These scans require a technical setup not suited to the situation in the operating room.

Contribution. We propose to use the pre-interventional MRI scans which are routinely taken to assess disease extent and inspect the contralateral breast. Based on these scans, we attempt to convey the target area to the breast surgeon during intervention. A prerequisite is to model the deformation of the breast from MRI in the prone position (using breast coils) to the supine position used in surgery to provide the current lesion position, which can then be displayed onto the breast using the scanner hardware that is employed to acquire the breast surface. The central element in this setup is the lesion tracing from the prone MRI scan into the current interventional position (see Figure 1).

Previous methods either lack speed or require extensive manual interaction as well as a sophisticated technical setup. Therefore, we propose an approach that combines automatic segmentation and meshing methods with a fast, real-time
capable deformation simulation. Our goal is to integrate an isotropic non-linear material law by adjusting the per-element elastic modulus in every simulation step. While anisotropic hyperelastic material laws further increase the physical accuracy of the results, they increase the computational effort significantly, and they require higher resolved meshes to represent the underlying biomechanical structures. Therefore, for a clinical setting, a compromise has to be made between fast calculations and highly realistic material behavior. We describe methods to systematically adjust the per-element elastic modulus between single steps of the simulation. Interactive speed of the simulation is maintained regardless of this addition, making it feasible for peri-interventional application.

Computationally, our approach has several beneficial properties. It avoids complicated measurements to determine non-linear material parameters, and by being based on Hooke’s law, it can be integrated into existing linear elastic code, thereby assuming non-linear or co-rotated strain formulations to enable large deformations. We approximate microscopic material tissue properties on a macroscopic scale by effectively stiffening the material under load to avoid unnatural behavior, e.g. inverted finite elements. The API to call the simulation code does not require to determine stress or strain values and therefore the method is transparent to the strain formulation (co-rotated or Green strain).

2 Material and Methods

Our proposed system starts with MR images taken in the prone position that are automatically segmented into deformable and fixed tissues using the methods proposed by Wang et al. [17]. The segmentation is used to setup the FE model, before the interactive deformation simulation starts in the operating room. A highly efficient FEM-based breast deformation approach is used to simulate the shape change from MRI to the current interventional positioning, even allowing for real-time repositioning at moderate mesh resolutions.

In the operating room, a surface scan of the patient is obtained using structured light scanning [9]. From the acquired point cloud, a mesh is built up, forming the target shape into which the MRI scan-based model is then fitted by optimizing parameters of the simulation. Finally, the display of the lesion position is achieved using the structured light projector.

MRI Data and Model Generation Standard non-fat-suppressed T2-weighted breast MRI data were obtained from five volunteers with voxel sizes in the order of $1 \times 1 \times 5$ mm, once in prone (facing down) and once in supine (facing up) position. All datasets were segmented into rigid and deformable tissue, where the breast parenchyma and the adipose tissue were considered elastic, and the thorax was considered fixed. A volumetric tetrahedral mesh was generated from a downsamples version of this data, resulting in meshes consisting of between 50k and 300k elements.

Breast Deformation Simulation Our approach is based on a multigrid finite element framework developed by Georgii and Westermann [5], which efficiently
simulates deformations of the breasts using the so-called co-rotated Cauchy strain formulation from Rankin and Brogan [14]. One novel aspect of our work is to update the per-element elastic modulus based on the shape change that the element experiences in a given simulation step. By this explicit per-element elasticity update, we effectively model a non-linear isotropic material law.

The deformation of a volumetric object is described by a displacement field \( u(x), \ u : \mathbb{R}^3 \rightarrow \mathbb{R}^3; \ x \in \mathbb{R}^3 \), which maps the reference configuration \( \Omega \) to the deformed configuration \( \{ x + u(x) \mid x \in \Omega \} \). Driven by external forces \( f \), a deformed solid is governed by the well-known Lagrangian equation of motion, \( M \ddot{u} + C \dot{u} + Ku = f \), where \( M, C, \) and \( K \) are respectively known as the mass, damping and stiffness matrices, \( u \) denotes the composition of the displacement vectors of all vertices, and \( f \) consists of the force vectors applied to these vertices. The stiffness matrix \( K \) is assembled from the so-called element stiffness matrices \( K^e \). Typically, every element in a finite element discretization has only a very small number of neighbors, and thus the resulting stiffness matrix is very sparse. The element matrices are precomputed with a fixed elastic modulus \( E_0 \). Due to the linearity of the underlying material law, the element matrix of a particular element can then be obtained by scaling \( K^e \) by the stiffness value of this element relative to \( E_0 \in \mathbb{R} \). Therefore, we can update the stiffness values within the assembling process at nearly no additional computational costs and thus achieve a fast update of stiffness values in the FE model analogously to previous approaches [4, 16]. To efficiently update the data structures of the numerical multigrid solver, we make use of a fast approach to compute sparse-sparse matrix products [6].

**Non-linear Material Modeling** We propose a novel measure of element deformation that is based on the change of element shapes. The shape of the element is considered in undeformed and deformed state, and the two states are compared. Note that the undeformed tetrahedra are dissimilar in shape, hence we calculate for each vertex of one element the largest relative change of distance to the three opposite edges, i.e. the maximum from twelve distances per element. By only considering contracting distances, this yields relative shape change values \( s \in [0, 1] \), from which a stiffness update has to be derived. Here, the initial hypothesis was that breast tissue is composed of lumps of stiff material that can move about with little friction, which requires a small elastic modulus on a macroscopic scale. However, under compression one observes a non-linear behavior, since now the stiff tissue parts determine the material behavior, i.e., the stiffness increases when the stress increases in our macroscopic model. The element relative elastic modulus is set to \( E_r = 1 + \alpha \cdot s \) with \( \alpha \in \mathbb{R} \) a user-defined scalar factor greater than zero that has been fixed at a value of 2.5 for all data in our experiments, and \( s \in \mathbb{R} \) the shape-based change value.

Explicit updates of the per-element elastic modulus in the simulation steps have to be performed carefully to ensure stability of the approach. This is due to the fact that internal forces in the body are mainly proportional to the elastic modulus, and thus updating this value while keeping the deformation increases the stored elastic energy. Therefore, we propose to use a dynamic simulation
model exhibiting damping and inertia, and to update the elastic modulus in small steps. This is accomplished by an automatically set stiffness damping coefficient proportional to the global average of element stiffness changes. In all our tests, with a time step of 0.033 s we achieved stable behavior with this damping scheme.

**Body Surface Scanning** A robust and proven structured light scanning system using binary reflected Gray code patterns was set up to allow for a setup 1 m from the patient. From the point cloud, a mesh is reconstructed and later used as a reference state into which to deform the breast as acquired in the pre-interventional MRI scan. This approach has been chosen over other technologies since it is easily scalable in terms of speed, coverage, and resolution, transportable, and requires minimal setup. To determine the orientation of the scanned surface in the world coordinate system, an inertia measurement unit (IMU) is attached to the structured light scanner setup. After calibrating to the horizontal plane, it tracks the orientation of the scanner and can thus provide the orientation of the camera and hence the direction of the gravity with respect to the observed scene. Point clouds were in this work simulated based on supine MRI.

**MRI-to-Surface Matching and Target Area Display** The intraoperative matching of the deformed MRI scan to the body surface observed with the structured light scanner is accomplished by adjusting the base elastic modulus and the orientation of the patient, which has been approximated by the IMU sensor data. The four degrees of freedom in this formulation may be optimized using any optimization algorithm. To accelerate the simulation, we apply gravity in the opposite supine direction to achieve a gravity-free state and simultaneously apply gravity in the direction corresponding to the interventional setting. While this process is generally not applicable in case of non-linear material laws, in our setting it gives reasonable results which is due to the heuristic approach to stiffen the material under compression.

In this contribution we assess only the performance of the deformation simulation in terms of the distance between the surface of the deformed prone MRI scan and the surface derived from the supine MRI. We use the method of Quinnan et al. to assess the distances, whereby distances are always measured to the closest point in the other mesh [12].

When the optimized breast deformation has been found, the surface of the deformed tetrahedral mesh is used to calculate the projection of the target position onto this surface. Our intention is to use information of a tracked surgical tool to define the viewing direction, and the intersection of this direction with the scanned surface will be used to visualize the target position on the patient’s skin.

### 3 Results

The shape-based stiffness update has been developed and evaluated on an artificial dataset prior to the application to breast MRI data. In these experiments, the shape based update criterion was able to produce plausible deformations of
the model which were in line with the observed real-world behavior. Based on literature values, a Poisson ratio of 0.43 and a base elastic modulus of $10^3$ Pa was chosen to characterize the breast tissue material (cf. [7]).

For specific combinations of base elastic modulus and the weight parameter $\alpha$ in the update function, however, oscillations can be observed due to the discrete update that causes jumps in the material modeling. This was addressed with the implementation of a damping method for the stiffness updates. Curves were plotted to track the average update of relative stiffness values from simulation step to simulation step. From these plots it was observed that an update using the shape-based distance metric yielded a more robust behavior than an analogous approach based on the von Mises stress norm [8]. The reason for this difference is the unboundedness of the von Mises stress, while the limit of the shape-based update is determined by the update function, since the values derived from the shape change are always in the interval $[0..1]$.

Simple volume mesh phantoms (cf. [8]) also allowed us to compare the different metrics to elucidate the reasons for the oscillations. For this experiment, the relative stiffness values were visualized for individual tetrahedra to assess the smoothness of the relative stiffness map. The resulting maps exhibit smooth transitions of relative stiffness values between neighboring tetrahedra, which is beneficial since larger jumps are known to affect the stability of FEM solvers.

Four of the five volunteer data sets were evaluated regarding the average distance of the meshes for the region encompassing the deformable part of the breast. In one data set, the supine scan was not segmented by the automatic algorithm. Fig. 2 shows pairs of images in undeformed and deformed state.

![Fig. 2. Distance measurements between original prone and supine surfaces derived from prone/supine MRI scans (top row) and simulated supine (from prone) and original supine surfaces (bottom row) for different data sets (from left to right). A uniformly scaled color map (green to red: 0 to 90 mm distance) was used for all images to visualize the distances.](image)

The performance of the implementation using a damped shape-based update of the elastic modulus is summarized in Table 3. Datasets were sampled to $5 \times 5 \times 5$ mm resolution. Note that by enlarging voxels to $10 \times 10 \times 10$ mm one can increase the speed by a factor of 8, thereby allowing real-time updates. For
the practical application, only the simulation of either the left or the right breast is required, approximately doubling the speed in both cases.

**Table 1.** Timing statistics. 25 simulation time steps (sim.) were performed, and the same number of per-element stiffness update operations (up.). Timings were measured on an Intel Core i7 Quad 3.0GHz CPU.

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**4 Conclusion**

Physically convincing breast deformations can be handled with our efficient framework. It is particularly beneficial for practical applicability, since after calibration and initialization no further parameters need to be adjusted. Also, the fast, automatic segmentation of the data and the easy setup and execution of the simulation will help to proceed with further pre-clinical tests of the setup.

There are a number of topics, however, that we wish to address in the future. At the moment, we use a simple approach to determine the gravity-free state, which does not account for non-linearities in the material behavior. We plan to refine the process of generating a gravity-free state similar to previous approaches, where the reference state was iteratively approximated from a loaded configuration [13]. In addition, we will have to consider the unknown compression introduced by the breast coil. Unfortunately, this will also increase the computational costs for the simulation.

Also, the simulation cannot adequately deal with deformations that are observed in volunteers with large or fatty-replaced breasts. Our ongoing work therefore introduces contact surfaces with frictional forces to model the physical situation between soft tissue and the rib cage more accurately; also we are examining a coupled simulation of a deformable body filled with a viscous fluid.

Last, the structured light scanning approach will have to prove its usability in a clinical setting, both regarding field of view and achievable contrast and speed. In case of a successful application, extensions can be made, for example regarding a simultaneous scan-display mode using a real-time capable structured light scanning approach [18].

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References