Generation and performance of patient-specific forward models for breast imaging with EIT

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Abstract. It has now been well established that accurate geometric conformity of the forward model for EIT reconstruction has significant benefits for artefact reduction and localisation of conductivity changes within the domain. The problems of generation of patient specific forward models need to be addressed as segmentation of volumetric data from CT or MRI is inadequate for time-critical clinical use. This group has pioneered methods of generating patient-specific surface models from known landmarks and electrode positions and have used this data to warp finite element models for EIT reconstruction. This paper presents a further application of these methods to use known electrode positions for breast imaging to generate an accurate B-Spline surface model of a subject and to warp an existing finite element model to the surface using elastic deformation. Results will show that a forward model can be generated, conforming more realistically to actual subject geometry, that will further enhance the performance of the reconstruction algorithm offering significant benefits to clinical EIT breast imaging.

1. Introduction
The geometric conformance between the subject and forward model is an important consideration for improving reconstruction of images in Electrical Impedance Tomography (EIT). It has been demonstrated that perturbation shape is improved and image artifacts are reduced when a geometrically accurate finite element (FE) mesh is generated for the forward model. This has been reported for both brain function [1] and lung function [2]. The most common approach for generating anatomically correct geometry is from segmented CT and MRI datasets. This is both labour-intensive and time-consuming and can only be considered as a means of providing generic models [3]. The alternative, which can produce more patient-specific models, is to warp (morph) simple geometry or generic models using surfaces built from known points on the patient such as electrode positions [4, 5].

The research group at the Dartmouth-Hitchcock Medical Center (Lebanon, NH) has developed a broadband high frequency EIT system for breast imaging [6] which has been used to image over 100 women in a clinical trial [7]. The forward models used in this work are FE meshes based on actual electrode positions but have idealized surface geometry based on a linear interpolation comprising simple geometric primitives such cylinders, cones, and ellipses to form a 3D shape that roughly resembles a breast shape matching the position of up to 4 rings of 16 electrodes. This paper reports on a work in progress that applies mesh warping methods described above to generate more accurate...
forward models based on ellipsoidal B-Spline surfaces.

2. Methods

2.1. Data Acquisition

EIT breast data is acquired using a dedicated data acquisition system and physical patient interface as shown in figure 1. The data acquisition system presents 64 channels for applying potentials and measuring resulting currents on a maximum of 64 electrodes, arranged to form four concentric rings of 16 electrodes each, stacked vertically with a separation of 3.25 cm. The physical patient interface consists of an examination bed where the patient lies prone. A hole in the surface allows the breast to be pendant under gravity below the level of the bed, where the data acquisition system is housed. Electrodes are mounted on the tip of motorized rods, which can be actuated to bring the electrodes in contact with the breast. All electrodes in a ring are actuated together to form a circular ring of a given diameter which is recorded by the system and later used to form the initial patient specific mesh. Depending on the size of the breast one or more electrode rings can be brought in contact and used; typically, 2 or 3 rings are used, that is 32 or 48 electrodes. The system adapts the number of applied voltage patterns based on the number of electrodes, and normally a protocol that uses the maximum number of linearly independent sinusoidal patterns is used, resulting in \( n-1 \) patterns for \( n \) electrodes.

2.2. Mesh Warping

One example of a FE mesh taken from the study described in [6] comprising 55037 elements, 11414 nodes and 3 electrode rings was used as the source model. A number of assumptions are made regarding the original geometry of the subject’s breast: 1) the original breast shape can be closely approximated as a semi-ellipsoid; 2) the major axis of the ellipsoid lies on the z-axis; 3) the minor axis lies at the breast/body interface and 4) that the ellipsoid passes through the largest z values of the electrodes in the 2nd and 3rd rings. The first ring is at \( z = 0.5 \) cm and the electrodes are 0.8 cm diameter. The ellipsoid is assumed therefore to pass through \( z = 7.4 \) and \( z = 10.65 \) cm. The first three assumptions are in line with those of other investigators in the field of breast mammography, for example [8]. The fourth is the best approximation is the absence of undeformed geometry data and serves to demonstrate the principle.

The vectors defining the orthogonal projection of the surface nodes to the ellipsoid are evaluated using methods previously described in [4] and these are used to define the displacement boundary conditions for the warping algorithm [5]. The resulting ellipsoid model is then warped back using the original electrode positions to define the displacement boundary conditions. To minimize element quality degradation, the latter warping process was executed four times; one for each ring of electrodes and a fourth to achieve the final model. Poisson’s ratio was taken to be 0.45. The whole procedure, coded in MATLAB, took under two minutes on a 3.16 GHz Intel Xeon processor under Windows XP.

2.3. Reconstruction

Image reconstruction is formulated as a traditional non-linear least squares inverse problem with
Tikhonov regularization as in equation (1). The forward model is based on a finite element implementation and uses the Complete Electrode Model; both conductivity and permittivity are reconstructed at the same time. While the meshes shown in figure 2 are used for computing the forward potentials, an interpolation scheme is used to represent conductivity on a coarser grid, having 1,200 elements. This arrangement reduces the number of parameters to be estimated, and it is based on lumping together the conductivity values of several elements in the fine mesh, to form a desired number of “coarse pixels”.

\[
\sigma_{\text{rec}} = \arg\min_{\sigma} \frac{1}{2} \| \mathbf{V}(\sigma) - \mathbf{V}_{\text{meas}} \|^2 + \alpha \frac{1}{2} \| \mathbf{L} \sigma \|^2
\]  

3. Results

3.1. Mesh warping
The mesh warping procedure produced a final model that was visually more representative of the patient’s breast, in-situ, with all electrodes applied. Element quality was reduced by the process: minimum stretch of the original model was 0.215 and for the final warped model it was 0.140. Stretch, in this case, is defined as the ratio of the inscribed radius in an element to its maximum edge length, normalised to lie between 0 and 1. The results are illustrated in figure 3.

3.2. Reconstruction
EIT data from a volunteer subject with a known cancerous mass of 1.3 cm diameter at 5 cm from the nipple was reconstructed using two different FE meshes. Figure 2(a) shows the first mesh built using the standard procedure used at Dartmouth described in §1. The same data was reconstructed over the warped mesh shown in figure 2(b). Contrast enhanced MRI images of the subject are shown in Figure 3.

![Figure 2: Warping process a) the original patient-specific model and b) final improved forward model](image)

![Figure 3: Contrast enhanced breast MRI images of a subject with a known cancerous mass. The 1.3cm mass, shown in brighter colour, is at a distance of 5cm from the nipple a) axial plane b) sagittal plane](image)

The two different EIT reconstructions are shown in figure 4; an increased conductivity, shown in red, corresponds to the location of the tumor.
4. Discussion and Conclusions

While the estimation of the original patient breast form was based on a relatively weak assumption, the selection of an ellipsoidal form is justified based on previous investigator’s work. The outcome of this case study clearly demonstrates, in principle, that a more precise and valid forward model can be generated within good time scales for clinical use. While it is difficult to assert that image improvements are significant from this case study, the results show promise that benefits could be obtained.

This leads the way to future investigations. The ideal scenario would be to obtain more accurate geometric data of the subject, *in-situ*, and prior to electrode application. There are many ways that this can be simply achieved within the existing system, for example: photographically or by acquiring electrode positions on the breast before applying any significant force.

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