Comparison of Electrical Impedance Tomography Reconstruction Techniques Applied to IMPETOM System

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***Abstract*— Electrical Impedance Tomography (EIT) reconstruction can estimate thorax fluid content. Its use in critically ill patients is promising and may prove clinically useful. Boundary voltages (16-electrode frames) were obtained with our 50 kHz-5 mA IMPETOM system. Comparison of the 492-element Newton-Raphson algorithm with EIDORS open source tool (NOSER & GREIT algorithms), applied to a healthy volunteer, suggests that anatomically adjusted 3D models give better results. Nevertheless for phantom imaging an initial uniform image yields more accurate reconstructions. The results help in the selection and implementation of the reconstruction method for systems similar to IMPETOM.**

I. INTRODUCTION

Estimation of alveolar fluid content and distribution is essential in the management of conditions such as cardiogenic pulmonary oedema, pleural effusions, pneumonia and adult respiratory distress syndrome (ARDS). Electrical impedance of tissues can be estimated by measuring voltages on the skin while applying high frequency currents ($>$ 20 kHz) whose amplitudes ($<$ 5 mA) are below perception thresholds. Processing electrical impedance matrices yields tomographic images [1]. This method, known as Electrical Impedance Tomography (EIT), is a low-cost, non-invasive, continuous-measurement method used to obtain low resolution images of the distribution of pleuro-pulmonary fluids and air. It is thus an attractive alternative to currently adopted imaging procedures of the thorax (X-rays, Ultrasound, Computer Tomography (CT) and Magnetic Resonance Imaging (MRI)). Following Barber and Brown's pioneering works [2], we developed a system called IMPETOM (**impe**dance **tom**ography) consisting in a series of prototypes [1], [3], [4], [5], [6] tested in phantoms and healthy volunteers. This paper compares different reconstruction methods (Figure 2) used to elaborate data obtained with IMPETOM. This comparison led to specification improvements of our pre-clinical prototype.

II. MATERIAL AND METHODS

A. Inverse Problem

1) Mathematical background: EIT is a technique used to derive conductivity within a region from measurements taken on its boundary. The region of interest Ω is part of space ($\Omega \subset \mathbb{R}^3$), the same electrodes are used to inject the currents and to measure the voltages. There are several possible configurations for an EIT system, including the frequency and waveform of injected current, the number of electrodes and their pairing (e.g. adjacent, Figure 1 or opposite electrodes). The problem to solve is the mapping $d \in D \mapsto m \in M$ where d is the potential along the boundary of the region Ω, and $D \subseteq H^{1/2}$, being $H^{1/2}$ the $1/2$ order Sobolev space [7], m represents the conductivity inside the region, and M is the subset of differentiable functions m such that $m > 0$. Since boundary information is used to reconstruct the interior of the region, this is an inverse problem. As shown later, it is also an ill-conditioned problem, which makes reconstruction not straightforward. To overcome the ill-conditioning we introduced a regularisation method.

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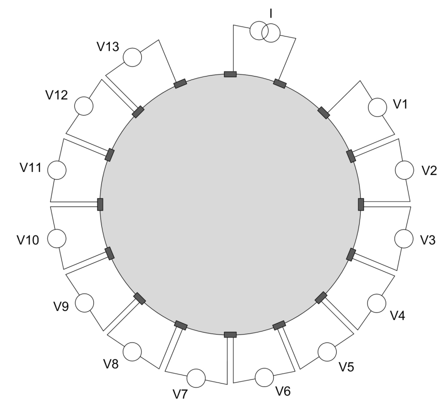

Fig. 1: EIT data collection system.

The system is governed by Maxwell'S equationsconsidering we are using low frequencies, the magnetic field is ignored. Given the that thorax Ω is closed and bounded, with a smooth boundary $\partial\Omega$, conductivity within the thorax σ is a function of the spatial variable x. The potential is ϕ and the electric field $\mathbf{E} = -\nabla\phi$. The current density is $\mathbf{J} = -\sigma \nabla \phi$ consistent with Ohm's law. Since there are not current sources within the body, Kirchoff's law yields $\nabla\cdot\sigma\nabla\phi = 0$ and the current density along the boundary $\partial\Omega$ can be expressed as $j = -\mathbf{J}\cdot\mathbf{n} = \sigma\nabla\phi\cdot\mathbf{n}$, where $\mathbf{n}$ is the outgoing normal vector to $\partial\Omega$. All currents entering the body Ω eventually exit: mathematically $\int_{\partial\Omega} j = 0$. The Dirichlet boundary condition states that, given the conductivity σ and knowing the potential along the boundary $\phi|_{\partial\Omega}$ is enough to determine the body potential ϕ. Similarily, the Neumann boundary condition states that σ,and the current density along the boundary, j, are enough to determine ϕ [9]. Using these two conditions we can define the mapping $\Lambda_\sigma : \phi|_{\partial\Omega} \rightarrow j$ and the problem to be solved is the inverse of

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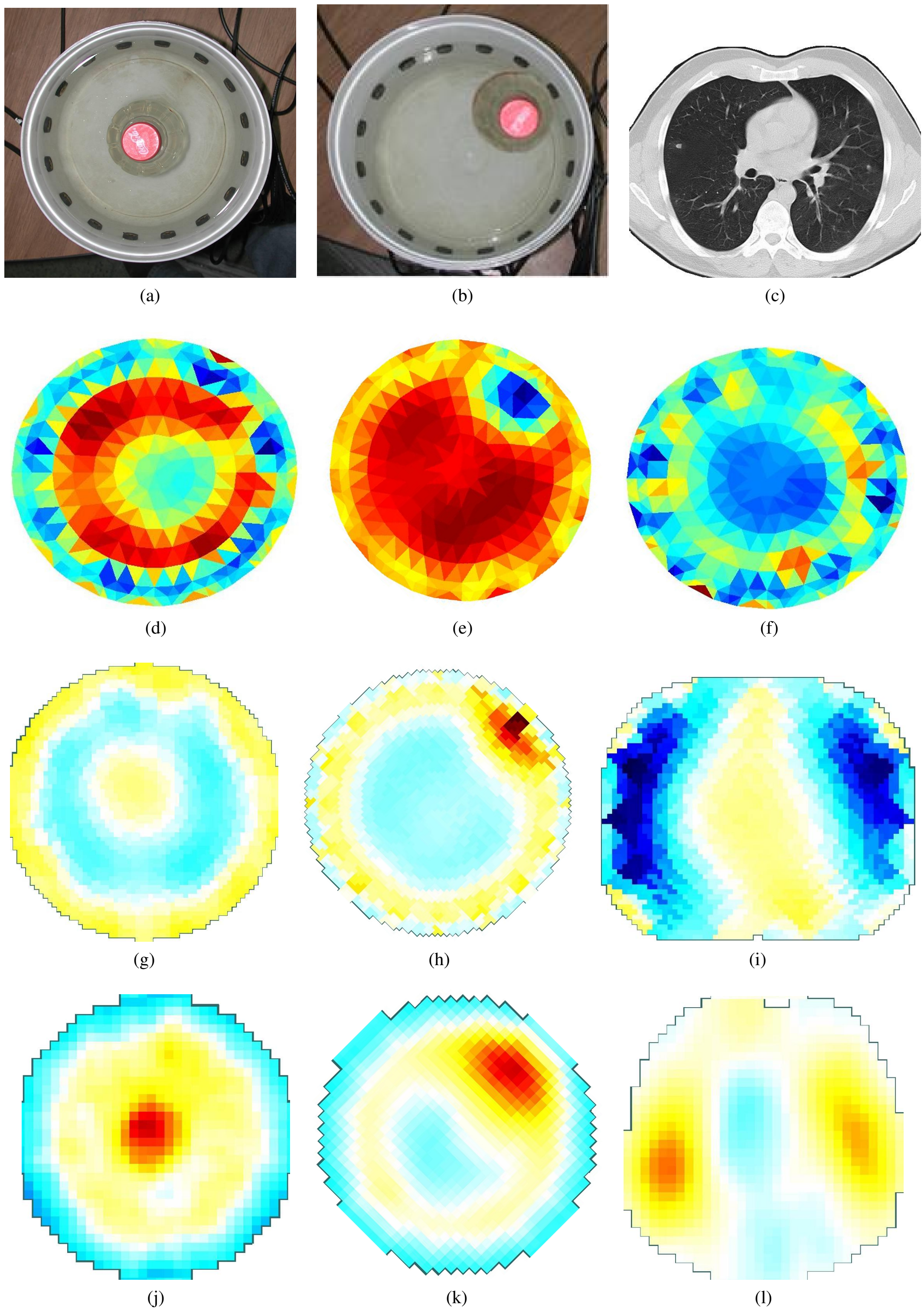

Fig. 2: Comparison of reconstruction methods: the first row shows the phantom or pacient to be imaged. The lines below show the output of IMPETOM (492 pixels), NOSER (equiv. 2048 pixels) and GREIT (1024 pixels) algorithms. First column is an empty plastic bottle in a saline filled tank. Second column shows the bottle placed near the edge of the tank. Third column is a sample CT scan and tomographic sections of a normal volunteer.

the mapping $\Lambda_\sigma \to \sigma$, which produces σ values, distributed conductivity.

But in EIT systems we only know the potential along the boundary $\phi|_{\partial\Omega}$ at the electrodes, and these potentials are represented by the measurement vector d.

2) First part of the solution: Forward Problem Solving: To solve the inverse problem it is necessary to know the forward problem, $h : M \to D$ which is the mapping of the conductivity inside the body Ω to the potentials along the boundary $d = h(m)$. Since it is a non-linear mapping we assume there must be a linear equivalent:

$$d = Hm \tag{1}$$

The determination of the forward operator h by mathematical analysis is only possible if the geometry and the conductivity of the EIT problem are simple (e.g. uniform conductivity in a cylinder) [8]. In all other cases, it is necessary to numerically discretise both the region and the conductivity with finite elements. It is also necessary to feed the method wiht initial experimental data to obtain a linear formula such as (1).

3) Second part of the solution: Regularisation: The inverse problem consists in the determination of the parameters m corresponding to the measurements d knowing the mapping H. The measurements errors and uncertainties involved are such that (1) does not hold. When this happens we have an ill-conditioned problem in the Hadamard sense [10] which implies that at least one of the following is true:

(i) The solution does not exist.
(ii) The solution is not unique.
(iii) The solution is a noncontinuous function of the data, such that small perturbations cause arbitrarily large errors in the reconstructed parameters.

With the least squares method the problem can be expressed as:

$$m_{LS} = \arg\min \|Hm - d\|^2 \tag{2}$$

Where m_{LS} being the discretise solution of σ. Facing the problem by means of the single value decomposition (SVD) [11, p. 52] of the matrix H it is possible to observe that in the presence of measurements noise the solution can become very unstable, and the resulting conductivity inaccurate.

Tikhonov modified equation (2) to confine the solution to a space of expected solutions: this is done by adding a term introducing information on where the solution might be.

$$m_{(\alpha,L,m_0)} = \arg\min(\|Hm - d\|^2 + \alpha\|L(m - m_0)\|^2) \tag{3}$$

Where m_0 is an initial estimation of m. Different forms of L can be used, such as Tikhonov's standard regularisation $L = I$ and NOSER regularisation $L = \mathrm{diag}(\sqrt{J^T J})$ where J is the Jacobian matrix whose elements $\frac{\partial d}{\partial m}$ are derived from an initial conductivity estimate. In (3) m is therefore the desired tomographic reconstruction of the region Ω.

B. IMPETOM Structure

IMPETOM, described in [1] includes: Waveform synthesis, Current source, Differential amplifier, Synchronous voltmeter, Multiplexers and Control block. IMPETOM [3], [5] uses a single ring of 16 electrodes, which applies a sinusoidal, 50 kHz-5 mA current injected in two adjacent electrodes. Voltage is measured on all other pairs of adjacent electrodes, obtaining 13 measurements. This pattern is repeated for every pair of electrodes, thus $13 \times 16 = 208$ voltages measurements are obtained. This vector is called a *frame* and contains the information used for reconstruction. Only one current source is multiplexed but 16 separated measurement channels work in parallel.

C. EIT Reconstruction Implementations

To obtain tomographic reconstruction, we consider three different implementation: the original IMPETOM [4], GREIT and NOSER regularisation, both obtained using the EIDORS package [12] which is an open source software suite for image reconstruction in EIT with several algorithms to solve both forward and inverse problems. IMPETOM original reconstruction software uses the Newton-Raphson method, with finite elements.

D. Experimental Data

The original data was obtained using IMPETOM circuitry [5] and was tested with both IMPETOM system and EIDORS [13]. This comparison is important for prototype development and deciding the best algorithm to adopt.

1) Cylindrical tank phantom: For develoment purposes an original phantom was build. It consists of a cylindrical tank 21 cm in diameter, filled with saline water. Inside the tank an empty 8 cm diameter plastic bottle was initially moved along the diameter of the tank, and subsequently around its boundary.

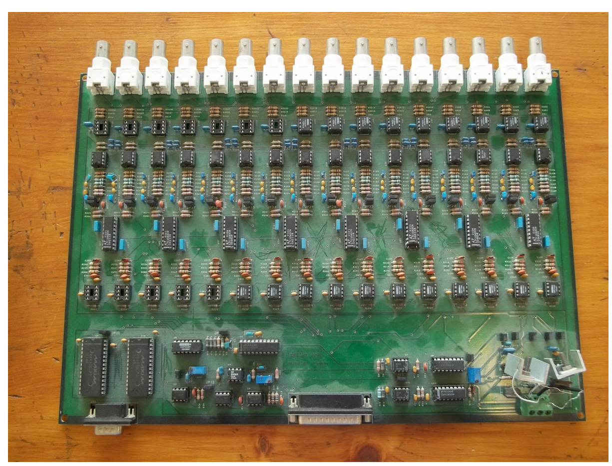

Fig. 3: IMPETOM circuit board as described by [3]. Upper coaxial connectors are used for the 16 skin electrodes. 16 voltmeters and one current source (lower left area) are visible. A DB25 connector (bottom centre) is used to connect to the A/D converter.

2) Healty volunteer: A healthy eight-years-old son one of the authors [5] was harnessed with the 16 electrode IMPETOM to obtain tomographic images of his chest.

3) Acquisition parameters: A/D converter of IMPETOM allowed to acquire images at a rate of $15 frames/sec$. The experiment with the phantom included 300 frames and 400 were acquired when the volunteer was imaged.

4) GREIT recontruction algorithm: One of the reconstruction algorithms we tested is GREIT, an open source, consensus linear reconstruction algorithm for lung EIT [14]. The specifications of GREIT are: *(i)* Single ring electrode configurations with adjacent current injection and measurement. *(ii)* Linear reconstruction of a 2D conductivity difference image, based on a 3D forward model. *(iii)* Reconstruction into a 32×32 pixel array for a single ring of 16, 12 or 8 electrodes. *(iv)* predefined shapes of: a) neonatal chest, b) male and female adult chest, and c) cylindrical tank. GREIT is useful because it includes performance figures of merit for EIT images reconstructions. Consensus figures of merit, in order of importance, are: *(i)* uniform amplitude response, *(ii)* small and uniform position error, *(iii)* small ringing artefacts, *(iv)* uniform resolution, *(v)* limited shape deformation, and *(vi)* high resolution.

5) Reconstruction parameters: Standard IMPETOM parameters were used: 492 elements, 2D circular mesh.

NOSER regularisation parameter was set by trial and error to 0.07 for phantom imaging (Figures 2g, 2h) and to 1.0 for the healthy volunteer (Figure 2i). For the cylindrical phantom 256 elements-2D circular mesh was used. A thorax shaped mesh was selected to image the volunteer.

The GREIT algorithm was trained wiht a uniform distribution of points for the chest and a heavy-centred distribution of points [14] for the phantom.

The initial image for the cylindrical tank is a uniform circle which gives a *frame* of simulated measurements with the forward soultion. This frame is substracted from the real measure *frame*. The difference was then used to fed the inverse problem in order to obtain tomographic section (Figure 2). This method is always used by IMPETOM. By contrast with NOSER and GREIT we use a different approach for the volunteer, which consist of using the mean of 400 real measurements as the initial *frame*. The difference between the mean *frame* and the real *frame* is fed to the inverse problem algorithm yielding the tomographic image.

III. RESULTS

The result of the comparison is shown in Figure 2. Considering that our goal was to develop a prototype for quantitative air/fluid chest distribution [1] and not an anatomically accurate image the results are promising. IMPETOM Figures 2d, 2e, 2f were not as good as GREIT, Figures 2j, 2k, 2l but where than NOSER, Figures 2g, 2h, except for the lung reconstruction in Figure 2i. With standard desktop computing power all three methods create images in less than a second.

The GREIT algorithm included 3D mesh to account for current leakages to adjacent planes, which is factor of quality.

There are two factors that enabled us to obtain better results for chest reconstruction (Figures 2i, 2l) with respect to the original IMPETOM reconstruction (Figure 2f): a chest shaped model and starting with the mean of 400 frames to feed the inverse problem solver.

A quantitative comparison between different methods would be desirable, and with the available data is possible to implement it in future works.

IV. CONCLUSIONS

Both the IMPETOM hardware and software must be improved using open source tools and anatomically adjusted 3D models. By doing so we expecto to improve our tool to clinically quantify lung water over time.

V. ACKNOWLEDGMENTS

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