VARIABILITY IN EIT IMAGES OF LUNGS: EFFECT OF IMAGE RECONSTRUCTION REFERENCE

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ABSTRACT: There is significant interest in Electrical Impedance Tomography for measurement of breathing. However, *Kunst et al.* (Physiol. Meas. 1998) have shown that variability in parenchymal density (in emphysema and haemodialysis patients compared to normals) had a large effect on the amplitude of EIT images for the same inspired volume. We hypothesize that this effect is due to the assumption made by EIT difference imaging that changes occur relative to a homogeneous conductivity distribution. To test this hypothesis, we developed a 3D finite element model of the thorax, and simulated EIT measurements for a small tidal volume at different levels of lung conductivity. Images were reconstructed using: 1) a homogeneous model, 2) a model with physiologically realistic conductivity levels, and 3) a model with conductivities matching the simulation model. Results show that the reconstructed image amplitude of the homogeneous model varies strongly with lung conductivity. The magnitude of the variations is compatible with the data of Kunst *et al*. The physiologically realistic model showed slightly less variation, and the matched conductivity model showed almost uniform amplitude response. These results suggest that the variability in EIT image amplitude of the lungs may be due to assumption of homogeneity made by difference EIT image reconstruction algorithms.

Keywords: Electrical Impedance Tomography, Lung Function, Image Reconstruction

1. INTRODUCTION

One of the most promising applications of Electrical Impedance Tomography (EIT) is for monitoring lung function [4], to measure the amount and distribution of inspired air. EIT images of the lung benefit from its large size and large conductivity contrast to other body tissues. There is also a linear relationship between the measured tidal volume and the amplitude of differential images (e.g. [7]). On the other hand, EIT images of the lung are subject to the movement of electrodes during breathing [2] and changes in posture [8]. EIT difference imaging has been a successful method for this application, because it is less sensitive to these uncertainties [2].

While differential EIT image amplitude is proportional to tidal volume, the constant of proportionality varies dramatically between subjects. In order to calculate quantitative lung volumes, it is typically necessary to calibrate EIT images with a known volume level (e.g. [3,5]). Kunst *et al.* [9] studied the variation in EIT image amplitude between subjects. Images of a given tidal volume were compared between normals and groups with high parenchymal density (haemodialysis patients) and low parenchymal density (emphysema patients). Results were measured in terms of the sum of image pixels per litre of tidal volume. The emphysema group had significantly lower impedance change (11.6±6.4) than normals (18.6±4.2), while the haemodialysis group showed a significantly larger impedance change (30.5±13.1). Furthermore, during dialysis, the latter group showed impedance changes much closer to the normals (21.4±8.6). These results show EIT in poor light: not only can measurements of the same tidal volume vary by a factor of three between patient groups, but even if calibration is performed, the calibration factor can undergo large changes rapidly.

We are interested in understanding the cause of this variability. In this paper, we hypothesize that this effect is largely caused by the assumption of homogeneity made in the formulation of difference imaging in EIT. In order to explore this effect, a finite element model of the thorax was constructed, and simulation data at different baseline lung conductivities generated.

Subsequently, images were reconstructed from these data under different difference imaging assumptions, and compared to the results of Kunst *et al.*[9].

2. METHODS

1.1. Finite Element Model

In order to simulate the effect of baseline lung conductivity, we developed a 3D finite element model (FEM) of the conductive properties of the thorax. The FEM uses 10368 tetrahedral elements and covers a region 15 cm vertically centred on the heart. Sixteen compound electrodes were simulated and spaced equally around the thorax at the level of the centre of the heart. A 2D slice through the centre of this FEM is the same as the model used in [2]. Conductivity values for tissue regions were interpolated from [6] and are shown in table 1.

Tissue	Condition	Conductivity (mS/m)
Bone		10
Soft tissue		480
Lung	Expiration	120
Lung	Inspiration	60

Table 1: Tissue conductivity values used for simulation model [6].

The FEM (F) calculates a vector of voltage measurements, $\mathbf{v}$. Conductivity values for all tissues except lung were held constant, while the lung conductivity (σ_L) was varied. Thus, we represent the measurements as a function of σ_L: $\mathbf{v}=F(\sigma_L)$. Simulation data were calculated to model the EIT difference measurements due to introduction of a small tidal volume (ΔV) at different levels of baseline σ_L. Since the relationship between σ_L and lung volume (V_L) is not precisely known *in vivo* [10], we model lung resistivity as proportional to V_L. For a small ΔV, we make the following approximation:

$$\sigma = \frac{\sigma_L}{1+\Delta V/V_L} \approx \sigma_L\left(1+\Delta V/V_L\right) \approx \sigma_L \exp\left(-\Delta V/V_L\right) \tag{1}$$

Using this approximation, tidal volume ΔV constitutes a constant decrease in log conductivity. Difference measurements are simulated for inspiration and expiration as:

$$\begin{aligned} v_{insp} &= F\left(\sigma_L \exp\left(-\Delta V/V_L\right)\right) \\ v_{expi} &= F\left(\sigma_L\right) \end{aligned} \tag{2}$$

Simulations were conducted for a small value of $\Delta V/V_L$ for a range of values of σ_L from 5 mS/m to 2000 mS/m. This large non-physiological range was simulated in order to clarify the trend of the results.

1.2. Image Reconstruction

EIT difference images were calculated using the 2D reconstruction algorithm of [1]. In order to ensure a different geometry for image reconstruction and simulations, a circular geometry with 256 finite elements was used for image reconstruction. The lung region was modelled as an inner circular region constituting 76% of the medium area. For each simulated level of σ_L, a difference image was calculated from the measurements $v_{insp} - v_{expi}$. An EIT estimate of tidal volume (ΔV_{EIT}) was calculated by summing all pixels in a region of interest incorporating the lungs. To obtain a unitless measure, ΔV_{EIT} was then normalized with respect its value for σ_L = 120 mS/m.

Typically, reconstruction algorithms for EIT difference images assume that the initial conductivity is homogeneous, and conductivity changes occur with respect to this baseline value. This assumption is clearly unwarranted for imaging of the thorax, where the lungs are significantly less conductive than other tissue. EIT imaging algorithms based on a finite element model can typically be modified to use a different initial conductivity assumption. For example, in [1], the EIT forward problem is linearized as

$$\mathbf{z} = \mathbf{H}\mathbf{x} + \mathbf{n} \tag{3}$$

where $\mathbf{z}$ is the vector of differential measurements, $\mathbf{x}$ is the conductivity change image, and $\mathbf{H}$ is a Jacobian (sensitivity) matrix calculated from a FEM ($F(\boldsymbol{\sigma})$), as

$$\mathbf{H} = \left.\frac{\partial F_i(\sigma)}{\partial \sigma_j}\right|_{\sigma = \sigma_0} \tag{4}$$

where $\boldsymbol{\sigma}_0$ is a vector representing the background conductivity distribution of each element in the FEM. For simplicity, in equation 4, we ignore the effect of parameterization of conductivity changes, $\mathbf{x}$. In order to modify the assumption of a homogeneous background conductivity, it is possible to modify $\boldsymbol{\sigma}_0$ to account for the lung conductivity σ_L, and then calculate the corresponding $\mathbf{H}$, and use it to calculate the reconstructed image. The EIT estimate of tidal volume using this modified reconstruction algorithm is represented as $\Delta V_{EIT,\sigma_L}$.

3. RESULTS

Simulation data were imaged, and ΔV_{EIT} calculated, using three different reconstruction algorithms: 1) using a homogeneous $\boldsymbol{\sigma}_0$ ($\Delta V_{\mathrm{EIT,homog}}$), 2) using $\boldsymbol{\sigma}_0$ with physiological values and σ_L at its inspiration value (60 mS/m) ($\Delta V_{\mathrm{EIT,insp}}$), and 3) using $\boldsymbol{\sigma}_0$ with physiological values and σ_L matching the simulated value ($\Delta V_{\mathrm{EIT,simul}}$). Figure 1 shows graphs of ΔV_{EIT} for each algorithm as a function of σ_L.

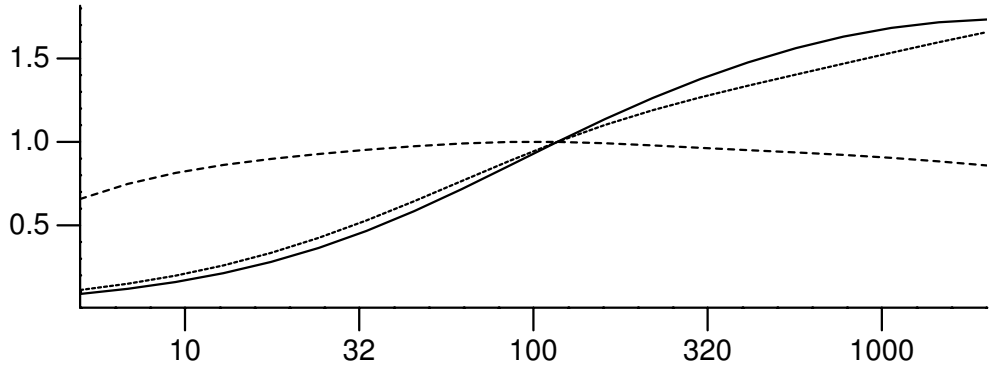

Figure 1: EIT difference image amplitude due to a small tidal volume as a function of baseline lung conductivity (σ_L) (mS/m). Image amplitude is normalized to a value of 1.0 when lung conductivity matches expiration (120 mS/m). *Solid:* images reconstructed with homogeneous background ($\Delta V_{\mathrm{EIT,homog}}$), *Dotted:* images reconstructed with lung region conductivity of 60 mS/m ($\Delta V_{\mathrm{EIT,insp}}$), *Dashed:* images reconstructed with lung region conductivity equal to the simulation model value (on horizontal axis) ($\Delta V_{\mathrm{EIT,simul}}$).

The results for $\Delta V_{\mathrm{EIT,homog}}$ (solid line) are consistent with those of Kunst *et al.* [9]. Image amplitude increases dramatically with increasing lung conductivity; there is a 70% increase in image amplitude as σ_L increases from 60 mS/m to 120 mS/m. Use of constant but physiologically

realistic values reduces the dependence on σ_L slightly (dotted line). Finally, the use of parameters that match the simulation (dashed line) results in significant decrease in the dependence on σ_L.

4. DISCUSSION

Our motivation for this study is to understand the causes of the results of Kunst *et al.*[9], in which different baseline lung conductivity levels introduced a dramatic difference in the magnitude of EIT images of the same tidal volumes. We have developed a rough simulation model of the effect of the assumption of homogeneous baseline lung conductivity on EIT images, which is able to account for the magnitude of the observed effect. This result suggests that the variability observed could possibly be eliminated by enhancements to EIT image reconstruction algorithms. This is supported by the graph of $\Delta V_{\mathrm{EIT,simul}}$ in which the variability was significantly reduced.

On the other hand, many other factors could contribute to the observed effect, such as: 1) breathing pattern differences (abdominal versus thoracic breathing), 2) size of thorax, 3) non-linear relationship of conductivity change to inspired volume, 4) movement of the chest with breathing, 5) changes in Cole-Cole parameters of lung tissue in patients with haemodialysis and emphysema. We postulate that the baseline conductivity effect is dominant, as most of the other factors would appear to be significantly smaller than the observed variability. For example, a simulation study of the movement of the chest with breathing showed changes due to movement of approximately 20% [2]. Modifications in baseline conductivity may also explain the variability in EIT images with changes in posture [8]

In conclusion, these results suggest that an important contribution to variability in the amplitude of EIT difference images of the lungs is the assumption of homogeneity of the background conductivity in difference image reconstruction; furthermore, modifications to image reconstruction algorithms may be able to reduce the magnitude of the variability.

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