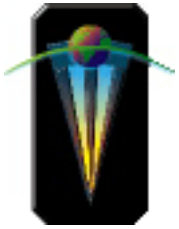
Variability in EIT Images of Lungs: Effect of Image Reconstruction Reference

Andy Adler
School of information and Technology, University of Ottawa, Ottawa Canada
adler@site.uottawa.ca

Brad Graham
School of Information and Technology, University of Ottawa, Ottawa Canada
graham.bm@sympatico.ca

School of Information Technology and Engineering



Introduction

Differential EIT image amplitude is proportional to tidal volume, however the constant of proportionality varies dramatically between subjects.

Kunst, Vonk Noordegraaf, Straver, Aarts, Tesselaar, Postmus, and de Vriesy [9] have shown that variability in parenchymal[1] density amongst subjects from different groups had a large effect on the amplitude of differential EIT images for the same inspired volume.

	Normals	Emphysema	Haemodialysis	Haemodialysis during dialysis
Image Amplitude	18.6 ± 4.2	11.6 ± 6.4	30.5 ± 13.1	21.4 ± 8.6
Image amplitude is measured in terms of image pixels per litre of tidal volume				

[1]Parenchyma: the internal functional tissues of an organ, as opposed to supporting or structural tissues

Image amplitude is calculated as the sum of the grey scale values of each pixel in the image. In a FEM a pixel refers to an element of the FEM and the grey scale value is the conductivity.

Images of a given tidal volume[2] were compared between normals and groups with high parenchymal density (haemodialysis[3] subjects) and low parenchymal density (emphysema subjects).

The emphysema group had significantly lower impedance change than normals, while the haemodialysis group showed a significantly larger impedance change. Furthermore, during dialysis, the latter group showed impedance changes much closer to the normals.

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.

[2]Tidal Volume is the amount of air inhaled and exhaled during normal breathing, normally about between 500 and 1000ml.

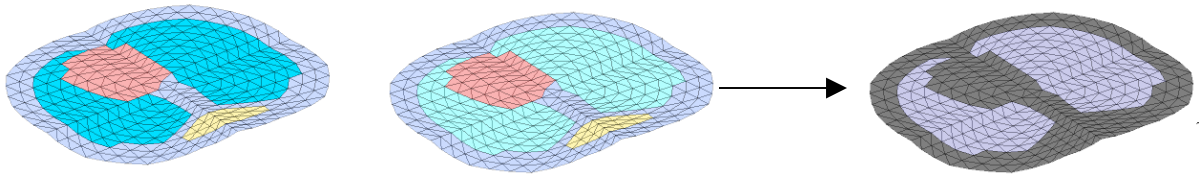
[3]Haemodialysis: the use of a machine to clean wastes from the blood after the kidneys have failed. The blood travels through tubes to a dialyzer, which removes wastes and extra fluid. The cleaned blood then flows through another set of tubes back into the body.

Goal of this Work

We hypothesize that this effect is due to the common assumption made by EIT difference imaging that impedance changes occur relative to a homogeneous background conductivity distribution.

The goal of this work is to test the hypothesis that the large variability in image amplitude described above is due to the use of algorithms that assume impedance changes occur relative to a homogenous background.

EIT Difference Imaging produces an image of the change in conductivity that occurs between two sets of measurements.



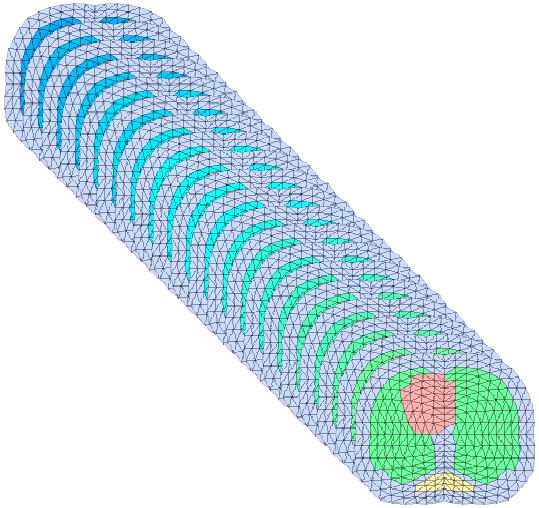
Conductivity at Time 1

Conductivity at Time 2

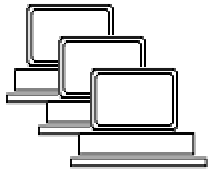
Conductivity Change between time1 and time2 that we want to recover through EIT

In order to explore this effect, simulation data were calculated to model the EIT difference measurements due to introduction of a small tidal volume at different background levels of lung conductivity.

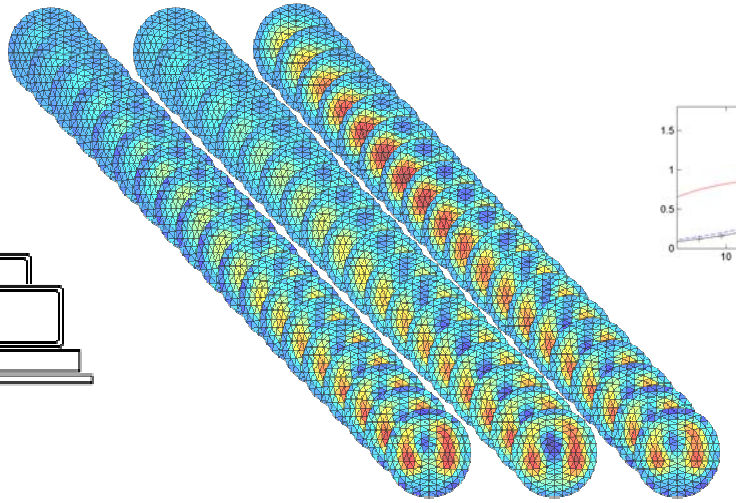
Impedance difference images were then reconstructed from this simulated data under three different difference imaging assumptions, and the resulting image amplitudes compared to the results of *Kunst et al [9]*.



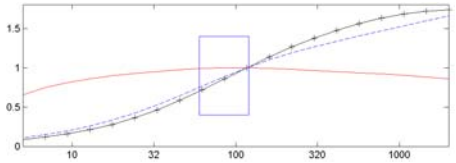
20 Sets of Simulated Data



3 Imaging Algorithms



3 Sets of 20 Reconstructed Images

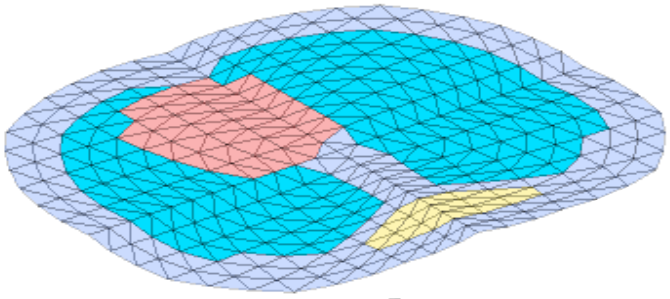


Comparison of Image Amplitudes

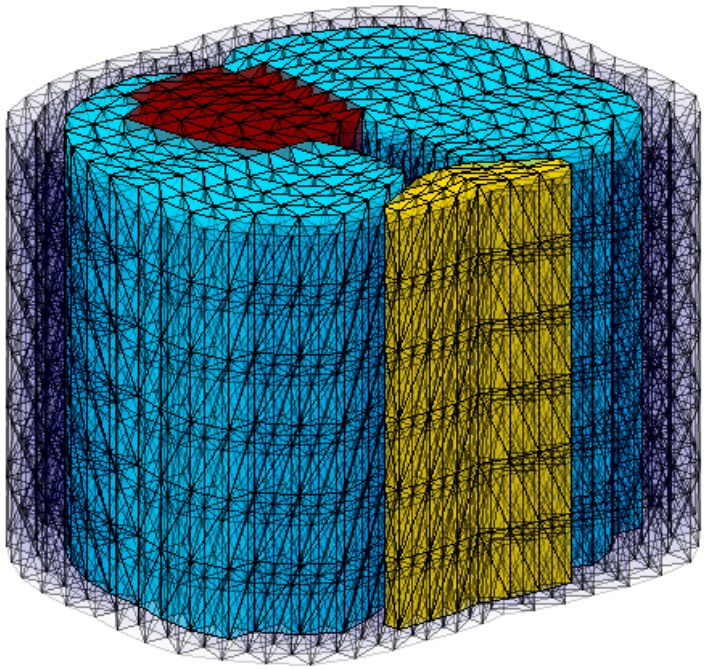
Simulating the Data

Simulated voltage measurements were generated based on a 3D finite element model (FEM) of the thorax. A 2D slice through the centre of this FEM is the same as the model used in [2].

The 3D FEM consisting if 10368 tetrahedral elements was constructed by vertically extruding the 2D FEM. The FEM is divided into 4 regions of differing conductivity: lung, heart, spine, and soft tissue.



2D FEM used in [2]

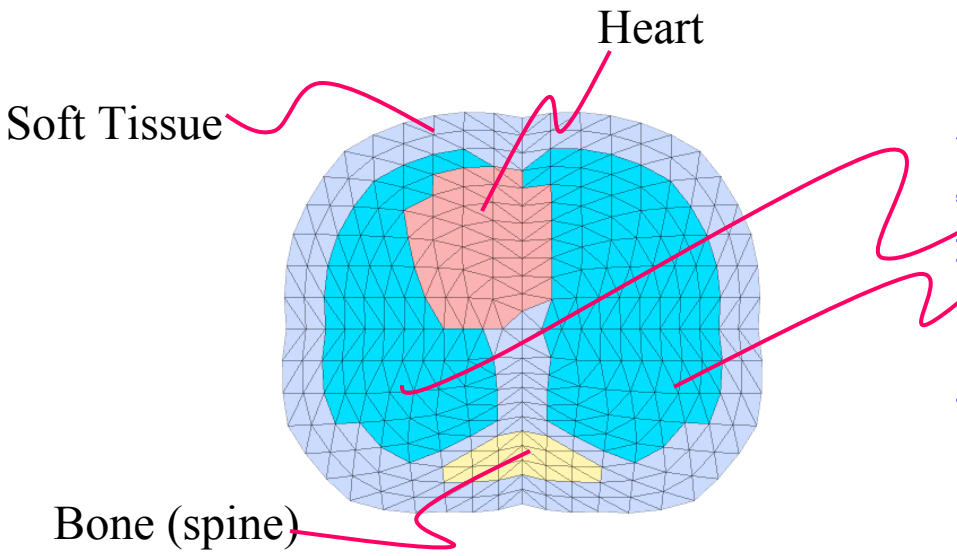


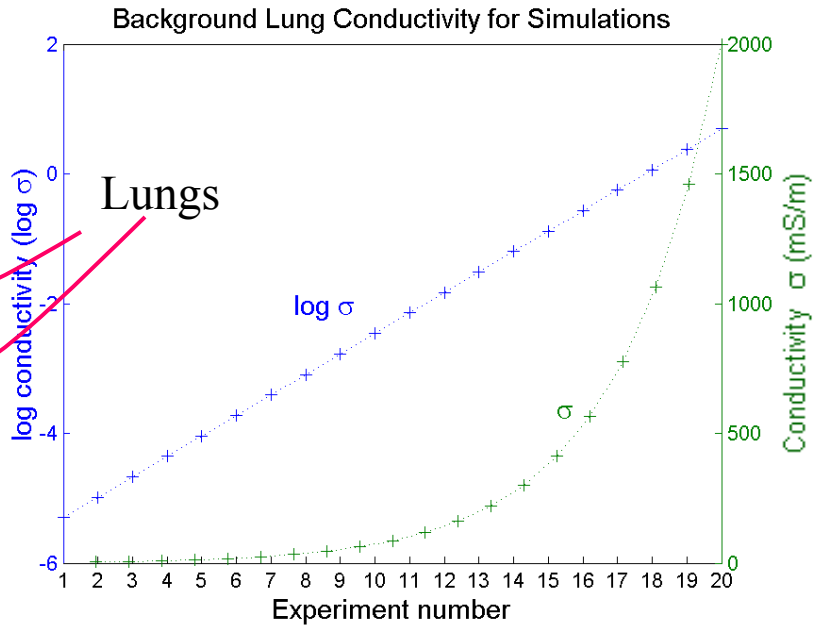
3D FEM used in this work

Physiologically realistic conductivity values for tissue regions were interpolated from [6].

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

Conductivity values for all tissues except lung were held constant, while the lung conductivity (σ) was varied.



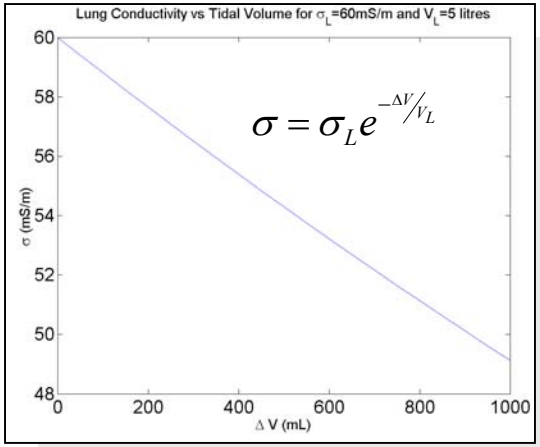


Since the relationship between lung conductivity, σ_L, and lung volume (V_L) is not precisely known *in vivo* [10], we model lung resistivity as proportional to V_L. For a small tidal volume, ΔV, we make the following approximation:

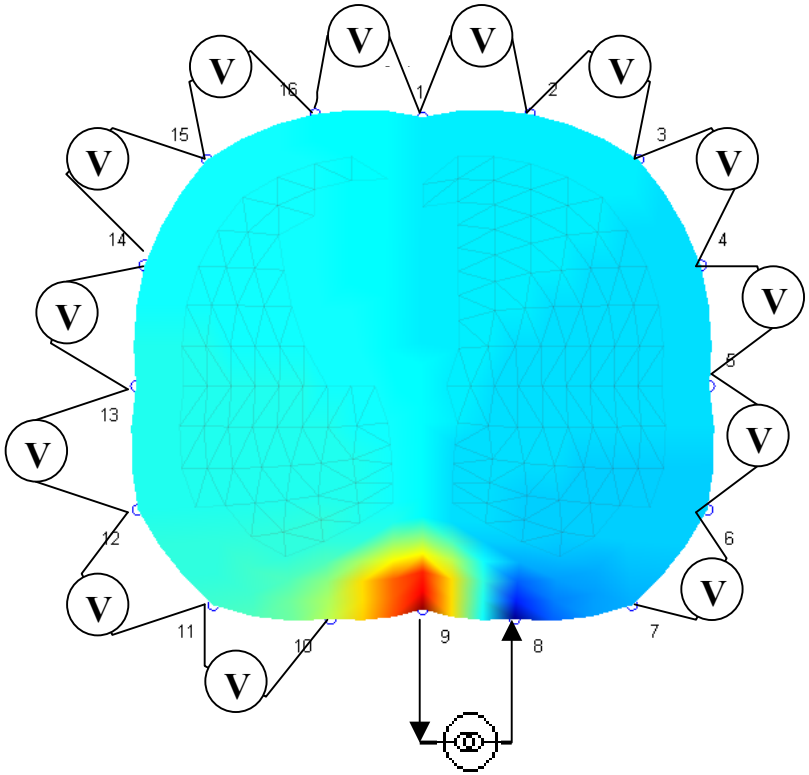
$$\sigma = \frac{\sigma_L}{1 + \Delta V / V_L} \approx \sigma_L e^{-\Delta V / V_L}$$

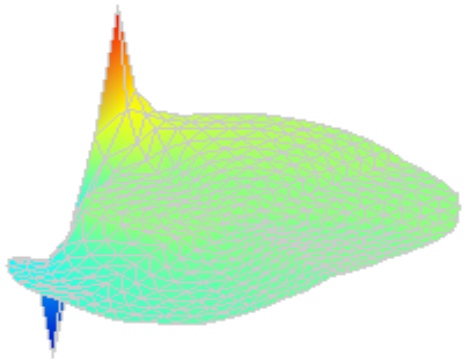
Using this approximation, tidal volume, ΔV, constitutes a constant decrease in log conductivity.

σ is conductivity
σ_L is lung conductivity
σ_0 is background lung conductivity
V_L is lung volume
ΔV is tidal volume
$F(\sigma)$ is the FEM, a function of conductivity and geometry



Sixteen compound electrodes were simulated and spaced equally around the thorax at the level of the centre of the heart. Using the adjacent drive current pattern each simulation resulted in a set of $16 \times 13 = 208$ voltage measurements.





3D Plot of voltages at the nodes of a 2D slice of a FEM

Current being driven in and out of electrodes 8 and 9 while 13 individual voltage measurements are taken at the other electrodes.

Voltages that would be measured from one of the current driving electrodes are not used due to the uncertainties introduced by the voltage drop across the electrode-skin contact impedance.

A complete set of 208 measurements is obtained by injecting currents at all 16 possible pairs of adjacent electrodes with each injection allowing 13 voltage measurements.

Each difference image required 2 sets of simulated voltages, $v^{inspiration}$ calculated at the initial lung conductivity and $v^{expiration}$, calculated after the lung conductivity change caused by the introduction of a small tidal volume. The subsequent EIT difference image was then calculated from the *dynamic signal*:

$$z = \frac{v^{\text{inspiration}} - v^{\text{expiration}}}{\frac{1}{2}\left(v^{\text{inspiration}} + v^{\text{expiration}}\right)}$$

where z, $v^{inspiration}$, and $v^{expiration}$ are vectors of length 208.

We performed 20 simulations in which conductivity values for all tissues except lung were held constant, while the lung conductivity was varied over a range of 5 mS/m to 2000 mS/m in 20 steps.

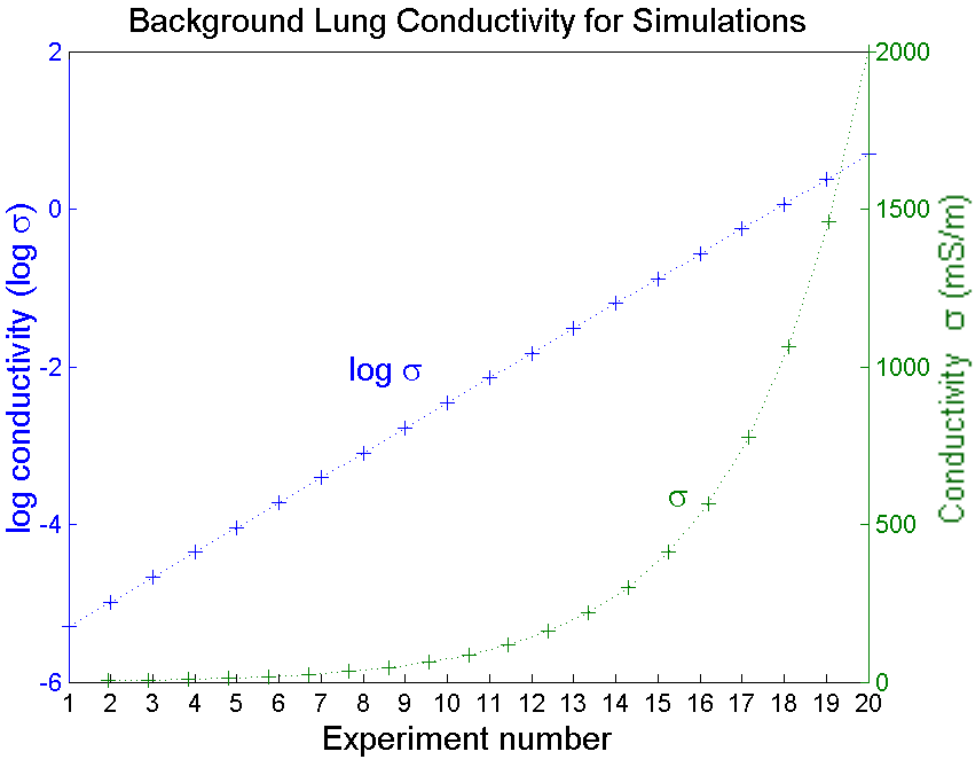
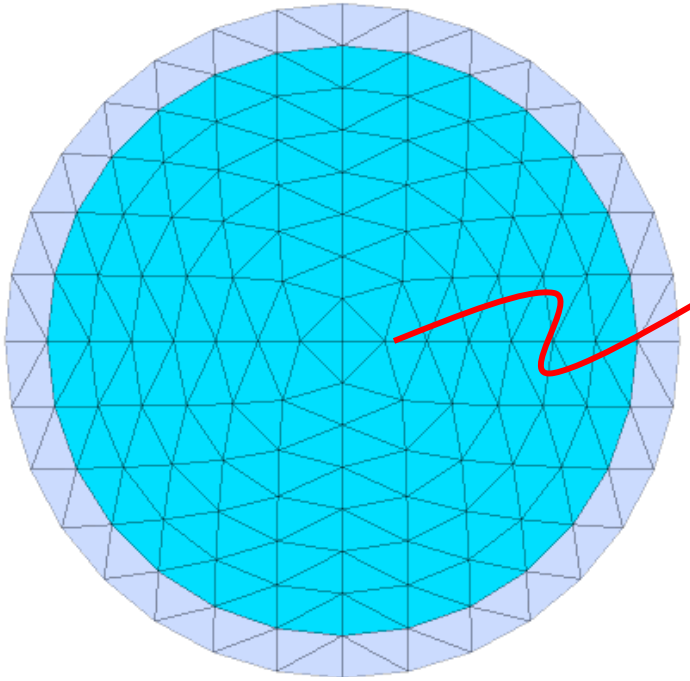


Image Reconstruction

In order to ensure a different geometry for image reconstruction and simulations[4], a circular geometry with 256 finite elements was used for image reconstruction. The lung region was modeled as an inner circular region constituting 76% of the medium area.



[4]An inverse crime occurs when the same (or nearly the same) theoretical ingredients are used to simulate and subsequently invert data in an inverse problem such as EIT Image reconstruction.

To avoid an inverse crime we use a different mesh (geometry and element density) than the one used to simulate the data in order to avoid possible cancellation of errors used in simulating the data through the forward model.

EIT difference images were calculated using the 2D reconstruction algorithm of [1]. In this algorithm the forward problem of EIT is linearized by finding a matrix H and a parameterization f such that $z=Hx+n$ where z is the vector of differential measurements, x is the conductivity change image[5], n is the measurement noise, and H is a Sensitivity matrix calculated from a FEM as

$$H_{ij} = \left. \frac{\partial F_i(\sigma)}{\partial \sigma_j} \right|_{\sigma=\sigma_0}$$

where $F_i(\sigma)$, a function of the conductivity, is the FEM, and σ_0 is a vector representing the background conductivity distribution about which impedance changes will occur as a result of changes in lung volume due to breathing.

[5]The most commonly used parameterization is the logarithm conductivity,

$$x = \ln(\sigma^1) - \ln(\sigma^2) = \Delta \ln(\sigma)$$

Which has the advantage of corresponding to a positive conductivity for any real value x

In [1] it is shown how H is derived from the dynamic signal

$$z = \frac{v^{\text{inspiration}} - v^{\text{expiration}}}{\frac{1}{2}(v^{\text{inspiration}} + v^{\text{expiration}})}$$

H relates the effect of a small change in conductivity to a small change in differential voltage measured. It is a function of the geometry of the FEM and the *a priori* assumption of the background conductivity distribution, σ_0 , about which conductivity changes occur.

We use a regularized version of *H* to solve for *x* given the dynamic signal *z*. The regularization scheme used is the Maximum *a priori* (MAP) algorithm of [1] in which we solve for *x* as

$$x = (H^T W H + \mu Q)^{-1} H^T W z$$

where $\sigma_n^2 W = R_n^{-1}$ is a function of the noise

and $Q = \sigma_x^2 F^T F$ is a high pass filter

However, the ideas presented in this work apply to any EIT difference imaging algorithm based on a sensitivity matrix.

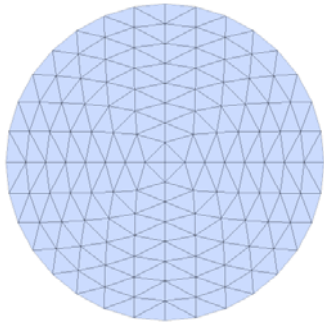
Reconstruction algorithms for EIT difference images often assume that the background conductivity of the region being imaged is homogeneous, and conductivity changes occur with respect to this baseline value. This assumption is clearly unwarranted for imaging of the thorax, where the lungs are significantly less conductive than other tissue.

In order to modify the assumption of a homogeneous background conductivity, σ_0, is altered to account for the conductivity of the various tissues in the thorax. The sensitivity matrix, *H*, is constructed from the modified σ_0 and used to calculate the reconstructed image.

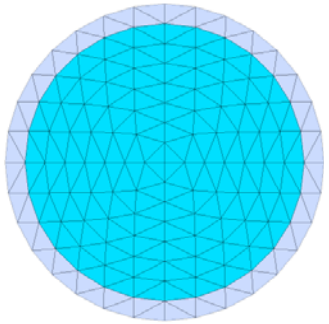
Images were reconstructed from the simulated data using three different reconstruction algorithms:

1) using a homogeneous σ_0 ($\Delta V_{EIT,Homog}$)
2) Using σ_0 with physiological values and σ_L at its inspiration value (60 mS/m)
3) Using σ_0 with physiological values and σ_L matching σ_L that was used to simulate the data

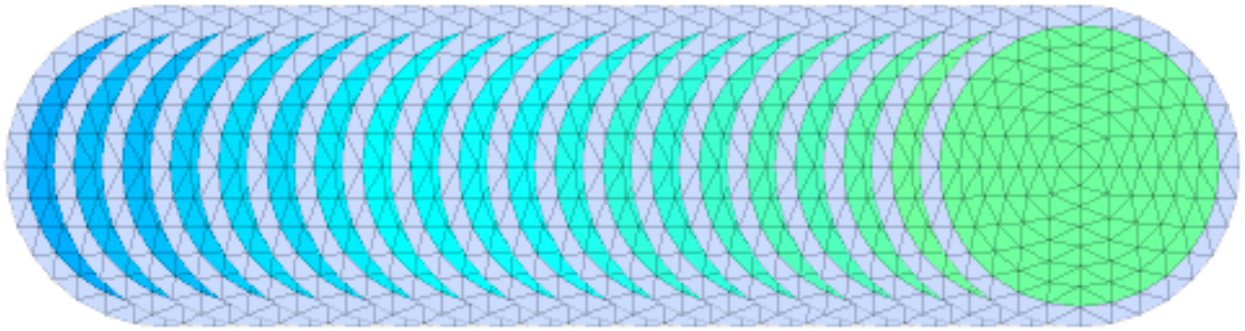
The 3 FEM Models used to construct the Sensitivity Matrices



Homogeneous background conductivity distribution



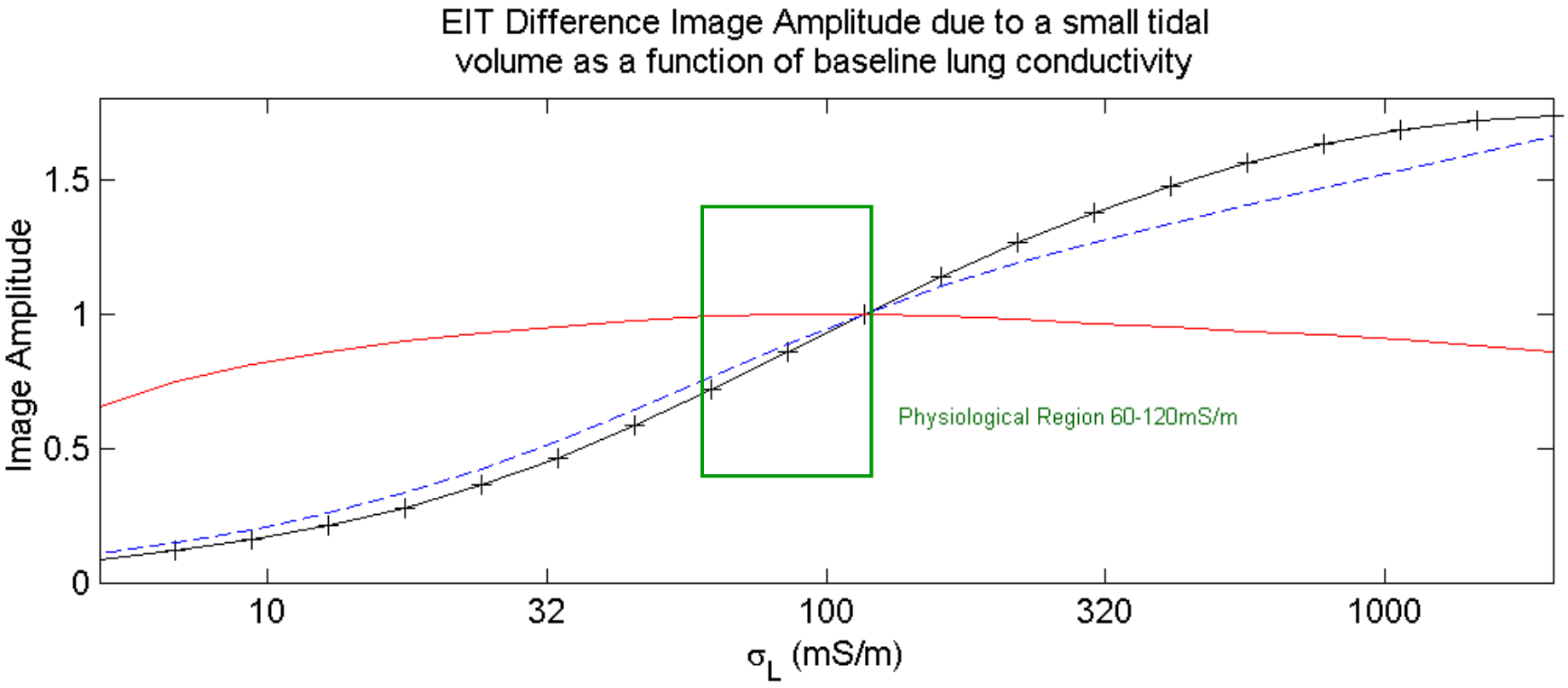
Background Conductivity distribution with Physiologically realistic value for lung, $\sigma_L = 60mS/m$



Background Conductivity distribution varies in step with value used to simulate data from 5mS/m to 2000mS/m

Results

An EIT estimate of tidal volume (ΔV_{EIT}) was then calculated by summing all pixels in a region of interest incorporating the lungs.



To obtain a unitless measure image amplitude, (ΔV_{EIT}), was normalized with respect its value for $\sigma_L = 120 mS/m$

The results for $\Delta V_{EIT,Homog}$ (solid black line) are consistent with those of *Kunst et al.* [9].

Image amplitude increases dramatically with increasing background/initial lung conductivity: there is a 70% increase in image amplitude as increases from 60 mS/m to 120 mS/m.

Use of constant but physiologically realistic values reduces the dependence on σ_L slightly (dashed blue line).

However, the use of parameters that match the simulation (solid red line) results in significant decrease in the dependence on σ_L.

Discussion

Our goal is to understand the causes of variability in EIT difference image amplitude reported in *Kunst et al.*[9].

Many factors could contribute to the observed variability in image amplitude, such as:

1) breathing pattern differences (abdominal versus thoracic breathing),

2) size of thorax,

3) nonlinear 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 background conductivity effect is dominant, as most of the other factors 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].

The results of the simulations suggest that the dominant contribution to the variability in 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.

Contact Information

Andy Adler
School of Information and Technology,
University of Ottawa, Ottawa, Canada;
adler@site.uottawa.ca
http://www.site.uottawa.ca/~adler/

Brad Graham
School of Information and Technology,
University of Ottawa, Ottawa, Canada;
graham.bm@sympatico.ca

References

1. Adler A, Guardo R, Electrical Impedance Tomography: Regularised Imaging and Contrast Detection, *IEEE Trans.Medical Imag.*, 15:170-179, 1996.
2. Adler A, Guardo R, Berthiaume Y. Impedance Imaging of Lung Ventilation: Do we need to account for Chest Expansion? *IEEE Trans. Biomedical Engineering*, 43:414-421, 1996.
3. Frerichs I, Hahn G, Hellige G, Thoracic Electrical Impedance Tomographic Measurements During Volume Controlled Ventilation-Effects of Tidal Volume and Positive End-Expiratory Pressure, *IEEE Trans. Medical Imag.,* 18:768-778, 1999.
4. Frerichs I, Electrical impedance tomography (EIT) in applications related to lung and ventilation: a review of experimental and clinical activities, *Physiol. Meas.* 21:R1-R21, 2000.
5. Frerichs I, Hinz J, Herrmann P, Weisser G, Hahn G, Dudykevych T, Quintel M, Hellige G, Detection of local lung air content by electrical impedance tomography compared with electron beam CT. *J Appl Physiol* 93: 660-666, 2002.
6. Gabriel C, Gabriel S, Corthout E, The Dielectric Properties of Biological Tissues: I. Literature Survey. *Phys. Med. Biol.* 41: 2231 - 2249, 1996.
7. Harris H D, Suggett A J, Barber D C, Brown B H, Applications of applied potential tomography in respiratory medicine, *Clin. Phys. Physiol. Meas.* 8:A155-A65, 1987.
8. Harris H D, Suggett A J, Barber D C, Brown B H, Applied Potential Tomography: A New Technique for Monitoring Pulmonary Function., *Clin. Phys. Physiol. Meas.* 9:A79-A85, 1988.
9. Kunst P W A, Vonk Noordegraaf A, Straver B, Aarts R A H M, Tesselaar C D, Postmus P E, de Vriesy P M J M, Influences of lung parenchyma density and thoracic fluid on ventilatory EIT measurements, *Physiol. Meas.* 19:27-34, 1998.
10. Nopp P, Rapp E, Pfutzner H, Nakesch H, Ruhsam C, Dielectric Properties of Lung Tissue as a Function of Air Content, *Phys. Med. Biol.* 38: 699 - 716, 1993.