Rongqing Chen*, and Knut Möller

Global Inhomogeneity Index Evaluation of a DCT-based EIT Lung Imaging

Abstract: *Purpose:* To evaluate a novel structural-functional DCT-based EIT lung imaging method against the classical EIT reconstruction. *Method:* Taken retrospectively from a former study, EIT data was evaluated using both reconstruction methods. For different phases of ventilation, EIT images are analyzed with respect to the global inhomogeneity (GI) index for comparison. *Results:* A significant less variant GI index was observed in the DCT-based method, compared to the index from classical method. *Conclusion:* The DCT-based method generates more accurate lung contour yet decreasing the essential information in the image which affects the GI index. These preliminary results must be consolidated with more patient data in different breathing states.

Keywords: EIT, lung imaging, DCT, GI.

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1 Introduction

Electrical Impedance Tomography (EIT) is a feasible non-invasive imaging method with great capability to visualise the regional ventilation distribution of lungs to assist clinician in adjusting proper PEEP levels for patients under mechanical ventilation [1], and further to avoid ventilator induced lung injury [2,3].

However, the spatial resolution of EIT is low, yet temporal resolution higher than traditional morphological imaging methods, such as computed tomography (CT). Therefore, our team have previous combined imaging modalities of CT and EIT to roughly restrict EIT image to CT generated anatomy [4]. This DCT-based EIT approach includes detailed prior information about both the thorax contour and lung shape obtained from the discrete cosine transformation (DCT) of the CT image, which as a side effect is resulting in improved interpretability for clinicians.

The objective of this work is to evaluate the plausibility of this DCT-based method against the traditional Gauss-Newton one step method in clinical settings. In a first step Global Inhomogeneity (GI) index is used for comparison, which indicates the difference of the volume distribution within a ventilation period [5].

1.1 Classical EIT Reconstruction

The reconstruction of an EIT image is ill-posed, therefore a regularization is needed to obtain the estimates of the conductivity distribution [6,7]. The actual conductivity distribution can be inhomogeneous, thus, a finite element model (FEM) is commonly used to discretize the domain into piecewise constant regions. In the following the reconstructing process using the one-step Gauss-Newton method for time-difference imaging is introduced.

The reconstruction of conductivity can be written as:

$$\hat{x} = \arg\min_{x} \left\{ \mathbf{J}\mathbf{x} - \mathbf{y}_2^2 + \lambda^2 \mathbf{R}\mathbf{x}_j^j \right\} \tag{1}$$

where $\hat{x}$ represents the reconstructed conductivity change, $\mathbf{J}$ is a Jacobian matrix that maps the conductivity change in a FEM element to the measured voltage variation. The regularization is defined as $\mathbf{R}$ and weighted by λ to obtain a reasonable solution.

With the widely accepted l_2 -norm, the distribution is calculated as:

$$\hat{x} = \left(\mathbf{J}^{\mathrm{T}}\mathbf{J} + \boldsymbol{\lambda}^2 \mathbf{R} \right)^{-1} \mathbf{J}^{\mathrm{T}} y = By \tag{2}$$

matrix $\mathbf{B}$ is the reconstruction matrix which calculates the impedance distribution variation from the measured boundary voltages. The reconstruction is obtained from a simple matrix multiplication which is so efficient to implement real-time EIT imaging.

1.2 DCT-based EIT Reconstruction

The DCT-based EIT lung imaging algorithm is based on prior anatomy information from CT images assessed via the Discrete Cosine Transformation (DCT) method. DCT is a widely implemented method in image processing, e.g. in JPEG image compression [4]. The concept of DCT is to represent the image with a sum of cosine functions in varying

* **Corresponding author: Rongqing Chen** Institute of Technical Medicine, Furtwangen University, Jakob-Kienzle-Str. 17, VS-Schwenningen, Germany. e-mail: chr@fs-furtwangen.de
Knut Möller Institute of Technical Medicine, Furtwangen University, Jakob-Kienzle-Str. 17, VS-Schwenningen, Germany. e-mail: moe@fs-furtwangen.de

frequencies, in x-direction and y-direction respectively. For a two dimensional image A with M rows and N columns we can define the DCT as:

$$\begin{cases} V_{p,q} = \alpha_p \alpha_q \sum_{m=0}^{M-1} \sum_{n=0}^{N-1} A_{m,n} \cdot F_{p,q} \\ F_{p,q} = cos \frac{(2m+1)p\pi}{2M} \cdot cos \frac{(2n+1)q\pi}{2N} \end{cases} \quad (3)$$

where

$$\alpha_p = \begin{cases} \frac{1}{\sqrt{M}}, p = 0 \\ \sqrt{\frac{2}{M}}, 1 \le p \le M-1 \end{cases}$$

and

$$\alpha_q = \begin{cases} \frac{1}{\sqrt{N}}, q = 0 \\ \sqrt{\frac{2}{N}}, 1 \le q \le N-1 \end{cases}$$

The matrix $\mathbf{V}$ consists of the DCT coefficients, of which a subset can recover a compressed image of A.

$$\tilde{A}_{m,n} = \sum_{m=0}^{M-1} \sum_{n=0}^{N-1} \alpha_p \alpha_q \tilde{V}_{m,n} \cdot F_{p,q} \quad (4)$$

where $\tilde{\mathbf{V}}$ is a sparse matrix of the same $\mathbf{V}$ size but only having several non-zero elements.

The dimensionality of the EIT inverse problem can be reduced significantly by concentrating on the most important subset of DCT coefficients. The implementation is achieved by multiplying the Jacobian matrix $\mathbf{J}^{n \times n}$ by a matrix $\mathbf{K}^{n \times DCT}$ derived from DCT coefficients, which are far less than the number of FEM elements. For this article the same number of frequencies (15) in both x-direction and y-direction were chosen, which sums to only 225 DCT coefficients.

The discussed DCT approach can be expressed as:

$$V_{p,q} = \sum_{m=0}^{M-1} \sum_{n=0}^{N-1} A_{m,n} \cdot D(p,q)_{m,n} \quad (5)$$

in which $\mathbf{D}(p,q)$ is the cosine function at frequencies p and q. We generate a matrix $\mathbf{C}(p,q)$ to introduce the prior anatomy information of the lung contour thus a proper distribution can be calculated.

$$C(p,q)_{m,n} = P_{m,n} \cdot D(p,q)_{m,n} \quad (6)$$

of which $P_{m,n}$ indicates whether the pixel belongs to the lung value.

$$P_{m,n} = \begin{cases} 1, pixels\ within\ lung \\ 0, pixels\ ouside\ lung \end{cases}$$

Each column k_j of $\mathbf{K}$ is created as follows:

$$k_j = T\left(\mathbf{C}(p,q)\right) \quad (7)$$

where T is a map to assign every pixel in $\mathbf{C}(p,q)$ to element in the FEM model which covers the pixel. The index j derives from the linear combination of the frequencies p and q.

Upon calculating the matrix $\mathbf{K}$, we can obtain a new Jacobian matrix, which further describes the reconstruction as:

$$\hat{x} = \left(\mathbf{J}_{DCT}^T \mathbf{J}_{DCT} + \lambda^2 \mathbf{R}_{DCT}\right)^{-1} \mathbf{J}_{DCT}^T y = \mathbf{B}_{DCT} y \quad (8)$$

the reconstruction matrix $\mathbf{B}$ maps the voltage variations to the DCT coefficients.

With the reconstructed x we can now recover the image $\mathbf{H}$, constrained of prior anatomical information derived from the original CT image:

$$H = \sum_{m=0}^{n_{xdct}} \sum_{n=0}^{n_{ydct}} \mathbf{C}(p,q) \cdot \hat{x}_{DCT,j} \quad (9)$$

1.3 Global Inhomogeneity Index

The Global Inhomogeneity (GI) was primarily introduced to simplify the intra-individual comparability of EIT images obtained from at different time points. The GI is usually calculated from a tidal image that derives from subtracting end-expiration from end-inspiration [5]. The calculation of GI requires the median value of a tidal image and the sum of the variation in every pixel, which describes as:

$$GI = \frac{\sum_{x,y \in lung} \left| DI_{xy} - Median\left(DI_{lung}\right) \right|}{\sum_{x,y \in lung} DI_{xy}} \quad (10)$$

where the DI_{xy} denotes the value of impedance variation in a tidal image, and the DI_{lung} is the pixel within a detected lung area. The lung area detection is calculated using the value which is 20 percent of the mean tidal value, or in the DCT-based EIT reconstruction algorithm, the anatomical information from the CT image is used for the lung contour.

In a clinical setting the GI value is associated with the disease state of a certain patient, which provides clinicians some indication how to set ventilation therapy [5].

2 Method

For a preliminary evaluation, EIT data from a former study was analyzed retrospectively. The EIT data was recorded by Pulmo Vista 500 (Dräger Medical, Lübeck, Germany). From this dataset two cycles of ventilation are included into the evaluation. For each cycle of ventilation, the sequence of frames was down sampled to a total of 30 frames. Among the 30 frames, we selected 6 frames to represent the different phase of ventilation, where the first frame is the end-expiration, and the last frame the end-inspiration.

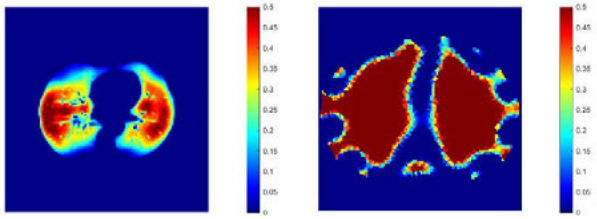

Figure 1: The tidal images generated by different methods. Left: DCT-based EIT tidal image. Right: Classical EIT tidal image.

For each cycle of ventilation, we implemented both one step Gauss-Newton algorithm for a classical EIT reconstruction, and our DCT-based EIT algorithm for a structural-functional image. For both reconstruction method, a tidal image is acquired, which are illustrated in Figure 1. For classical EIT construction, we obtained an estimated lung area using a threshold at 20 percent of the mean tidal variation, while for CT-EIT method, the lung area is derived directly from the CT. Both lung regions are shown in Figure 2.

Figure 2: Lung regions calculated by different methods. Left: DCT-based EIT generated lung region. Right: Classical EIT generated lung region.

For each breathing cycle the 5 selected frames, which represent the different phase of a ventilation cycle, we calculated the median and sum of the variation within the lung area, and further calculated the GI value for every frame for both methods. We further normalized the GI value of each method for every cycle with the largest GI value in that group. The GI value is listed in Table 1 for comparison.

3 Discussion

In this preliminary evaluation between the classical EIT reconstruction and the DCT-based EIT reconstruction, complex time-difference impedance images were represented by an index, the GI. Generally, the GI index varies between the two algorithms, i.e. the GI value seems to depend on the reconstruction algorithm, which could be expected. But unexpectedly, the qualitative results differ. For classical EIT reconstruction, the GI value have a tendency to decreases as the frame reaching the end-inspiration, while for DCT-based EIT reconstruction, the GI index is more stable, and shows cyclic behaviour.

Absolute differences in GI index with respect to the reconstructions can be explained by the different constraints imposed on the images. The Tikhonov prior smoothes images, while the DCT use only less degrees of freedom to reconstruct and a cosine base function. In this study, only the lowest 15 frequencies for x-direction and y-direction, which is in total 225 elements, were used for reconstruction. Compared with the elements of the classical FEM, which usually exceeds thousands of degrees of freedom (parameter values), the reconstruction process is more stable and less artefacts are produced as visible, e.g. as in the lung area estimation in the classical approach.

Table 1: Normalized GI Index Values Calculated from Different Methods

Ventilation Cycles	Frames	GI for DCT-based EIT Method	GI for Classical EIT Method
1	10	0.9953	1.0000
	15	0.9276	0.8430
	20	0.9516	0.7541
	25	0.9857	0.6957
	30	1.0000	0.6859
2	10	1.0000	1.0000
	15	0.8872	0.9132
	20	0.8521	0.8213
	25	0.8856	0.7149
	30	0.9142	0.6792

The DCT method must be further investigated to gain insight into the potential of structural-functional image reconstruction. By fusing CT morphological information naturally to the EIT image, clinicians will be given a better

intuition how to interpret dynamics during ventilation. However, as GI index is considered to provide important information associated with inhomogeneous tidal volume distribution in the lung, a standard GI index must be generated. Thus, more reconstruction methods and other ventilation indices will be included in further steps of this evaluation.

One of the limitation of this study is that only two ventilation cycles from one patient are analysed, which results may not generalize. However, the comparison between different frames already provide an indication that considerable differences can be found in further investigations including multiple patients with diverse pathological states. The DCT-based EIT reconstruction generates images are much easier interpretable for clinicians as the anatomically lung contour is clearly illustrated, and the distribution of air is better allocated to defined lung areas compared with the classical EIT reconstruction.

4 Conclusion

The DCT-based EIT lung imaging method is feasible to reduce the artefacts outside the lung region, and to reconstruct a more clinician-friendly image during ventilation for the patient. The GI index evaluation indicates that the reconstruction process has influence on the distribution of regional gas allocation. Therefore, we will further investigate the influence of different reconstruction methods to clinically relevant indices.

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