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Phase-contrast functional EIT images

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Abstract: We describe a phase-contrast functional EIT measure which is useful for measurements on patients receiving HFOV. The measure calculates the regional phase offset, which appears to provide useful information on regional lung mechanics.

1 Introduction

High frequency oscillatory ventilation (HFOV) is a type of mechanical ventilation which uses rapid pressure oscillations (up to 15 Hz) around a constant distending pressure. Since rapid oscillations permit low tidal volumes, HFOV is understood to act as a lung protective mode of ventilation, and is therefore seeing increasing use, especially for the delicate lungs of preterm infants. One concern with HFOV is that the actual volume delivered to the patient is very difficult to monitor, and EIT has shown significant promise for this application [1], since modern EIT hardware is fast enough to capture the relevant volume changes.

Lung mechanics is typically characterized by parameters of compliance ($C = \frac{\text{L}}{\text{kPa}}$) and resistance ($R = \frac{\text{kPa}}{\text{L·s}}$). EIT-derived measures of volume are mostly proportional to the dynamic compliance, but yeild no information on R. Increases in tissue resistance reflects narrowing of airways and changes in parenchyma. The time constant $\tau = RC$ of tissue introduces a delay in the ventilation signal, which results in a change of phase in the regional EIT signal.

We describe a measure of the regional phase change and an algorithm to calculate it efficiently. We then show an analysis of phase change during pneumothorax [2].

2 Methods

Using phasor notation, measurable quantities correspond to the real component of signals. For a HFOV frequency $\omega = 2\pi f_{\text{HFOV}}$, ventilator pressure (Airways Opening) $P_{\text{AO}} = P_{\text{MAP}} + \Delta P e^{j\omega t}$. For an EIT image voxel v, $V_v - V_{\text{MAP}} = C_v \Delta P e^{j\omega(t-\tau_v)} = C_v \Delta P (e^{-j\omega\tau_v}) e^{j\omega t}$ (assuming appropriate calibration). Thus, EIT amplitude is proportional to the regional C_v, while EIT phase $e^{-j\omega\tau_v}$ is related to the regional time constant.

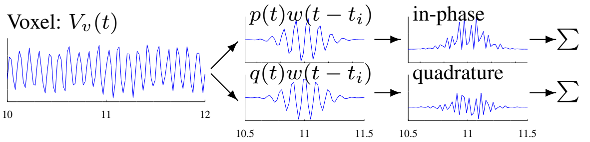

Figure 1: Phase detection: voxel waveform (*left*) multiplied by windowed p, q (*centre*) to produce (*right*) which are summed.

An EIT-phase fEIT image is calculated by processing EIT waveforms as follows (fig. 1).

- Calculate the HFOV frequency and phase, either from the ventilator or from the global EIT signal, from which in-phase ($p(t) = \cos\omega t$) and quadrature ($q(t) = \sin\omega t$) references are calculated.
- If the ventilator and EIT system are not perfectly synchronized, we recommend detecting $p(t)$ via a narrow-band filter, and $q(t) = H(s(t))$ via the Hilbert transform, $H(\cdot)$.
- A window $w(t - t_i)$ is chosen (triangular with 1 s width); at each time of interest t_i, calculate: $p_w = w(t-t_i)p(t-t_i)$ and $q_w = w(t-t_i)q(t-t_i)$
- For EIT voxel v with waveform V_v, $P(t_i) = \int p_w(t-t_i)V_v(t)dt$, and $Q(t_i) = \int p_w(t-t_i)V_v(t)dt$.
- $v(t_i)$ has amplitude and phase of $P(t_i) + jQ(t_i)$.

Phase is not calculated (set to zero) for low amplitude voxels. This approach may be accelerated in the frequency domain, using the relationship $P(t_i) = \int p_w(t-t_i)V_v(t)dt = p_w \otimes V_v$, which can be represented as the multiplication of Fourier transforms. Thus $P(t_i), Q(t_i)$ are samples of a windowed narrow-band filter of V_v.

3 Results and Discussion

Fig. 2 analyses the data of [2] to calculate functional images before and after the onset of pneumothorax. As expected, a large volume of air enters the thorax, as shown in the *Aeration* image. The distribution of ventilation, *Compliance*, shows small changes. Interestingly, there is an contrast in the *Phase* image before the onset of pneumothorax, in a location which predicts the eventual gas buildup.

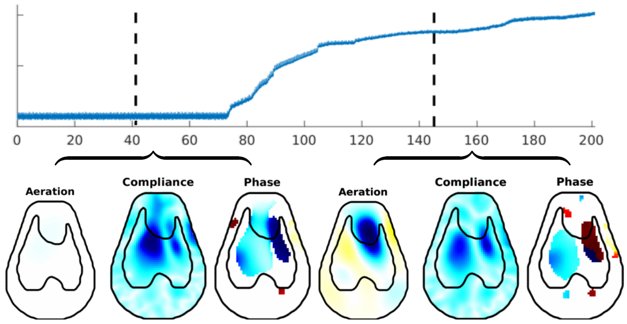

Figure 2: Functional EIT images from data of [2]. *Top:* global EIT signal vs. time (s). Vertical bars indicate time points at which fEIT images are calculated (each scaled individually to maximum value). *Aeration:* ΔEIT with respect to $t = 0$; *Compliance:* Amplitude $\|P + jQ\|$; *Phase:* Phase of $P + jQ$.

In summary, we describe the calculation of phase functional EIT images from EIT data during HFOV ventilation. Such images promise additional information on the distribution of lung mechanical properties. Our example shows a promising application, which we are pursuing in a larger set of data.

A number of engineering challenges to this analysis remain, such as the phase unwrapping of EIT-phase signals, since phase must be consistent across a boundary. We currently do not unwrap phase, since our attempts to use classic approaches did not yield useful images.

References

[1] GK Wolf *et al*, *Ped Crit Care Med*, 11:610–615, 2010

[2] Miedema M *et al*, *Am J RCCM*, 194:116–118, 2016.