Short Distance Impedance Pneumography

Abstract: Gold-standards for biosignal acquisition require body-spanning sensor positioning which is contradictory to the high integration of modern wearable medical monitors. In applications where obtrusiveness can decrease accuracy, as in sleep monitoring, compact sensor configurations are not only a matter of convenience. To acquire respiratory signals, most systems rely on nasal cannula pressure sensors or inductance plethysmography. Another well-established method is the impedance pneumography, where we aim to contribute to the field of short distance electrode configurations. Evaluating distances down to 8 cm we report linear correlations above 0.85 with respect to a pneumotachometer reference. We estimate the respiratory rate with an error below 0.2 bpm. Inspiratory and expiratory phase detection is possible with an error below 2.5%. Using a first order polynomial model we estimated the respiratory flow with a relative error of down to 19% at 8 cm. We conclude that short distance impedance pneumography is feasible and rough flow and volume estimates are possible using linear models. Further research regarding shorter distances and calibration is of great interest.

Keywords: impedance pneumography, short distance, closely spaced electrodes, unobtrusive, medical monitoring, wearables, respiration, flow, volume, rate, phase, sleep

1 Introduction and Related Work

Home sleep monitors became vital in both pre-test scenarios and in certain cases as diagnostic tools. It is now possible, if not typical, that the monitor itself is much smaller than the area covered by its sensors. Gold-standards oftentimes require body-spanning sensor positioning. As to develop truly unobtrusive medical instrumentation, this limitation needs to be overcome by identifying novel, compact sensor sites.

Recording of respiration often relies on obtrusive belts or nasal cannula pressure sensors [1]. We showed that the respiratory rate can be obtained from the electrocardiogram with electrode distances down to 24 mm [2]. For sleep monitoring, however, the estimation of respiratory phases, flow and volume is desirable. A widely accepted method for respiratory signal estimation is the impedance pneumography (IP). However, optimizing electrode configurations is still a subject of ongoing research. Khambete at al. conducted a theoretical study [3]. They concluded that a high sensitivity can be achieved when the electrodes are located in different horizontal planes. Seppä et al. contributed an analysis of electrode configurations for a highly linear relation between impedance and lung volume change [4]. The authors proposed placing two electrodes on the arms and two electrodes at the left and right midaxillary lines. Wang et al. analyzed four different electrode configurations [5]. The authors concluded that placing electrodes on the left and right midaxillary line at the height of the fifth and seventh rib resulted in the most robust IP. Recently, Jeyhani at al. analysed possible short distance electrode locations [6]. The authors proposed a configuration spanning the left half thorax between the upper sternum and the 6th costal line on the left midaxillary line.

We conducted three experiments with electrode distances down to 8 cm and analysed correlation measures, respiratory rate estimation, respiratory phase detection, signal delay as well as flow and volume approximation. We aim to contribute to the further development of short distance IP configurations.

2 Materials and Methods

2.1 Acquisition System

Using the Analog Devices AD5933 network analyzer, we implemented a 1 kHz to 100 kHz variable-frequency bioimpedance measurement system. We extended the device’s frontend by implementing a current source in bridge configuration. Thus we were able to acquire the
bioimpedance using a tetrapolar current injection measurement scheme, bypassing electrode and interface impedances. We were using the airflow measured by a synchronously recorded pneumotachometer (PNT) as a respiratory reference.

2.2 Measurement and Preprocessing

We conducted three experiments with two different electrode positions and six subjects each (25±5 years, 3 female) in an upright sitting position. The configurations are given in Table 1 and visualized in Figure 1. An Electrode pair was defined as on injecting and one measuring electrode. In all experiments, the first electrode pair stayed fixed at the sternal line at the height of the 2nd and 3rd intercostal space (ICS2 and ICS3). In the first experiment, we measured the IP over the left half of the torso, termed \( hT \) hereafter. The second electrode pair was located at the left midaxillary line at ICS4 and ICS5. For the second and third experiment we moved the second electrode pair to a distance of exactly 8 cm from the first electrode pair, approximately to the left midclavicular line. To evaluate the influence of a relative electrode movement, we conducted two 8 cm measurements. One with the electrodes being fixed against each other, named \( 8\text{fixed} \), and one without relative fixation, referred to as \( 8\text{free} \). The measurements were carried out using a 100 kHz sinusoidal current with a 10 µA peak at 500 samples/s.

Table 1: Measurement names and electrode configurations. For all experiments, the electrode pair 1 is located at sternal line, ICS2/3.

<table>
<thead>
<tr>
<th>Name</th>
<th>Electrode Configuration and Fixation</th>
</tr>
</thead>
<tbody>
<tr>
<td>( hT )</td>
<td>Pair 2: left midaxillary line,ICS4/5 (free)</td>
</tr>
<tr>
<td>( 8\text{free} )</td>
<td>Pair 2: left midclavicular line,ICS4/5 (8cm, free)</td>
</tr>
<tr>
<td>( 8\text{fixed} )</td>
<td>Pair 2: left midclavicular line,ICS4/5 (8cm, fixed)</td>
</tr>
</tbody>
</table>

In all scenarios, the subjects performed paced breathing of 1 minute breathing at 15 breaths per minute (bpm), 30 seconds apnoea simulation and 1 minute breathing at 30 bpm.

Following the acquisition, the raw data were low-pass filtered at 4.5 Hz. The signal baselines were removed by subtracting a 6000 point moving average. The PNT signal was used directly as the flow reference. As to obtain a volume reference, the PNT signal was integrated. The absolute impedance was used as a volume signal surrogate. A flow signal surrogate was obtained using the first derivate of the absolute impedance. Both the surrogate flow and volume signals were low-pass filtered at 1 Hz and their baseline was removed by subtracting a 3500 point moving average.

2.3 Correlation and Respiratory Rate

The correlation between the impedance and the integrated PNT signal as well as between the first derivate of the impedance and the PNT signal was assessed using the Pearson linear correlation coefficient \( \rho \). Signals with a \( \rho < 0.5 \) as well as apnea signals have been rejected from further analysis.

We estimated the respiratory rate (RR) in 30 s segments from the absolute impedance with 50 % overlap. The RR estimation was obtained using the maximum of the DFT between 6 and 36 bpm. For the analysis, points with a distance from the median greater than three times the median absolute deviation are considered outliers and are excluded. We use the mean absolute error (MAE) and the mean error (ME) as performance metrics, the latter being an indicator for the bias.

2.4 Respiratory Phase and Signal Lag

Throughout this work, inspiratory and expiratory respiratory phases are defined using the extrema in the volume signals. We analyse the accuracy of the fiducial point detection by means of false positives (FP) and misses (MISS) in percent of the total amount of points.

In addition to the respiratory phases, we derived the signal lags between the flow and volume signals and their surrogates estimated from the impedance by calculating the maximum absolute cross correlations and the corresponding lags.

2.5 Respiratory Flow and Volume

For estimating the respiratory flow and volume in absolute values, a model-based calibration between the impedance and PNT data is needed. We employed first order polynomial models for both flow and volume estimation. The flow model given in Equation 1 was fitted between the first
derivate of the absolute impedance and the PNT reference signal. The volume model given in Equation 2 was fitted between the absolute impedance and the integrated PNT reference signal. For a brief outlook, we also evaluated the performance of higher order polynomials. Prior to the fits, the signal lag had been removed.

\[
\begin{align*}
\text{Flow}_{\text{est}}(t) &= a_r \frac{d}{dt} \sqrt{Z^2_{\text{real}}(t) + Z^2_{\text{imag}}(t) + b_f} & (1) \\
\text{Vol}_{\text{est}}(t) &= a_v \sqrt{Z^2_{\text{real}}(t) + Z^2_{\text{imag}}(t) + b_v} & (2)
\end{align*}
\]

We performed a subject-specific fit using 90% of the data drawn randomly from the measurements and used the remaining 10% of the data for testing in a 10-fold cross validation. We calculated the MAE, ME as well as two relative performance metrics. Firstly, the signal agreement (SA) between the reference \text{Ref} and the estimation \text{Est} as used by Seppä et al. and given in Equation 3 [7]. Secondly, a normalized mean squared error (NMSE) in percent as given in Equation 4.

\[
\text{SA} = 1 - \frac{1}{N} \sum_{i=1}^{n} \frac{[\text{Est}(i) - \text{Ref}(i)]^2}{\text{Var(Ref)}}
\]

\[
\text{NMSE} = \frac{\sum_{i=1}^{n} [\text{Est}(i) - \text{Ref}(i)]^2}{\sum_{i=1}^{n} [\text{Ref}(i)]^2} \times 100
\]

3 Results

3.1 Correlation and Respiratory Rate

The average, measurement specific Pearson correlation coefficients for both the raw flow and volume estimates, (i.e. before applying the model fits) are given in Table 2 together with the MAE and ME of the RR estimation. The flow correlation was slightly higher than the volume correlation. The \text{hT} measurement resulted in a higher linear correlation than the 8 cm measurements, the latter were comparable.

<table>
<thead>
<tr>
<th>Name</th>
<th>(r_{\text{flow}})</th>
<th>(r_{\text{vol}})</th>
<th>(\text{RR}_{\text{MAE,bpm}})</th>
<th>(\text{RR}_{\text{ME,bpm}})</th>
</tr>
</thead>
<tbody>
<tr>
<td>\text{hT}</td>
<td>0.94±0.06</td>
<td>0.92±0.08</td>
<td>0.12±0.09</td>
<td>-0.02±0.05</td>
</tr>
<tr>
<td>\text{8free}</td>
<td>0.88±0.05</td>
<td>0.85±0.10</td>
<td>0.10±0.03</td>
<td>-0.03±0.04</td>
</tr>
<tr>
<td>\text{8fixed}</td>
<td>0.88±0.04</td>
<td>0.86±0.07</td>
<td>0.13±0.02</td>
<td>-0.03±0.04</td>
</tr>
</tbody>
</table>

The relation between the estimated RR and the PNT reference rate for the two 8 cm measurements is illustrated in Figure 2 by a Bland-Altman plot. The respiratory rate was slightly underestimated by -0.03 bpm. The standard deviation below 0.1 bpm. No proportional error was found.

![Figure 2: Bland-Altman plot of the estimated respiratory frequency versus the reference rate in both 8 cm measurements.](image)

3.2 Respiratory Phase and Signal Lag

The results of the fiducial point detection and signal lag analysis are given in Table 3. The false positive rate remained under 1.5% for all measurements. The number of missing points was below 0.5% for the \text{hT} measurement but increased in the 8 cm measurements. The signal lag was below 100 ms and comparable in both the \text{hT} and \text{8fixed} measurement. The lag increased in the \text{8free} measurement to roughly 150 ms.

<table>
<thead>
<tr>
<th>Name</th>
<th>FP%</th>
<th>MISS%</th>
<th>\text{Lag}_{\text{flow,ms}}</th>
<th>\text{Lag}_{\text{volume,ms}}</th>
</tr>
</thead>
<tbody>
<tr>
<td>\text{hT}</td>
<td>1.23±1.19</td>
<td>0.35±0.61</td>
<td>47±35</td>
<td>53±42</td>
</tr>
<tr>
<td>\text{8free}</td>
<td>0.56±0.64</td>
<td>4.26±7.71</td>
<td>143±60</td>
<td>145±61</td>
</tr>
<tr>
<td>\text{8fixed}</td>
<td>1.32±0.99</td>
<td>2.42±4.84</td>
<td>79±103</td>
<td>83±113</td>
</tr>
</tbody>
</table>

3.3 Respiratory Flow and Volume

The average results of the 10-fold cross validation using the first order polynomial models for flow and volume estimation are given in Table 4 and Table 5 respectively. In both the flow and volume estimation, the \text{hT} measurement outperformed both 8 cm measurements which showed comparable results. The signal agreement is slightly higher for the \text{hT} measurement than for the 8 cm measurements, which differ inconsistently with regard to flow and volume estimation. The ME is small for all models and measurements, thus the method shows only a small bias. The flow estimation is more accurate than the volume estimation. Figure 3 shows example waveforms for the reference and estimated respiratory flows in an \text{8fixed} measurement setup.
Using third order polynomial models, the relative error for the flow estimates decreased to 13 % for the hT and 17 % for the 8 cm positions. For the volume estimations, the error decreased to 19 % for the hT and 25 % for the 8 cm positions. Higher order polynomials showed no performance increase.

### 4 Summary and Discussion

We found a high linear correlation of 0.94 for flow and 0.92 for volume signal estimation for the hT measurement. The correlation decreased with distance to about 0.86. We propose two main reasons therefore. Firstly, the relevance of cardiogenic oscillations in the impedance increases with decreasing distance, thus requiring for more rigorous filtering. A solution was proposed by Seppä et al [8]. Secondly, the linearity of the IP is highly dependent on the electrode positioning [4]. Respiratory rate estimation was highly accurate with a very small bias below -0.03 bpm and an MAE below 0.2 bpm. The accuracy was not affected by the position.

The fiducial point detection for respiratory phase estimation showed good results for the hT measurement with false positive rates below 1.5 % and missing point rates below 0.5 %. The false positive rate was not consistently affected by the measurement position. The missing point rate increased to 2.4 % for the fixed and 4.3 % for the free 8 cm configuration. A signal lag below 100 ms was found in both the hT and 8fixed measurements which increased to 150 ms in the 8free case.

The first order polynomial flow and volume calibration showed small biases. The hT measurement flow and volume estimation showed a signal agreement of 0.87 and 0.85 which is comparable with the literature. The relative error was 16 % and 20 %. For the 8 cm measurements, the flow estimation error increased slightly by 3 %, the volume estimation error by 9 %. Third order polynomial models reduced the error by about 3 to 5 % in all measurements. Higher order models showed no significant performance increase. The remaining error is largely caused by low-frequency in-band components. higher calibration accuracies might be possible by following the findings of Młyńczak et al. [9]. An additional finding of the authors was the long-term usability of a single calibration.

The signal agreement dropped inconsistently between 8free and 8fixed measurements. The biases remained unaffected. The high relative error and low bias point towards an increasingly nonlinear relationship between the surrogate and physiological signals with decreasing distance. Thus, nonlinear models should be considered. Młyńczak et al. proposed the use of feedforward neural networks [10].

![Figure 3: Example signals for a fixed electrode 8 cm measurement.](image_url)

### 5 Conclusion

Respiratory rate estimation and phase detection are possible to a high degree of accuracy in IP measurements with 8 cm electrode distance. Respiratory flow and volume estimation using a first order polynomial model are feasible for rough estimates. More advanced and possibly nonlinear models are recommended for a more accurate estimation. Fixation showed no consistent influence on the results.

### Author Statement

Research funding: The author state no funding involved. Conflict of interest: Authors state no conflict of interest. Informed consent: Informed consent has been obtained from all individuals included in this study. Ethical approval: The research related to human use complies with all the relevant national regulations, institutional policies and was performed in accordance with the tenets of the Helsinki Declaration, and...
has been approved by the authors' institutional review board or equivalent committee.

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