NON-INVASIVE VENTILATION MONITORING DURING CARDIOPULMONARY RESUSCITATION BY MEANS OF THORACIC IMPEDANCE

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Abstract: Recent studies report suboptimal quality in the performance of cardiopulmonary resuscitation (CPR). The continuous measurement of ventilation performance during CPR would therefore be desirable in order to identify deviations relative to the current guidelines and facilitate improvement of the overall quality of CPR. We studied the relationship between lung volume changes and resulting thoracic impedance changes measured by a modified defibrillator. This was done to evaluate the potential of using thoracic impedance for estimation of ventilation rate, inspiration time and tidal volume (TV). A mean correlation coefficient ρ of 0.971 was found between the respiratory and impedance waveforms of a ventilation cycle before noise filtering. The similarity in wave shape shows that the technology has potential for use in estimation of ventilation rate and inspiration time. Our results also show that thoracic impedance is not suitable for TV estimation because of large variability between patients in the assumed linear relationship between TV and resulting impedance change.

Introduction

Recent studies [1, 2, 3, 4] report that the quality of out-of-hospital cardiopulmonary resuscitation (CPR) performed by medical personnel is low compared to the recommendations [5]. Even in the optimal in-hospital environment of highly trained and experienced health care professionals, CPR often does not meet the current guidelines. Aufderheide et al recently reported that rescuers consistently tend to hyperventilate out-of-hospital cardiac arrest patients [4], and a pig model has shown that increased tidal volumes adversely affect cardiac output [6]. Wik et al [7] demonstrated that a too short inspiration time was a common problem in a manikin model where emergency medical service personnel delivered mouth to mouth ventilations. To overcome the problem of suboptimal CPR, Handley et a [8] suggested to incorporate a feedback system in automated external defibrillators (AED), facilitating optimal ventilations through automated regulation of ventilation rate, inspiration time and tidal volume.

This calls for the continuous measurement of ventilation rate, tidal volume and inspiration time during CPR. Thorax impedance has been used for ventilation monitoring since the mid sixties [9]. The thorax impedance changes increases with the amount of air in the lungs. The increase is thought to arise from the thinner and longer conduction path caused by an increase in air volume in the lungs . Pellis et al [10] measured ventilations in pigs by thorax impedance through defibrillator pads. They suggest that the same method could be used to monitor human ventilation activity during resuscitation. We therefore wanted to study the relationship between patient ventilations and the corresponding impedance changes measured by a modified AED with standard defibrillator pads in lead II position. The relationship between ventilations and TI changes was established through correlation and regression analysis on measurements from 32 unconscious patients.

We first present the patient material and the equipment used for the respiratory and thorax impedance measurements, which is followed by a description of the methods used to quantify different aspects of the relationship between the two biomedical signals. We then present the results of our study, followed by a discussion and conclusion.

Materials and Methods

The study was carried out at an emergency department of a tertiary care university hospital. The study procedures were in accordance with the ethical standards of the responsible committee on human experimentation.

Data was collected from 32 patients (23 male, median age of 50, interquartile age 44-59) in hemodynamically stable controlled mechanical ventilated

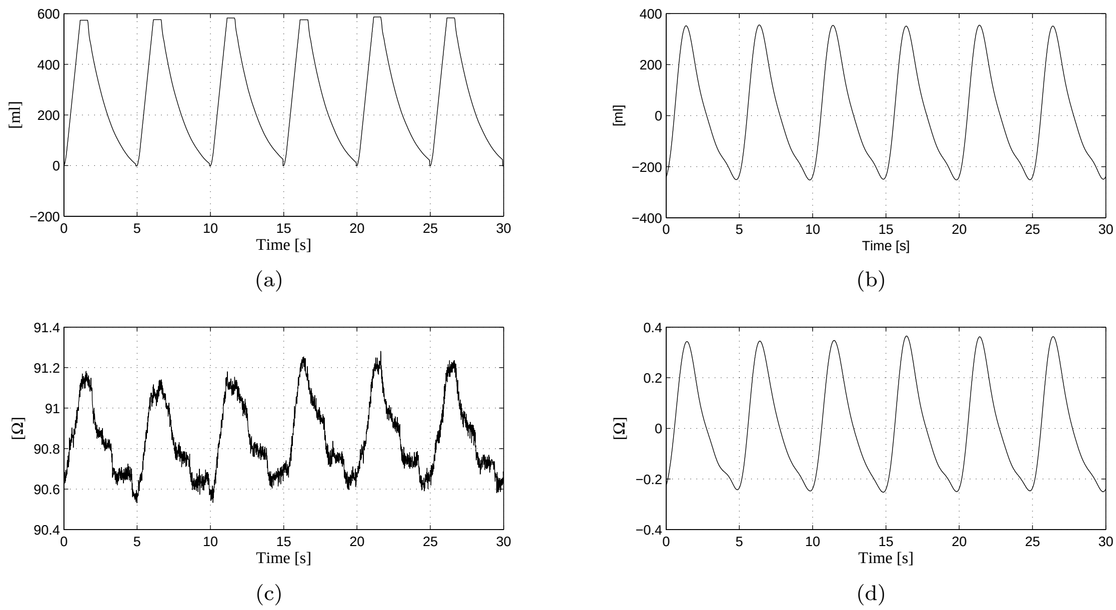

Figure 1: An example volume trace of 30 s (a) before and (b) after filtering, and resulting impedance trace (c) before and (d) after filtering. The close relationship is easily observed.

conditions at the University Clinic of Vienna municipal hospital. Weight ranged from 50 to 120 kg (median 76, interquartile weight 65-89.5). All patients were ventilated with a ServoI® Ventilator system (Version1.2, Siemens Medical Group, Frankfurt, Germany), which also continuously measured the respiratory volume. Sampling rate was 100 Hz and dynamic range was 16 bits. The resolution was 0.1 ml/bit. An investigational defibrillator, based on a commercially available defibrillator (Heartstart®4000SP Laerdal Medical Cooperation, Stavanger, Norway) was used for the impedance measurements in the study. A 32 kHz sinusoidal excitation current, 3mA peak-to-peak, was applied between the defibrillation pads, and the resulting impedance was registered. The resolution of the defibrillator impedance measurement system was 0.74 mΩ/bit. Sampling rate was 500 Hz and dynamic range was 16 bits. Heartstart®4000SP with the necessary analysis software were provided by Laerdal Medical. The thorax impedance measurements were recorded using commercially available self-adhesive electrode defibrillator pads (Heartstart Pads®, Philips Medical Systems, Seattle, WA, USA). The impedance measurements were downsampled to 100 Hz to be compatible with the volume measurements.

The patients were ventilated at tidal volumes (TV) of 400, 600, 800 and 1000 ml, for 2 minutes at each TV. TV is the amount of air inhaled and exhaled during one respiratory cycle. The ventilation and impedance measurements were imported into MATLAB® (The Mathworks®, Inc., Natick, MA, USA) for further processing. The impedance measurements were downsampled to 100 Hz to be compatible with ventilator data.

According to Baker et al [9, 11], the relationship between lung volume change and impedance change for an individual is essentially linear. The potential of TI-based estimation of ventilation rate, volume and inspiration time is strengthened by a linear relation between the volume and the TI measurement. We therefore want to quantify the linearity of the relationship. For each ventilation cycle, we let $\mathbf{v} = [v(0), v(1), ..., v(n), ..., v(N)]^T$ represent the volume measurement and $\mathbf{z} = [z(0), z(1), ..., z(n), ..., z(N)]^T$ represent the impedance measurement. The respiration cycle is represented with N samples. It is made sure that both the volume and impedance measurement starts at 0 by subtracting $v(0)$ from $\mathbf{v}$ and $z(0)$ from $\mathbf{z}$. We then solve the equation

$$\mathbf{z}\alpha = \mathbf{v} \tag{1}$$

to find the regression coefficient α that best describes the relationship between $\mathbf{v}$ and $\mathbf{z}$ in a least squares sense. The worst case sample deviation between the volume and the regression line, e_{max}, is found as

$$e_{max} = \max_{\text{all } n} |v(n) - z(n) \cdot \alpha| \tag{2}$$

It is used as a measure of the strength of the linearity assumption. e_{max} gives an impression of how much the estimated volume curve will deviate from the true volume curve, if we know $\mathbf{z}$ and α. The correlation coefficient ρ of the two signals is calculated for each ventilation cycle to further quantify the shape similarity. It is found as

$$\rho = \frac{(\mathbf{v} - \mu_v)^T (\mathbf{z} - \mu_z)}{\sqrt{\sigma_v \sigma_z}} \tag{3}$$

where μ_v and σ_v is the mean and variance of $\mathbf{v}$, and μ_z and σ_z is the mean and variance of $\mathbf{z}$. For each

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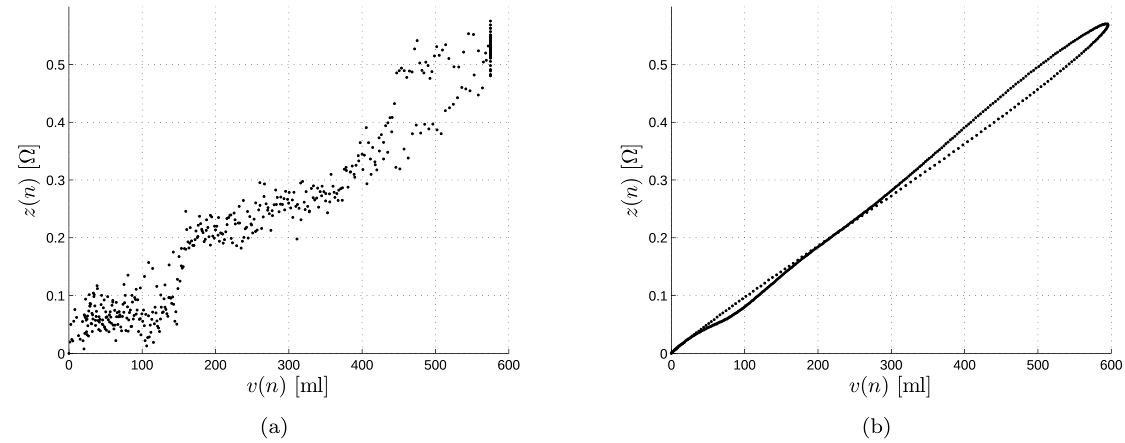

Figure 2: An example ventilation trace of the volume, represented by **v**, and the impedance, represented by **z**, of one ventilation cycle plotted against each other (a) before and (b) after filtering.

patient and we calculate e_{max} and ρ at each TV for an equal number of ventilation cycles. We then calculate the mean and the standard deviation of each parameter to get an impression of their distribution.

We also want to explore the improvement in correlation between **z** and **v** when removing pulse artifacts [10, 7] and baseline drift in the impedance channel. For this purpose we use a finite impulse response equiripple bandpass filter designed using MATLAB®. The pass band of the filter is from 0.06 Hz to 0.66 Hz, with stop band below 0.01 Hz and above 0.71 Hz. The attenuation of the stop bands is -20 dB, which gives a filter of 2836 filter taps. The long delay introduced by the filter makes it unsuitable for real-time application, and it is only designed to isolate the respiratory-related information found in the impedance channel. The filter is used on both the impedance and the volume measurements to impose the same effects on both signals. The calculations of e_{max} and ρ are then carried out on the filtered signals in the same manner as described above.

The potential of estimating the tidal volume based on thorax impedance measurement is also dependent on the variability of the assumed linear relation between patients. We therefore find the linear relationship for each respiration cycle, represented by α, as

$$\alpha = \frac{z(n_{max})}{\mathrm{TV}}[\Omega/\mathrm{ml}] \quad (4)$$

where n_{max} is the sample number of the maximum value of **z**. $z(n_{max})$ is the impedance change used for TV estimation if α is known. α is termed the impedance coefficient [9]. We study the distribution of α within each patient to evaluate the TV estimation accuracy of the method over time if calibrated for each patient. It has been shown that α is correlated with body weight [12], and we therefore divide the α by the patient's weight to obtain the specific impedance coefficient, α_w $[\Omega/ml/kg]$. We then calculate α and α_w for each patient as the mean of the coefficients calculated

Table 1: The mean and standard deviation of e_{max} after filtering at different tidal volumes

TV	Mean (std. dev.) of e_{max} [ml]	
	Before filtering	After filtering
400 ml	85.51 (32.09)	6.46 (5.37)
600 ml	120.79 (58.17)	10.39 (9.74)
800 ml	143.81 (78.35)	13.33 (9.01)
1000 ml	185.46 (84.71)	16.38 (12.53)

for each respiration cycle. We find α and α_w for all ventilations, and for ventilations at 400, 600, 800 and 1000 ml. The distribution of α and α_w for the whole patient group is then studied, which gives us an impression of the generality when considering the potential for TV estimation based on impedance measurements.

Results

Figure 1 shows a volume trace from a patient and the corresponding thorax impedance signal before and after filtering. In Figure 2, the volume trace is plotted against the resulting impedance before and after filtering, and the linear behavior of the relationship is easily observed along with the improvement in correlation from filtering. We then calculated e_{max} for all ventilations at tidal volumes of 600 ml before and after filtering. The mean e_{max} was 120.79 ml, with a standard deviation of 58.17 ml, while this value was reduced to mean e_{max} of 10.39 ml, with standard deviation 9.74 ml, after filtering. The means and standard deviations of e_{max} before and after filtering are presented in Table 1. We then calculated the mean and standard deviation of ρ between the tidal volume waveforms and the impedance waveforms. The mean and standard deviation before filtering was 0.971 and 0.027, and 0.999 and 0.001 after filtering.

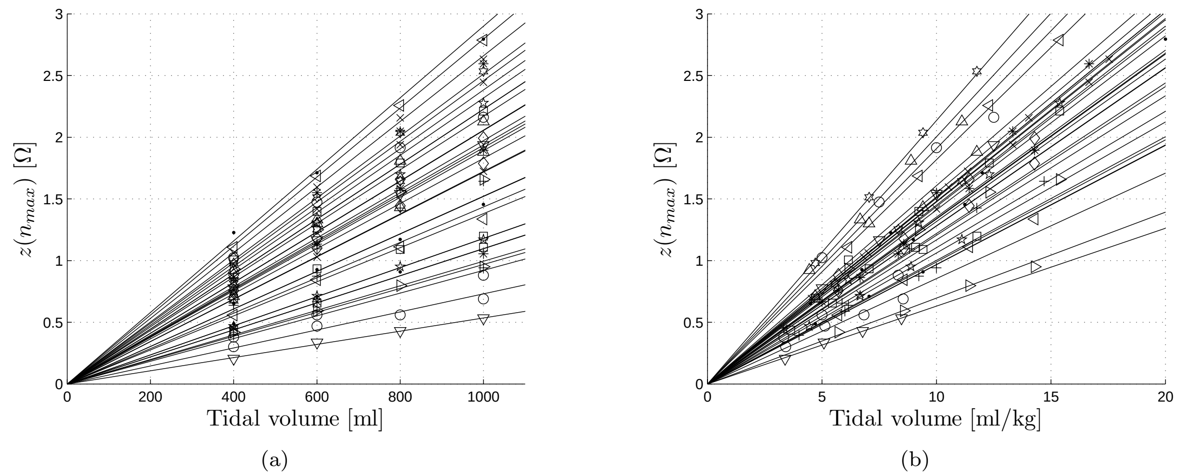

Figure 3: Mean thorax impedance change for each patient at tidal volumes of 400, 600, 800 and 1000 ml, with the tidal volumes expressed in ml (a) and ml/kg (b). The lines represent the estimated linear relationship between tidal volume and impedance change. Each patient is represented with one regression line and one set of markers for each tidal volume the linear relationship is estimated for.

We then study the distribution of α and α_w. The relative standard deviation of α has a mean of 0.074 within each patient. The distribution of α and α_w within the patient group are illustrated in Figure 3, where a line represents the estimated linear relationship between TV and $z(n_{max})$ for one patient. The mean (standard deviation) was 0.00195 (0.00066) Ω/ml for α and 0.148 (0.035) $\Omega/ml/kg$ for α_w. It is seen that the percentwise standard deviation for α_w (23.6%) is lower than for α (33.7%).

Discussion

As indicated by the example in Figure 3(a) and the results in Table 1, the volume and impedance waveform are very similar in shape. The similarity can be further improved by use of filtering. There is therefore potential for development of reliable detection of ventilatory activity. The similarity in waveform gives us an easy and accurate way of estimating the inspiration time once the ventilation cycle has been detected, along with the ability to give feedback on the ventilatory rate. The measurements have however shown that the impedance signal is very sensitive to movement and has a significant baseline drift. Chest compressions and handling of the patient will therefore generate noise and make the detection of ventilations difficult. Similar noise issues have been addressed in connection with ECG analysis during compressions [13], where adaptive filtering techniques are used to remove compression artifacts.

The low variability of α over time within each patient suggests that once α is known, TV can be estimated based on impedance changes with fair accuracy. This can be illustrated with an example. We find that $\alpha = 0.002$ for a patient, and observe a impedance change of 1.0 Ω for a ventilation cycle. The TV is estimated to be 500 ml, and having the estimate of α's relative standard deviation of 0.074, we find the standard deviation of our TV estimate to be approximately 40 ml. The high variability of α seen in Figure 3(a) shows that the technology does not have good generality when used for TV estimation. The results when comparing the relative standard deviation of α versus α_w indicates that the generality is slightly improved when considering estimating TV in ml/kg instead of in ml. If we compare tidal volume control using impedance with tidal volume control using chest rise [14], chest rise will result in an average volume of 400 ml (range 200 - 600 ml). If we let an impedance change above 0.6 Ω indicate a ventilation of sufficient ventilation, our data show that this will give an equal average volume of 400 ml, but the range will be much larger (range 200 - 1100 ml). The variability is however diminished some by using α_w. If we follow the above example, we will get an average volume of 4.8 ml/kg (range 2.8 - 9.5 ml/kg). The variability is however still large, and chest rise will give a tidal volume that is more accurate than using impedance change.

Conclusions

The good correlation and near linearity of the relationship between lung activity and TI change gives the technology good potential for rate and inspiration time estimation in a CPR setting. The low variability of α within each patient makes the behavior of the relationship predictable once it is known, and tidal volumes can then be estimated with a fair accuracy. The high variability of α between patients will however make a general model for TV estimation based on TI changes inaccurate.

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