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Signal-to-Motion Artifact Ratio Versus Frequency for Impedance Pneumography

Javier Rosell and John G. Webster

***Abstract*—We measured transthoracic impedance between 12.5 and 185 kHz in nine adults. We used a system with two impedance channels, both simultaneously detecting the real part of impedance at two different frequencies. We used only two electrodes in the midaxillary line, connecting both channels in parallel. The amplitude relation between the two channels was measured for different maneuvers and frequencies. Results show for normal breathing an increase of the signal of 20% and a decrease in motion artifacts from 12.5 to 185 kHz. We conclude that, for the maneuvers studied, it is better to work at higher frequencies than the ones commonly used. Also, we suggest a method to further increase the signal-to-motion artifact ratio based on measurement at two frequencies.**

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J. Rosell is with the Departament d'Engiyeria Electrònica, Universitat Politècnica de Catalunya, Barcelona, Spain.

J. G. Webster is with the Department of Electrical and Computer Engineering, University of Wisconsin, Madison, WI 53706 USA.

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I. INTRODUCTION

Transthoracic impedance is widely used for monitoring ventilation in infants and adults. A major problem is the presence of motion artifacts [1], [2]. Some experimental systems use a four-electrode measurement technique to reduce the artifacts, but the improvement of the signal-to-artifact ratio is not obvious [2] and, in fact, commercial monitors use only two electrodes. The origin of the signal is also not well-established. There are at least two major sources that could produce a change in the transthoracic impedance: the first is a change of the volume and geometry of the ribcage during breathing [3], and the second is the change in pulmonary tissue impedance [3]–[5] due to the changing fraction of air in the lungs and also to the lung perfusion. But there are other possible sources, such as changes in electrode position, electrode impedance, or diaphragm movement.

Impedance measurement at multiple frequencies is a very active research area [6]. It has been used to evaluate fat-free mass and intracellular-extracellular compartments in the body. In the trunk, measurements over the frequency range from 9.6 kHz to 614 kHz have been reported using a four-electrode method between the upper arm and the calf [7] and also of transthoracic impedance at two frequencies [8]. The feasibility of *in vivo* spectroscopy using techniques of electrical impedance tomography is also being investigated [9].

Pneumograph measurements could be made over a wide range of frequencies. The amplitude of the total transthoracic impedance decreases with increasing frequency [7], [8]. Also, Pacela reported in [10] that the change in impedance increases with increasing frequency, but the data are unpublished. The increase of electrode-skin impedance at lower frequencies makes it more practical to work at greater than 10 kHz. Frequencies well over 100 kHz are not desirable because of possible radio frequency interference and artifacts arising from the coupling of the subject to ground [10]. Thus, commercial equipment operates at a single frequency between 10 kHz and 100 kHz.

This study examines the effect of frequency on the amplitude of the ventilatory signal and the signal-to-motion artifact ratio in order to guide the choice of working frequency for impedance pneumography and to see if more ventilatory information could be gained using multiple frequencies.

II. METHODS

We measured nine subjects in a supine position (eight males and one female) with no known respiratory abnormalities who were between 22 and 44 years of age and weighed from 52 to 87 kg. Two Ag/AgCl disposable electrodes were placed at the midaxillary lines one centimeter under the nipples. The electrodes used (Signa II, Burdick Corp.) exhibit an electrode-skin contact impedance of less than 1 kΩ at 12.5 kHz.

We used a custom two-channel impedance plethysmograph with coherent in-phase detection. One of the channels was set at a fixed frequency (57 kHz) and the other one was able to inject currents at 12.5, 25, 100, or 185 kHz. Current injected was 0.07 mA_{rms} at each frequency. The measured phase error was less than one degree at all the frequencies. The bandpass for both channels (−3 dB) was from 0.03 Hz to 10 Hz.

The protocol consisted of seven maneuvers:

1) no breathing (10 s),
2) normal breathing (20 s),
3) arm movement without breathing (15 s),

4) ribcage breathing (30 s),
5) leg movement without breathing (15 s),
6) abdominal breathing (30 s), and
7) simulated obstructive apnea (15 s).

This protocol was repeated four times. Channel #1 was always at 57 kHz and channel #2 at 12.5, 25, 100, and 185 kHz successively.

During maneuver #1, a 2-Ω resistance in series with one electrode was short-circuited four times using a reed relay in order to calibrate the sensitivity of both channels with the same load conditions. Also, the cardiogenic signal was measured during this interval.

For maneuver #3, subjects were instructed to slowly raise the arms until the hands reached the head and then return the arms to the sides. The leg movement consisted of raising both legs, bending the knees until the upper legs were perpendicular to the torso, and then returning to the original position.

For the simulated obstructive apnea maneuver, subjects were instructed to close their glottis and attempt to move air from the abdomen to the ribcage and back again by moving the muscles of their diaphragm.

For maneuvers #4 and #6, the subjects were instructed to concentrate on the ribcage movement or on the abdominal muscles, respectively, to perform the ventilation.

The analog output of both channels was acquired using a Macintosh II computer with an A/D card (National Instruments NB-MIO-16X). The sampling frequency was 50 Hz per channel. LABVIEW was used for storage and data analysis.

For each subject, maneuver, and pair of frequencies, the amplitude relation between the real part of the impedance signal at 57 kHz and the other impedance at a variable frequency was calculated using a linear least square fit. The calibration signal allows us to cancel the gain differences in both channels. In this way, we obtain for each subject the relative impedance sensitivity (S) versus frequency referred to the impedance at 57 kHz

$$S = \frac{\Re[Z(f_1)]}{\Re[Z(57 \quad kHz)]}. \tag{1}$$

III. Results

Fig. 1 shows the typical waveforms for a pair of frequencies and the relative impedance sensitivity (S) for the different maneuvers in our protocol. The peak-to-peak amplitude of the signal for normal breathing was in the range of 2 Ω to 6 Ω depending on the subject. For the same subject, there were no large amplitude differences between normal, ribcage, and abdominal breathing. The artifact waveform was similar to the normal breathing signal because we instructed the subjects to move the limbs with a repeated interval similar to the breathing rate. The signal amplitude produced by limb movement showed large intersubject variations, from 0.6 to 7 times the amplitude for normal breathing.

In all the subjects and frequencies, arm movement yielded the biggest motion artifact. Simulated obstructive apnea yielded the smallest artifact. The leg movement artifact was always smaller than the arm movement artifact, but there were large intersubject differences. We attribute these differences to the fact that some people in this maneuver contract the abdomen more than others, and some of them also move the upper thorax, resulting in a combination of different motion artifacts sources. For two subjects, the protocol was repeated using another type of electrode (Silver Sircuit Premie from Sentry). The results were similar to the ones obtained with the Signa II electrodes.

Table I shows the relations between the amplitudes at 57 kHz and the other frequencies for breathing and motion artifacts (mean ± standard deviations).

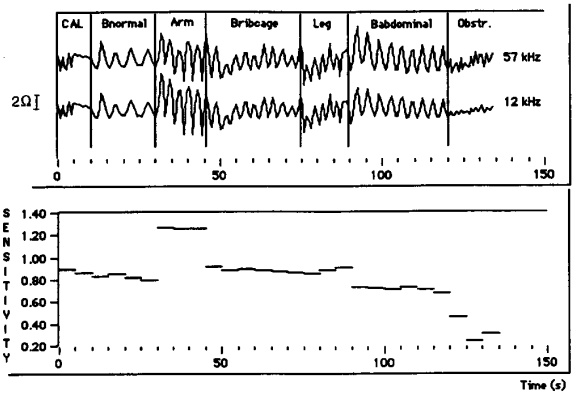

Fig. 1. Transthoracic impedance signals for breathing and motion artifacts. (Top) Real part of impedance at 57 kHz and 12.5 kHz. (Bottom) Relative impedance sensitivity S.

TABLE I
Relative Impedance Sensitivity S Versus Frequency for Different Movements. Mean ± Standard Deviation

Freq.	Ventilation			Motion artifact		
(kHz)	Normal	Ribcage	Abdomin.	Arms	Legs	Obstruct.
12.5	0.83±0.05	0.89±0.08	0.82±0.04	1.59±0.50	1.23±0.36	0.78±0.23
25	0.93±0.10	0.96±0.04	0.90±0.05	1.17±0.12	1.07±0.11	0.99±0.32
57	1.00	1.00	1.00	1.00	1.00	1.00
100	1.07±0.04	1.07±0.06	1.07±0.04	0.94±0.02	0.96±0.04	1.07±0.13
185	1.11±0.04	1.09±0.04	1.10±0.06	0.84±0.10	0.95±0.08	1.21±0.16

A clear decrease with frequency is observed for motion artifacts, especially for arm movement. For simulated obstructive apnea, the dispersion is very large; for some people, it decreases with frequency, while for others it increases. Also, in some cases, the signal amplitude is so small that the correlation between the signals at different frequencies is near zero.

Fig. 2 shows the improvement of signal-to-artifact ratio (SAR) at higher frequencies for arm and leg movements. For arm movement, which is the most important artifact, we obtain an improvement of SAR of around 30% working at 57 kHz instead of 12.5 kHz, and it is possible to increase it 14% more working at 185 kHz.

The cardiogenic signal is almost constant over the frequency band used. Nevertheless, using a four-electrode technique with sensing electrodes at the level of the nipples, we note a decrease of the amplitude with frequency in accordance with [8]. We note also that the amplitude of the cardiogenic signal and its shape at all frequencies depend on the air volume in the lungs.

IV. Discussion

These results show that working at higher frequencies improves the signal-to-artifact ratio. The upper frequency will be limited by the problem of subject-to-ground coupling and for direct coupling between the wires. Possible radio-frequency interference in the AM band also imposes a limit.

The increase of the ventilatory signal as frequency increases is opposite to the results of [7]. In that work, the thoracic impedance between the legs and the arms was measured, instead of the transtho-

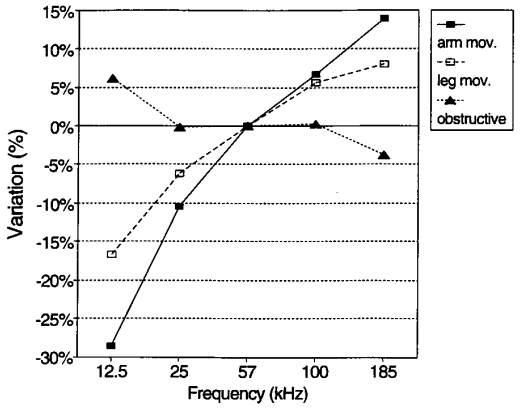

Fig. 2. The signal-to-artifact ratio (SAR) versus frequency for different motion artifacts.

racic impedance we measured. Nevertheless, to find out possible measurement errors in our system, we measured the same longitudinal impedance in two people and we also found a decrease of the ventilatory signal.

In our case, the increase of the ventilatory-related signal with frequency could be produced by the decrease of the lung impedance at higher frequencies [5], allowing more current to flow through the lungs and thus increasing the sensitivity to impedance changes in the lungs produced by changes in air volume. The decrease of motion artifacts with frequency could be explained by the smaller electrode-to-skin impedance at higher frequencies.

Table I shows that at low frequencies the sensitivity for abdominal breathing differs from that of ribcage breathing, a t-test for dependent samples on the row data give a difference of 0.07 ($p = 0.06$) at 12.5 kHz and 0.057 ($p = 0.02$) at 25 kHz. A single frequency system would require pairs of electrodes for both the ribcage and the abdomen to decrease the error in the estimation of ventilation volume. By using two frequencies, we obtain information about the type of breathing and this could be used to decrease the error.

For obstructive apnea, the origin of the different frequency behavior for different people is not clear. We hypothesize that the level of the electrodes with respect to the diaphragm and the differences in the way the maneuver is done could be the reason for these discrepancies. Further work has to be done using different electrode locations to test these hypotheses.

Fig. 3 shows that we can use a system with two frequencies to distinguish some motion artifacts from normal breathing based on their different impedance changes. The method consists of obtaining the amplitude relation (S) for normal breathing for a given person and then multiplying the 57-kHz signal by S and subtracting the low-frequency signal. For normal breathing, this will yield a signal close to zero, but in the presence of motion artifacts, S will change and we will obtain a signal related to the movement. The intersubject variability on S for different maneuvers only has effect on the amplitude of the output signal during movement artifacts. Furthermore, an increase of SAR could be obtained using this signal as a noise reference in an adaptive filter.

V. Conclusions

We found that the ventilatory signal increases and the motion artifacts produced by limb movement decrease from 12.5 kHz to 185 kHz. Thus, in order to optimize the signal-to-artifact ratio

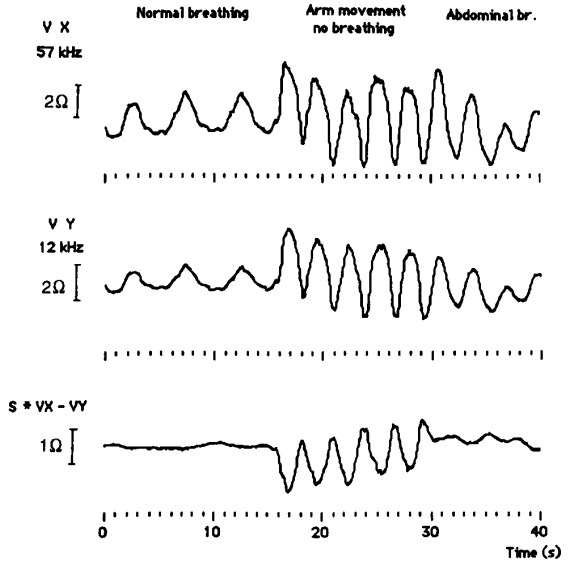

Fig. 3. Detection of motion artifacts using two impedance channels at different frequencies. The amplitude of the output signal (bottom) increases during motion.

(SAR) using two electrodes for electrical impedance pneumographic systems, the frequency must be as high as possible in this band.

We have shown that the signals from two channels at different frequencies are not identical and this could be used to detect motion artifacts and increase the SAR in dual-frequency pneumographic systems. The advantage of such a system is that it does not require more than two standard electrodes and common cables and connectors.

Multifrequency measurements could be used to more accurately determine the sources of the ventilatory signal and motion artifacts.

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