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Cancellation of Respiratory Artifact in Impedance Cardiography

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***Abstract*—Impedance cardiography is a noninvasive technique for monitoring stroke volume, based on sensing variation in the thoracic impedance, $z(t)$, due to blood flow. In this technique, first derivative of impedance, dz/dt, is used to calculate two parameters: ventricular ejection time and $(-dz/dt)_{max}$. Respiration and motion artifacts cause base line drift in the sensed impedance waveform, mainly during exercise, and this drift results in errors in the estimation of the two parameters. Ensemble averaging of $z(t)$ signal suppresses motion artifacts but it introduces distortion in the signal. In this study, simultaneously acquired respiratory signal is used for cancellation of corrupting respiratory artifact from the recorded $z(t)$ signal, using adaptive noise cancellation technique.**

***Keywords*—Adaptive cancellation, Respiratory artifact, Impedance cardiography.**

I. Introduction

IMPEDANCE cardiography is a noninvasive technique for monitoring stroke volume, based on sensing the changes in the electrical impedance of the thorax, $z(t)$, caused by variation in blood volume during the cardiac cycle [1]-[16]. Time derivative of the thoracic impedance ($-dz/dt$) is known as impedance cardiogram (ICG). The parameters required for estimating the stroke volume, using Kubicek, Bernstein, or Sramek formulas [1], [2], [11] are left ventricular ejection time (T_{lvet}) and maximum change in the impedance cardiogram, $(-dz/dt)_{max}$.

Fig. 1 shows ICG and other related waveforms. Ventricular ejection time is defined as the time difference between points B and X in ICG waveform. The point B, often occurring at onset of rapid upstroke of ICG, corresponds to opening of the aortic valve. While the aortic valve closure is identified as point X, associated with the minimum following the maximum in ICG. Point X corresponds to second heart sound in phonocardiogram (PCG). Point B is generally identified as base line crossover point before maximum peak of ICG. DeMarzo *et al* [8] observed that B-point can occur at any point on the ascending segment of the waveform in some patients. Further, sensing of the variation in thoracic impedance due to blood flow is influenced by respiration and motion artifact [2], [5], [10], [14], [15]-[20]. These artifacts have large amplitude as compared to impedance variation due to cardiac cycle, and can cause a large base line drift. ICG ($-dz/dt$) signal bandwidth extends over 0.8-20 Hz. Respiratory related artifact extend over 0.04-2 Hz, while motion related artifact have band of 0.1-10 Hz [9], [15]-[17]. Therefore spectra of the motion and respiratory artifacts may partly overlap with that of the ICG. This causes errors in detection of T_{lvet} and $(-dz/dt)_{max}$ from ICG waveform.

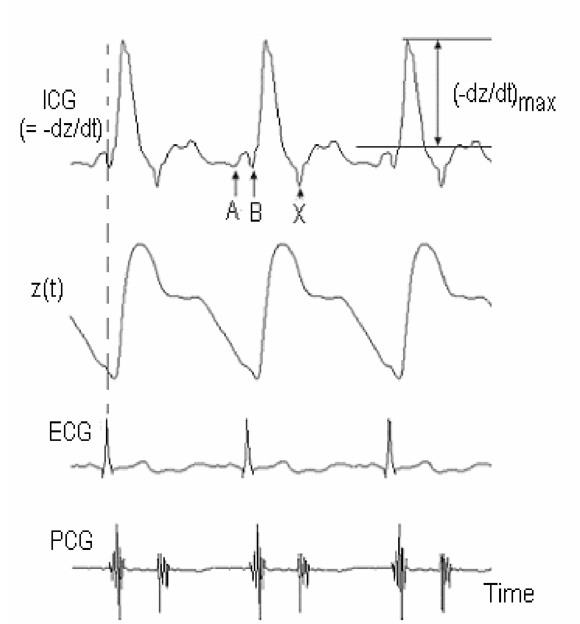

Fig. 1. Typical impedance cardiogram (adapted from [3]). $z(t) = Z_o - Z(t)$ where $Z(t)$ is thoracic impedance and Z_o is the basal impedance.

Holding the breath during recording can be used to avoid respiratory artifact. However, it may change the stroke volume and holding of breath is generally difficult during or after exercise.

Ensemble averaging is most commonly used technique for suppressing artifact in the ICG signal [5], [6], [10], [15]. In ensemble averaging, time frames are decided with respect to some internal reference point or with respect to a reference point in another waveform. Hence simultaneously acquired electrocardiogram (ECG) R-points are used to decide time frames and ICG is ensemble averaged on beat-by-beat basis, synchronized with the R-point of ECG. However, ensemble averaging suppresses beat-to-beat relation and transient changes in ICG signal. Because of heart rate variability, ensemble averaging tends to blur or suppress the less distinctive point B of the waveform and may result in error in its detection [10], [11], [17]-[20]. A method, based on cross-correlation analysis for estimation of ventricular ejection time from PCG has been investigated [11]. It may be more reliable than the estimation from the ensemble averaged ICG waveform, particularly during exercise, or post-exercise duration, when cardiac activity is rapidly changing. Further, the time difference between B-point of ICG and R-point of ECG may change, resulting in smudging of ICG peaks.

Yamamota *et al.* [17] have used a narrow band pass IIR digital filter, centered around the heart rate. However, it introduces nonlinear phase distortion and attenuates high frequency components of ICG signal. Raza *et al.* [19] used a high pass IIR digital filter with voluntary cardio-respiratory synchronization. In this technique, a high pass digital filter is programmed for a cutoff frequency that varies as a function of heart rate. Forward filtering followed by backward filtering is used to reduce phase distortion. The disadvantage of this technique is the possibility of distortion of the ICG signal acquired during exercise and post-exercise relaxations.

Barros *et al.* [20] have used an adaptive filter with scaled Fourier linear combiner for removal of movement artifacts. ICG signal was expressed as a Fourier series, with a period determined by R-R interval of ECG. A metronome was used to adjust respiratory and movement artifacts at different frequencies. However, if breathing and movement artifact frequencies coincide with the ICG signal then this method fails to cancel the respiratory and movement artifacts.

In impedance cardiography, there is partial overlap between $z(t)$ and respiratory artifact spectra and statistical properties of both vary. Hence non-adaptive digital filter is not effective in removing respiratory artifact from $z(t)$. Spectral subtraction method [21] which involves estimating noise magnitude spectrum and subtracting it from the contaminated spectrum, can not be applicable here because respiratory artifact spectrum can not be dynamically estimated using average or quantile based estimation. However, an adaptive filter [22], [23] can be employed to dynamically change its transfer function to remove respiratory artifact.

Respiratory artifact is the variation in the sensed thoracic impedance, caused primarily by change in the dimension of thoracic cage during inhale and exhale phases of respiration. Hence air flow during respiration is directly related to the respiratory artifact and can be used to provide a reference for adaptive cancellation of respiratory artifact from the $z(t)$ signal. The respiratory signal acquired from a respiratory sensor, simultaneously with $z(t)$ signal, is taken as reference input, contaminated $z(t)$ being the primary input. The objective is to adaptively remove the respiratory artifact from the $z(t)$ signal without having to control the respiration. Advantage of this method is that it does not require any control of respiration. Respiratory signal can be acquired during exercise. Hence beat-by-beat stroke volume calculation is possible, even if respiratory artifact has large variation. Also, beat-by-beat variation in stroke volume and T_{lvet} may provide additional diagnostic information about cardiovascular functioning. It is to be noted that the technique suppresses additive respiratory artifact. The stroke volume itself gets modulated by the respiratory cycle. This modulation should remain unaffected by the proposed technique.

II. Method

A. Adaptive Cancellation Scheme

A schematic of adaptive respiratory artifact canceller is shown in Fig. 2. Signal $z(n)$ is the thoracic impedance signal contaminated by respiratory artifact from the thoracic impedance sensor. Reference signal $r(n)$ is obtained from the respiratory sensor. The respiratory sensor used here senses the airflow at nostrils, and hence the reference signal has a delay with respect to the respiratory artifact because of the movement of thorax cage. To compensate for this, a delay of n_d samples is introduced in signal path of $z(n)$. Number of taps in the FIR filter should be large enough for the adaptive filter to shift the coefficients to match for the difference in the delay. Adaptive filter output $\hat{r}(n)$ is subtracted from contaminated input $z(n)$ so that the total output error $e(n)$ is progressively minimized.

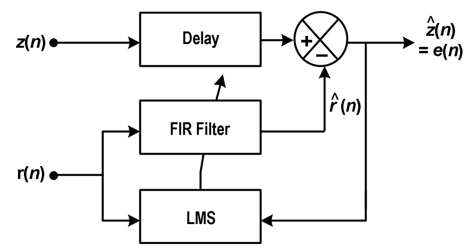

Fig. 2. Adaptive respiratory artifact canceller

For adaptation, we use the least mean square (LMS) algorithm, also known as stochastic gradient algorithm [22], [23]. The LMS algorithm is used to control the digital FIR filter. The output, for M-tap filter with coefficients $w_n(k)$, is given as

$$\hat{r}(n) = \sum_{k=0}^{M-1} r(n-k)\, w_n(k)\,. \tag{1}$$

The error $e(n)$ is the same as the cleaned signal $\hat{z}(n)$, and is given as

$$\hat{z}(n) = z(n-n_d) - \hat{r}(n)\,. \tag{2}$$

The filter coefficients can be updated according to

$$w_{n+1}(k) = w_n(k) + 2\,\mu\,\hat{z}(n)\,r(n-k) \tag{3}$$

where μ is the coefficient satisfying the condition $0 < \mu < 2/l_{max}$ where l_{max} is the largest eigen value of the autocorrelation matrix of its input. This technique allows dynamic adaptation to adjust filter coefficients on sample-by-sample basis, such that the output has minimum artifact in least-square sense.

B. Experimental Set-up

Recordings were done on several normal subjects with no known history of cardiovascular disorders. Impedance cardiograph instrument developed at IIT Bombay [12], [24] was used for recording $z(t)$ waveforms. Signal $z(t)$ was sensed by injecting a high frequency (≈ 100 kHz), low intensity (<5 mA) current into the thorax. Four-electrode configuration, with disposable spot electrodes, was used for reducing the effect of skin-electrode contact impedance. In the physical arrangement of outer pair, one electrode was placed around abdomen at the lateral side of the lower ribs and the other around upper part of the neck. For the inner electrode pair, one electrode is placed around the thorax at the level of joint between xiphoid and sternum and the other around the lower part of the neck. The respiratory signal was recorded to sense inhale and exhale phases by placing a thermistor based respiratory sensor close to the nostrils. Waveforms of the varying thoracic impedance $z(t)$, ECG, PCG, and respiration were simultaneously acquired at a sampling rate of 500 Sa/s with 12-bit resolution using a data acquisition unit interfaced to PC through USB port. The signals were recorded for 7 s duration at 5 min. intervals in normal resting conditions and post-exercise resting conditions.

C. Signal Analysis

The recorded thoracic impedance signal was low pass filtered to attenuate high frequency noise. A 2-point differentiation was used to obtain ICG waveform. The resulting ICG waveform exhibited strong presence of respiratory artifact, particularly in the post-exercise recordings. The adaptive cancellation scheme of Fig. 2 was implemented for different combination of filter tap length and delay in the primary signal path. Based on maximum eigen value of autocorrelation matrix of the input signal $z(n)$, $\mu = 0.0044$ was used for adaptation. Most satisfactory results were obtained for tap length $M \approx 200$ and sample delay ≈ 70. The filter coefficients were found to get stabilized in approximately 500 samples.

III. Results

Fig. 3 shows the recorded $z(n)$, ECG, PCG, and respiration waveforms from a post-exercise recording on one of the subjects. Fig. 4 shows the processed outputs. First waveform in this figure is the ICG obtained by 2-point differentiation of $z(n)$. It is seen to be contaminated by respiratory artifact, making it difficult to detect T_{lvet} and ICG peaks appear to change from cycle to cycle. The second waveform in Fig. 4 is the impedance waveform after adaptive cancellation of the artifact. ICG obtained by differentiating this waveform shows almost no effect of respiration, making it easy to detect the B and X points and these are found to be consistent with simultaneously acquired PCG. Values of ICG peaks are found to be stable. Fig. 5 shows one cycle of this ICG. For an estimation of effectiveness of the adaptive cancellation in removing the

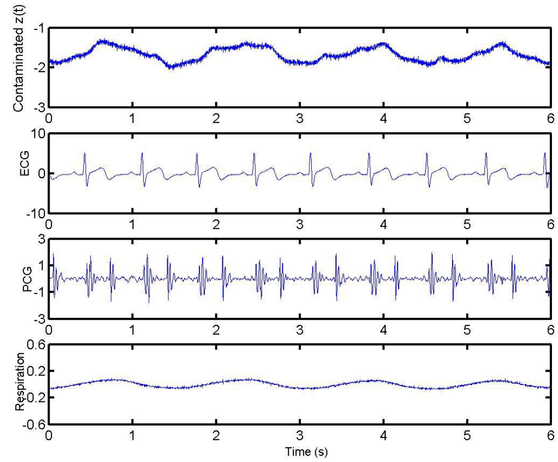

Fig. 3. Recorded z(t), ECG, PCG and respiration from subject ‘VKP’ in post-exercise condition.

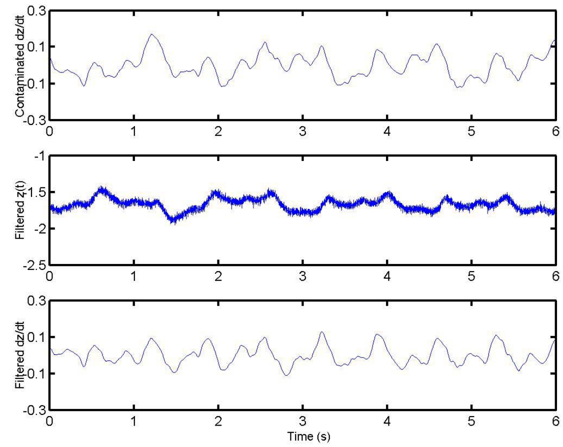

Fig. 4. Processing outputs for waveforms in Fig. 3: (a) ICG from $z(t)$, (b) $z(t)$ after adaptive cancellation, (c) ICG obtained after the removal of respiratory artifact.

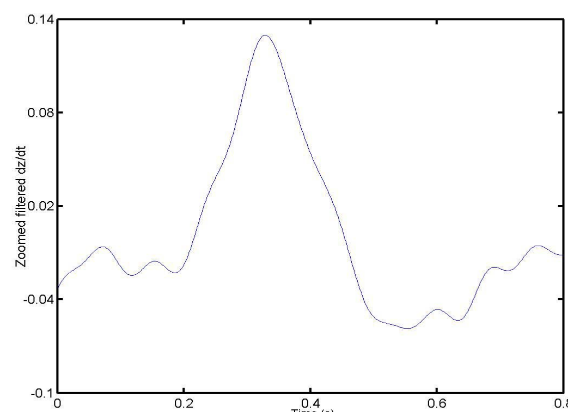

Fig. 5. One cycle of ICG waveform obtained after adaptive cancellation of respiratory artifact.

respiratory artifact, spectra of the original ICG, the respiratory reference signal, and ICG obtained from cleaned waveform were compared and these are shown in Fig. 6.

These were computed for the last 3 s duration of waveform segment in which the filter weights were almost stable The first few spikes in original ICG frequency spectrum are mainly due to respiratory artifact, which is much smaller in case of adaptive filtered ICG. Comparison of spectra of original ICG and ICG obtained from cleaned waveform shows attenuation in spectral peaks of respiratory artifact occurring at 0.22, 0.44, 0.66, 0.88, 1.10 and 1.32 Hz. In cleaned ICG, maximum attenuation of respiratory component was 16.5 dB. Similar analysis was carried on several subjects by sensing $z(t)$ from various parts of thorax and similar results were obtained.

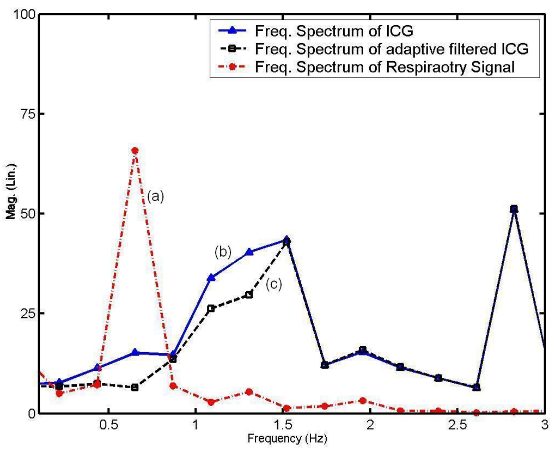

Fig. 6. Spectra of signals in Fig. 3 and Fig. 4: (a) Respiratory reference signal, (b) ICG obtained from original impedance signal, (c) ICG obtained from cleaned impedance signal.

IV. Conclusion

For attenuating respiratory artifact present in ICG signal, LMS based adaptive respiratory artifact canceller is implemented. This method uses signal from respiratory sensor as the reference. Advantage of this method is that it does not require any control of respiration and can be used for cancellation of respiratory artifact, particularly during or post exercise, when cardiac activity is rapidly changing. The thoracic impedance signal, after cancellation of respiratory artifact may be used for beat-by-beat stroke volume calculation.

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