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IMPROVED DETECTION OF VENTRICULAR EJECTION TIME FOR IMPEDANCE CARDIOGRAPHY

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Abstract- Impedance cardiography is a noninvasive technique for monitoring stroke volume (SV), based on sensing variation in the thoracic impedance 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, and this results in errors in the estimation of the two parameters. Ensemble averaging of ICG signal suppresses motion artifacts but it introduces distortion in the signal. In this study, simultaneously acquired phonocardiogram (PCG) signal is used to estimate the time difference between the first and second heart sounds as a measure of ventricular ejection time. Since PCG signal is less influenced by respiratory and motion artifacts, it gives a relatively error free measurement.

Key Words

Impedance cardiography; Stroke volume; Ventricular ejection time; Phonocardiogram.

1. Introduction

Impedance cardiography is a simple, cost effective and noninvasive technique for monitoring stroke volume (SV), based on sensing the changes in the electrical impedance of the thorax, caused by variation in blood volume during the cardiac cycle [1]-[13]. Time derivative of the thoracic impedance (*-dz/dt*) is known as impedance cardiogram (ICG). The parameter required for estimating the stroke volume, using Kubicek, Bernstien, or Sramek formulas [5] are left ventricular ejection time (T_{lvet}) and maximum change in the impedance cardiogram, $(-dz/dt)_{max}$.

As shown in Fig. 1, 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, defined as the minimum following the maximum in ICG. Points B and X correspond to first and second heart sounds respectively. There is no standard method for detecting point B of waveform. In most of the analysis, B-point is identified as base line crossover point before maximum peak of ICG. DeMarzo *et al* [2] observed that B-point can occur at any point on the ascending limb of the waveform in some patients. Further sensing of the variation in thoracic impedance due to blood flow is influenced by respiration and motion artifacts [4], [8], [11], [15], [16], [17]. These artifacts have large amplitude as compared to impedance variation due to cardiac cycle, and can cause a large base line drift. ICG signal bandwidth extends over 0.5-20 Hz. Respiratory related artifacts extend over 0.03-10 Hz, while motion related artifacts have band of 0.1-10 Hz [3], [11], [15], [17]. Therefore frequency band of the motion artifacts may partly overlap with that of the ICG. This causes errors in detection of ventricular ejection time from ICG waveform.

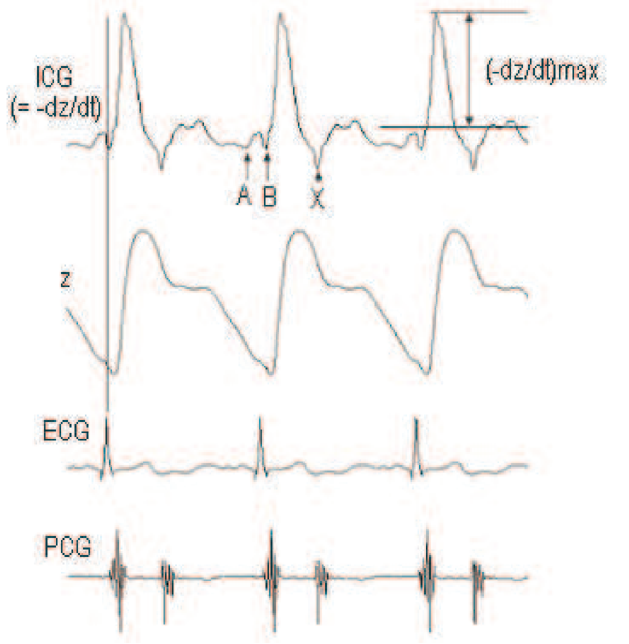

Fig. 1 Typical impedance cardiogram (adapted from [9]). $z(t) = Z_o - Z(t)$ where Z*(t)* is thoracic impedance and Z_o is the basal impedance.

Ensemble averaging is most commonly used technique for suppressing artifacts in the ICG signal [4], [8], [11], [13], [16], [17]. From ensemble averaged ICG, ventricular ejection time and maximum change in the waveform is measured. 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 supress the less distinctive point B of the waveform and may result in error in its detection [4], [11], [13], [14].

Here we present phonocardiogram (PCG) as an alternative signal source to measure ventricular ejection time for stroke volume calculation. The relationship between heart sounds in the PCG with the B and X point of ICG has been studied and reported earlier [5], [7], [8], [9], [10], [11]. As shown in Fig. 1, points B and X of ICG waveform are synchronous to first and second heart sounds respectively. Hence time difference between first and second heart sounds can be used as ventricular ejection time T_{lvet}. The PCG signal is less affected by motion artifacts, hence it can give relatively error free estimation of T_{lvet}. We have used cross-correlation technique to estimate the time interval between the first and second heart sounds in PCG signal, acquired simultaneously along with ICG. The estimation of T_{lvet} from ensemble averaged ICG as well as PCG are computed and compared.

2. Method

As mentioned above, the points B and X of the ICG waveform are used to calculate ventricular ejection time, and they are synchronous to first and second heart sounds. Intra-subject variability results in event latency which introduces distortion in ensemble averaged waveform. Visual inspection of ensemble averaged waveform shows the ambiguity in detecting B and X points. Therefore it is important to accurately detect B, C and X points, independent of base line variation. PCG simultaneously acquired with ICG and ECG can be used for estimating the interval between these points, because PCG is not much affected by motion artifacts.

The PCG signal received from human heart in one heart beat has four sounds [18], [19], [20]. Generally only two heart sounds are perceived. Closure of tricuspid and mitral valves generates the first heart sound while closure of aortic and pulmonary valves corresponds to the second heart sound. Third and fourth heart sounds have very low intensity, and hence most of the time these are not audible. Opening of pulmonary and aortic valves coincides with the closure of tricuspid and mitral valves. Hence the time interval between first and second heart sounds corresponds to ventricular ejection time.

In this study, PCG signal is simultaneously acquired with varying impedance $z(t)$, basal impedance Z_o, and ECG for calculating ventricular ejection time T_{lvet} and comparing the same with that obtained from ICG waveform. The first and second heart sounds are segmented and then cross-correlation is used to measure time interval between them. For separating heart sounds, energy envelope of PCG is calculated and then by threshold detection, first and second heart sounds are segmented. The location of peak in cross-correlation gives the time difference between the two heart sounds, and hence ventricular ejection time T_{lvet}.

Recordings were done on five normal subjects with no known cardiovascular disease. Impedance cardiograph instrument developed at IIT Bombay [12], [21] was used for recording ECG and ICG waveforms. ICG was sensed by passing a high frequency (96 kHz), low intensity (<5 mA) current into the thorax. Four-electrode configuration, with spot electrodes, was used for reducing the effect of skin- electrode impedance. In the physical arrangement of outer pair, one electrode was placed around abdomen 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 PCG was recorded to sense heart sounds by placing a phono-transducer (Pamtron, Mumbai, India) on intercostal space just to the left of the sternum.

3. PCG Analysis

Waveforms Z_o, $Z(t)$, ECG, and PCG are simultaneously acquired at sampling rate of 1 k Sa/s using a data acquisition unit (USBDAQ-9100-MS, manufactured by Adlink Technology, Taiwan) interfaced to PC through USB port. As shown in Fig. 2, first PCG signal is passed through a band pass filter (Butterworth low pass filter with f_{c1} = 7.5 Hz and Butterworth high pass filter with f_{c2} = 100 Hz, cascaded together) to attenuate high and low frequency noise and physiological interferences. Further squaring is done and squared waveform is low pass filtered (f_c = 12.5 Hz) to get energy envelope.

Peaks in the energy envelope are located by dynamic thresholding to give the position of first or second heart sound. The energy envelope is processed as windowed segments, with window length corresponding to approximately one heart sound duration (300 ms). If the peak within the window is supra threshold, it is accepted as indicative of heart sounds, and 60% of the peak is set as the threshold for the next segment. Next the decision process identifies first and second heart sounds. In the decision process, the index of detected peaks are compared with previous peak. If the time interval between peaks labeled *i* and *i*-1 is less than those labeled as *i*-1 and *i*-2, then the peak *i* is marked as second heart sound peak (corresponding to point X in ICG) and the peak *i*-1 is marked as first heart sound peak (corresponding to point B in ICG). After separating first and second heart sounds, cross-correlation between the corresponding energy envelopes is used to find the delay. The location of peak of cross-correlated waveform corresponds to the delay between first and second heart sounds, and gives the ventricular ejection time. Cross-correlation between the segments of the signal waveform as well as those of the energy envelope was used, and it was found that the energy envelope based estimate was much more consistent. Fig. 3 shows a typical PCG signal acquired from subject 'NSM' and output waveform after each processing step.

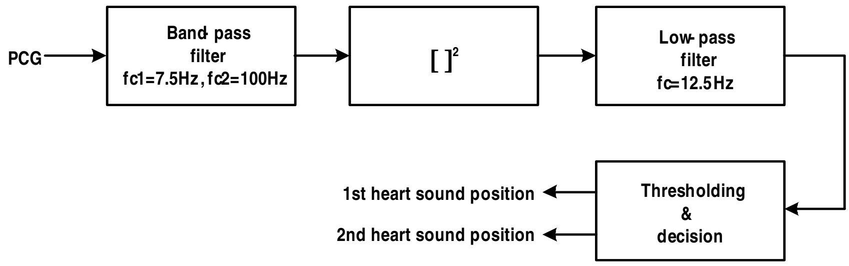

Fig. 2 Processing stages for separating first and second heart sound

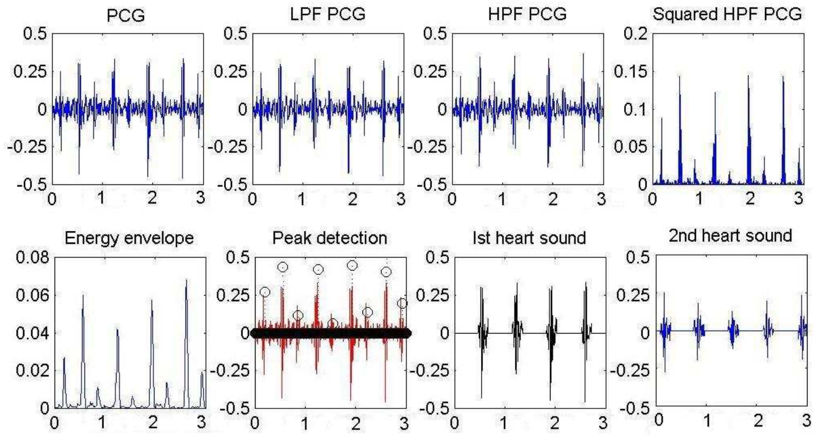

Fig. 3 Output waveforms from processing of PCG, for recording from subject 'NSM' (x-axis: time in s, y-axis: arbitrary units)

4. Results

Processing as discussed in the previous section was used to calculate ventricular ejection time from the ICG waveform as well as from PCG signal. The recordings were done in normal conditions and post exercise relaxation condition at intervals of 5 min. The number of ensemble averaging cycle for ICG waveform was taken as 8.

Table 1 Estimated values (mean, s.d.) of Z_o, $(-dz/dt)_{max}$, and T_{lvet} from ICG and PCG, for 5 subjects under resting condition. The number in parentheses indicates the s.d.

Subject	Beat /min	Z_O (Ω)	$(dz/dt)_{max}$ (Ω/s)	T_{lvet} (ms) from ICG	T_{lvet} (ms) from PCG
NSM	78 (1.18)	40.06 (1.75)	5.82 (0.22)	637 (98)	649 (79)
VKP	72 (3.12)	20.51 (0.27)	5.33 (0.87)	391 (22)	380 (23)
SP	70 (0.89)	22.66 (0.89)	4.49 (0.33)	340 (3)	359 (5)
PKL	83 (3.52)	24.17 (0.42)	12.08 (0.75)	305 (18)	304 (14)
MS	68 (2.10)	22.60 (0.47)	3.33 (1.59)	381 (13)	360 (9)

Table 1 gives the estimated value for the parameters for stroke volume calculation: Z_o, $(-dz/dt)_{max}$ and T_{lvet}. The

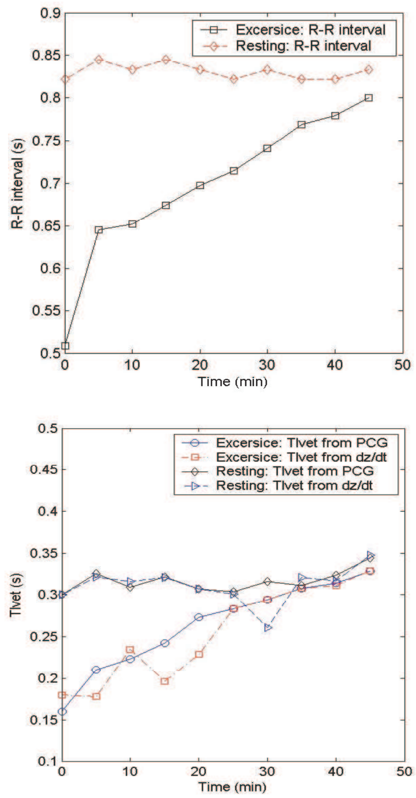

Fig. 4 R-R interval and T_{lvet} vs time plot in resting and post-exercise rest condition for subject 'VKP'.

values of T_{lvet}, obtained by the two methods, closely match, and standard deviations for the two also are similar However visual inspection of plots of these values as a function of time indicates fluctuation in T_{lvet} estimated from ICG. This was much more noticeable during the plots for post-exercise measurements. Fig. 4 shows a plot of R-R interval and T_{lvet} for a typical subject. During resting condition, we see a few deviation in T_{lvet} values and these could be related to error in locating B and X points in ensemble averaged ICG. These deviations are much more visible in the post-exercise plot. This may be because of shift in the B and X points due to ensemble averaging when stroke volume and cardiac activities are changing.

5. Conclusion

A method for estimation of ventricular ejection time from PCG has been investigated. 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.

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