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 \( \frac{dz}{dt} \) is used to calculate two parameters: ventricular ejection time and \( (\frac{dz}{dt})_{\text{max}} \). Respiration and motion artifacts cause baseline 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 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 \( (\frac{dz}{dt}) \) is known as impedance cardiogram (ICG). The parameter required for estimating the stroke volume, using Kubicek, Bernstein, or Sramek formulas [5] are left ventricular ejection time \( T_{\text{Joa}} \) and maximum change in the impedance cardiogram, \( (\frac{dz}{dt})_{\text{max}} \).

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]). \( z(t) = Z_c - Z(t) \) where \( Z(t) \) is thoracic impedance and \( Z_c \) 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 suppress 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 \( T_{ve} \). The PCG signal is less affected by
motion artifacts, hence it can give relatively error free
estimation of \( T_{ve} \). 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 \( T_{ve} \)
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 \( T_{ve} \) 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 \( T_{ve} \).

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 \( f_1 = 7.5 \text{ Hz} \) and Butterworth high pass filter with \( f_2 = 100 \text{ Hz} \), cascaded together) to attenuate high and low
frequency noise and physiological interferences. Further
squaring is done and squared waveform is low pass
filtered (\( f = 12.5 \text{ Hz} \)) 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 approxi-
mately 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.
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 $T_{vett}$ from ICG and PCG, for 5 subjects under resting condition. The number in parentheses indicates the s.d.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Beat/min</th>
<th>$Z_o$ (Ω)</th>
<th>$(-dz/dt)_{max}$ (Ω/s)</th>
<th>$T_{vett}$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NSM</td>
<td>78 (1.18)</td>
<td>40.06 (1.75)</td>
<td>5.82 (0.22)</td>
<td>637 (98)</td>
</tr>
<tr>
<td>VKP</td>
<td>72 (3.12)</td>
<td>20.51 (0.27)</td>
<td>5.33 (0.87)</td>
<td>391 (22)</td>
</tr>
<tr>
<td>SP</td>
<td>70 (0.89)</td>
<td>22.66 (0.89)</td>
<td>4.49 (0.33)</td>
<td>340 (3)</td>
</tr>
<tr>
<td>PKL</td>
<td>83 (3.52)</td>
<td>24.17 (0.42)</td>
<td>12.08 (0.75)</td>
<td>305 (18)</td>
</tr>
<tr>
<td>MS</td>
<td>68 (2.10)</td>
<td>22.60 (0.47)</td>
<td>3.33 (1.59)</td>
<td>381 (13)</td>
</tr>
</tbody>
</table>

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

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).

Fig. 4: R-R interval and $T_{vett}$ vs time plot in resting and post-exercise rest condition for subject 'VKP'.
values of $T_{rot}$ 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 $T_{rot}$ 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 $T_{rot}$ for a typical subject. During resting condition, we see a few deviation in $T_{rot}$ 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.

References