Variable impedance cardiography waveforms: how to evaluate the preejection period more accurately

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Abstract. Impedance method has been successfully applied for left ventricular function assessment during functional tests. The preejection period (PEP), the interval between Q peak in ECG and a specific mark on impedance cardiogram (ICG) which corresponds to aortic valve opening, is an important indicator of the contractility state and its neurogenic control. Accurate identification of ejection onset by ICG is often problematic, especially in the cardiologic patients, due to peculiar waveforms. An essential obstacle is variability of the shape of the ICG waveform during the exercise and subsequent recovery. A promising solution can be introduction of an additional pulse sensor placed in the nearby region. We tested this idea in 28 healthy subjects and 6 cardiologic patients using a dual-channel impedance cardiograph for simultaneous recording from the aortic and neck regions, and an earlobe photoplethysmograph. Our findings suggest that incidence of abnormal complicated ICG waveforms increases with age. The combination of standard ICG with ear photoplethysmography and/or additional impedance channel significantly improves the efficacy and accuracy of PEP estimation.

1. Introduction

Electromechanical systole includes two functionally important systolic time intervals (STI): the pre-ejection period (PEP) and left ventricular ejection time (LVET). Evaluation of PEP and LVET, as well as the ratio PEP/LVET, remains to be a subject of particular physiological and clinical importance for noninvasive assessment of cardiac function [1]. In impedance cardiography, LVET is usually used for stroke volume evaluation [2], while PEP proved to be an important indicator of the state of contractility and its neurogenic control. Assessment of PEP and LVET intervals require accurate identification of ejection onset, which is often problematic due to individual and physiological variability of the B-point [3-5]. In part, it can be achieved using other ‘events’ in impedance cardiogram (ICG) also related to the beginning of ejection [4-6]. Another side of this problem concerns the peculiar shapes of the ICG waveforms encountered in patients [7-9]. Some authors [7, 9] pointed out that the abnormalities in the ICG waveform *per se* can be a sign of cardiovascular pathology, but the underlying mechanisms of such ‘pathologic’ changes remain unclear. Unfortunately, few studies are focused at the peculiarities of ‘pathologic ICG waveforms’. In any case, the significant corruption of dZ/dt waveform leads to ambiguity in detecting the onset of ejection even at rest. The shape of Z-signals can be drastically distorted not only by motion artifacts, but also due to physiological changes in central hemodynamics triggered by exercise. All these factors make evaluation of PEP by ICG method rather uncertain and
inaccurate. Therefore, the development of novel and simple methods improving reliability of PEP measurements during active motion conditions is urgent. To meet this end, we introduced: 1) an additional channel of Z-signal led from the neck region and 2) the earlobe photoplethysmography (PPG).

2. Methods

Twenty eight healthy volunteers, aged 19 to 64, and 6 patients with coronary heart disease (CHD; aged 57 to 76) were studied at rest (standing or sitting) and during treadmill or bicycle exercise test. ICG signals from the aortic and neck regions and ECG in CM5 lead were recorded simultaneously with a dual channel impedance electrocardiograph RPKA-2-01 (MEDACC, Moscow). A tetrapolar ICG-electrode configuration was used in both impedance channels, which worked independently at the frequencies of 30 and 50 kHz. The band-type current electrodes fixed to the forehead and a leg were common for both tracts. A pair of voltage spot electrodes was located at the upper thorax along the ascending aorta projection [5]. Two other spot electrodes used for the second channel were placed in the lower and upper part of the neck. As a supplementary method of tracing PEP an earlobe PPG was used, according to [10].

The signals were continuously sampled at the rate of 1 kHz with a 16-bit analogue-to-digital converter (National Instruments, Texas) and stored on a hard disk. The first and the second derivatives of the impedance signal were calculated point-by-point using a 3-rd order polynomial fitting over 61 points (Savitzky-Golay method [11]). The time coordinates of \( \frac{d^2Z}{dt^2} \) max were automatically determined, and the time interval between the points of the ECG R wave peak and \( \frac{d^2Z_{Ao}}{dt^2} \) max was calculated in each cardiac cycle. These values were used for PEP estimation from impedance aortogram [5, 12-13]. Similarly, PEP

Neck

and PEP

Ear

intervals were determined from other both signals, as specified above, by \( \frac{d^2Z_{Neck}}{dt^2} \) max and \( \frac{d^2PPG}{dt^2} \) max, respectively. A method of synchronous averaging of pulse waves [14] for 5-20 cardiac cycles was used to eliminate the motion artefacts. The changes in ICG and PPG waveforms, as well as the trends of RR interval, PEP

Ao

, PEP

Neck

, and PEP

Ear

during functional tests were analysed. In order to show the related blood volume changes, \( \Delta Z \) curves and their derivatives are shown inverted on the plots, i.e., the positive deflections of the curves correspond to the decrease in impedance.

3. Results and Discussion

3.1. Abnormal ICG waveforms in elderly people and in cardiologic patients

All healthy subjects at the age below 40 years demonstrated canonical ICG waveform (figure 1, a and b) depicted in the frequently cited publications [2, 3]. Most of the healthy subjects under 60 years also had near canonical ICG waveform (figure 1, c). In virtually all subjects above 60, the characteristic abnormalities in the shape of both \( \Delta Z \) and \( \frac{dZ}{dt} \) ICG waves were clearly seen (figure 1, d). Similar alterations were also typical of patients with CHD (figure 1, e-f). The percentage of abnormal Z-waves observed at rest in the groups of various age is shown in table 1. The characteristic feature of such age- and pathology-dependent changes is a remarkable stepwise preejection wave superposing on the ejection wave front (figure 1, d-f, middle traces). By our hypothesis, there are two major processes contributing to formation of the impedance pulse wave: the changes in the heart geometry and surrounding vessels produced by ventricular contraction, which precede the ejection onset, and an increase in the volume of aorta and other major arteries during ejection. The former process initiates the preejection wave and the latter triggers the ejection wave.

When the preejection changes become especially pronounced, the superposition of the two waves gives rise to abnormal \( \Delta Z \) waveforms. As a result, both \( \frac{dZ}{dt} \) and \( \frac{d^2Z}{dt^2} \) curves become double-humped (d-
f, bottom traces), and the first peak in both of these derivatives, associated with the preejection wave, becomes nearly equal or even greater than the second one, corresponding to the ejection wave. No doubt, such waveforms are extremely complicated for establishing the exact time of PEP termination.

Note that identification of B-point, which is commonly used as a mark of PEP end, hits many problems even in simpler ICG waveforms [3-5, 13]. The alternative approaches based on seeking for maximum of either $dZ/dt$ [6] or $d^2Z/dt^2$ [5, 12, 13] also appeared to be problematic in such cases because it cannot be determined definitely which peak in $dZ/dt$ or $d^2Z/dt^2$ corresponds to the beginning of ejection. In that case an independent method of central pulse recording is urgent.

![Figure 1](image)

**Figure 1.** The shape of ICG waves in healthy subjects of various age (a, b, c and d) and in patients with CHD (e, f). *From top to bottom:* ECG, $\Delta Z$ (Ohm), $dZ/dt$ (black, left scale, Ohm/s) and $d^2Z/dt^2$ (grey, right scale, Ohm/s²).

Manifestations of the preejection processes were most evident in the elderly subjects and in patients, however, some distortions of $dZ/dt$ waveform can also be discerned in the young subjects (figure 1, b and c). We have not found any indications on the age-dependent changes of ICG waveforms in literature. The factors which could affect the shape of preejection waves in our study were the locality of impedance recording from the aortal area and a vertical posture. Previously, the alterations in the ICG shape with change in posture were observed in patients with heart disease [15].

### 3.2. Corruption of ICG waveform during exercise

In some healthy subjects, initially normal ICG waveforms were drastically corrupted during physical exercise, Valsalva maneuver or other tests. An example of such phenomenon is presented in figure 2.
A healthy man (aged 52) performed a prolonged (30 s) expiration accompanied with abdominal muscles contraction. The trend of PEP estimated by $d^2Z/dt^2$ maximum (the thick red line on the upper trace) displayed a sudden and significant decrease in its value during this maneuver, which could be attributed to an abrupt rise in the left ventricular contractility. In fact, it was an artifact related to the enormously raised prejection wave which masked the ejection upstroke. As a result, the second peak of $d^2Z/dt^2$ which normally indicates the onset of ejection disappeared and PEP was erroneously estimated by the first $d^2Z/dt^2$ peak, corresponding to the beginning of the prejection wave (the left panel). This was confirmed by the absence of changes in PEP trend assessed with PPG signal recorded simultaneously from the earlobe. The PPG trend (the thin blue line on the upper trace) is shown taking into account the time correction for the pulse wave propagation from aorta to the ear. On the right panel showing the corresponding PPG waveforms and their derivatives, it is clearly seen that the time of the major $d^2$PPG/$dt^2$ maximum remained unchanged during the maneuver.

**Figure 2.** Corruption of ICG waveforms during an expiratory maneuver resulted in erroneous PEP estimation while the shape of PPG signals remained normal. Upper panel shows trends for PEP$_{Ao}$ and corrected PEP$_{Ear}$. The waveforms of $dZ/dt$ (black) and $d^2Z/dt^2$ (red) for ICG recorded before and during the expiration are displayed on the left panels below. The corresponding waveforms of $d$PPG/$dt$ (black) and $d^2$PPG/$dt^2$ (violet) are given on the right panels. R shows the time of ECG R peak; the circles indicate the second derivative maxima used for PEP determination.

In another subject (age 55), pronounced alterations in the shape of ICG waveforms were observed during cycling at a constant load of 60 W (figure 3). As a rule, PEP abruptly shortened at the beginning of cycling due to effect of Starling mechanism [16]. It was followed by a few minute period of slow PEP changes coming to a prolong plateau that persisted up to the end of cycling. Similar course of PEP changes could be observed for the intervals determined from ear PPG or additional impedance recording from the neck region (upper trends in figure 3). By contrast, PEP trend for impedance aortogram continued to decline throughout the exercise reaching significantly lower values to its end (the lowest, black trend). The difference between the trends can be attributed to transformation of the ICG waveforms recorded from the aortic region: the main peak of $d^2Z_{Ao}/dt^2_{max}$ became double-humped and the B-point disappeared completely (bottom panels in figure 3), which results in erroneously shorter PEP estimates due to premature detection of the ejection onset by any algorithm. The shapes of the signals recorded from the neck and the ear (not shown) did not changed noticeably. It is worth to note
that the transformations of the aortic ICG waves depicted in figures 2 and 3 were of hemodynamic origin since the noisy component in both cases was negligible.

### 3.3. The usage of Z-neck or PPG-ear channels in treadmill or cycling tests

Analysis of more than 30 records of dual-channel ICG and PPG signals performed during treadmill and cycling tests demonstrated pronounced motion artifacts in the neck Z-signals and to a lesser extent in aortic ICG. The most noise-immune was the earlobe PPG. This implies that PPG can be a good assistant to aortic ICG when PEP is evaluated during exercise. Previously, Spodick [10] demonstrated efficacy of the earlobe PPG method for STI evaluation. His method assumed a relative constancy of the time shift between the pulse waves arriving to the carotid artery and to the ear. In fact, it is not exactly so. Our findings indicated that, in subjects over 50 years old, a delay between the aortic impedance and ear PPG waves increased (by about 30 % in average) in the early stage of treadmill test, and it decreased (by 10 %) in the recovery period. The latency variations were less pronounced during the bicycle test (< 10 % of its baseline value).

![Figure 3](image.png)

**Figure 3.** Changes in aortic ICG waveforms during cycling (in a healthy man). *Upper traces:* $\text{PEP}_{\text{Ao}}$ (Z-1), $\text{PEP}_{\text{Neck}}$ (Z-2) and $\text{PEP}_{\text{Ear}}$ (PPG). *Below:* the aortic waveforms of $\Delta Z$, $dZ/dt$ and $d^2Z/dt^2$ at rest (Bgr-2'), during load (on 5, 12 and 17 min from the start of the record), and during recovery (on 18 and 21 min).

Thus, PPG method can not be simply substituted for the original ICG method because the corresponding pulse transition time is not stable during the exercise. However, the PPG signals can improve PEP assessment when the strong motion artifacts in aortic or neck ICG pulse waves cannot be eliminated by synchronous averaging. After evaluation of the time lags between ICG and PPG signals at rest and during the recovery period, PPG method can provide a satisfactory approximation to PEP changes during the tests. Also, PPG signals can help to avoid ambiguity in tracing the time course of PEP responses when the detection of the end of PEP with a computer algorithm is confused due to abnormal or corrupted ICG waveforms. If the proper ‘mark’ in the ICG signal pointing to the aortic valve opening could be identified using somewhat other technique (e.g., echocardiography) just before the start of the exercise, then the PPG signals would help in tracing PEP changes during motion. Another way to approximate the start point of ejection is to record the neck impedance. In contrast to
aortic ICG, the neck impedance has a simple pulse waveform even in elderly humans and patients. Also, it has a much smaller and stable latency, compared to the ear PPG. Together with aortic ICG, the neck signal yields the most accurate evaluation of PEP in the cases of complicated ICG waveforms, though it can be often useless in moving subjects due to strong artifacts. So, we consider that this channel is useful mainly for appropriate identification of PEP termination ‘mark’ in complicated ICG waveforms at rest, in the recovery period, and during other stages with relatively small motion artifacts in the neck impedance.

4. Conclusion
Thus, a combination of standard ICG with earlobe PPG and/or additional impedance channel can significantly improve the efficacy and accuracy of PEP estimation, especially in cardiologic patients and elderly people or in the cases with pronounced motion artifacts. PPG is most useful during treadmill or bicycle exercise tests, because it is far less sensible to motion-related disturbances. The time shift between PPG and ICG signals is not stable. The method requires time-to-time corrections for these changes. The neck impedance is more preferable to use under the rest conditions or when a maneuver does not require a strong physical activity or neck muscles tension. The advantage of the latter is a small latency between the neck and the aortic pulses.

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