Simulation of ‘pathologic’ changes in ICG waveforms resulting from superposition of the ‘preejection’ and ejection waves induced by left ventricular contraction

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Abstract. The impedance cardiography (ICG) is widely used for beat-to-beat noninvasive evaluation of the left ventricular stroke volume and contractility. It implies the correct determination of the ejection start and end points and the amplitudes of certain peaks in the differentiated impedance cardiogram. An accurate identification of ejection onset by ICG is often problematic, especially in the cardiologic patients, due to peculiar waveforms. Using a simple theoretical model, we tested the hypothesis that two major processes are responsible for the formation of impedance systolic wave: (1) the changes in the heart geometry and surrounding vessels produced by ventricular contraction, which occur during the isovolumic phase and precede ejection, and (2) expansion of aorta and adjacent arteries during the ejection phase. The former process initiates the preejection wave \( W_{pE} \) and the latter triggers the ejection wave \( W_{Ej} \). The model predicts a potential mechanism of generating the abnormal shapes of \( \frac{dZ}{dt} \) due to the presence of preejection waves and explains the related errors in ICG time and amplitude parameters. An appropriate decomposition method is a promising way to avoid the masking effects of these waves and a further step to correct determination of the onset of ejection and the corresponding peak amplitudes from ‘pathologically shaped’ ICG signals.

1. Introduction
A reliable and accurate method for detection of the ejection onset is crucial for evaluation of the pre-ejection period (PEP), an important indicator of the changes in contractility during functional tests. To this aim, many algorithms [1, 2] have been elaborated but none of them is universal. The problem of correct and unambiguous detection of the start of ejection by ICG is not completely solved, and the automatic detection of specific ‘landmarks’ in ICG signals (such as B point) remains rather complicated. Moreover, in the case of specific ICG waveforms met in the cardiologic patients and elderly people [3, 4], the conventional algorithms and ‘guidelines’ used for finding the demarcating points (and even C peaks) usually fail. So, this problem turns to be not simply technical but fundamental in character.

The characteristic feature of such ‘pathological’ waveforms is a remarkable stepwise preejection wave superposing on the ejection wave front. By our hypothesis, there are two major processes contributing to the formation of impedance pulse wave: (1) the changes in the heart geometry and surrounding vessels which occur during the isovolumic phase of the left ventricular contraction, and (2) expansion of aorta and adjacent arteries during the ejection phase. The former process initiates the preejection wave \( W_{pE} \) and the latter triggers the ejection wave \( W_{Ej} \). Using a simple mathematical model, we analyzed some effects of summation of these processes on the \( \frac{dZ}{dt} \) waveform and associated ICG parameters.
2. Model

By our assumption, the ICG impedance waveform represents an algebraic sum of the preejection wave and the ejection wave, i.e., \( \Delta Z = Z_{pk} + Z_{Ej} \). An asymmetric bell-shaped function \( W(A, b, c, t) = A \cdot e^{-\frac{t - c}{b}} \) was chosen to simulate the contribution of each of these waves (figure 1, A: \( W_{pk} \) – blue curve, \( W_{Ej} \) green curve). The result of summation of the big bell \( W_{Ej} \) with the small bell \( W_{pk} \) can be seen on panel B. The first and the second time derivatives of all curves are shown below (\( A \) and \( B \)).

The formulas describing the analytical functions and their derivatives are given on the right panel (figure 1). Note, that the first derivative \( dW/dt \) of each bell is biphasic. It attains the maximum at a time \( t_{max} \), which points to the steepest part of the bell front, crosses zero line at \( t_b \) (corresponds to bell’s maximum), then runs down to the minimum at \( t_{min} \) indicating the steepest site on the downslope, and, afterwards, goes again to zero. Such a trend of \( dW/dt \) curve is typical for any bell-shaped curve with sigmoid up- and downslopes. Noteworthy, that the formulas for calculation of the derivatives of the chosen function \( W(A, b, c, t) \) are very simple (figure 1). Another intriguing property of this function is that the time \( t_b \) lies strictly on a half way from \( t_{max} \) to \( t_{min} \). This allows obtaining realistic-like \( dZ/dt \) curves by setting not the abstract parameters \( A, b, \) and \( c \), but the more meaningful time and amplitude characteristics: \( t_{min}, t_{max}, \) and \( (dW/dt)_{max} \). For the ejection wave these parameters have the following meaning: \( t_{min} \) (or, more exactly, the period from \( t = 0 \) to \( t_{min} \)) is an analog of the left ventricular ejection time, \( (dW/dt)_{max} \) is an analog of the C-wave amplitude, and \( t_{max} \) indicates the position of \( (dW/dt)_{max} \) point on the time scale and relates to the steepness of C-wave front. In our model the start of the big bell \( (W_{Ej}) \) was fixed at 100 ms on the time scale, while the time shift \( \Delta t \) between the small and big bells (starting from the point \( t = 0 \) set for each bell function separately) varied from -100 to 0 ms (i.e., function \( W_{pk} \) started earlier than \( W_{Ej} \) by \( \Delta t \)).

The impact of the preejection waves on the shape of the resultant wave \( \Delta Z \) and its derivatives, as well as the associated changes in the timing of \( dZ/dt \) and \( d^2Z/dt^2 \) peaks and their amplitudes were analyzed.

3. Results

3.1. ICG waveform distortion induced by preejection waves

The model predicts the appearance of double-humped derivatives \( dZ/dt \) and \( d^2Z/dt^2 \) of the resultant wave. Figure 2 (a-g) shows the changes in the simulated ICG waveform representing the sum of two
bell-shaped functions with the fixed $dW/dt$ peak heights (1:5), as the small bell (preejection) approaches and superimposes onto the big bell (ejection). Although little changes are evident in the resultant $\Delta Z$ waveform, the shape of $dZ/dt$ alters from $b$ to $f$ remarkably. The most pronounced changes are seen in the shape of $d^2Z/dt^2$. Thus, even small amplitude preejection waves, if they are rather steep, can produce a noticeable distortion of the resultant $dZ/dt$ and $d^2Z/dt^2$ waveforms. Such phenomenon we observed earlier in some young and middle-age people, especially, during functional tests [4]. Summation of $W_{pe}$ and $W_{ej}$ waves can also lead to elevated B-point in $dZ/dt$ (figure 2, d-f).

When the amplitude of C-wave decreases or, vice-versa, the preejection wave increases and $W_{pe}$ downslope becomes steeper or comparable to the upstroke of $W_{ej}$, the resultant $\Delta Z$ waves will acquire the features typical of the cardiologic patients, with the first (‘preejection’) peak in both derivatives ($dZ/dt$ and $d^2Z/dt^2$) being nearly equal or even greater than the second peak corresponding to ejection (figure 2, g-h). In such situations most algorithms used for ICG processing fail to work. Some decomposition technique will be better to use in these cases.

It is important that the combination of this two-bell model with the non-linear fitting of the curves based on Levenberg-Marquardt algorithm [5] offers two-bell simulation of a large body of really recorded abnormal ICG $dZ/dt$ shapes with a good precision ($r^2 > 0.99$).

![Figure 2](image1.png)  
**Figure 2.** Transformation of ICG waves (red) induced by preejection waves (blue) started at various times before the ejection waves (green).

### 3.2. Changes in timing of ejection onset and amplitudes of $dZ/dt$ and $d^2Z/dt^2$ peaks

The waves $W_{pe}$ not only mask but shift the ‘mark’ points which are used for detection of PEP end. Figure 3, A shows the changes of the resultant $dZ/dt$ waveforms as wave $W_{pe}$ approaches to wave $W_{ej}$. The colored lines are the traces of the corresponding ‘landmarks’: B-point, defined by a complex algorithm (red) and by maximum of the 3-rd derivative [1] (magenta), maximum of the $d^2Z/dt^2$ ($d2\, max$, green), and maximum of the $dZ/dt$ (blue) traces. Three black traces on the left indicate the start (by Ono et al [2]), peak and the minimum points of $dW_{pe}/dt$. The other black traces correspond to the big bell $W_{ej}$ which does not move. They indicate the start point and peaks of $dW_{ej}/dt$ and $d^2W_{ej}/dt^2$. Below, on panel B, a top view of the scaled-up plot A is shown. In figure 3, panel C the changes (in %) of C-wave (blue) and $d2\, max$ (green) amplitudes are plotted as functions of the time shift $\Delta$ between the starts of waves $W_{pe}$ and $W_{ej}$. The bottom plot (C panel) shows the time lag between the $d2\, max$ peak (the green...
line in $B$) and the corresponding peak of the ejection wave (the black vertical line in $B$). The magenta lines on these plots reflect behavior of the absolute maximum of $d^2Z/dt^2$ when $d_2$ max corresponding to C-wave becomes smaller than the earlier peak related to $W_{pE}$.

Thus, the presence of preejection waves can lead to considerable errors in the time and amplitude ICG parameter measurements resulting in erroneous assessment of stroke volume and contractility indices.

**Figure 3.** Changes in ICG parameters with changing time shift between the bell’s starts. Step in changing time shift (time interval between horizontal lines in $A$ and $B$) is 2 ms.

4. Conclusions
A simple two-bell model can adequately depict the peculiarities of actual ICG waveforms correlating to the early phase of ventricular systole which are typical of cardiologic patients and some elderly people. The model predicts a potential mechanism of generating the abnormal shapes of $dZ/dt$ due to the presence of the preejection waves and explains the related errors in the impedance time and amplitude parameter measurements. An appropriate decomposition method is a promising way to avoid the masking effects of these waves and a further step to correct determination of the start of ejection and the corresponding peak amplitudes from ‘pathologically shaped’ ICG signals.

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