An Introduction to Electrical Cardiometry™

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The continuous measurement of thoracic electrical bioimpedance (TEB) obtains an intriguing waveform signal: the portion of the bioimpedance waveform related to the cardiac cycles resembles to a good part an arterial pressure waveform. For many decades, efforts were undertaken to derive stroke volume and cardiac output of this waveform or its derivatives. These efforts relied on a model that contributes the rapid change of bioimpedance which occurs shortly after aortic valve opening to the expansion of the compliant ascending aorta (Figure 1), assuming that more blood volume temporarily stored in the ascending aorta contributes to a decrease in bioimpedance (or an increase in electrical conductivity of the thorax). Impedance Cardiography (ICG) is the most prevalent method relying on the model that volumetric changes in the ascending aorta are the origin of the characteristic change in bioimpedance following aortic valve opening.

In 2001 Bernstein and Osypka[1] challenged this traditional model of interpreting the bioimpedance signal. They developed a new model (Electrical Velocimetry™), which leads to a new method called *Electrical Cardiometry*™.

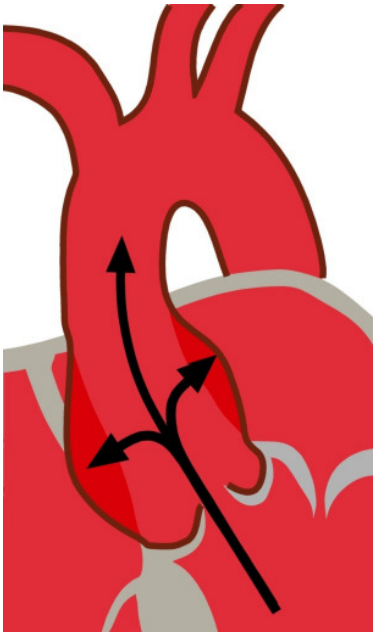

Figure 1: Volumetric expansion of the ascending aorta shortly after aortic valve opening

[1] Bernstein DP, Osypka MJ. Apparatus and method for determining an approximation of stroke volume and cardiac output of the heart. US Patent No. 6,511,438.

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Measurement of Thoracic Electrical Bioimpedance (TEB)

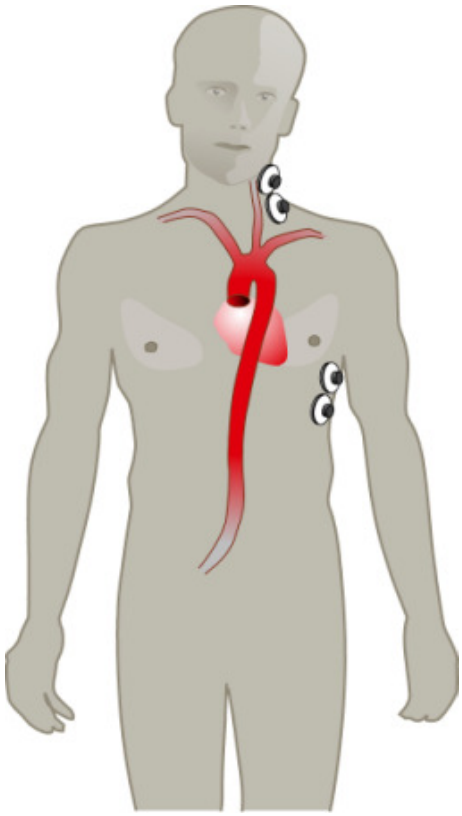

Figure 2: Electrode array for measuring TEB

The Electrical Cardiometry monitors AESCULON® and ICON® determine an estimate of the stroke volume (SV) by measuring the thoracic electrical bioimpedance (TEB), and in particular the changes thereof related to the cardiac cycle.

For this purpose an array of 4 surface ECG electrodes is attached to the left side of the neck and the lower thorax (approximately at the level of the xiphoid process; Figure 2). An electrical alternating current (AC) of constant amplitude is applied via the pair of outer electrodes to the thorax and in particular – because blood is the most conducting tissue in the thorax – the ascending and descending aorta. The resulting voltage and a surface ECG are obtained via the inner pair of electrodes. The ratio of applied current and measured voltage equals the conductivity, which is recorded over time.

The measured bioimpedance over time can be expressed as the superposition of three components:

$$Z(t) = Z_0 + \Delta Z_R + \Delta Z_C$$

where Z_0 is the quasi-static portion of the impedance, also referred to as the base impedance, mostly determined by thoracic fluids including the thoracic blood volume, ΔZ_R are the changes of impedance related to respiration, and ΔZ_C are the changes of impedance related to the cardiac cycle. ΔZ_R is considered an artifact to the estimation of stroke volume and, thus, suppressed.

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The timely course of the measurement of ΔZ_C reveals a signal waveform with a shape akin to an arterial blood pressure waveform. Figure 3 illustrates a set of waveforms recorded in parallel.

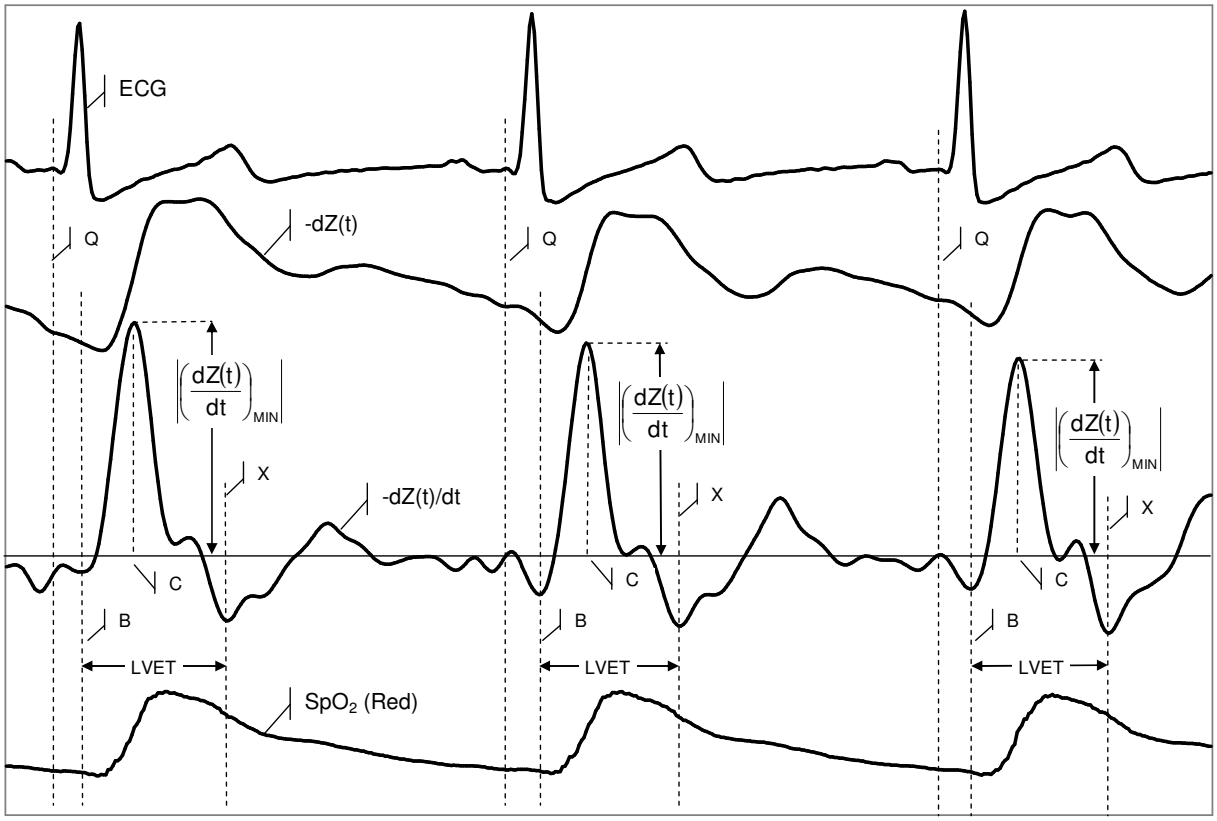

Figure 3: Signal waveforms of the surface electrocardiogram (ECG, on top), the inverted change of impedance related to the cardiac cycle ΔZ_C, here labeled as –dZ(t), the calculated time-derivative of –dZ(t), here labeled as –dZ(t)/dt, and the pulse plethysmogram obtained by finger pulse oximetry. Note that the –dZ(t)/dt is an artificial signal obtained by calculation and not a physiological signal.

The –dZ(t)/dt waveform is analyzed to determine the following landmarks:

- Opening of aortic valve (Point B) – begin of flow time (FT) / left-ventricular ejection time (LVET);
- Peak aortic blood acceleration (Point C, occurring in time with aortic dP/dt_{MAX});
- Closure of aortic valve (Point X) – end of flow time (FT) / left-ventricular ejection time (LVET).

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The Model of Electrical Velocimetry™ (EV™)

Most hemodynamic parameters can not be measured directly but require a theoretical model for interpretation of the measurement results. For instance, measurements of electrical voltages on or in the body require the model of the electrocardiogram (ECG) to become useful. Estimating stroke volume from thoracic bioimpedance measurements is no exception.

A hemodynamic parameter obtained can only be as accurate and reliable as the theoretical model. Thus, knowledge or proper assumption of the origin of the signal measured is the key to modeling.
The method of *Electrical Velocimetry*™ is based on the fact that the conductivity of the blood in the aorta changes during the cardiac cycle.

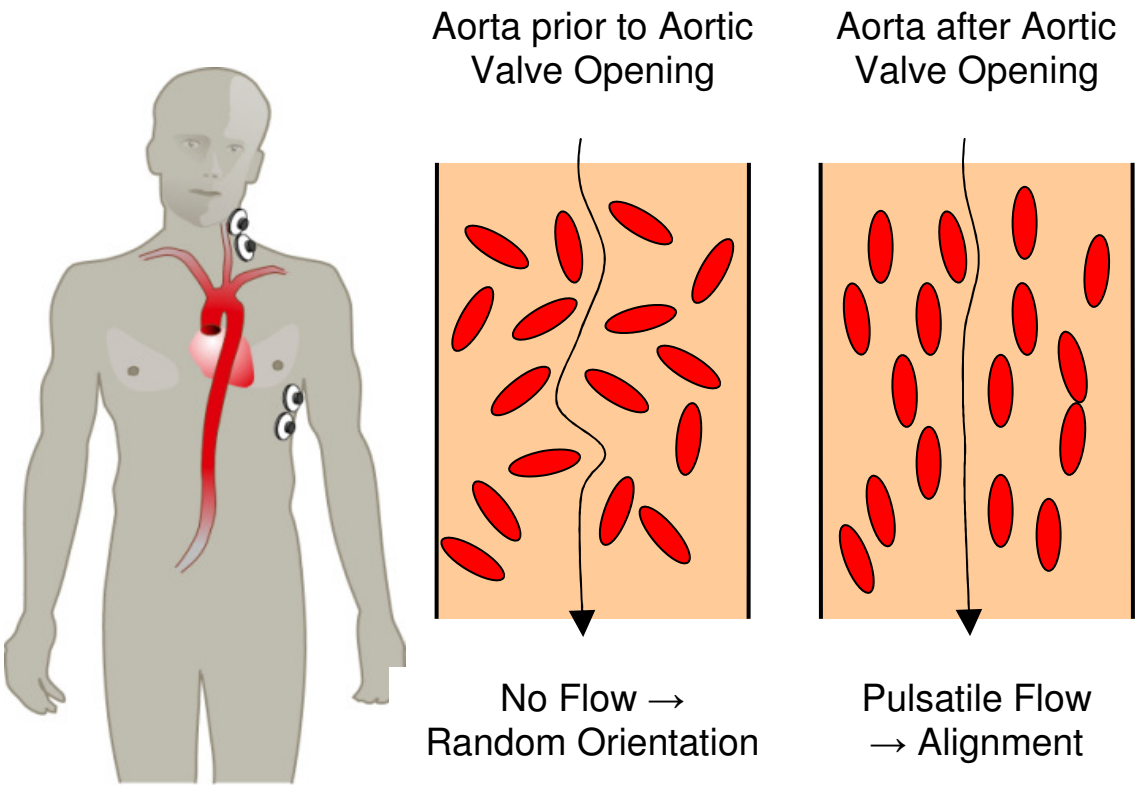

Figure 4: Electrode arrangement and orientation of red blood cells (erythrocytes) in the aorta prior and shortly after aortic valve opening

Prior to opening of aortic valve, the red blood cells (erythrocytes) assume a random orientation – there is no blood flow in the aorta (Figure 4). An electrical current applied must circumference the red blood cells for passing through the aorta, which results in a higher voltage measurement and, thus, lower conductivity.

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Very shortly after aortic valve opening, the pulsatile blood flow forces the red blood cells to align in parallel with the blood flow (mechanical properties of the disc-shaped blood cells). Now an electrical current applied passes the red blood cells more straight forward and easily, which results in a lower voltage measurement and thus in a higher conductivity.

Note: For technical reasons, impedance rather than conductivity is measured. Impedance is reciprocal to conductivity. It can be shown that the shape of the inverted change-of-impedance waveform, –dZ(t), is akin the change-of-conductivity waveform. Since the –dZ(t) signal waveform also shows resemblance with an arterial pressure waveform, it has become common use to display the inverted change-of-impedance waveform, –dZ(t), and read it as a change-of-conductivity waveform.

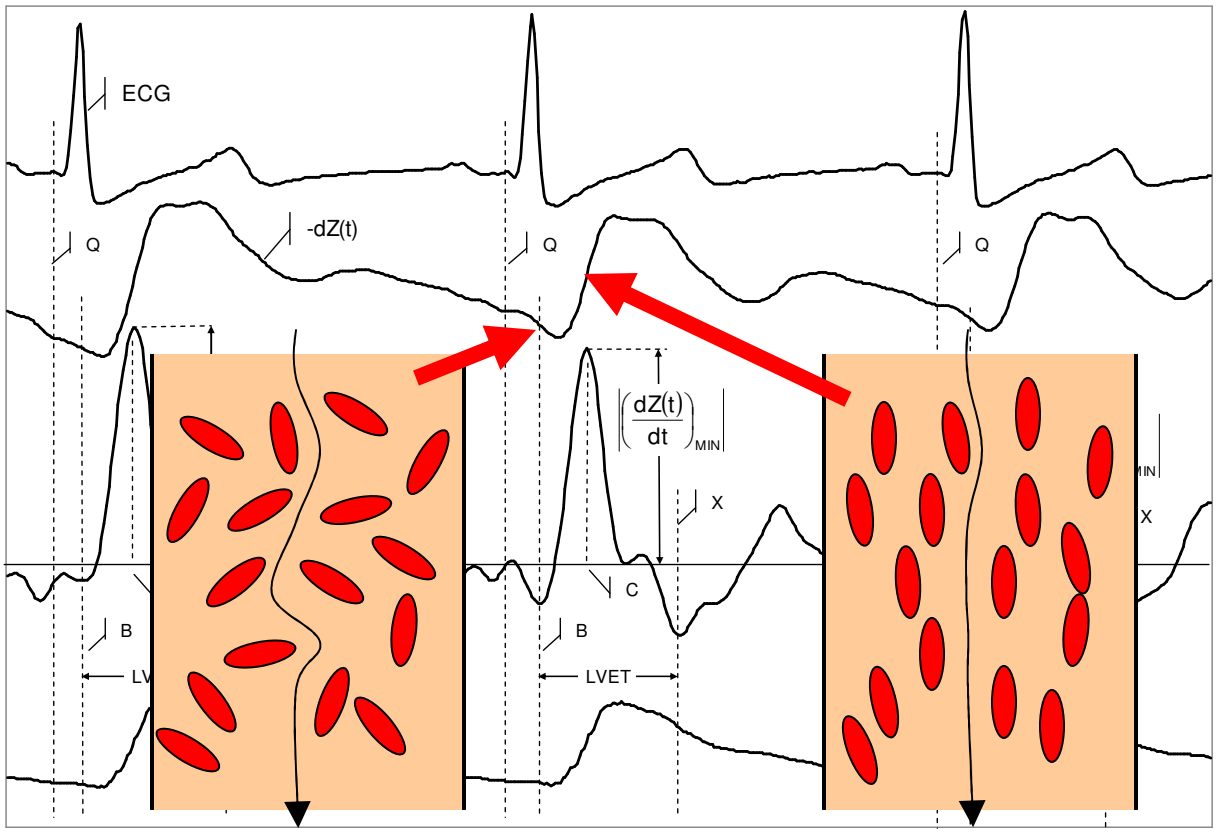

Figure 5: Timely course of parallel recordings of ECG, impedance waveforms and pulse plethysmogram

Figure 5 illustrates the course of the surface ECG (waveform on top), –dZ(t) (second waveform from top), the calculated, artificial -dZ(t)/dt signal (third waveform from top) and the pulse plethysmogram (obtained by pulse oximetry; waveform at bottom).

The change from random orientation to alignment of red blood cells upon opening of aortic valve generates a characteristic steep increase of conductivity (corresponding to a steep decrease of impedance) – beat to beat (see red arrows pointing to the two states shown in the change-of-conductivity signal).

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The steeper the slope of the –dZ(t) signal, or the higher the peak amplitude of -dZ(t)/dt, the quicker the alignment process and, thus, the higher the contractility of the heart.

Note: *Doppler Velocimetry* 'looks' at the erythrocytes in the aorta to determine their velocity. Electrical Velocimetry™ 'looks' also at the erythrocytes but determines their change in orientation to derive blood velocity.

The model of *Electrical Velocimetry*™ considers the peak amplitude of -dZ(t)/dt divided by the base impedance Z_0 as an index for peak aortic acceleration, and as an index of contractility (ICON™):

$$ICON = \frac{\left| \left(\frac{dZ(t)}{dt} \right)_{MIN} \right|}{Z_0} \cdot 1{,}000\,.$$

The general equation for estimating stroke volume (SV) by means of thoracic electrical bioimpedance calculates the product of a patient constant C_P (in ml), the mean blood velocity index $\bar{v}_{FT}$ (measured in s^{-1}) during flow time FT, and flow time (FT; measured in s):

$$SV_{TEB} = C_P \cdot \bar{v}_{FT} \cdot FT$$

The model of *Electrical Velocimetry*™ derives the mean blood velocity index $\bar{v}_{FT}$ from the measured index for peak aortic acceleration[2]:

$$\bar{v}_{FT} = \sqrt{\frac{\left| \left(\frac{dZ(t)}{dt} \right)_{MIN} \right|}{Z_0}}$$

The higher the mean blood velocity $\bar{v}_{FT}$ during flow time, the more SV the left ventricle ejects.

The model of *Electrical Velocimetry*™ 'corrects' the flow time for heart rate, i.e., applies a corrected flow time (FT_C) to computation of stroke volume:

$$FT_C = \frac{LVET}{\sqrt{T_{RR}}},$$

where LVET is the measured left-ventricular ejection time and T_{RR} is the measured R-R interval. The longer FT_C or LVET, the more SV the left ventricle ejects.

The model of *Electrical Velocimetry*™ uses the 'volume of electrically participating tissue' (V_{EPT}) as the patient constant. The V_{EPT} is derived primarily from body mass. Thus it is important that the body mass of a person subject to monitoring is measured or estimated as accurately as possible.

[2] U.S. Patent Number 6,511,438 and international patents

Any error in the measurement or estimation of weight translates into an error of similar magnitude in stroke volume.

The model of *Electrical Velocimetry*™ calculates stroke volume as

$$SV = V_{EPT} \cdot \sqrt{\frac{\left| \left(\frac{dZ(t)}{dt} \right)_{MIN} \right|}{Z_0}} \cdot FT_C$$

and cardiac output (CO; measured in L /min) as the product of stroke volume (SV; measured in ml) and heart rate (HR; measured in 1 /min):

$$CO = \frac{SV}{1{,}000} \cdot HR$$

Summary: The model of *Electrical Velocimetry*™ assumes that the alignment of red blood cells (erythrocytes) in the aorta contributes to the significant change in the impedance soon after aortic valve opening. The model relies on a *change* in impedance of the aortic blood. Continuous blood flow, such as artificially added blood flow by a gravity pump, does not contribute to a change of impedance and as a consequence is not accounted for.

The model of *Electrical Velocimetry*™ estimates stroke volume based on an input of body mass, a mean velocity index empirically derived from a peak amplitude measurement assumed to be an index of peak aortic acceleration of blood flow, and a measurement of flow time.

Note: We distinguish between the model of *Electrical Velocimetry*™ and the method of noninvasive hemodynamic monitoring, *Electrical Cardiometry*™.

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Publications (Validation):

[1] Zoremba N, Bickenbach J, Krauss B, Rossaint R, Kuhlen R, Schälte G. Comparison of electrical velocimetry and thermodilution techniques for the measurement of cardiac output. Acta Anaesthesiol Scand 2007: 51: 1314-1319.

[2] Suttner S, Schöllhorn T, Boldt J, Mayer J, Röhm KD, Lang K, Piper SN. Noninvasive assessment of cardiac output using thoracic electrical bioimpedance in hemodynamically stable and unstable patients after cardiac surgery: a comparison with pulmonary artery thermodilution. Intensive Care Med, 2006 Dec; 32 (12): 2053-8. Epub 2006 Oct 13.

[3] Norozi K, Beck C, Osthaus WA, Wille I, Wessel A, Bertram H. Electrical velocimetry for measuring cardiac output in children with congenital heart disease. Br J Anaesth, 2007, EPub Nov 16.

[4] Osthaus WA, Huber D, Beck C, Winterhalter M, Boethig D, Wessel A, Sümpelmann R. Comparison of electrical velocimetry and transpulmonary thermodilution for measuring cardiac output in piglets. Pediatric Anesthesia 2007 Aug 17, 8, 749-755

[5] Schmidt C, Theilmeier G, Van Aken H, Korsmeier P, Wirtz SP, Berendes E, Hoffmeier A, Meissner A. Comparison of electrical velocimetry and transesophageal Doppler echocardiography for measuring stroke volume and cardiac output. Br J Anaesth, 2005 Nov; 95 (5): 603-610. Epub 2005 Sep 9.

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Frequently Asked Questions

***Electrical Cardiometry* and *Impedance Cardiography* (ICG) utilize measurements of thoracic electrical bioimpedance (TEB). What have the two methods in common, where are the differences?**

Electrical Cardiometry and Impedance Cardiography (ICG)

- measure TEB,
- determine a base impedance Z_0,
- extract the change of impedance related to the cardiac cycle ΔZ(t) (Figure 6),
- calculate the rate of change of impedance $\frac{dZ(t)}{dt}$, i.e., the time derivative of ΔZ(t),
- determine a peak amplitude $\left|\left(\frac{dZ(t)}{dt}\right)_{MIN}\right|$,
- identify through analysis of the $\frac{dZ(t)}{dt}$ signal waveform the timely occurrences of opening (point 'B') and closure (point 'C') of the aortic valve, and determine left-ventricular ejection time as the time period in between, and
- calculate an estimation of stroke volume according to $SV_{TEB} = C_P \cdot \bar{v}_{FT} \cdot FT$ with the patient constant C_P (in ml), the mean blood velocity index $\bar{v}_{FT}$ (measured in s^{-1}) during flow time FT, and flow time (FT; measured in s).

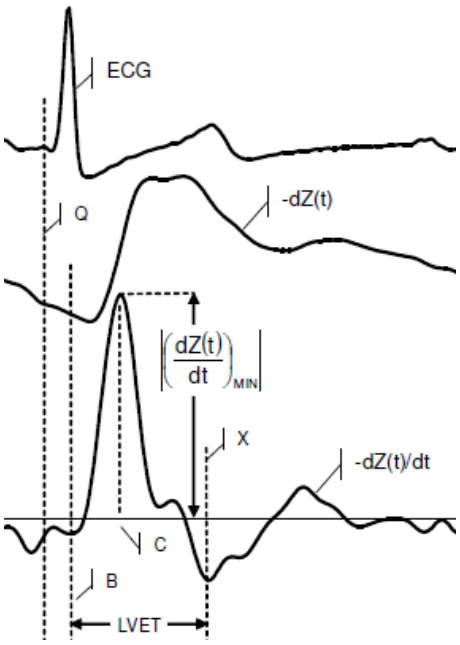

Figure 6: Tracing of surface ECG, change of impedance -ΔZ(t), and the rate of change of impedance $-\frac{dZ(t)}{dt}$ for one cardiac cycle

Electrical Cardiometry and *Impedance Cardiography* (ICG) differ in the model applied to the impedance measurements, in particular to the interpretation of the origin of the significant change in impedance shortly after aortic valve opening.

1. The model of ICG contributes this decrease in impedance (or increase in conductivity) solely to the volumetric expansion of the ascending aorta. In other words, the portion of the ΔZ(t) signal waveform following aortic valve opening resembles the increase in volume in the aorta, or the wall motion of the aorta. Following this model, the calculation of $\frac{dZ(t)}{dt}$ results in

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determining the velocity of volumetric change (in radial direction but not in direction of the flow!), and $\frac{\left|\left(\frac{dZ(t)}{dt}\right)_{MIN}\right|}{Z_0}$ is an index of peak velocity (of the volumetric change).

2. In contrast, the model of *Electrical Cardiometry* (also referred to as *Electrical Velocimetry*™) interprets the sharp decrease in impedance (or increase in conductivity) as a result of the alignment of the red bloods due to the pulsatile flow in the aorta. In other words, the significant change in impedance correlates with a change in blood velocity (in the direction of the blood flow). Following this model, the calculation $\frac{dZ(t)}{dt}$ of results in determining the acceleration of the blood flow (in the direction of the flow!), and $\frac{\left|\left(\frac{dZ(t)}{dt}\right)_{MIN}\right|}{Z_0}$ is an index of peak acceleration (of the blood flow).

In a nutshell, the ICG model considers $\frac{\left|\left(\frac{dZ(t)}{dt}\right)_{MIN}\right|}{Z_0}$ as an index of peak velocity while the *Electrical Velocimetry*™ interprets it as an index of peak acceleration. Because the equation for stroke volume requires the entry of a velocity index, *Electrical Velocimetry*™ first applies a nonlinear transformation to the index of peak acceleration to obtain a mean velocity index and then enters this mean velocity index into the stroke volume equation.

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