Bioelectrical Impedance Analysis and its Clinical Application

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ABSTRACT

Introduction: Electrical behavior of body tissues is being studied for the past 100 years and the method has been progressively improvised for various clinical applications in the human body. Bioelectrical impedance analysis is a noninvasive and low-cost tool commonly used for cardiovascular monitoring and body composition analysis.

Method: In this method, a small magnitude constant amplitude sine-wave current is applied to body surface as a carrier of information. The signal developed along the current path is sensed from the body to yield electrical impedance of the body segment (commonly known as bio-impedance). Anatomical factors influence the magnitude of bio-impedance, whereas physiological processes such as blood circulation and respiration modify the change in impedance as well as a rate of change of impedance as a function of time.

Clinical applications: Though some of the physiological processes measured using bio-impedance since its introduction in 1940 have taken back seat in view of evolution of other noninvasive methods like color Doppler, it is a reliable noninvasive method for continuous monitoring of cardiac output in physiological laboratories and intensive care units, study of blood flow variability as a substitute to blood pressure variability, and body mineral and fat analysis. This review article describes various concepts and methods of bio-impedance analysis used in clinical practice.

Keywords: Bio-impedance analysis, Blood flow variability, Body composition measurement, Impedance plethysmography, Vector impedance cardiography.

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INTRODUCTION

Electrical impedance is the result of the opposition of a material to the flow of an alternating electric current. Impedance (Z) measured comprises of the sum of two vectors, resistance (R), and reactance (Xc), which is measured at a selected frequency and is mathematically represented by the equation Z2 = R2 + Xc2. The resistance of a material depends upon its physical property known as resistivity, which inversely relates to the product of the number of charge carriers available in the material per unit volume, charge (in coulombs) per charge carrier and mobility of the charge carrier.1,2 In the case of the human body, reactance is mainly contributed by skin and body electrode interface. Capacitive (storage) effects of cell membranes, tissue interfaces, and structural features3,4 have a small but significant contribution to the reactance of the body segment under consideration.

Impedance measurement in biological material with the application of steady electric field leads to polarization effect due to capacitance formed by the electrodes with the biological specimen as the dielectric. This difficulty is reduced to a large extent by employing time-varying electric field, usually sinusoidal, in place of direct current (DC). The frequency of the time-varying field is chosen between 20 kHz and 200 kHz for the measurement of biological materials to limit the above-described effects5 to 4% and also to avoid bio-originated electrical signals like electrocardiogram, electromyogram, and electroencephalogram.

A typical bio-impedance measuring system comprises a carrier current generator (including sine-wave oscillator followed by the voltage to current converter and output electrodes), an impedance sensing circuit, and an electronic processor as shown in Figure 1. Sinusoidal current from a carrier current generator is passed through the body using carrier electrodes I1 and I2. The developed voltage along the body segment is sensed using sensing electrodes V1 and V2 and electronically processed to yield electrical impedance (Z) of the body segment, which is directly proportional to the sensed voltage signal. A change in the impedance of the body segment

Fig. 1: Block diagram of a typical impedance measurement system

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Bioelectrical Impedance Analysis and its Clinical Application

caused by physiological processes such as blood circulation and respiration can be obtained by subtracting the initial value of the impedance from the instantaneous impedance and is known as \( \Delta Z(t) \) signal. The first-time derivative of \( Z \) commonly known as \( \Delta Z/dt \) is also obtained by the electronic processor. Since \( \Delta Z(t) \) and \( \Delta Z/dt \) are produced by the physiological processes, it is possible to extract the changes produced by one particular process by either suppressing the other process or by signal processing techniques.\(^6\)

Measurement of physiological processes from these impedance signals is a vast field known as impedance plethysmography (IPG), impedance cardiography (ICG), vector ICG (VICG), and blood flow variability analysis (BFVA). Anatomical factors influence the magnitude of bio-impedance and its study is separately known as body composition analysis (BCA). It makes use of multiple frequencies of carrier current to selectively ensure its passage through extracellular or intracellular space as depicted in Figure 2.

Some of the physiological processes measured using bio-impedance since its introduction in 1940, such as measurement of peripheral arterial blood flow and peripheral venous circulation, have been overtaken by other noninvasive methods like color Doppler. However, it has strategic relevance in the continuous monitoring of cardiac output in intensive care units, the study of blood flow variability as a substitute to blood pressure variability, and BCA. These clinical applications are briefly described in the following sections.

CLINICAL APPLICATIONS

Impedance Plethysmography

IPG is an indirect assessment of blood volume changes in any part of the body from changes in the electrical impedance of the body segment. It was first introduced by Nyboer\(^8\) in 1940. He later proposed "parallel conductor theory" for the estimation of peripheral blood flow,\(^9\) as follows:

\[
\Delta V = \rho_b \frac{L^2}{Z_0} \Delta Z \quad (1)
\]

where \( \Delta V \) is the change in the blood volume, \( \rho_b \) is the resistivity of blood, \( L \) is the length, \( Z_0 \) is the gross impedance of the body segment in situ, and \( \Delta Z \) is the change in impedance caused by physiological processes. This principle was later extended for the estimation of stroke volume (SV) by Kubicek et al. with an introduction of \( dZ/dt \) waveform\(^10\) as follows:

\[
SV = -\rho_b \frac{L^2}{Z_0} (\Delta Z / \Delta t)_{SV} \cdot T_{LVET} \quad (2)
\]

where \( \Delta V \) and \( \Delta Z \) in equation (1) have been replaced by SV and product of (\( dZ/dt \))\(_m \) and \( T_{LVET} \) respectively, and \( L \) is the distance between the base of the neck and the xiphisternum. \( (dZ/dt)_m \) is the height and \( T_{LVET} \) is the duration of the systolic peak in \( dZ/dt \) signal. This paved the way for continuous monitoring of cardiac output (CO) noninvasively.

For the measurement of peripheral blood flow, the venous occlusion principle was employed to take care of low signal to noise ratio of Nyboer’s technique. However, this not only interfered with the venous circulation but also made the measurement off-line. Parulkar et al.\(^11\) and Jindal et al.\(^12\) extended Kubicek’s method of SV estimation using \( dZ/dt \) for online estimation of blood flow index (BFI), differential pulse arrival time (DPAT), and pulse terminating time (PTT) in various segments of the limb.\(^11\) These online parameters were very useful for the diagnosis and posttherapeutic assessment of peripheral vascular occlusive diseases (PVOD) as follows:

- Decrease in BFI at and below a segment in the upper or lower extremity suggests a significant reduction in the blood flow in the corresponding segment.
- Increase in DPAT in a segment indicates an arterial occlusion in the proximal segment in contrast to a decrease in DPAT suggesting generalized narrowing.
- At least, 50% (of control value) value of BFI in the segments distal to arterial occlusion indicates good collateral circulation around the occlusion.
- No further decrease in BFI in distal segments of the limb having arterial occlusion indicates good collateral circulation.
- Normal BFI in supine and a significant increase (>30%) in BFI on elevating the extremity by 45° indicates venous stasis due to primary varicosity.
- Decreased BFI in supine followed with a significant increase (>30%) on elevation suggests varicosity secondary to deep vein thrombosis in the respective extremity.

Rich information provided by IPG on PVOD was slowly overtaken by the color Doppler technique during the 1980s due to the simplicity of the later.\(^4\) However, the color Doppler was unable to provide the following information which was freely available with IPG:

- Status of collateral circulation due to small cross-sectional area and low blood velocity in the collateral vessel.
- Detection of multiple blocks in the limb due to low blood velocity in the distal limb segment.
- Total blood flow in the limb segment, which is important to know for the survival of the limb.

Impedance Cardiography

ICG reveals blood volume changes in the thoracic cavity consequent to rhythmic contraction of heart, ejection of blood to the aorta, and relaxation of the heart. For this measurement, the electrodes are applied around the neck (I1, V1) and around the abdomen (I2, V2) as shown in Figure 3A, commonly known as the neck–abdomen configuration. Typical ICG recording and its main characteristic points are shown in Figure 4. A, B, C, X, Y, O, and Z are the main
opening of aortic valve (as verified by M-mode echocardiography); point C corresponds to instant of maximum ejection rate of left ventricle (evidenced by electromagnetic flowmetry); X and Y coincide with closure of aortic valve and pulmonary valve respectively (shown by M-mode echocardiography); O is coincident with mitral valve opening and Z is coincident with the end of rapid filling phase.

Excellent temporal correlation of ICG with various events of the cardiac cycle has led to the assessment of systolic time index (STI), myocardial contractility (Heather’s index), aortic regurgitation fraction (ARF), and mitral regurgitation fraction (MRF) with reasonable precision. A study conducted by Barde et al. in 2006 reports the comparison between two methods of electrode placement in ICG. They have compared the horizontal positioning of electrodes and vertical positioning of electrodes and concluded that the horizontal method using ICG with four surface ECG electrodes is equally good for the measurement of cardiac output in humans.

Since ICG gives beat-to-beat stroke volume [as given by equation (2)], it becomes the method of choice for continuous monitoring of cardiac output noninvasively. Figure 5 shows the graphic-user interface of this application for noninvasive cardiac output monitor (NICOM) developed by Bhabha Atomic Research Centre (BARC). Cardiac output is computed by multiplying the SV value by heart rate. For continuous monitoring, the spot electrodes can be used in place of the conventional band electrodes as suggested by Barde et al. For more precise estimation, the user can mark B, C, and X points on the waveform for any number of cardiac cycles up to 10 and click on “compute” to obtain average cardiac output of 10 cardiac cycles.

Vector ICG
Vector electrocardiography (VICG) was introduced in 1913. It displays P, QRS, and T complexes in the form of a “loop”. In a similar

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Fig. 3A and B: Positioning of electrodes for recording in neck abdomen configuration (A); all electrodes are tied as a loop around the body segment in ICG. Horizontal positioning electrodes in trans-chest configuration for VICG (B); carrier electrodes (I1 and I2) are band electrodes while sensing electrodes (V1 and V2) are spot electrodes.

Fig. 4: Characteristic points on the ICG waveform

Fig. 5: Graphic-user interface for continuous monitoring of cardiac output noninvasively
this network, any blood volume change in aorta or vena cava gives maximum contribution to change in total impedance. However, blood volume change in pulmonary arteries does not contribute significantly due to the presence of $Z_T$ in series.

On the contrary, as shown in Figure 6B, impedance due to the main pulmonary artery ($Z_{RPA}$, $Z_{LPA}$, and $Z_{AO}$) cover the entire segment in trans-chest configuration; aorta and vena cava join serially with tissue impedance $Z_T$ and connect in parallel to the above network; impedances due to pulmonary veins ($Z_{PV}$) join with the left atrium ($Z_{LA}$) to cover the entire length between the sensing electrodes; heart’s contribution is minimal as it does not cover the entire sensing area. Thus, the change in impedance in the pulmonary arteries and the pulmonary veins are reflected maximally in the total impedance, whereas a change in the impedance of vena cava, aorta, and heart has a negligible contribution.

Figure 7 illustrates VICG waveforms recorded in the neck–abdomen and trans-chest configuration termed as VICG(V) and VICG(H), respectively, in patients with isolated mitral regurgitation (Fig. 7A) and patient with functional tricuspid regurgitation complicating mitral regurgitation (Fig. 7B), respectively. As can be seen from the figure, the VICG(V) waveform looks similar in both the cases, whereas the VICG(H) waveform is distinctly different in Figure 7B. The diastolic wave is much larger than the systolic wave representing gross malfunction of the pulmonary circulation.

manner, VICG was introduced by Pedhnekar et al.\(^\text{19}\) by recording impedance changes in trans-chest configuration as shown in Figure 3B. Figure 6 gives the impedance analysis of thoracic cavity in conventional neck–abdomen (A) and trans-chest (B) configurations. In (A), impedance recorded between the neck and the xiphisternum is a result of series and parallel combinations of several impedances of contents of the thoracic cavity as follows. Impedances due to right and left pulmonary arteries ($Z_{RPA}$ and $Z_{LPA}$) join parallely and then with general impedance ($Z_T$) in series; impedances due to superior vena cava ($Z_{SVC}$), inferior vena cava ($Z_{IVC}$), and right atrium ($Z_{RA}$) join in series and get parallely connected to the pulmonary artery network; aorta with its superior branches cover the entire length between the electrodes and, therefore, contributes an impedance $Z_{AO}$ in parallel with the above network. Remaining tissues and other chambers of heart contributing respective impedance $Z_T$ and $Z_H$ join in series and then parallely to the above network. As evident from

![Diagram of the thoracic cavity](image)

**Figs 6A and B:** (A) Electrical equivalent circuit of the thoracic cavity in the neck–abdomen configuration, it is dominated by impedance changes in the aorta, right ventricle, and vena cava. (B) Electrical equivalent circuit of the thoracic cavity in trans-chest configuration, it is dominated by impedance changes in the pulmonary arteries.

![VICG waveforms](image)

**Figs 7A and B:** VICG waveforms recorded from a patient with isolated mitral regurgitation (A) and with tricuspid regurgitation complicating mitral regurgitation (B). No significant difference is noted between VICG(V) waveforms in (A) and (B). VICG(H) waveform in (B) is markedly different from that of (A) due to the presence of tricuspid regurgitation. The diastolic wave is much larger than systolic wave in VICG(H) waveform in (B).
Body Composition Measurement

Body composition measurement is based on the measurement of electrical behavior of the tissue and cells in the body. As described in Introduction section, different body tissues contribute differently to the total body impedance and, therefore, can be used for composition analysis. The measurement is carried out with single as well as multiple frequencies. Single-frequency BIA were found to be potentially useful in the industry up to 1990; subsequently, several multifrequency analyzers have been introduced. The utilization of BIA as a bedside technique is convenient and safe due to its noninvasive nature. Simplicity and its ability to produce quick reproducible outcomes has enhanced clinical usage of this technique for BCA. Many regression equations are available for estimating total body water (TBW) and fat-free mass (FFM) as a function of total body impedance, weight, height, gender, and age.

BIA measurement can be carried out by whole body approach as well as segmental approach. In whole-body approach, the measurement is taken between the wrist and the ankle as shown in Figure 8; however, it has a limitation as the body impedance is measured on one side (left or right). This limitation is overcome by the segmental approach as shown in Figure 9. In this approach, the body is represented by five segments (e.g., arms, trunk, and legs) and impedances and phase angles of each segment are measured separately by placing the electrodes as shown in Figure 9. From the segmental impedance and the phase angle values, the respective values of segmental resistance and reactance can be calculated. On the basis of segmental impedance values, reactance, and phase angle of the total body can be calculated by vector summing the parameters of each segment.

In a recent study, Patil et al. have validated BIA fat mass of the body against X-ray absorptiometry in Indian adults. Fat mass (FM) was measured in 131 subjects using dual energy X-ray absorptiometry (DEXA) and using commercially available segmental multi-frequency bioelectrical impedance analyzer as 28.11 ± 9.30 kg and 28.12 ± 9.11 kg, respectively. Patil et al. have also given following prediction equation for bone mineral content (BMC) in human subjects which is helpful in screening for osteoporosis as follows: \[ \text{BMC} = -3.5268 + (0.0279 \times h) + (0.01456 \times w) + (184 \times \left(\frac{h^2}{Z_{50}}\right)) - (1.08 \times (w \times \left(\frac{h^2}{Z_{6.25}}\right))) - (0.0032 \times \text{age}) - (0.103 \times \text{sex(men = 1, women = 0)}), \]

where \(h\) is the height in cm, \(w\) is the weight in kg, age in years, \(Z_{50}\) is the body impedance at 50 kHz, and \(Z_{6.25}\) is the body impedance at 6.25 kHz.

Physiological Variability

Heart rate variability (HRV) has emerged as a potential tool for assessment of autonomic nervous system (ANS) and for critical care monitoring since the late 1970s, to the extent that it has become a routine feature of state-of-the-art bedside monitors. Usually, for HRV monitoring, electrocardiogram of the patient is recorded for 5 minutes and the same is used to derive consecutive RR intervals (time elapsed between two consecutive R-waves in the ECG signal) or the heart rate on a beat-to-beat basis. The Fourier transform of the RR-interval series yields HRV spectrum, which comprises three peaks: very low frequency (VLF), low frequency (LF), and high frequency (HF). These are related to baroreceptor/renin–angiotensin, sympathetic nervous system, and a parasympathetic component of ANS. Long-term (24 hours) HRV studies are also in practice which yield ultra-low-frequency peaks relating to circadian rhythm, thermos-regulation, etc. Since peripheral blood flow signal is consequent to cardiac contraction, it can also be used for obtaining HRV. This signal from IPG undergoes the same process as that for electrocardiogram to obtain HRV.

There is growing recognition that assessment of physiological variability (e.g., heart rate and arterial pressure) across multiple time scales may provide important insight into acute and long-term clinical outcomes, such as the risk of stroke, myocardial infarction, and organ damage from hypertension. The variability analyzer
Bioelectrical impedance analysis has been assisting the medical fraternity right from its inception for the measurement of central and peripheral blood flow. Its unique advantage of assessing collateral circulation and distal arterial runoff and continuous monitoring of cardiac output has no parallel in noninvasive instrumentation. Vector impedance modeling can help in the assessment of selected parts of the body. It is more versatile than color Doppler as it can give BCA and physiological variability analysis.

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**References**