Measurement of Cardiac Output using Bioimpedance Methd

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ABSTRACT

Cardiac output (CO), in conjunction with hemoglobin concentration and arterial oxygen saturation are the cornerstones of oxygen transport. Cardiac output is the functional expression of cardiovascular performance and can be used to confirm the need for, or usefulness of treatment. Determination of cardiac output is an important procedure in interventional cardiology and also used in cardiothoracic surgery. Existing invasive cardiac output measurement methods are reasonably accurate, but their use is limited to the intensive care units. The risk of infection, blood loss or other complications associated with arterial catheters is also a matter of concern. Thus, a robust noninvasive alternative is considered to be desirable. Impedance cardiography (ICG) is one such method, which is noninvasive, easy to use, provides continuous CO measurement and has a better accuracy than the existing minimally invasive techniques. Traditionally, ICG involves applying a current field longitudinally across a segment of thorax by means of a constant magnitude, high frequency, and low amplitude alternating current. Most commonly implemented methods utilize the dependency of the peak systolic upslope of $\Delta Z(t)$ on the volumetric changes. However,change in the bioimpedance due velocity-induced blood resistivity variations has also been recently explored.This paper presents a detailed study of impedance cardiography as a tool for measuring the cardiac output and the issues related with it.

Keywords
Cardiac output, impedance cardiography, non invasive.

1. INTRODUCTION
Cardiac output, stated simply, is the volume of blood pumped by the heart per minute. It is expressed in l/sec and normal range of cardiac output is 4-8 l /min. It is equal to the volume pumped out by the heart in one contraction, known as stroke volume, multiplied by heart rate. Cardiac output is the functional expression of cardiovascular performance and can be used to confirm the need for, or efficacy of treatment. Determination of cardiac output is an important procedure in interventional cardiology and also used in cardiothoracic surgery. Cardiac output is one of the most important parameters for cardiac function monitoring, providing an estimate of whole body perfusion oxygen delivery and allowing for an understanding of the causes of high blood pressure [1].

There are various techniques available to measure the cardiac output. These are Fick's principle, dye dilution, thermodilution, pulse pressure methods and ultrasonic methods. But all of these are either invasive or minimally invasive techniques. Till today the pulmonary artery catheter thermodilution (PATD) technique is considered as a gold standard for CO measurement in the intensive care units and operation theatre. These requires catheterization of the patient which itself adds to morbidity and sometimes mortality of the patient.

Bioimpedance technique is a non-invasive technique which estimates the stroke volume based on the impedance changes occurring inside the human arterial system due to change in blood volume. This technique is sometimes referred to as impedance cardiography. Cardiac output is easier to measure by impedance cardiography than by thermodilution with a pulmonary artery catheter. It can be applied quickly and easily. In addition, it does not pose a risk of infection, blood loss or other complications associated with arterial catheters. Further, continuous CO monitoring is not possible with these all catheter related invasive methods. Non–invasive impedance cardiography appears to be a solution for all these problems. This paper studies impedance cardiography that measures volume changes of blood in chest, calf or any other region of the body by sensing electrical resistance changes in respective area to estimate the cardiac output.

2. THEORY OF BIOIMPEDANCE MEASUREMENT
Bioimpedanceis the response of a living organism to an externally applied electric current. It is a measure of the opposition to the flow of this electric current through the tissues. The measurement of the bioimpedance (or bioelectrical impedance) of the humans and animals has proved useful as a non-invasive method for measuring various physiological parameters such as blood flow (often referred to as bioimpedance plethysmography). The technique of measuring volume changes of an organ is called Plethysmography. The volume may be due to blood, water or air. Blood flow measurement in arteries is termed as Impedance Plethysmography (IPG), while that in veins is termed as Occlusive Impedance Plethysmography (OIP). Impedance Cardiography (ICG) measures blood flow in major vessels of cardiac region from which stroke volume is obtained. Plethysmography finds a wide range of application in medical diagnosis, right from measurement of haemodynamic parameters[2] upto measurementof *vatta, pitta* and*kapha* in ayurveda for diagnosis of various diseases[3].

Cardiac output is calculated as the product of the stroke volume and heart rate. The stroke volume can be defined as the volume of the blood pumped by the heart during every

cardiac cycle. This volume is inversely proportional to the resistance offered by the vascular bed. Thus measuring the changes in this impedance would offer a good estimate of the changes in stroke volume. Methods measuring the variations in the bioimpedance along the thoracic region, brachial artery and from the forearm/limbs have been detailed in this paper.

2.1 Thoracic Bioimpedance Method (TEB)

Thoracic bioimpedance (TEB) is the least invasive of the CO monitors. The technology was first developed by Kubicek et al. [4] in the 1960's, with the initial testing being carried out on astronauts [5]. The basis for its use was later established and improved by researchers over the following decades, based on animal and human research [5, 6, and 7].

CO is directly proportional to the mean pressure drop over complete systemic arterial-venous circuit (i.e., mean arterial pressure (MAP) – central venous pressure (CVP)) and is inversely proportional to the total peripheralresistance (TPR) of the circuit.

CO ≈ (MAP – CVP) / TPR

This relationship is analogous to the Ohm's law where current is analogous to cardiac output (flow), pressure is analogous to the voltage and vascular resistance is analogous to the resistance.

$$V = IR \rightarrow R = \frac{V}{I}$$

If the current, *I* is kept constant, the changes in the voltage (ΔV) would directly reflect the changes in the resistance (ΔR). For alternating current flow, resistance is known as impedance (Z) and is complex, frequency-dependent parameter. For alternating current we can say that:

$$V = IZ \rightarrow Z = \frac{V}{I}$$

Z is measured in ohms. If current, *I*remains constant, then periodic changes in voltage (ΔV) produce concurrent changes in impedance (ΔZ)

$$\Delta V \approx \Delta Z$$

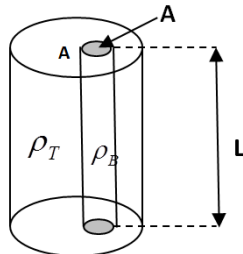For simplicity, the thorax can be modeled as a concentric cylindrical conductor of length L.

Fig. 1 :Simple Concentric cylinder model of human thorax

The outer conductor is the thoracic cavity while the inner conductor with the smaller diameter represents the enclosed greater vessels as shown in Fig.1.

Let Z_B and Z_T be the impedances of the inner and outer conductor respectively. The total impedance Z of the thoracic region can be considered to be the parallel combination of Z_B and Z_T.

$$Z = \frac{Z_B Z_T}{Z_B + Z_T} \qquad (1)$$

$$\frac{dZ}{dZ_B} = \frac{Z_T^2}{(Z_B + Z_T)^2} = \frac{Z^2}{Z_B^2}$$

Impedance of a conductor is given by

$$Z_B = \frac{\rho_B L}{A_B} \rightarrow A_B = \frac{\rho_B L}{Z_B} \qquad (2)$$

$$Volume = Length * cross\,sectional\,area = AL = \frac{\rho L^2}{Z} \qquad (3)$$

$$Z_B = \frac{\rho_B L^2}{V_B}$$

Let F_B represent the volume of the blood flowing through the inner cylinder.

$$\frac{dF_B}{dZ_B} = \frac{d(A_B L)}{dZ_B} = -\frac{\rho_B L^2}{Z_B^2} \quad (4)$$

Substituting the value of $\frac{dZ_B}{Z_B^2}$ from equation (1), the flow rate can be calculated as :

$$\frac{dF_B}{dt} = -\frac{\rho_B L^2}{Z^2}\frac{dZ}{dt} \quad (5)$$

Where Z is the mean value of the measured impedance. It is also referred to as non pulsatile basal impedance.

Thus maximum change in in the stroke volume during one cycle can be given as :

$$SV = -\frac{\rho_B L^2}{Z^2}\min_{t \in cycle}\frac{dZ}{dt}T_E (6)$$

where T_E is the ventricular ejection time.

Kubiceket. al[4] developed the impedance cardiograph where four-electrode technique (Fig. 2) was used. This technique would enable to minimize the noise caused by variation of electrode-to-skin contact resistance dueto movements of patient. Special band electrodes put around the body as shown in Fig. 2.
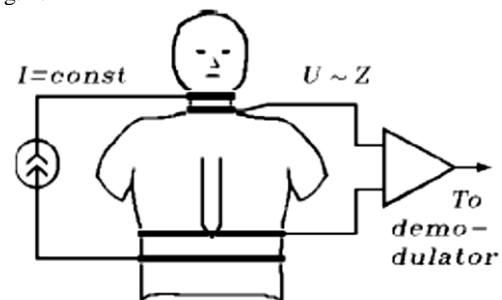

Fig. 2:Kubicek'sArrangement of electrodes at the recording of theelectrical impedance for SVestimation[4].

As the output impedance of the current source and the input impedance of the voltage amplifier can be very high, the effects of the changes of the contact resistance would be minimal. Use of band electrodes faced a lot of criticism due to the noise introduced by them.

Sramek's[8] spot electrode placement helped in eliminating this problem. 8 spot electrodes were placed at convenient points on the lines where the band electrodes were laid (Fig 3)
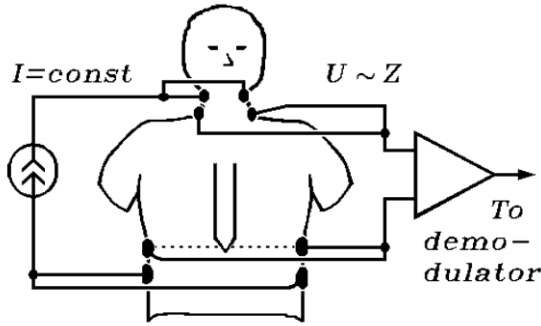

Fig. 3:Sramek's arrangement of electrodes at the recording of theelectrical impedance for SVestimation[8].

The recent TEB setup connects the patient to the monitor via a patient cable attached to eight solid-gel, disposable electrodes (Fig. 3). The TEB measurement current is passed through the thorax in a direction parallel with the spine between the beginning of the thorax (the line at the root of the neck) and the end of the thorax (at the level of diaphragm – the xiphoid process level). Four dual sensors with eight lead wires are placed on neck and chest. The "current" electrodes, deliver an alternating current of 4 mA at a frequency in the range of 100KHz to 400 KHz. Current transmitted by outer electrodes seek the path of least resistance i.e.the blood filled aorta.
Baseline impedance is measured using inner electrodes. With each heartbeat, blood volume and velocity in the aorta changes. Corresponding change in impedance is measured. Baseline and changes in impedance are used to measure and calculate hemodynamic parameters. These four electrodes also detect four different vectors of the ECG signal. The Heart Rate (HR) is derived from the R-R intervals of the ECG signal. Due to the anatomical shape of the thorax, a preferential placement for all eight electrodes is along the frontal plane – the widest thoracic dimension.

2.1.1 Limitations
TEB is affected by a number of factors [9]:
(i)   Changes in tissue fluid volume
(ii)   Respiration-induced changes in the volume of pulmonary and venous impedances.
(iii)   Changes in electrode contact or position
(iv)   Arrhythmia, as the ventricular ejection time is determined using the interval betweenQRS complexes
(v)   Acute changes in tissue water, for example, pulmonary or chest wall edema or pleural effusions
(vi)   Noise from electrocautery, mechanical ventilation and surgical manipulation
(vii)   Changes in myocardial contractility, for example, from anesthetic drugs or ischemia.
(viii) Extremely oily skin and sweating.

2.2 Brachial Velocimetry
The Bioimpedance method relates the changes in the stroke volume with the change in the impedance. However, this change in impedance is not only because of change in volume but also due to the change in blood velocity[10][11]. I. C. Henry et.al.[12] devised a new approach of stroke volume measurement from the brachial artery based on this principle. The performance of the TEB measurement highly deteriorates in the presence of the excess extravascular lung water, namely, pulmonary edema. Chemlaet. al[13] showed that the peak aortic blood acceleration is highly correlated to peak brachial artery acceleration (r = 0.79).

The upper arm, more specifically the brachium, is considered to be a cylinder and the impedance across the vessel length is given by Eq.3

Assuming that variation in the length of the vessel w.r.t. to time is of trivial magnitude and that all other variables are continuously differentiable functions of time;

$$\frac{dZ(t)}{dt_{max}} = \frac{L^2}{V_B}\frac{d\rho_B(t)}{dt_{max}} - \frac{\rho_B L^2}{V_B^2}\frac{dV_B(t)}{dt_{max}} \tag{7}$$

Eq.7 expresses the variation of the impedance firstly, due to variation in the velocity and secondly due to variation in the volume of the blood within the vessel(V_B). The term $d\rho_b(t)/dt_{max}$ is the peak rate of change of the red cell velocity. The first derivative on the right hand side of (7) is given in Ω s^{-2} and the second derivative on the right hand side of the equation Ω s^{-1}. By comparative time domain analysis for dZ/dt_{max}, obtained from the transthoracic approach, it was determined that dZ/dt peaks in the time domain of acceleration [10,11], and therefore,

$$\frac{dZ(t)}{dt_{max}} = \frac{L^2}{V_B}\frac{d\rho_B(t)}{dt_{max}} \tag{8}$$

This means that the transbrachialdZ/dt_{max} represents the peak rate of change of the transbrachial specific resistance of blood (Ω s-2) inducedby change in blood velocity.The square root transformation of the quotient of dZ/dt_{max} and $Z_0$was necessary to obtain mean ohmicvelocity.

$$Mean\ Ohmic\ Velocity = \sqrt{\left(\frac{L^2}{V_B}\frac{d\rho_B(t)}{dt_{max}}\frac{1}{Z_0}\right)} = \sqrt{\left(\frac{dZ(t)}{dt_{max}}\frac{1}{Z_0}\right)} \tag{9}$$

For impedance measurement, two current flow electrodes are placed, one deep in the left axilla, and the second on the distal brachium, which is medial to the antecubital fossa. Voltage sensing electrodes are placed near each current flow electrodes and spaced apart at approximately 5 cm. A proportionality constant between the brachial velocimetry-derived SV and ICG-derived SVneeds to be derived, in order to accurately determine the cardiac output. This can be obtained by implementing a variation of the equation of continuity for conservation of mass flow.

$$Area\ (cm^2) \times \bar{v}\left(\frac{cm}{s}\right) = volume\ (ml) \times \sqrt{\left(\frac{dZ(t)}{dt_{max}}\frac{1}{Z_0}\right)}\left(\frac{1}{s}\right) \tag{10}$$

$$Volume_{brachial} = Volume_{TEB} \cdot \frac{\sqrt{\left(\frac{dZ(t)}{dt_{max\ (ICG)}}\frac{1}{Z_0}\right)}}{\sqrt{\left(\frac{dZ(t)}{dt_{max\ (brachial)}}\frac{1}{Z_0}\right)}} \tag{11}$$

The stroke volume obtained using this technique shows a percentage error of 28.9 % when compared with the Doppler echocardiography using the Bland-Altman analysis [12]. Impedance-derived SV obtained from the transbrachial approach has potential advantages over the transthoracic approach. First, the brachial artery is less likely to be corrupted by competing, extraneous volume and velocity

signals.Also, the noise introduced in the measurement due to extravascular lung water has been eliminated, which is the major hurdle in thoracic ICG measurements in impaired patients. The impedance change, its rate of change and its peak value, dZ/dt_{max}, can be considered to be purely blood velocity-induced blood resistivity changes, as the change in the brachial artery diameter is around 2-3% throughout the flow interval.

2.3 Peripheral pulse analysis

The limitations of the thoracic bioimpedance method inspired a lot of research in electrode placement for impedance measurement. P. Barde[14] et al placed the electrodes in a horizontal manner on the thoracic area.
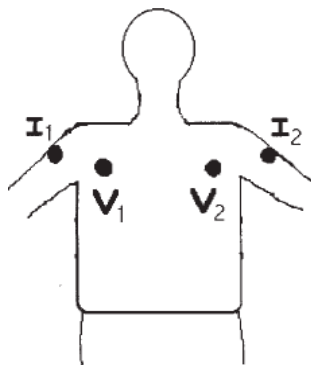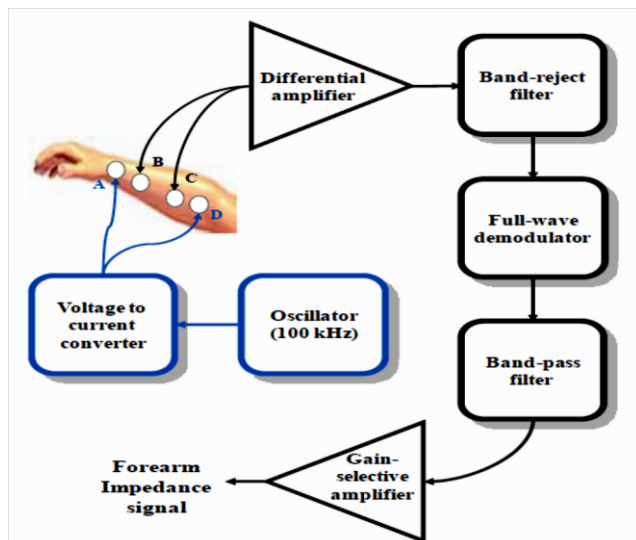

Fig 4: Horizontal placement of electrodes [14]

Fig 4: Placement of electrodes on the forearm [15]

Two current injecting spot electrodes (I_1, I_2) were placed on mid arm laterally one on right (I_1) and other on left (I_2) side and the voltage sensing spot electrodes were placed at 2nd intercostal space along the anterior axillary line which anatomically corresponds the level of aortic arch (Fig. 4). The outer pair transmitted the ac current of 4 mA, 100 KHz, and the changes in the impedance were measured from the outer electrodes.

Jia-Chug Wang et.al [15] measured the impedance by placing the electrodes on the forearm as shown in the Fig.4. Both the methods use the Kubicek's equation (Eq.3) for calculation of stroke volume. The cardiac output is estimated by multiplying this SV with the heart rate after multiplication by a suitable proportionality constant.

3. VALIDATION STUDIES

Sharma et. al[16] compared the TEB technique with the thermodilution showing a good correlation between the estimated values .The Bland Altman analysis comparing the two methods showed an error of 19.3 %. Peacock et al. [17] evaluated the impact of impedance cardiography in 89 dyspneic patients. The investigators concluded that impedance cardiography data resulted in significant changes in diagnosis and therapeutic planning during the evaluation of such patients. Deepak et al [18] showed that the correlation coefficient between the TEB and the invasive Fick's method is 0.9. Several investigators found thatTEB compared favorably with PATD in varying settingsincluding during cardiac catheterization, surgical patients,and emergency room patients [19]. Van DeWateret al. [20] found the TEB compared favorably with PATDin post cardiac surgical patients. Koobi et al. [21], usingwhole-body impedance cardiography in Coronary artery bypass grafting patients,reported excellent repeatability which allowed for continuousmonitoring.Spiess et al. [22] used BioZ (SonoSiteInc,Bothell,Wash,USA) intraoperatively for patients undergoingCoronary Artery Bypass Grafting and found that the technique initially comparedwell with PATD, but postoperatively, the Bland-Altman analysis was not as robust.C. Schmidt et. al.[23] showed that the agreement between TBEV transoesophagealDoppler echocardiography is clinically acceptable, and these two techniques can be used interchangeably.Cotter et.al.[24] compared TEB with PATD in patients with acute heart failure concluding that the correlation between the two methods is fairly good.

Table 1 presentsvarious published studies validating data yielded fromimpedance measurement to standard invasive methodologies(the presumed "gold standard").

4. CONCLUSION

The desirable characteristics of the cardiac output monitoring techniques are accuracy, reproducibility, operator independency, fast response, continuous and easy to use. The bioimpedance methods and its variants described in this paper offer the advantages of ease of use, operator independency and continuous measurement. Validation studies carried out by researchers have shown a better reproducibility of the results. Accuracy and fast response are still an issue of concern. Several instruments are available commercially, which apply these principles for measurement of cardiac output, but most of them requires external calibration. A robust estimate of the proportionality constant needs to be developed in case of brachial and peripheral impedance measurement. The measurement of cardiac output is extremely important in case of haemodynamically unstable patients. Therefore, the measuring device must be robust against the noise introduced due to the movement of the patient (restlessness), increased cardiac activity (angina pain, fibrillation) and excessive sweating. Moreover, the effect of metal inside the body on the impedance measurement also needs to be investigated.

Author	Patient target group	Sample size	Method	Results and conclusion
Sharma et.al[16]	Age 39-73 yrs, post coronary artery bypass graft patients	46	TEB vs TD	Corrleation between ICG and Td is fair (0.548) during the volume control mode and weak during spontaneous breathing (0.469)
Peacock et. al [17]	Emergency department, Age >65 yrs, dyspnea, Heart failure, Pulmonary disease	89	TEB	ICG data changed diagnosis in 5.4%, medication in 23.6%, and dosage in 25% of patients.
Deepak et. al. [18]	Patients undergoing cardiac Catheterization	10	TEB vs Fick	Good correlation (r=0.9) between the cardiac index obtained by impedance cardiography and that derived invasively after cardiac catheterization
Gujjar et. al.[19]	Post-cardiac surgery patients	35	TEB vs TD	Correlation coefficient : 0.856, with a percentage error of 26.44%
Van DeWater et al.[20]	Post-coronary artery bypass graft patients	53	TEB (Kubicek, Sramek,Sramek-Bernstein) vs TC	Correlation coefficient of CO Kubicek and TD : 0.309 CO Sramek with TD : 0.361 CO Sramek-Bernstein : 0.469
Koobi et al[21]	Patients undergoing a coronary artery bypass grafting operation.	74	whole-body impedance cardiography vs TD	The repeatability value (rv = 2.83 x SD) for whole-body impedance cardiography (rv = 0.46 L/min) was considerably better than for the thermodilution method (rv = 1.05 L/min).
Spiess et al.[22]	Patients undergoing primary elective coronary artery bypass surgery	47	TEB vs TD	Overall correlation coefficient : 0.71
C. Schmidt [23]	Patients scheduled for coronary artery surgery	35	TBEV vs Doppler	Correlation Coeffcient : 0.86
Cotter et. al. [24]	Acute heart failure	31	TD vs ICG	Correlation Coefficient : 0.85 Similar good correlation to CI at baselineand follow-up with vasodilation therapy. Highcorrelation for whole-body impedance, 1.4 L/min precision.
HK Jorgensen et. al. [25]	Patients scheduled for elective abdominal aortic surgery	12	TEB vs Doppler	Correlation coefficient : 0.15. The bias was +1.5 l/minute with 95% limits of agreement between -2.1 and 5.1 l/minute
Belardinelli et. al. [26]	Ischemic cardiomyopathy, No pulmonary disease, No hypertension,	25	TD vs ICG	Correlation Coefficient: 0.89 High correlation to CO at rest and during exercise.
Yung GL et. al. [27]	Pulmonary artery hypertension, Clinically stable	39	TD vs ICG TD vs Fick	Correlation Coefficient : 0.8

TEB : Thoracic Bioimpedance, TD : Thermodilution, ICG: Impedance Cardiography

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