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Stroke Volume Obtained from the Brachial Artery Using Transbrachial Electrical Bioimpedance Velocimetry*

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Abstract--**Stroke volume (SV) is the quantity of blood ejected by the cardiac ventricles per each contraction. When SV is multiplied by heart rate, cardiac output is the result. Cardiac output (CO), in conjunction with hemoglobin concentration and arterial oxygen saturation are the cornerstones of oxygen transport. Measurement of CO is important, especially in sick humans suffering from decompensated heart disease and systemic diseases affecting the contractility or loading conditions of the heart. Although reasonably accurate invasive cardiac output methods are available, their use is restricted to those individuals hospitalized in the intensive care units. Thus, a robust noninvasive alternative is considered desirable. Impedance cardiography (ICG) is one such method, but in patients with severe heart disease and/or excess extravascular lung water, the method is inaccurate. This paper concerns the introduction of a new method, transbrachial electrical bioimpedance velocimetry (TBEV). The technique involves passage of a constant magnitude, high frequency, and low amperage ac from the upper arm to the antecubital fossa. In all other respects, the operational aspects of TBEV are consistent with ICG. There is good evidence suggesting that the TBEV waveform and its derivatives are generated by blood resistivity changes only.**

I. Introduction

The determination of noninvasive cardiac output (CO) encompasses a range of pulsatile techniques and methods which measure left ventricular stroke volume (SV) [1]. Several approaches have been introduced over the past several decades, but none have proved sufficiently accurate to modulate therapy in sick patients. One such method that has been repeatedly studied is impedance cardiography (ICG). As summarized in several extensive meta-analyses, most studies show fair to good correlation and percent error approximately ±40% [2]. The technique performs at its worst in the presence of excess extravascular lung water, namely, pulmonary edema.

Introduction of equations defining SV in more robust biophysical terms have recently been introduced [1], and have been shown to provide more accurate approximations of SV along a full range of hemodynamic perturbations. However, the problem with excess extravascular lung water has not been solved.

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This paper concerns the rationale for a new bioimpedance method, transbrachial electrical bioimpedance velocimetry (TBEV). The impetus for the new method derived from data extracted from the work of Chemla *et al*. [3]. They showed that, peak aortic blood acceleration is highly correlated to peak brachial artery acceleration (r = 0.79). They also showed that peak brachial artery blood acceleration is not affected by downstream vasoactivity, but was only responsive to β_1 adrenoceptor stimulation of the heart [4].

The following study comprised two parts. The first exercise consisted of determining the magnitude of the TBEV volume conductor (V_C). The second study consisted of comparing TBEV-derived SV with that computed from Doppler echocardiography.

II. Theory

Consider the upper arm, more specifically the brachium, to be a cylinder comprising a smaller cylindrical blood vessel embedded within an encompassing cylindrical tissue impedance. If the brachial artery is considered a thick-walled, blood-filled, conductive conduit of length L (cm) and blood volume, V_b, and static specific resistance, ρ_b, the static impedance across the vessel length to an applied ac field in end-diastole is

$$Z = \frac{\rho_b L}{A} \equiv \frac{\rho_b L^2}{V_b}. \quad (1)$$

If Z becomes time variable with respect to blood velocity, which is represented by a change in the blood specific resistance $\rho_{b,}$ and a change in vessel blood volume V_b, displacing alveolar gas, the following results,

$$\Delta Z(t) = \frac{\Delta\rho_b(t) L^2}{\Delta V_b(t)}. \quad (2)$$

Where $\Delta Z(t)$ is a time variable impedance comprising a velocity component, eliciting a change in the specific resistance of blood $\Delta\rho_b(t)(\Omega\text{cm s}^{-1})$ and $\Delta V_b(t)$ $(\Omega(t))$, causing a change in vessel volume. With the exception of vessel length L (cm), if all variables on the left and right side of (2) are continuously differentiable functions of time,

$$\frac{dZ(t)}{dt} = \frac{L^2}{V_b}\frac{d\rho_b(t)}{dt} - \frac{\rho_b L^2}{V_b{}^2}\frac{dV_b(t)}{dt}, \quad (3)$$

where the first derivative on the right hand side of (3) is given in $\Omega\ s^{-2}$ and the second derivative on the right hand side of the equation $\Omega\ s^{-1}$. 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 [5,6], and therefore,

$$\frac{dZ(t)}{dt_{\max}} = \frac{L^2}{V_b}\frac{d\rho_b(t)}{dt_{\max}} \tag{4}$$

Which means that transbrachial dZ/dt_{max} represents the blood velocity induced, peak rate of change of the transbrachial specific resistance of blood ($\Omega\ s^{-2}$). As it was found, the square root transformation of the quotient of dZ/dt_{max} and Z_0 was necessary to obtain ohmic mean velocity.

$$\bar{v}_{ohmic} = \sqrt{\left(\frac{L^2}{V_b}\frac{d\rho_b(t)}{dt_{\max}}\frac{1}{Z_0}\right)} = \sqrt{\left(\frac{dZ(t)}{dt_{\max}}\frac{1}{Z_0}\right)} \tag{5}$$

where ohmic mean velocity is given as s^{-1}. When the extreme right side of (5) is multiplied by a V_C appropriately constructed for determining SV from the brachium and systolic flow time (SFT), transbrachial SV is obtained.

III. Methods

A. Determination of a TBEV Volume Conductor

In order for TBEV-derived SV to equal an accurately determined ICG-derived SV, a variation of the equation of continuity for conservation of mass flow was implemented. Solving for $V_{C(TBEV)}$;

$$V_{C(TBEV)} = V_{C(ICG)} \cdot \frac{\sqrt{dZ/dt_{\max(ICG)} \cdot Z_{0(ICG)}^{-1}}}{\sqrt{dZ/dt_{\max(TBEV)} \cdot Z_{0(TBEV)}^{-1}}} \tag{6}$$

In the pilot study, performed primarily to determine a constant of proportionality, such that $SV_{(TBEV)} = SV_{(ICG)}$, 38 healthy subjects were recruited for study. Informed and written consent was obtained from each participant. To obtain brachial artery impedance changes, two current flow electrodes were placed, one deep in the left axilla, and the second current flow electrode placed on the distal brachium, medial to the antecubital fossa. Voltage sensing electrodes were placed, each proximate to the current flow electrodes and spaced apart at approximately 5 cm. Subjects were measured while supine on an exam table, breathing freely. TBEV impedance data were obtained from a custom analog circuit generating a voltage output that was digitized in real time using a BIOPAC UIM100C system (BIOPAC, Goleta, CA, USA), and then stored on a personal computer (Sotera Wireless, San Diego, CA, USA). All waveforms were sampled at 16-bit resolution and 500 Hz. The system was powered by a 12 V battery to expunge any 60 Hz common mode artifacts. Satisfactory TBEV signals were obtained in all 38 subjects. ICG transthoracic impedance data, namely, dZ/dt_{max} and $Z_{0,}$ were obtained from a commercially available ICG device (Lifegard II, Analogic Corp., Peabody, MA, USA). The technique of obtaining SV via the transthoracic method, using an eight spot electrode, tetrapolar montage is fully described elsewhere (6). Paired data from both the TBEV and ICG techniques were collected over a five minute period and then stored on a personal computer. Upon finishing the calibration study, a mean value for the aggregate of $V_{C(ICG)}$ collected from all 38 subjects was also obtained. Equation 6 was then implemented to find a mean value for $V_{C(Brachium)}$. Bland-Altman analysis was used to find the mean bias, precision and percent error of the ohmic mean velocities obtained from both the transthoracic and transbrachial approaches. The following ICG equation was employed to determine SV from the transthoracic approach [5,6]:

$$SV_{(ICG)} = V_{ITBV} \cdot \sqrt{\frac{dZ/dt_{\max}}{Z_0}} \cdot T_{LVE} \tag{7}$$

Where V_{ITBV} (mL) is the volume conductor, allometrically equivalent to intrathoracic blood volume (ITBV) by body mass (kg), where the density of blood is $\cong$ 1.05 g mL^{-1} ($V_{ITBV} = 16W^{1.02}$) [6], $[(dZ/dt_{max} \cdot Z_0^{-1})]^{0.5}$ is a dimensionless ohmic equivalent of mean velocity (s^{-1}), and T_{LVE} is left ventricular ejection time (s).

B. Doppler/echocardiography-derived SV

SV obtained from the transbrachial electrical bioimpedance velocimetry technique was compared to SV obtained from interrogation of the combined transthoracic Doppler velocimetry/echocardiographic method. The study population consisted of 29 subjects, comprising 18 males and 11 females who had no apparent clinically significant heart disease.

Determination of Doppler/echo-derived SV was obtained by means of standard pulsed Doppler and tissue harmonic 2D image data. A Philips SONOs 5500 with D.0 software and transducer 3 was employed. Image frames and loops of relevant measurements were stored to an optical disk. According to standard guidelines [7], SV was obtained from the left ventricular outflow tract (LVOT), just proximal to the aortic valve leaflets, while the subject lay quietly on their left side. From this site both the systolic (time) velocity integral (cm) and LVOT diameter (cm) were obtained. The following equation was used to determine SV:

$$SV_{Doppler} = \pi\left[\frac{D}{2}\right]^2 \int_{t_1}^{t_2} v(t)dt \tag{8}$$

Where D (cm) is the diameter of the LVOT just beneath the aortic valve leaflets, and the integral is the systolic or time velocity integral (cm).

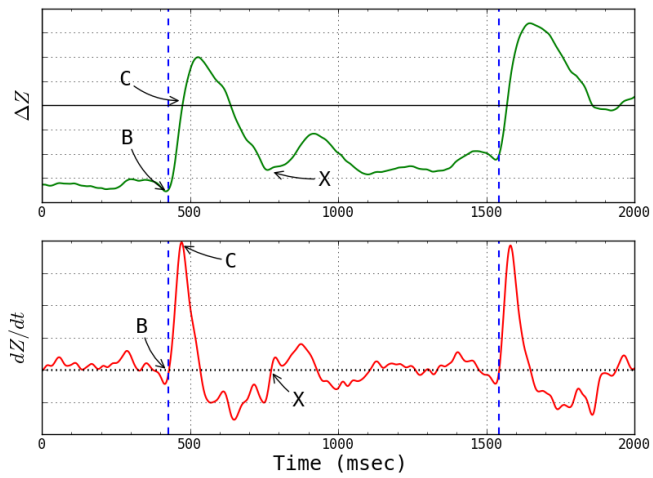

Figure 1. Impedance change and first derivative of change over two cardiac cycles, obtained from the brachial artery. Beginning of flow, B; point of maximum flow, dZ/dt_{max}, C; and end of flow, X; B→X is systolic flow time (T_{SF})

The left brachial artery imaging study employed a SONOs 5500, D.0 software, using an 11-3 MHz ultraband vascular linear array transducer. In the supine position satisfactory interrogation of the brachial artery was achieved in all 29 subjects. The vascular transducer was placed, such that the brachial artery was visualized in the transverse plane. The transducer was adjusted in order to image the brachial artery segment showing clear anterior and posterior intimal lumen interfaces. Images of the arterial segments were stored for post-processing in order to enhance artery boundaries. A Hough transform was used to identify the inner edge of the arteries. Mean diameters, mean diameter changes, and percent mean diameter changes were determined. In the same vessel segment real-time color-flow Doppler images were recorded and stored.

C. TBEV-derived SV

TBEV data were obtained from a custom analog circuit, generating a voltage output that was digitized in real-time using a BIOPAC UIM100C system (BIOPAC, Goleta, CA USA) and then stored on a personal computer (Sotera

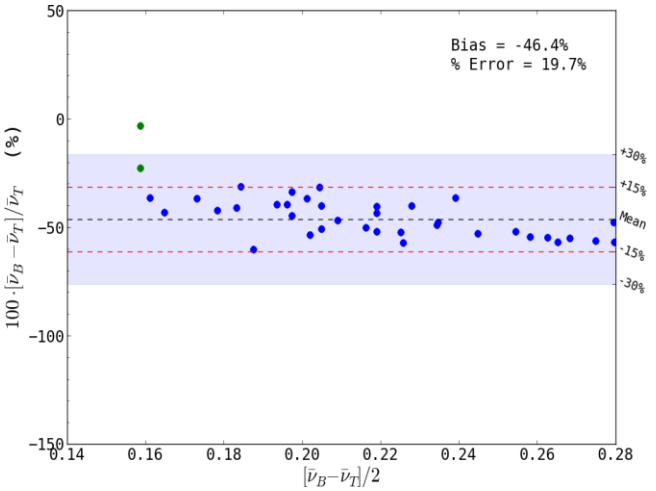

Figure 2. Bland and Altman plot of the brachial ohmic mean velocity versus thoracic ohmic mean velocity.

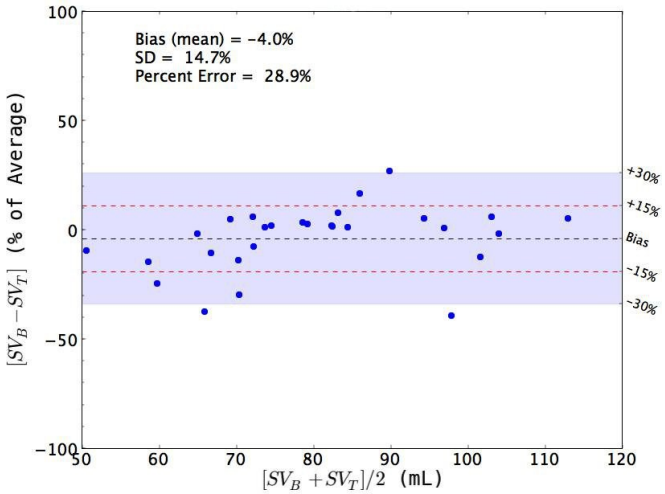

Figure 3. Bland and Altman plot of stroke volume obtained from the brachial artery versus stroke volume obtain by cardiac Doppler echo.

Wireless, San Diego, CA USA). All waveforms were sampled at 16-bit resolution and 500 Hz. The system was powered similarly to that used in the III A. Satisfactory TBEV signals were obtained in 28 of 29 subjects.

Operationally, the technique used in the calibration pilot study was virtually identical to that used in the SV comparison study of TBEV vs. Doppler/echo. In both studies, the four lead electrode montage featured a constant magnitude driving current of 4 mA rms at 70 kHz, producing a current field. The voltage measured between the voltage sensing electrodes was amplitude-demodulated by rectification and filtering. The raw voltage was split and processed, with a high pass filter with cutoff frequency of 0.1 Hz to obtain the change in voltage ΔU(t) (volt) and a low pass filter with cutoff frequency of 10 Hz to obtain a quasi-static voltage U_0. The raw ac and dc voltages were converted to the corresponding impedances Z by Ohm's Law. Digital signal processing was applied to the ac impedance ΔZ(t), to remove low and high frequency noise from dZ/dt. For each patient, TBEV dZ/dt signals were averaged over the measurement cycle to obtain dZ/dt_{max}. Fiducial landmarks on TBEV dZ/dt were identified to compute mean T_{SF} (Fig. 1). With the exception of the magnitude of the volume conductor, TBEV-derived SV is computed precisely as in (7).

D. Statistical Methods

Bland-Altman analysis [8] was used to show the mean difference and percent error between ohmic mean velocities obtained from both the ICG and TBEV methods. Bland Altman analysis was also used to determine the mean bias, standard deviation and percent error between the TBEV and Doppler/echo-derived SV.

IV. Results

The results of the TBEV volume conductor calibration exercise (III A) is shown in Fig. 2. It shows a bias of ~50% (-46.4%) and percent error = 19.7%. It is to be noted that, 36

Table 1. Mean brachial artery diameter and change over the cardiac cycle

	Males		Females	
	Mean ± SD	*[range]*	*Mean ± SD*	*[range]*
Mean (mm)	4.03 ± 0.48	[3.20 – 4.80]	3.10 ± 0.50	[2.40 – 3.90]
Change (mm)	0.10 ± 0.06	[0.04 – 0.23]	0.08 ± 0.05	[0.03 – 0.19]
% Change	2.36 ± 1.42	[0.91 – 5.59]	2.42 ± 1.18	[0.86 – 4.80]

of 38 data points are equal to or less than 15% from the mean. These data confirm a 2:1 ratio of the magnitude of the V_{TBEV} volume conductor to the V_{ICG} volume conductor. The transbrachial volume conductor is thus, $32W^{1.02}$

Bland-Altman results from the TBEV vs. Doppler/echo SV comparison study (Fig. 3) show a mean bias of 0.4% (-2.5 mL), standard deviation of 14.7% (precision 11.9 mL), and percent error 28.9% (limits of agreement 23.8 mL).

Results from the brachial artery study show that the percent change in mean diameter from end-diastolic dimension to peak systolic expansion is between 2 to 3 % [Table 1].

V. Discussion and Conclusions

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, as is true using the transthoracic method [9]. Secondly, it may be less affected by extravascular lung water, such as pulmonary edema and pleural effusions [10,11], which tends to confound thoracic ICG measurements in impaired patients. Thirdly, since the brachial artery diameter changes little (2-3%) throughout the flow interval, the impedance change, its rate of change and its peak value, dZ/dt_{max}, are virtually pure examples of the blood velocity-induced blood resistivity changes demonstrated in the rigid tube experiments provided by Gaw et al. [12]. As is true for transthoracic-derived dZ/dt_{max}, TBEV-derived dZ/dt_{max} is an ohmic acceleration analog. Results from these experiments show that transthoracic ohmic mean velocity is, on the mean, and with small variance, twice that of the transbrachial method. Therefore, by the equation of continuity for mass flow, the TBEV volume conductor is twice that of the transthoracic V_C. Results from the Doppler/echo-derived SV compared to TBEV-derived SV show that, the equation of continuity for mass flow correctly predicted the TBEV V_C. Results from the Doppler/echo vs. TBEV SV comparison also confirm interchangeability of the two methods according to accepted criteria [13]. Additional studies are needed to test the reliability of the measurement in a wider range of subjects, who may present conditions that typically interfere with ICG techniques.

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