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AORTIC BLOOD FLOW FROM INTRA-THORACIC IMPEDANCE

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The intra-thoracic electrical impedence change may be caused as much by aortic blood flow as by aortic movement,[1] contradicting the statement by Mitchell and Newbower[2] that all of the impedance change is due to movement alone. With two impedance analyzers, however, the aortic movement component of the intra-thoracic impedance may be cancelled out enabling a more accurate measurement of the aortic blood flow.

Methods. In the simplified anatomical model (Fig. 1), the aorta is represented as an infinite cylinder of conductivity $\sigma_2(v)$ and radius R passing through the thorax represented as an infinite homogeneous medium with electrical conductivity σ_1. A six-electrode oesophageal probe is aligned parallel to the lumen of the aorta at a distance A from it. B is the half-spacing of the 100 kHz current source electrode pair. C_1 and C_2 are the half-spacings of the two voltage sensing electrode pairs, connected respectively to the Z_1 and Z_2 impedance analyzers. For the patient, these impedance outputs are connected via an impedance calibrator switch box and two adjustable-offset amplifiers to a multi-channel chart recorder. The offset adjustment is used to expand the impedance scale, enabling precise measurement of Z_1 and Z_2. An equation linking the oesophageal impedance (Z) with blood velocity (v) may be derived from the model by the application of Maxwell's equations,[3, 2] giving

$$Z_i(v) = \frac{1}{2\pi\sigma_1}\left[\frac{2C_i}{B^2-C_i^2} + \frac{\sigma_2(v)-\sigma_1}{\sigma_2(v)+\sigma_1}\cdot I_i\right] \quad (1),$$

where $I_i = \frac{1}{\sqrt{X^2+(B+C_i)^2}} - \frac{1}{\sqrt{X^2+(B-C_i)^2}}$ and $X = 2A + A^2/R$.

Because A and R are identical for both electrode pairs, the equations specified by (1) for the two impedance signals Z_1 and Z_2 can be solved simultaneously to eliminate A and R. The notation can be simplified by rewriting equation (1) as

$$Z_i(v) = k_i + H(v)I_i \quad (2),$$

where $k_i = \frac{1}{2\pi\sigma_1}\cdot\frac{2C_i}{B^2-C_i^2}$ (3), and $H(v) = \frac{1}{2\pi\sigma_1}\cdot\frac{\sigma_2(v)-\sigma_1}{\sigma_2(v)+\sigma_1}$ (4).

For the two probe signals this gives,

$$Z_1(v) = k_1 + H(v)I_1 \text{ and } Z_2(v) = k_2 + H(v)I_2 \quad (5).$$

Thus $\frac{Z_1-k_1}{I_1} = H(v) = \frac{Z_2-k_2}{I_2}$ (6), and $\frac{I_1}{I_2} = \frac{Z_1-k_1}{Z_2-k_2}$ (7).

Given Z_1, Z_2, B, C_1 and C_2, (7) can be solved numerically for X, and then (6) for H(v). Defining $H^*(v) = 2\pi\sigma_1 H(v)$, equation (4) becomes

$$\sigma_2(v) = \frac{\sigma_1[1+H^*(v)]}{1-H^*(v)} \quad (8).$$

Electrical conductivity of blood increases by at least 10% for peak flows,[3, 4] and the approximate relationship (adapted from Visser *et al.*[3]) is

$$\sigma_2(v) = \sigma_2(v=0)\,[1-a[1-e^{-b\left(\frac{dv}{dt}\right)}]] \quad (9),$$

where $\sigma_2(v=0)$ refers to the electrical conductivity of blood during the diastolic (no flow) phase. For a haematocrit of 45%, and a peak conductivity change of 12.5%, $a \cong -0.17$ and $b \cong 0.15$ $(cm/s)^{-1/2}$. By rearranging equation (8), changing from v to t as the independent variable, and substituting for $\sigma_2(v)$ from (9), we get the instantaneous velocity of blood flowing in the aorta,

$$v(t) = \left[\frac{1}{-b}\ln\left(1+\frac{1}{a}\left(\frac{\sigma_1}{\sigma_2(t=0)}\left(\frac{1+H^*(t)}{1-H^*(t)}\right)-1\right)\right)\right]^2 \quad (10).$$

It is assumed that v=0 when t=0 at the start of each cardiac cycle.

Results and Discussion. Equation (10) does not depend on A or R. This is in sharp contrast to the single impedance analyzer method[1] where it is not possible to distinguish unambiguously the contributions to the measured intrathoracic impedance made by aortic movement and aortic blood flow. It must be remembered that the model used here is only an approximation to the real measurement. At the very least the contribution of noise from changes in A will be reduced by using two impedance analyzers instead of one. Fig. 2 shows dual impedance data for one cardiac cycle from a patient following cardiac surgery, together with the resulting velocity profile calculated using the given method. Assuming R = 1.25 cm, numerical integration gives a stroke volume of 57 ml. Cardiac output which was measured simultaneously by thermal dilution gave a stroke volume of 59 ml.

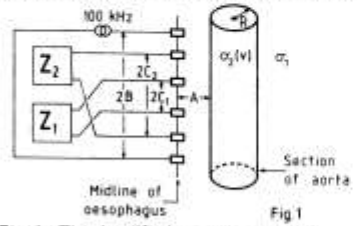

Fig.1

Fig. 1. The simplified anatomical model.

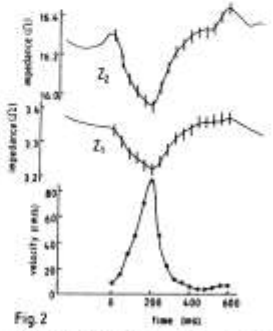

Fig.2

Fig. 2. An aortic blood velocity profile calculated from impedances Z_1 and Z_2 sampled at 40 ms intervals (bars). For this data, B = 5 cm, C_1 = 1 cm, C_2 = 3 cm, σ_1 = 1,600 Ω^{-1} cm^{-1}, σ_2 (v=0) = 1/155 Ω^{-1} cm^{-1}.

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