ELECTROMETRIC METHOD OF CONTINUOUS RECORDING OF THE VOLUME VELOCITY OF THE BLOOD FLOW AND CHANGES IN THE TOTAL VOLUME OF BLOOD IN THE HUMAN LUNGS

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In this study an attempt was made, from the theoretical standpoint to examine the possibility of using measurements of the electrical conductivity of the lungs for the quantitative determination of the regional blood flow, to find the simplest possible formulas for the calculations, and to compare the results obtained with the values determined by direct measurment of the blood flow during perfusion of a lobe of the lung at a known rate.

The following assumptions were made in order to solve the problem.

1. The lung was regarded as a three-component structure (air, tissue, and blood) with known or measurable specific resistances of its components and with volume ratios which varied from cases to case.

2. It was assumed that the air in the alveoli consists of spheres of uniform size, not conducting electricity, and surrounded by the tissue and blood of the organ.

3. On the basis of ideas concerning the anatomical structue of the network of blood vessels in the lungs, and of details of the electrical resistance of two-component isotropic and anisotropic media given in the literature [1, 4], it was assumed that the coefficient dependent on the orientation of the vessels relative to the electric field arising at the moment of measuring will not be significantly greater than 1.

4. By analogy with the technique of measuring the electrical conductivity of the lungs [2], it was assumed that the large vessels and bronchi of the lobe of the lung do not introduce any significant distortion into the accepted structural picture.

By the use of these assumptions an equation can be deduced for the relationship between the electrical and volume values in a chosen phase of cardiac activity.

For diastole:

$$\gamma_1 = [\gamma_B b_1 + \gamma_T (1 - b_1)] \frac{2a_1}{3 - a_1}, \qquad (1)$$

for systole:

$$\gamma_2 = [\gamma_B b_2 + \gamma_T (1 - b_2)] \frac{2a_2}{3 - a_2}, \qquad (2)$$

where γ_1 and γ_2 are the specific resistance of the organ (in $\Omega \cdot$ cm) in the period of diastole and systole; γ_B and γ_T the specific resistance of the blood and the tissue of the organ (in $\Omega \cdot$ cm); b_1 and b_2 the relative volume of blood in the period of diastole and systole; a_1 and a_2 the relative volume of blood and tissue in the period of diastole and systole.

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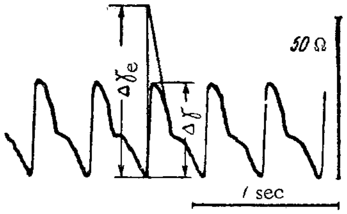


Fig. 1. Record of the pulse variations in electrical conductivity of the lungs (by the method of graphic extrapolation). γ – electrical equivalent of the pulse increase in blood volume per unit volume of organ. γ_e – electrical equivalent of the systolic injection of blood per unit volume of organ.

The values of the relative volumes are given by:

$$a = \frac{V_B + V_T}{V}, \quad b = \frac{V_B}{V_B + V_T},$$

where V_B, V_T, and V are the volumes of blood and tissue and the total volume of the organ, including the air, respectively.

It may easily be seen that

$$a_2 = a_1 + \Delta a, \quad b_2 = b_1 + \Delta b.$$

Subtracting equation (1) from equation (2) and making the necessary substitutions and transformations, we obtain:

$$\Delta\gamma = \gamma_2 - \gamma_1 = \frac{2a_1}{3 - a_1}\cdot(\gamma_B - \gamma_T)\cdot\left(\Delta b + \frac{3\cdot\Delta a}{(3 - a_1)\cdot a_1}\cdot\Delta b\right) + \frac{3\cdot\Delta a}{(3 - a_1)\cdot a_1}\cdot\gamma_1.$$

Expressing Δa, Δb through ΔV (the increase in blood volume with the pulse per unit volume of the organ)

$$\Delta a + \Delta V(1 - a_1), \quad \Delta b = \Delta V\frac{1 - b_1}{a_1}.$$

After transformations we obtain:

$$\Delta V = \frac{(3 - a_1)\cdot\Delta\gamma}{2(\gamma_B - \gamma_T)(1 - b) + 3\gamma_1\cdot\frac{1 - a_1}{a_1}}. \tag{3}$$

In practice it is difficult to use equation (3) because of the necessity of determining several of the values included in it. However, by taking into account the real values of these terms which are difficult to determine ($a_1 \approx 0.3$, $b_1 \approx 0.15$, $\gamma_T \approx 0.00074\ \Omega^{-1}\cdot cm^{-1}$at body temperature) and the way in which they are linked together in equation (3), it can be shown that as a first approximation this equation reduces to:

$$\Delta V = \frac{\Delta\gamma}{\gamma_B}. \tag{4}$$

Since $3 \gg a$, γ_B, γ_T, $1 \gg b$ and γ_1 is directly proportional to a, small deviations of a, b, and γ_T from the mean values have no significant effect on the end result.

Equation (4) can be deduced in another way [3], although its approximate character is less clearly apparent:

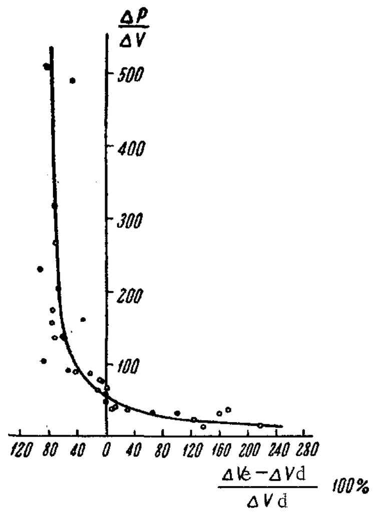
$$\Delta R = \frac{R_1\cdot R_2}{R_1 - R_2},$$

where ΔR is the resistance of the pulse increase in blood volume; R_1 and R_2 are the resistance of the organ in the periods of diastole and systole respectively, but

$$\Delta R = \rho_B\cdot\frac{l}{S} = \frac{\rho_B\cdot l^2}{\Delta V},$$

where ρ_B, l, S, and ΔV are the specific resistance, the "length," the "area of cross section," and the pulse increase in blood volume respectively

$$\left(\Delta V = \frac{\rho_B\cdot l^2}{\Delta R}.\right)$$



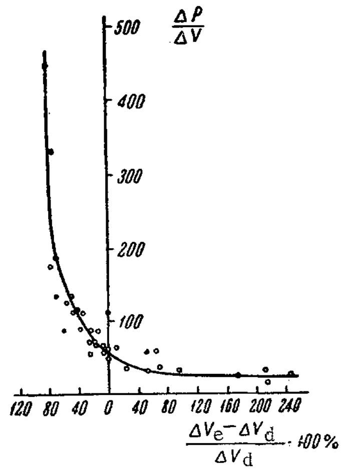


Fig. 2. Effect of the elastic properties of the lungs on the magnitude of the blood flow determined by the electrometric method. Along the axis of abscissas—discrepancy between the results of determination of the blood flow by the direct (ΔV_d) and the electrometric (ΔV_e) methods (in %). On the left, ΔV_e was determined by equation (6), on the right by equation (5); along the axis of ordinates—a value proportional to the modulus of volume elasticity (in mm Hg/ml).

Mean Values of Minute Volume of the Circulation (MVC) in the Human Lungs (in ml/100 cm^3 of Organ)

Number of observations	Normal conditions			Pathological conditions	
	lower lobes	middle lobes	after pneumonectomy	partial collapse of the lung	mitral stenosis
Number of observations	15	6	3	4	9
MVC	122.2	85.8	214.1	91.0	86.6

Since in the present case the pulse increase in blood volume is calcultaed per unit volume of organ, then $l = 1$, but

$$\frac{\Delta R}{K} = \frac{1}{\Delta\gamma} \quad \rho_B = \frac{1}{\gamma_B}, \quad \text{то} \quad \Delta V = \frac{\Delta\gamma}{\gamma_B}, \tag{4}$$

where K is the electrode coefficient used for converting the measured values into specific values.

Since the pulse amplitude of electrical conductivity is the electrical equivalent of the pulse increase in the volume of blood in the lungs during systole, the value of ΔV must be less than the value of the stroke volume, because a small volume of blood leaves the lung actually during systole. Like Nyboer [3], who measured the blood flow in the human limbs, we used the method of graphic extrapolation to determine the electrical equivalent of the stroke volume (Fig. 1). Substituting in equations (3) and (4) the value of the electrical equivalent of the volume—$\Delta\gamma_e$—for the value of $\Delta\gamma$, we obtain the stroke volume:

$$\Delta V_e = \frac{(3 - a_1)\cdot\Delta\gamma_e}{2(\gamma_B - \gamma_T)(1 - b) + 3\gamma_1 \frac{1 - a_1}{a_1}}, \tag{5}$$

$$\Delta V_e = \frac{\Delta\gamma_e}{\gamma_B}, \tag{6}$$

where $\Delta\gamma_e$ is the volume of blood (in ml) flowing through 1 cm^3 of the organ during a cardiac contraction. In practice it is more conveient to use a value 100 times greater (per 100 cm^3 of the organ).

The value of the stroke volume thus determined must depend on the elastic properties of the organ. This is evidently allowed for only partly in eqations (5) and (6). It was therefore considered important to try to find the magnitude of the correction for elasticity for the general case, when the elastic properties of the organ may be significantly modified.

In connection with the need for checking the assumptions made and for introducing a correction for the elastic properties of the organ, experiments were carried out on isolated lobes of the lung of dogs and cats in which the stroke volume was determined simultaneously by the direct and the electrometric method. The results of these measurements are given in Fig. 2.

It can be concluded from these graphs that both equations, the approximate (6) and the more exact (5), yield roughly identical results. In the general case, when the elastic properties of the organ may vary within wide limits, a correction for elasticity is essential when calculating the volume blood flow. By introducing an analytic expression into the graph, it can be shown that the approximate equation for determination of the blood flow, taking account of elasticity, may be written in the simple form:

$$\Delta V_B = Q \cdot \frac{\Delta P}{\Delta V} \cdot \Delta V_e, \tag{7}$$

where ΔV_B is the volume of blood (in ml) ejected by the right ventricle during one systole per unit volume of the lungs; Q is a constant coefficient equal to 1.79 ml/mm Hg; ΔP is the systolic increase in pressure in the pulmonary artery (in mm Hg); ΔV is the pulse increase in the blood volume in the lungs (in ml) per unit volume, determined by equation (4), and ΔV_e is the stroke volume (in ml) per unit volume of the elastic properties.

The use of equation (6) for calculating the blood flow thus assumes that the ratio between the pulse increase in pressure and the pulse increase in volume is constant.

The results of the direct measurement of the minute volume thus agreed with those claculated by the equations to within 20% in the case of most of the measurements. However, we can assume that in fact the degree of accuracy will be higher, because the comparison with the direct method of determination of the blood flow is based on the hypothesis that the blood flow is uniform throughout the volume of the lobe; as a first approximation this may be so in the conditions of the natural circulation but it is very difficult to ensure during artificial perfusion of the lungs, used as a method of control. The calculated values of the pulse increase in blood volume and of the stroke volume are probably a little on the low side, because the coefficient dependent on the orientation of the vessels, mentioned above, must in fact be greater than 1. However, this also could not be demonstrated reliably in the experiments with perfusion of the lungs.

The measurement of the blood flow by the electrometric method was carried out in clinical conditions. The results of some observations are given in the table.

The purpose of compiling this table was merely to illustrate the possibility of using the proposed method in clinical conditions.

So far as the assessment of the dynamics of the blood filling of the lungs—determination of the amount of blood injected or ejected (in ml/100 cm^3 of the organ)—is concerned, this is based on the use of equation (4). Since it was assumed, when equation (3) was deduced, that the volume of blood expelled during one systole is small by comparison with the volume of the lungs, the suitability of equation (4) for assessment of the dynamics of the blood filling is limited to cases when the volume of blood accumulated is small by comparison with the volume of the lungs.

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