ELECTROPLETHYSMOGRAPHIC DETERMINATION OF THE REGIONAL RELATIVE BLOOD VOLUME OF THE LUNGS

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On the basis of a theoretical investigation of a biophysical model of the lung structure and subsequent experimental verification of the results, a method of calculating the regional relative blood volume of the lungs from the data of electroplethysmographic investigation of the lungs and the results of measurement of the specific electrical conductivity of the blood is suggested. Equations for calculating the regional relative blood volume of the lungs are derived and their working range is established so far as the use of the proposed method of calculation in clinical and physiological investigations of the lungs is concerned.

KEY WORDS: lungs; regional blood volume of the lungs; electroplethysmography.

Electroplethysmography (rheography, impedance plethysmography) is nowadays a method widely used in clinical and physiological investigations of the peripheral circulation.

One of the modifications of this method, of importance on its own account, is regional electroplethysmography of the lungs [3], by means of which certain indices of the regional pulmonary hemodynamics can be estimated quantitatively [4]. Quantitative assessment of the results of electroplethysmographic investigation in general and of the lungs in particular is a most important but, at the same time, a most difficult task.

This paper describes a method of quantitative evaluation of the regional relative blood volume of the lungs (K_V), which is taken to mean the volume of blood (V_b) present per unit volume of lung (V), including blood, lung tissue, and air, i.e.:

$$K_v = \frac{V_b}{V}. \tag{1}$$

Irrespective of the relative volumes of the components, the type of relationship between the specific electrical conductivity of the lung structure (γ) and the specific electrical conductivities of the blood (γ_b), the lung tissue (γ_t), and air (γ_a) and the relative volumes (B_v, T_v, and A) of the respective components (blood, tissue, and air), can be represented by the known law of connection in parallel, provided that a correcting factor (F) is introduced:

$$\gamma = (B_v \cdot \gamma_b + T_v \cdot \gamma_t + A \cdot \gamma_a) \cdot F. \tag{2}$$

If the frequency of the current is 5 kHz, it can be taken that $\gamma_a = 0$. This not only shortens equation (2), but also enables the factor F to be represented as the product of two coefficients allowing for deviation from the parallel state for components of the "blood–tissue" structure (F_b) and for components of the "blood + tissue–air" structure (F_a):

$$\gamma = (B_v \cdot \gamma_b + T_v \cdot \gamma_t) F_b \cdot F_a. \tag{3}$$

The values of B_v and T_v are dependent: $B_v + T = S$, where S is the relative volume of the "blood–tissue" structure in the lungs (S + A = 1). Furthermore, $B_v = SB$ and $T_v = ST$, where B and T are the relative volumes of blood and tissue in the "blood–tissue" structure. Hence: $T_v = T \cdot B^{-1} \cdot B_v$. Solving equation (3) for B_v and making this substitution, we obtain:

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TABLE 1. Value of Ratio $E = \gamma b/\gamma \cdot B \cdot (1 - A)$ Depending on Relative Content of Blood and Air in the Lungs

B	A					
	80	75	70	60	50	40
20	1,00	1,02	1,07	1,13	1,4	1,21
25*	1,08	1,14	1,21	1,27	1,42	1,33
31	1,17	1,25	1,35	1,42	1,44	1,45
42	1,19	1,40	1,42	1,38	1,42	1,48
45	1,27	1,40	1,47	1,45	1,36	1,44
47	1,29	1,39	1,48	1,48	1,48	1,51
48	1,30	1,43	1,37	1,48	1,53	1,63
63	1,32	1,33	1,46	1,52	1,65	1,44

$M \pm m = 1{,}4 \pm 0{,}02$; $S = \pm 0{,}123$

*Value of B given for which the value of E was obtained by interpolation between values B = 0.20 and B = 0.31.

Legend: B) relative blood volume (in % by weight), A) relative volume of air (%).

$$B_v = \frac{\gamma}{(\gamma_b + T \cdot B^{-1} \cdot \gamma_t)F_b F_a}. \tag{4}$$

If this equation is used for practical calculations, the values of F_a, F_b, and γ_t must be estimated. To estimate the coefficient F_a it is permissible to regard the lungs (within the limits of the region of investigation) as a model structure consisting of an electrically conducting medium (blood + tissue; specific conductivity γ_{bt}) and, scattered in it, nonconducting inclusions which are approximately spherical in shape and of equal size (alveolar air), so that it can be described by the Velick–Gorin equation [9], which, allowing for the hypotheses and assumptions adopted, will have the form:

$$\gamma = 2A \cdot (3 - S)^{-1} \cdot \gamma_{bt}.$$

However, since the value of γ_{bt} can be expressed as: $S \cdot \gamma_{bt} = (B_v \cdot \gamma_b + T_v \cdot \gamma_t)F_b$, by considering equation (3) we obtain: $F_a = 2 \cdot (3 - S)^{-1}$. Clearly the value of F_a is almost independent of the value of S, so that according to physiological [7] and morphological [2] data $S \approx 0.3$; consequently, a range of values of S(A) can be distinguished for which $F_a \approx$ const. The complete estimation of the coefficient F_b is more difficult; however, as follows from data in the literature [1], the relationship between F_b and B is similar to that of F_a and S, so that we can evidently pick out a range of values of B for which $F_b \approx$ const. The mean value of the blood volume of the lungs (B) must be taken to be 40% by weight (for the human and the cat's lungs) [8].

The value of γ_t is not measured directly, but it is known that γ_t is about an order of magnitude or less below the value of γ_b [4].

The results of estimation of the values of B and γ_t enable the required equation to be simplified still more, for after transfer of γ_b to the denominator of the right hand side of equation (4) we are left in parentheses with the expression $(1 + T\gamma_t/B\gamma_b) = N$, the mean value of which $N = 1 + 0.15$ is almost independent of variations in the values forming the second term, and so $N \approx$ const.

Representing $(FN)^{-1} = E$, we can rewrite equation (4) in the form:

$$B_v = E\frac{\gamma}{\gamma_b}. \tag{5}$$

To obtain the final equation from equation (5) it is necessary to determine experimentally the value $E = E_0$ and the range of values of B and A for which it can be taken with a satisfactory degree of accuracy that E_0 = const. Experimental estimation of the value of E_0 is based on the use of equation (5) in the form: $E = \gamma_b \cdot \gamma^{-1} \cdot B(1 - A)$, the parameters of which were determined electroplethysmographically (γ and γ_b) and by direct (colorimetric, volumetric, and gravimetric) measurements of the relative volumes of blood, tissue, and air in experiments on the isolated posterior lobes of the lungs of seven anesthetized cats. The results of these experiments are summarized in Table 1, which shows how the coefficient E depends on the blood volume of the lungs (B) and

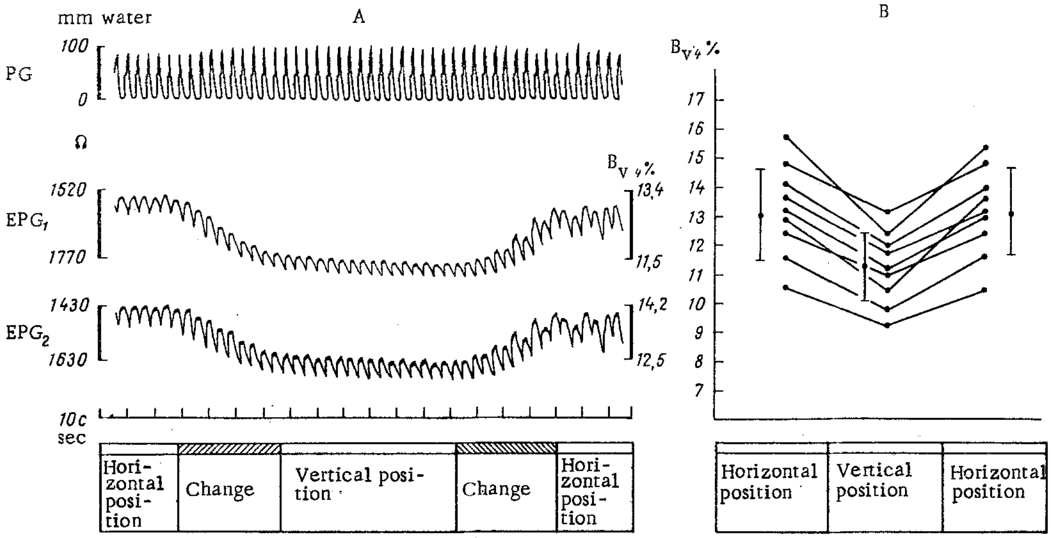

Fig. 1. Dynamics of regional relative blood volume of lungs during change in position of body in space. A) Change in relative blood volume of symmetrical dorso-basal regions of right and left lungs with a change in position of body from horizontal (lying supine) to vertical (head uppermost) and vice versa; B) relative blood volume of dorso-basal regions of lungs in horizontal and vertical positions recorded in nine animals during change from horizontal to vertical position and vice versa.

the relative volume of air (A). The standard statistical analysis of these data shows that within the working range of (B) from 0.25 to 0.63 (by weight) and of (A) from 0.4 to 0.8, the value of $E_0 = 1.4 \pm 0.02$; the mean error of measurement is ±12%, and the maximal error is ±23%.

The regional relative blood volume of the lungs can thus be calculated by the equation:

$$B_V = 1.4 \frac{\gamma}{\gamma_b}, \tag{6}$$

if the relative blood volume of the lungs (B; by weight) is not below 25% and if the relative volume of air is not more than 80% and not less than 40%. In units of electrical resistance (ohms) equation (6) can be written:

$$B = 1.4 \frac{\rho_b}{\rho}, \tag{7}$$

where ρ_b and ρ are the specific resistance of blood and the lungs respectively. To express the value of B_V in milliliters of blood per 100 cm^3 volume of the lungs or, what amounts to the same thing, in volumes per cent, the right hand sides of equations (6) and (7) must be multiplied by 100%:

$$B_V = 140 \frac{\gamma}{\gamma_b} (\%) \tag{8}$$

and

$$B_V = 140 \frac{\rho_b}{\rho} (\%). \tag{9}$$

In conclusion, as an example of the use of the suggested method to calculate the regional relative blood volume of the lungs, Fig. 1 shows a section of a simultaneous recording of two electroplethysmograms from symmetrical regions of the lungs and the value of the regional relative blood volume of these regions of the lungs, recorded in experiments to study postural hemodynamic responses of the pulmonary circulation [5, 6].

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SIMULTANEOUS RECORDING OF THE TRANSPULMONARY PRESSURE AND ELECTROMYOGRAM OF THE DIAPHRAGM

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To record the hysteresis loop and electromyogram of the diaphragm simultaneously it is recommended that a standard probe of the sort used to record the intraesophageal pressure, on which silver electrodes are mounted, be used. This method provides fuller information on the work of the respiratory muscles.

KEY WORDS: transpulmonary pressure; electromyogram of the diaphragm; electrode-probe.

The methods nowadays most widely used to investigate the work of the respiratory muscles are calculation of the work done in overcoming the elastic and nonelastic resistance to respiration from the pressure–volume curve (hysteresis loop) and electromyography of the respiratory muscles. The first method yields data which are integral indices of all forces taking part in the mechanics of the respiratory act; the second method gives an idea of the relative role of particular respiratory muscles in the work of respiration in the different phases of the respiratory cycle.

Usually the two methods are used separately, but this makes comparison of their results difficult. It seemed worthwhile to combine these clinical-physiological methods of investigation in order to be able to obtain fuller information on the biomechanics of respiration.

For this purpose the writers have used an esophageal probe, forming part of the set of instruments used to study the elastic properties of the lungs and the work of respiration and, in particular, part of the "compliance test" apparatus manufactured by the firm of Godart. The pressure measured in the lower third of the esophagus corresponds most adequately to the true intrathoracic pressure [1-5]. With the pressure transducer located there the coefficient of correlation between the intraesophageal and intrapleural pressure is 0.99 [6]. Usually the intrathoracic pressure is measured by a differential manometer relative to the pressure in the mouth. In this case the resistance of the instrument is automatically subtracted and the manometer shows the true intrathoracic, or so-called transpulmonary, pressure, one of the most important parameters for the calculation of many indices of the biomechanics of respiration.

When the probe is located in the lower third of the esophagus, after slight modification it can also be used as a bipolar electrode for electromyography of the diaphragm. The modification is as follows. Two silver electrodes 5 mm in diameter and 0.3 mm thick are glued to the thin-walled rubber balloon of the standard probe for recording the intraesophageal pressure, at a distance of 10 and 30 mm from its "blind" end. A lead made from copper wire 0.06 mm in diameter, with Viniflex insulation, is soldered to each electrode. The wires run

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