The Model of Vascular Bed for Estimation of Human Systemic Hydrohemodynamics

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Abstract—**Existing methods of impedance cardiography for estimating blood flow parameters are based on measuring amplitude and temporal characteristics of pulse waves in specific points. Our aim was to improve the algorithm of blood flow analysis using continuous representation of cardiocycle. The suggested algorithm of the vascular blood flow estimation during systolic and diastolic phase of pulse wave is proposed which uses the continuous "integral amplitude" of rheogram. Also, we have developed the basic hemodynamics equation for our heterogeneous model taking into account anthropometric, morphologic and anisotropic properties of various body segments. The resulting formulas were adapted for the estimation of peripheral blood flow in major body segments (head, arms, legs and torso, including abdominal and thoracic regions). A method of controlled systemic hemodynamics estimation is represented which utilizes an automatic commutation scheme with application of 5 electrode pairs on head and extremities. We describe the results of registration of blood flow redistribution between body segments during active orthostatic test with simultaneous control of a checksum, i.e. a difference between the volumes of central and peripheral blood flows.**

Keywords—**impedance cardiography, model of vascular bed, systemic hemodynamics.**

I. Introduction

The calculation of peripheral (and central) volume of blood flow is based on Kedrov's hemodynamics equation for homogeneous objects:

$$V(t) = Z(t)(V/Z) \tag{1}$$

It contains static (V/Z) as well as dynamic $(Z(t))$ rheogram components. For example, estimation of the stroke volume SV, proposed by Kubicek in 1970, is performed by the equation $SV = \rho_K \cdot \left(L^2/Z^2\right) \cdot Ad \cdot Ti$, where $Ad \cdot Ti = \Delta Z_a$ is the measured rheographic amplitude $\Delta Z(t)$ approximation [1], [2].

The value ΔZ_a is based on a simple assumption that the flow velocity during cardiac output is constant. Graphically, the value $Ad \cdot Ti$ corresponds to the square of a rectangle with sides Ad and Ti, which is nearly twice as much as the square under the real differential rheogram.

Example 1: For differential rheogram with the amplitude $Ad = 2.5$ Ohm/s and evacuation time $Ts = 0.3$ s, the value ΔZ_a is $\Delta Z_a = 0.75$ Ohm.

The measured volume V in the static component of (1) is calculated using an expression proposed by J. Nyboer in 1970: $V = V_g = \rho_K \cdot \left(L^2/Z\right)$ [3]. It represents "electrometric" (homogeneous) thoracic volume, i.e. a region between the neck and the xiphoid process with the length $L = 22$ cm. This region is filled in with blood that has specific resistance $\rho(b) = 150$ Ohm $\cdot$ cm, and its interelectrode impedance is $Z = 25$ Ohm. Then, $V_g = 150 \cdot (22 \cdot 22/25) \cong 2900$ cm^3, which is far less than "geometric" volume V_s. The volume V_s, at thorax perimeter $Q_g = 95$ cm and neck perimeter $Q_s = 41$ cm (average perimeter $Q_m = 68$ cm), will be $V_s = \left(Q_m^2 \cdot L\right)/4\pi \cong 8100$ cm^3, which is 2.8 times greater than V_g value.

For this example, the stroke volume will be $SV = 150 \cdot (22 \cdot 22/25 \cdot 25) \cdot 0.75 \cong 87$ ml, and the cardiac output $CO = 5.6$ l/min at $HR = 65$ per minute, which corresponds to normal. Thus, the measurement error for static component arising due to measured thoracic volume V_g (which is approximately 2.8 times less than V_s) is compensated by an error of a dynamic component ΔZ_a (which is greater by nearly a factor of three).

The difference between V_s and V_g is explained by the fact that total specific resistance of thorax $\rho(s)$ is not equal to blood specific resistance $\rho(b)$ since the tissues with higher specific resistances are located in the thoracic region, too $[\rho(s) \gg \rho(b)]$. The value of total heterogeneous specific resistance is $\rho(s) = \left(Q^2 \cdot Z\right)/(4\pi \cdot L) = 418$ Ohm $\cdot$ cm (presumably, not more than 150 Ohm $\cdot$ cm). Therefore, the ratio $\rho(s)/\rho(b) \cong 2.8$ is similar to the ratio $V_s/V_g \cong 3$.

It should be mentioned that direct calculation of SV according to the Kedrov's expression (1) allows to arrive at the same result $SV = 8100 \cdot (0.2/25) \cong 64.8$ ml even if the amplitude $\Delta Z = 0.2$ Ohm is equal to the maximal rheogram value.

Thus, both components of the traditionally used equation for homogeneous model of hemodynamics do not directly reflect the measured regions of the body.

II. Methodology

For the analysis of human vascular hemodynamics, a compartmental electric model was developed for 7 regions:

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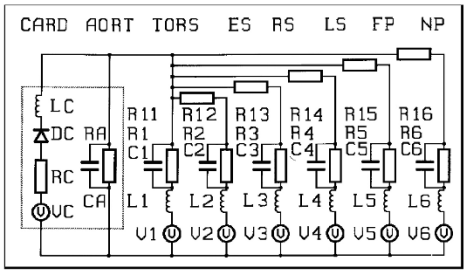

Fig. 1 The electric model of human regional hemodynamics

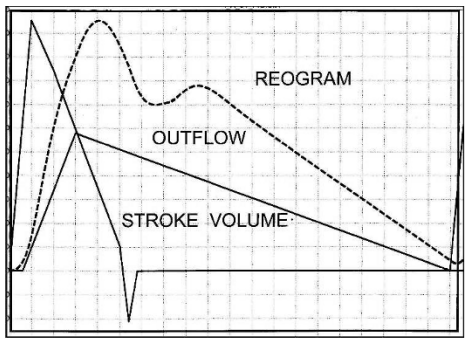

Fig. 2 The example of rheogram modelling of inflow and outflow phases

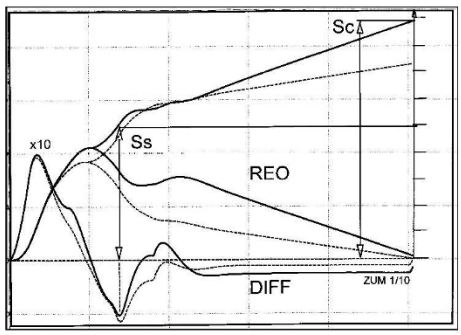

Fig. 3 The example of the rheogram "integral amplitude" detection

head (ES), right and left arms (RS, LS), right and left legs (FP, NP), trunk (TR), as well as heart and aorta regions. The further results of measurements are achieved according to the method of human systemic hydro- and hemodynamics evaluation developed in the Scientific Research Centre "Medass" [4], [5].

Figure 1 shows an electric scheme which models human regional hemodynamics. Each of 6 branches of this circuit which corresponds to peripheral regions of the body is connected to the parts modelling heart and aorta and contains elements V1–V6 which describe the volumes of regional blood vessels, R1–R6, C1–C6, L1–L6 modelling elastic properties, and R11–R16 which describe resistive vessels' characteristics.

Dynamic pulse-wave component – rheogram $\Delta Z(t)$ distinctly reflects two constituents. They are: systolic, produced by cardiac output, and diastolic, which reflects the outflow part of a vascular circulation pulse wave, which passes from arterial to venous bed (it is not seen at a rheogram).

The form of rheogram in the course of pulse wave reflects all the features of intravascular hemodynamics (Fig. 2). Maximal rheogram value just in the perfect single moment of a pulse wave might differ significantly in form and amplitude during systolic, as well as diastolic phase despite the same amplitude. So, it is necessary to process the whole pulse wave continuously integrally.

Figure 2 demonstrates the two-component model for rheogram forming, where an inflow phase reflects a cardiac output that happens during expulsion time, and outflow phase occupies the whole rheogram period. Blood is running out along vessels during systole, too. On the figure 2 the rheogram is smoothed by "elastic" inertial elements of the model.

The results of the peripheral and aortal (dotted line) rheograms modelling, and their "amplitude" integral values during systole S_s and in the course of the whole cycle S_c, are on figure 3.

Calculation of the circulation volume velocity might be proper when fits the following conditions.

- The whole rheogram reflects dynamic blood volume redistribution in a vascular bed during permanent measurement of its amplitude and time characteristics.
- Blood inflow is equal to the blood outflow during the whole period, when the transitory oedema is absent.

The volume of blood flow is calculated as an area under the rheogram curve.

$$S_c = \frac{1}{2}\int_0^{T_C} |\Delta Z| dt \qquad (2a)$$

$$S_S = \int_0^{T_S} |\Delta Z| dt \qquad (2b)$$

where S is the integral amplitude, T_C and T_S are the duration times of rheographic cycle and systolic phase, respectively. For the peripheral regions an integral amplitude S_C is used. The equation (2b) is used for cardiac output modelling.

III. Theoretical results

The calculation of the circulation volume velocity is based on the hemodynamic expression for heterogeneous model:

$$\Delta V_K = \left(V_K + \frac{\rho_K}{\rho_T} V_T \right) \frac{\Delta Z^{MF}}{Z^{LF}} HR, \qquad (3)$$

where ρ_T and V_T are the characteristics of electro conductive tissues.

Regional blood flow volume velocity RVB is calculated according to:

$$RVB = (K_A V_g + K_M V_s)(\Delta Z / Z) \qquad \text{or}$$

$$VB = \left(K_A \frac{\rho_K L^2}{Z^{LF}} + K_M \frac{Q^2 L}{4\pi} \right) \frac{\Delta Z^{MF}}{Z^{LF}} HR \qquad (4)$$

where the static component for the heterogeneous model contains the homogeneous volume V_g (the first part of expression), with respect of anisotropic region peculiarities – coefficient K_A. The second part of equation contains the corrective constituent considering morphological characteristics – coefficient K_M. For the calculation of the regional blood flow absolute values, measurement of a length and perimeter of all regions is needed.

The integral rheogram amplitude $\Delta Z^{MF} = S_S$ is registered at a traditional average frequency, for example 50 kHz. Interelectrode impedance is measured at low frequency 10 kHz, as soon as Z value at 50 kHz is partially shunted by the resistance of cellular liquid.

Specific peripheral blood flow SRB is detected according to the expression

$$SRB = K_A \cdot HR \cdot 60 \cdot 100 \cdot \left[S_C / Z^{LF} \right], \qquad (5)$$

where K_A – anisotropy coefficient, S_S – integral rheogram amplitude during single cardiac cycle – T_C, HR – heart rate per minute.

Detection of the cardiac output value and its derivatives: minute volume, total peripheral flow resistance would be detected according to the expression (3) under condition $\Delta Z = S_S$.

IV. An example of measurements and discussion

For the systemic evaluation of human body hemodynamics quite comfortable electrode system is used. It consists of 4 pairs of electrodes in the form of standard electrocardiographic clips. Current and potential electrodes are fastened in pairs, and binary head electrode is situated on the spring-loaded ear.

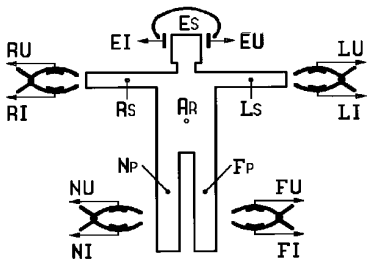

Fig. 4 Electrode placement for the evaluation of system hemodynamics

Table 1 The example of obtained data

Region	V_{S1}	V_{S2}	RVB	SVB	TW	STW
ES	7	4.9	0.98	20	3.76	65
RS	6	4.2	0.252	6	2.43	45
LS	6	4.2	0.252	6	2.43	45
NP	15	10.5	0.525	5	7.1	56
FP	15	10.5	0.525	5	7.31	56
Peripheral	49	34.3	2.534	7	23.2	33
Abdomin	38	26.6	2.926	11	14.4	66
Thoracic	13	9.1	1.729	19	37.7	77
Torso	51	35.7	3.927	11	22.2	66
Aortal	11.5	8.05	**5.46**	68	5.96	77
All body	100	70	5.46	100	45.5	65.0

Electrodes are placed on wrists, ankles and on frontal or temporal head regions (Fig. 4).

Electrode commutation insures the measurement, registration and calculation of blood flow in all mentioned above human body regions.

As an illustration, the example of healthy person examination is considered. This example corresponds to an "average man" weighing 70 kg and having body length 170 cm, arterial pressure 120/80, and a blood circulation minute volume 5.5 l/min in a supine position.

The results of examination are presented in Table 1. The relative (% of body mass) and absolute (kg) volumes of body regions are shown in columns V_{S1} and V_{S2}, respectively. Values of regional specific blood flow (ml/min per 100 g of tissue) are in column SVB. Absolute values of regional blood flow (l/min) are shown in column RVB.

Data in Table 1 show that the sum of blood flow volumes (RVB) in the whole body makes up 5.46 l/min (the last row). It can be seen that the blood circulation minute volume, which is RVB in the aortal region, makes up 5.46 l/min, too. Thus, the balance is maintained, the "control sum" is

achieved by measurement of the systemic and regional blood supply.

Verification of our suggested method on systemic hemodynamics evaluation using conventional methods is complicated or practically impossible. However, reliability of the obtained results can be tested using the "control sum" approach. Equality of the systemic and peripheral blood flow is proved, as soon as the minute volume of systemic circulation is equal to the sum of regional blood flows absolute values.

In Table 1 specific volumes are given. Specific indices have been always used as normalized values in invasive investigations of hemodynamics.

It is already accepted to express the result of cardiac index measurement as a cardiac output relative to body surface. So, these values in, e.g., adult athletes and children can be comparable.

V. Conclusion

Compartmental model for 7 body regions (head, extremities, abdominal and thoracic regions) is suggested which enables to study relations between the parameters of central and peripheral hemodynamics. The reliability of obtained estimates is illustrated by the close proximity of the blood minute volume to the sum of regional blood flows.

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