RESEARCH, DESIGN, AND TECHNOLOGY

A RHEOCARDIOGRAPHIC STUDY OF MODELED STROKE VOLUME ACCURACY

É. P. Baluev and V. B. Parashin

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In many areas of medicine, information on the heart's stroke and minute volume is required to evaluate the patient's condition. The basic requirements for methods of measurement are minimal trauma for the patient, the possibility of measuring stroke volume during each cardiac cycle, and a margin of error no greater than 10-15%.

The rheographic method [1, 2] meets the first two requirements to a very high degree; therefore, questions of evaluating the method's accuracy and ways to increase it [3] are of timely importance.

It is known [1, 2] that a rheocardiogram represents the relation of pulsation in chest cavity electrical impedance in a range of 0.05-1 Ω at frequencies of 0.2-0.3 Hz; measuring is done at 20-150 kHz with a four-electrode system. Methods of determining cardiac stroke volume (CSV) according to various rheocardiogram parameters are shown in [4-16].

A biophysical analysis of the connection between the heart's pulse volume and rheocardiogram parameters shows that they are influenced by a number of factors aside from pulse volume: torso shape and dimensions, heart shape, heart coordinates and orientation in the chest cavity, shifts in the heart's position during contractions, presence in the chest cavity of other organs and tissues with different electrical conductivity (large blood vessels, lungs), and changes in chest cavity volume during heart contractions and respiration.

From this it is obvious that the task of accurately measuring CSV by rheocardiogram is rather complex and can be approached in a number of ways.

A comparison of results of rheographic measurements of CSV in humans with measurements by other means is available in the literature (electromagnetic fluorometry or tracer distribution is usually used as a control).

The present approach has certain inherent limitations. Due to the random nature of CSV values measured and the great number of factors involved, a reliable comparison can be made only in a large group of subjects. The control methods used are themselves indirect and do not meet the above metrological requirements. The electromagnetic fluorometry method requires placing a sensor on the exposed aorta and is usable only in surgical interventions. The tracer method does not permit uninterrupted measurements, and is indirect and insufficiently accurate.

The discrepancy of results of measuring CSV by rheographic and other methods is 10-50%, according to various authors' data.

It has been noted [3] that the Kubicek formula [14] gives the most reproducible results among rheographic methods, and many types of measuring equipment have been developed and manufactured based on it.

Mathematical modeling of the process of determining CSV by rheocardiogram allows us to eliminate the influence of many incidental factors and thus is useful in studying methodical errors [17]. The basic limitation of this approach is that, due to computational difficulties, only simple models can be designed.

It is known [1, 2] that in general the relationship between stroke volume and electrical impedance pulsations in the chest cavity can be described by four parameters: \( R_0 \), the constant component of resistance between measuring electrodes; \( \Delta R \), the resistance pulsation; \( V_0 \), the volume of the chest cavity between electrodes; and \( \Delta V \), the stroke volume. Considering that \( \Delta R \ll R_0 \), \( \Delta V \ll V_0 \), any nonlinear relationship between \( \Delta R \) and \( \Delta V \) in the first approach can be made linear and represented in the form
Fig. 1. Schematic design of a device for checking accuracy of pulse volume measurement by rheoplethysmogram.
1) SAUS-6000 pneumatic drive; 2) reservoir; 3) pulsating volume; 4) RPC2-02 rheoplethysmograph; 5) "Argus-3" spirometer; 6) SI-18 oscillograph; 7) N338-4P recording system; a, b, c, d) point electrodes (a, d - current; b, c - potential); k, l, m, n) tape electrodes (k, n - current; l, m - potential).

Fig. 2. Basic dimensions of reservoir and pulsating volume.

Fig. 3. Experiment with a spherical volume.

Fig. 4. Experiment with cardiac massager.

\[
\frac{\Delta R}{R_0} = K \frac{\Delta V}{V_0} ,
\]

where \( K \) is the relative sensitivity of change of resistance to CSV, generally dependent on all the above factors.

In the simplest case, for a homogeneous cylindrical lead, we have

\[
R = \rho \frac{L}{S} ,
\]

where \( R \) is the resistance, \( \rho \) is the specific resistance, \( L \) is the length, and \( S \) is the cross-sectional area. Differentiating the formula and considering that \( V = S \cdot L \), where \( V \) is the volume of the lead, we arrive at (without considering the signs of \( \Delta R \) and \( \Delta V \))

\[
\frac{\Delta R}{R_0} = \frac{\Delta V}{V_0} (K = 1)
\]

or

\[
\Delta V = \Delta R \cdot \frac{L^2}{R_0^2} .
\]

Attempts to use this simplified model to determine CSV did not yield satisfactory results.

A model from two volumes \( V_1 \) and \( V_2 \) with specific resistances \( \rho_1 \) and \( \rho_2 \), where \( V_1 \gg V_2 \), \( \rho_1 \gg \rho_2 \), is more complex and more closely approaches natural measuring conditions. The