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Pulse Wave Contour Analysis in Automated Mode

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Based on the fundamental principles of contour analysis of central pulsograms and the method of detecting the characteristic points of a pulse wave, an improved approach to pulse wave contour analysis in 10 characteristic points is described – the method of automatic determination of amplitude-and-time informative parameters. This method makes it possible to detect the characteristic points of a pulse signal in its low amplitude segments. It can be used for detailed comparative analysis of the shape of the radial artery pulsograms detected at the conventional palpation points.

Introduction

Effective mathematical algorithms (e.g. spline approximation, regulating algorithms, etc.) in processing of biomedical experimental information provide enhanced accuracy of estimation of characteristics of the processes of interest. The measuring device–computer system according to the algorithms mentioned above provides effective analysis of biomedical information. The goal of this work was to consider the use of such algorithms for pulse wave contour analysis in automated mode, identification of characteristic points of pulsograms, and further analysis of pulsogram parameters, whose accuracy has a considerable effect on the accuracy of diagnosis.

Substantiation of Method for Contour Pulsogram Analysis

It was demonstrated in [1, 2] that pulsogram shape could be restored based on experimental information and methods of spline approximation and regularization. A method for contour analysis of a pulsogram (sphygmogram) based on such mathematical apparatus is suggested. The method should take into account all information parameters suggested by different authors. The method of Valtneris and Yauya [3] was selected as a prototype of the contour analysis method for a central pulsogram.

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The method is based on detection of amplitude–time parameters in 10 informative points of a pulsogram typical of not only phases (time intervals) of cardiac cycle, i.e. 9 temporal intervals of the cardiac cycle corresponding to projections on time axis described in [4], but also pulsogram shape shown in Fig. 1. It is widely believed that detection of the phase of the cardiac cycle is optimal when performed from a high-speed kinetocardiogram (HSKCG) containing reference synchronous curves (polycardiographic method, PCM), ECG, phonocardiogram (PCG), and sphygmogram of the carotid artery (SPGca) [4]. The HSKCG allows the following time intervals to be measured: cardiac cycle time (R-R), asynchronous contraction phase (AC), isometric contraction phase (IC), fast ejection phase (Em), slow ejection phase (Er), protodiastole (P), isometric relaxation phase (IR), fast filling phase (Fr), slow filling phase (Dy), and systole of atrium (Sa).

Absolute values of ordinates of informative points of the pulsogram and 9 angles characterizing amplitude– time relations of paired points ((1) and (2), (2) and (3), (3) and (4), ..., (9) and (10)) are suggested in this work to be discussed in addition to time intervals. A single pulsation wave period or cardiac cycle period T is equal to the R-R interval time and is determined by points 1 and 10. Point 10 (pulsation wave end) corresponds to the start of the next pulsation wave, point 1. The physiological sense of the selected parameters of the pulsation wave is described in [5].

The shape of the pulsation wave in a healthy patient is shown in Fig. 1. The typical coordinates of characteristic points are easy to determine. The amplitude–time parameters of the points allow hemodynamic parameters

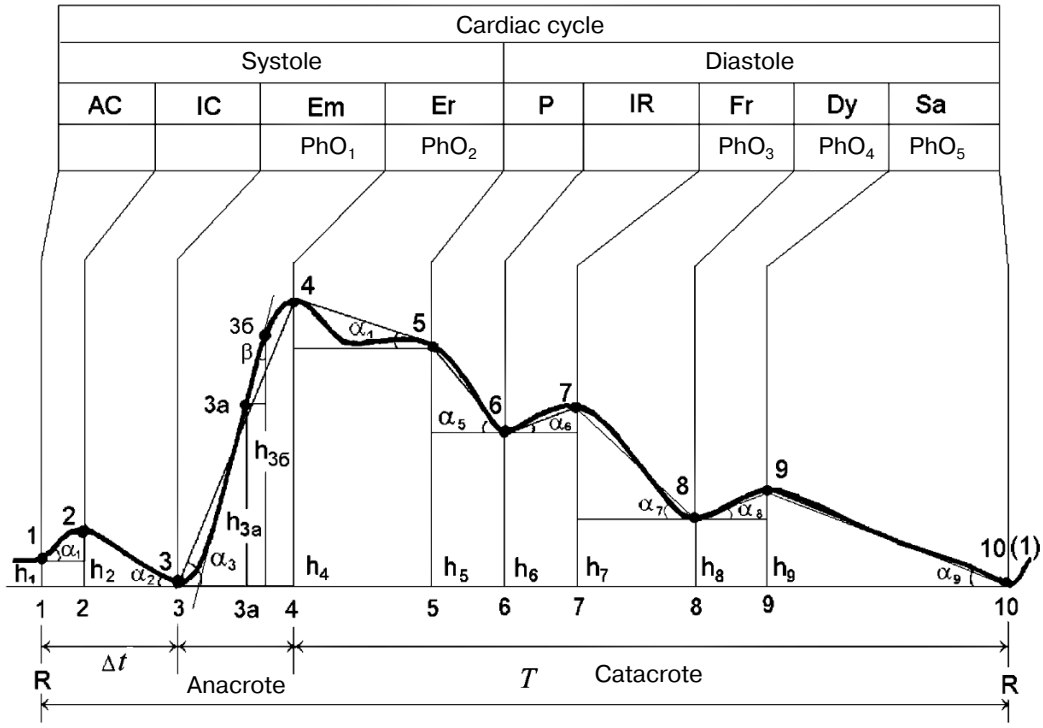

Fig. 1. Advanced variant of contour analysis of central pulsogram.

of the human cardiovascular system to be estimated using the mathematical model of circulation in the mode of high fluidity (Poedintsev and Voronova model), as well as the physiological status of the human body to be evaluated.

On the other hand, some segments of the pulsogram wave are poorly expressed in some pathologies (e.g., apex of the main pulsogram systolic wave in point 4). In this case, according to the method of Valtneris and Yauya [3], it is determined at the distance of 0.08 sec from the beginning of fast ejection phase (point 3, Fig. 1), which is not always correct. Other parameters of the pulsation wave (e.g., incisure point 6 in Fig. 1) are estimated similarly. The method suggested by J. Friart [6] provides an estimate of coordinates of this point at 0.3 sec from the systolic wave apex. It should be noted that the accuracy of determination of typical points specifies the accuracy of evaluation of hemodynamic parameters and ratio of heights $h_{3a} : h_4 : h_5$ and $h_6 : h_7 : h_9$ (Fig. 1) (characterizing the shape of the systolic and diastolic waves), as well as ratio of heights $h_4 : h_6$ (representing peripheral vascular resistance [5]), absolute value of h_6 (dependence of dicrotic wave on the arterial wall tonus) [7], etc.

Thus, the amplitude–time analysis of actual pulsation wave is possible if typical points are unambiguously determined. This problem can be solved using the amplitude–time analysis of the pulsation wave according to the algorithm described below. This algorithm includes isolation of single waves from the temporal sequence with further application of the model of a single pulsation wave.

Isolation of Single Pulsation Waves

Single pulsation waves corresponding to single cardiac cycles were isolated using the algorithm based on the properties of the first derivative of the cardiac cycle signal. We used for this purpose the local B-spline approximation method [1] with the following calculation method.

Let x_i be the sequence of an actual pulsation signal, $i = 0, 1, 2, ..., N - 1$; N is the number of realizations (steps of sampling). Therefore, x_i', x_i'' are the first and second derivatives of the pulsation signal. Let A_k be the array of local signal maximums, $k = 0, 1, 2, ..., M - 1$ (M is the number of previously unknown single pulsations in the realization).

Single pulsation waves are isolated using the following procedure. Initially, the pulsation wave (first single wave) with maximal amplitude is isolated from the realization (Fig. 1, point 4). The amplitude obeys the following condition:

$$A_0 = \max_{i=0}^{N-1} \{x_i\}, \text{ where } k = 0. \quad (1)$$

The start of the single wave is determined by the following condition:

$$x_i' = 0 \text{ at } i < i_k, \quad (2)$$

where i_k is the ordinal number (abscissa) of the amplitude maximum (Fig. 1, point 4) of the pulsation wave.

This condition unambiguously determines the first pulsation wave maximum (Fig. 1, point 4) at the left of local minimum of the initial signal (Fig. 1, point 3). This minimum is assumed to be the start point of a single pulsation wave (end of preceding pulsation wave).

The next pulsation wave with maximum amplitude from the rest of the single pulsations ($M - 2$) is searched using the same algorithm, etc. The empirical rule for such search for local maximums A_k (starting from $M - 2$) was derived from experimental results obtained in 30 patients and depends on the shape of the pulsation signal and disease nosology:

$$(1.1...1.25) \times A_k \geqslant A_{k-1}. \quad (3)$$

The search is finished when condition (3) becomes unattainable. After the pulsation waves are isolated, they are ranked according to the time of their succession.

Analysis of this method demonstrated that it was correct in 95% of cases, which was due to pulsation wave stability.

Points 1 and 2 (Fig. 1) supplement the pulsation wave to make a complete cardiac cycle. They are determined as the last two points of the preceding cardiac cycle according to the algorithm implemented using an automated pulsodiagnostic apparatus (APDA) [8] (below). Points 5-10 are determined similarly.

Single Pulsation Wave Mapping

The heuristic algorithm based on derivatives of the initial signal was used for isolation of single waves and determination of characteristic points. The characteristic points are the extremums and inflexion points and can be identified using the first and second derivatives.

Computation of characteristic points in single waves is based on the mapping model containing four parameters:

L_i – mean time intervals of cardiac cycles reduced to mean duration of ECG *R-R* waves determining coordinates of centers of zones of characteristic points. This parameter was calculated from kinetocardiograms of sampled representative data. Mean values of parameter L_i are consistent with [9];

dL_i – standard deviation of time of normalized phase intervals determining the width of the zone of characteristic points;

Cv_i – parameter taking values 0, 1, 2 for the curve of characteristic point (initial signal and its first and second derivatives, respectively);

Ty_i – parameter taking values 0 or 1 determining the character of the characteristic point in the curve (0 or local extremum).

The algorithm of identification of characteristic points is described below.

1. The first characteristic point is initial point of the pulsation wave (point 3, Fig. 1).

2. The time of the period of single wave $Lw = (R - R)$ is calculated in absolute units (msec). The coordinate P_i of the center of the zone of search for the *i*th point is determined in absolute units (msec) from:

$$P_i = \sum_{j=1}^{i-1} Lw \cdot L_j,$$

where j is ordinal number of phase intervals, $i = 2, 3, ..., 10$ – number.

The zone size is determined by:

$$Pl_i = P_i - (Lw \times dL_i);$$

$$Pr_i = P_i + (Lw \times dL_i)$$

for the left and right boundary points, respectively.

3. The nearest point to the center of the search zone Ty_i is determined at the curve of parameter Cv_i.

4. With successful fulfillment of step 3, this point is assumed to be *o* (characteristic point). For failure of step 3, the searching zone is increased by 1 to the left and to the right, and steps 3 and 4 are repeated.

The calculation algorithm includes procedures for evaluation of specific and urgent cases and processing of data in these cases, thereby increasing the algorithm confidence for various signals. The method of single wave mapping is effective in 70% of cases. Attempts are made for further increase in efficiency.

Estimation of Accuracy of Determination of Time Phases of the Cardiocycle

The method is implemented as a part of the integrated software of the automated pulsodiagnostic apparatus (APDA). Two modes of operation of the apparatus are available: semiautomatic operation with visual monitoring of pulsation signal mapping performed by a skilled

physician of the Functional Diagnosis Department of the Republic Military Hospital, and automatic operation according to the algorithm described above.

The accuracy of the automatic method of evaluation of a pulsation wave was estimated using calculation of durations of cardiocycle phases in a group of patients (21 males and females). The parameters, as well as polycardiographic results, were averaged over the group as well. The results were compared with model data. The quantitative deviation was estimated using the parameter δ:

$$\delta = \sum_{i=0}^{N-1} (x_i - m_i)^2,$$

where x_i is the array of calculated parameters, $i = 0, ..., N-1$; m_i are model data, $i = 0, ..., N-1$.

Results of comparative analysis of cardiocycle durations are shown in Fig. 2. The durations were calculated from a high-speed kinetocardiogram synchronized with the polycardiographic method (curve 1 – model) and radial artery sphygmogram (curve 2 – SPGra).

The measure of deviation δ was 0.0034, which was satisfactory. Thus, we conclude that this method at given parameters provides satisfactory results at the systolic phase. In the diastolic phase, the phase intervals *P*, *Fr*, and *Sa* are most different from the model. The experimental error is due to small amplitude of pulsation diastolic signal and poor sensitivity of spline interpolation.

The results of two methods of functional diagnosis are statistically consistent according to the Student *t*-test [10]. The variances of the experimental data were assumed to be unequal:

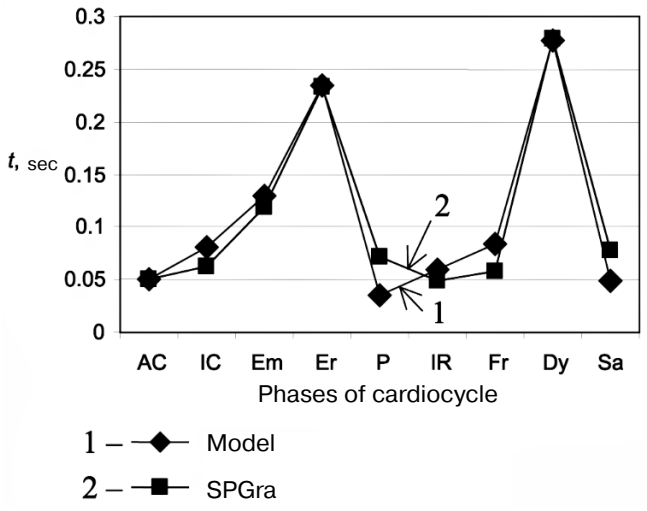

Fig. 2. Mean time of intervals of cardiocycle phases calculated using different methods.

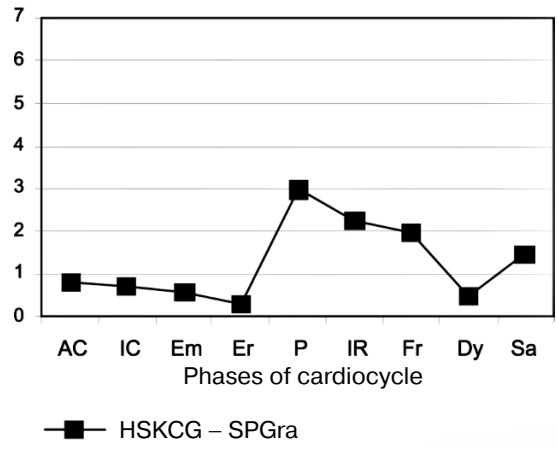

Fig. 3. *t*-Statistical distribution for different cardiocycle phases.

$$t = \frac{|\bar{x} - \bar{y}|}{\sqrt{m_1 + m_2}},$$

where $\bar{x}$ and $\bar{y}$ are mean values of experimental samples; $m_1 = \sigma_1^2/m$ and $m_2 = \sigma_2^2/n$; σ_1 and σ_2 are standard deviations of samples; m and n are their dimensions, respectively. The number of dimensions Q of *t*-distribution was calculated from

$$Q = \frac{\left(\sigma_1^2/m + \sigma_2^2/n\right)^2}{\frac{\left(\sigma_1^2/m\right)^2}{m-1} + \frac{\left(\sigma_2^2/n\right)^2}{n-1}}.$$

The *t*-statistical distribution for different cardiocycle phases is shown in Fig. 3. These times were calculated from high-speed kinetocardiograms recorded synchronously with the reference curves of the polycardiographic method and radial artery sphygmogram.

The results of cardiocycle time determination were statistically confident. High statistical confidence was observed in diastolic phase. Low statistical confidence was observed in incisure (*P* and *IR* intervals). However, the *t*-criterion was less than the critical level over the whole cardiocycle.

Practical application of the algorithm of time intervals and calculated of hemodynamic parameters of the heart was reported in [11-13].

Conclusion

Contour analysis of central pulsograms and a method for isolation of characteristic points of a pulsation wave provided a basis for upgrading the contour analysis

method in 10 characteristic points in automatic mode with further calculation of amplitude–time information parameters. This method provides isolation of characteristic points even in the case of signal smoothing and uncertain resolution of some segments. This allows the method to be used for estimation of cardiac activity by comparative analysis of pulsogram shape in the radial artery.

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