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OPTICAL METHODS

Nondestructive Method for Testing Elasticity of Walls of Human Veins and Arteries

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Abstract—Methods of testing the elasticity of walls of human veins and arteries are considered. The problems that arise when using them are noted. The necessity of developing nondestructive methods for testing the elasticity of the walls of human veins and arteries is substantiated. A technique for monitoring changes in their elasticity using noninvasive methods of pulse wave recording is proposed. The results of comparing data on changes in the elasticity of the walls of veins and arteries obtained using various methods of pulse wave recording and invasive methods are presented. The advantages and disadvantages of the proposed technique using various pulse waves are noted.

Keywords: elasticity, nondestructive testing, wall, vein, artery, laser radiation, pulse wave, structure, time interval, measurement error

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INTRODUCTION

Environmental degradation, acceleration of the life rhythm, and an increase in the level of stress load negatively affect human health [1–4]. Most people experience accelerated wear of various organs under such conditions. Human veins and arteries are no exception in this case. The wear of their walls leads to the formation of various diseases associated with deterioration of blood flow [5, 6]. Currently, a large number of methods have been developed to monitor the flow of blood in humans through veins and arteries, of which contactless methods have received the greatest use: MRI (magnetic resonance imaging) and ultrasound diagnostics [7, 8]. The use of these methods makes it possible to detect dilation or narrowing in the veins and arteries, formation of blood clots, etc. One of the problems that cannot be solved using these methods is monitoring the changes in elasticity of the walls of veins and arteries. Various invasive methods have been developed to determine the elasticity of veins and arteries. Their use requires the insertion of a catheter into a vein or artery with a hollow needle. A sensor (for example, a piezoelectric or induction one) is placed at the catheter outlet to record pressure changes dP. In this case, $E = dP/dV$ (dV is the change in the volume in the catheter associated with the change dP). The error in determining E is approximately 3–5% when matching reference times for the pressure "curve." To ensure this coordination, it is necessary to additionally use other devices, and this creates extra problems.

In addition, the use of these methods is associated with damage to a vein or artery. Therefore, it is not recommended to use them often (it takes time to repair damage to a vein or artery). In people who are regularly injected with intravenous medications, there are extremely few suitable places to implement this procedure. It should be noted that the procedure of measuring E by these methods can be painful.

In the process of treating diseases of veins and arteries, data on changes in the state of their elasticity is extremely important. This is especially true for people who are undergoing treatment at home, for example, for primary monitoring of the effect of medications or various procedures during outpatient treatment in a polyclinic. Monitoring changes in the elasticity of veins and arteries is also necessary in a number of other cases (for example, a person takes medications that contain components that affect the structure of the walls of veins and arteries, and during this period he/she needs to insert a catheter). This can lead to severe damage to veins or arteries with a low value of their elasticity.

Consequently, the development of new methods in the field of noncontact nondestructive methods that are now used in various fields of science and technology [9–11], especially for testing the changes in veins and arteries, is an urgent task (more and more people are susceptible to these diseases). In addition to the necessary accuracy, the main requirement for these methods is the possibility of application without limitation on the number of measurements.

One of these methods is pulse oximetry. A large number of devices using various sensors and detectors have been developed to record a pulse wave in various parts of the human body from the pulsation of veins, arteries, and vessels. Currently, devices using an acoustic or induction sensor have received the greatest application. Acoustic sensors can be divided into several types, which are mainly related to the design of the device they are part of. For example, a sports watch-tonometer that is placed on the wrist or a pulse oximeter that is placed on the shoulder and forearm or devices for use in specialized medical institutions. The principles of operation in these types of sensors remain unchanged (only the shape of the sensor design, operating voltages, etc. vary); this affects the accuracy of measuring pulse wave parameters and sensitivity to changes in its structure.

Fiber-optic sensors based on the Fabry–Perot interferometer are more rarely used. For example, in the Polyspectrum SPV device (NeuroSoft, Russia), piezoelectric or acoustic sensors are used to record a pulse wave on the carotid and radial arteries and volumetric sphygmography for the femoral artery, where induction or acoustic sensors are used. In addition, these devices often use referencing to the ECG signal to determine the beginning of a pressure wave. According to the delay of the femoral artery wave contour relative to the carotid artery contour, the pulse wave propagation time (or the pulse wave propagation velocity (PWPV)) is determined. The following relation is used [12] to determine E:

$$c = \frac{1}{\sqrt{1-\mu^2}}\sqrt{\frac{E}{\rho}\frac{h}{D}}, \tag{1}$$

where c is the PWPV, μ is Poisson's ratio, ρ is the blood density, h is the vessel wall thickness, and D is the vessel diameter.

There are many questions about relation (1) from the point of view of the physical foundations of its derivation from the equations of hydrodynamics with various approximations for a flowing fluid. For example, these include considering blood as a Newtonian fluid, which in fact is not such, and assuming a uniform flow velocity; there are also questions about determining vessel wall thickness and its diameter (they differ for each person due to the physiological characteristics of the body). Globally, it is customary to measure the elasticity E (or the pliability $1/E$) in units of Torr/mL or in Pa/mL (in some cases, H m^{-5} is used). Formula (1) results in other units of measurement for E that relate to Young's modulus. There is no explanation of how to move from Young's modulus to values of E measured, for example, in Torr/mL. It is impossible to compare the values of E obtained using (1) with the elasticity data obtained from the measured values of dP and dV.

It should also be noted that a significant disadvantage of these measurements is that the shape of the recorded pulse waves is affected by the presence of various interference (for example, electromagnetic, electric, and physiological). This leads to large measurement errors that are difficult to compensate for.

Methods of measuring the central pressure (CP) are also used to monitor the change in E. For example, the OMRON HEM-9000AI device (OMRON, Japan) uses a multielement applanation tonometer worn on the wrist to read the pressure signal from the radial artery. It should be noted that these methods are indirect measurements of elasticity E and require frequent calibration. In addition, they should be implemented under the supervision of a specialist in the field of cardiovascular diseases, since many factors that affect the CP level increase in a person with age (for example, "hidden" or "false" load on the myocardium, systolic and diastolic dysfunction of the left ventricle, etc.). All this considerably limits the use and possibilities of CP measurement methods for E monitoring.

The technique of analyzing the shape of a peripheral pulse wave recorded by photoplethysmography is also used to assess the elasticity E of arteries. Currently, two types of this technique are used (reflected and transmission pulse oximetry). Transmission pulse oximetry with the use of two recorded absorption signals with a wavelength of infrared (IR) and red spectrum of laser radiation transmitted through the blood vessels of the finger or earlobe has received the greatest application.

Two indices are used to assess the artery elasticity E. The reflection index RI is the percentage ratio of the height of the diastolic component of peripheral pulse wave to the height of the systolic component (the index reflects the state of the tone of small arteries and the value of the pulse wave of reflection). The rigidity index SI evaluates the pulse wave velocity and is calculated as the ratio of the patient's height to the time interval Δt between the systolic and diastolic components of the wave. Further, the value of E is determined based on these values of RI and SI using calibration dependencies obtained earlier for an average person. Recently, this method leads to very large errors, since changes in our lives (ecology, working conditions, etc.) have reduced the concept of an average person to nothing.

There are also a number of combined methods, for example, using special devices for simultaneous recording of pulse waves from the carotid artery and eyeballs, etc. These methods have not been widely used in practice.

A comparison of all methods shows that the use of transmission pulse oximetry by a person is the most accessible for multiple monitoring of both the artery elasticity and the general condition of the body in express mode at the required time without assistance. The analysis of the results obtained by various scientists [1–4, 7, 12–16] using this method shows that it has a number of disadvantages associated with the inertia of photodetectors, the presence of thermal noise in the photodetector during prolonged operation, a wide spectrum of recorded optical radiation leading to the occurrence of additional difficult-to-compensate light noise during measurements, the presence of a bias voltage that must be maintained with high stability, the absence of a mathematical model to describe the shape of a pulse wave and determine the position of maxima and minima, etc. All this increases the measurement error Δt, reduces the sensitivity of the method to determining the minimum change in the elasticity of arteries and veins, and degrades the reliability of the results obtained. This substantially limits the possibilities of this method, especially for monitoring changes in the state of elasticity of arteries and veins. One of the solutions to the problems considered is presented in this article.

METHOD OF MEASURING THE ELASTICITY OF HUMAN VEINS AND ARTERIES

Depending on the method of recording a pulse wave using laser radiation, its shape changes. Figure 1 shows one period of a pulse wave obtained using a volumetric photoplethysmogram (PPG) and a differential photoplethysmogram (DPPG), as well as an electrocardiogram, which is necessary to determine the starting point (the moment of the beginning of the formation of a new pulse wave period) obtained using the PPG and DPPG.

In the case of recording two pulse waves, the time intervals necessary for determining SI can also be found using the signal obtained using the DPPG (Fig. 1c). This increases the accuracy of determining SI. On the other hand, with a small dicrotic rise, the change of which is also associated with a change in the value of the E of arteries and veins, problems arise with determining t_4 (the position of the peak on the time axis t). This leads to an increase in the error of measuring t_4. In addition, studies have shown that when recording a pulse wave using piezoelectric, induction, and fiber-optic sensors, as well as photodetectors with photodiodes, the normal-wave systolic peak (Fig. 1a, amplitude A_1, time t_2) consists of two small peaks (ambiguous with respect to each other). This feature in the pulse wave structure is revealed when processing the pulse wave peak [17, 18]. This leads to an increase in the uncertainty in measuring the peak amplitude determining the position of the maximum on the time axis (time t_2). The error of measuring t_2 increases. A change in the value of E leads to a shift of these two peaks relative to each other, both in time and in amplitude. The position and parameters of the third peak on the fall of the pulse wave front also change; this can also be used to monitor the value of E. All this leads to the fact that to monitor the change in the value of E, it is extremely difficult to implement measurements of time intervals $\Delta t_1 = t_2 - t_1$, $\Delta t_2 = t_3 - t_2$, $\Delta t_3 = t_4 - t_3$, $\Delta t_4 = t_5 - t_4$, $\Delta t_5 = t_6 - t_5$, and $\Delta t_6 = t_7 - t_6$ with an error of less than 2%. Therefore, there are currently no methods using these parameters to monitor the value of the elasticity of arteries and veins. When assessing the rigidity index SI in clinical settings, a cardiogram -based estimation of the scale on the time axis is often used (Fig. 1b). The time interval between the two maxima is determined by the DPPG at points C_1 and C_2 (Fig. 1c), when the amplitude of the pulse wave signal is zero. A person will not be able to implement such a procedure independently.

Therefore, in order to solve the problem of monitoring the value of E using only the PPG, which a person can perform independently at the time he/she needs, we have developed the following.

A photodetector based on a charge-coupled device (CCD) developed by the present authors was used to record absorption signals at two wavelengths. In the new CCD design, the photosensitive layer was thinned to 100 μm (the standard value is 300–350 μm) and doped with bromine (the doping concentration 6%). Four hidden channels under the photosensitive layer were used to transmit information. This made it possible to increase the signal-to-noise ratio when recording a pulse wave signal whose structure is formed in this case in the form of steps corresponding to the levels of cell filling with charge (quantization of the shape of pulse wave fronts). Figure 2 shows the recorded pulse wave signal using the CCD. In contrast to the previously recorded laser radiation absorption signals using photodetectors (see Fig. 1a) or acoustic and induction sensors, in this case the process manifests a physics that is associated with the

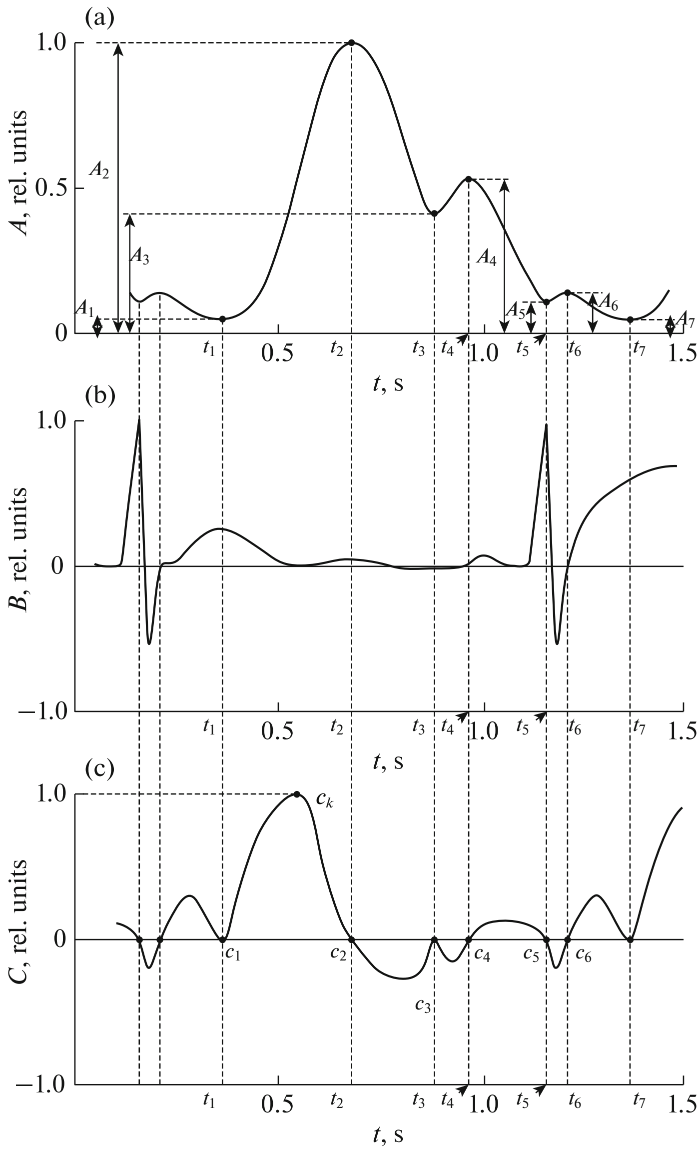

Fig. 1. Simultaneously recorded pulse wave signals using (a) an induction sensor or photodetector, (b) differential photoplasmogram, and (c) electrocardiogram.

pulsation of artery walls during the passage of blood flow during contraction of the heart muscle, the elasticity of arteries and veins, as well as the blood composition (the concentration of hemoglobin in the blood of people varies). Our research confirms this. For different people, the parameters (amplitude and duration) of the steps as well as their number in a pulse wave vary.

In addition, our studies have made it possible to conclude that the value of the signal-to-noise ratio of a pulse wave signal will be maximum if the direction of propagation of laser radiation is perpendicular to the blood flow in the vessel. At the same time, choosing the correct configuration of the sensor that is installed on the finger or earlobe allows one to increase the signal-to-noise ratio by at least 5–10%; in some cases this can affect the accuracy of diagnostics with a weak absorption signal (very thin blood vessels) [19–21].

In the course of these studies, it was found that the wavelength λ has the greatest influence on the signal-to-noise ratio of recorded laser absorption signals. In modern industrial devices, two laser radiation sources with $\lambda_1 = 660.2 \pm 0.4$ nm and $\lambda_2 = 940.2 \pm 0.4$ nm are used to record a pulse wave. These wavelengths were previously calculated for the parameters of an average person. As we have already noted, this term should be abandoned.

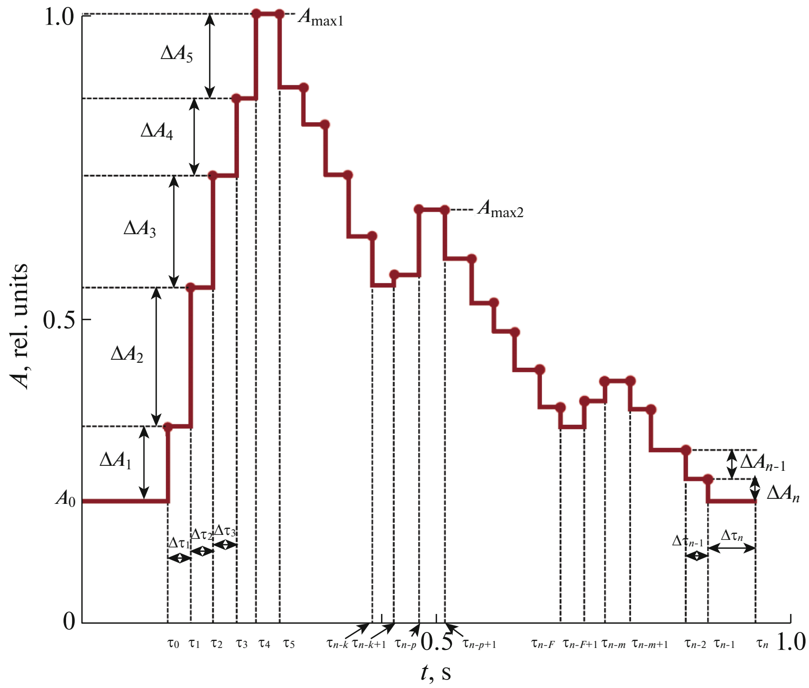

Fig. 2. Pulse wave shape when recording absorption signals using CCD array.

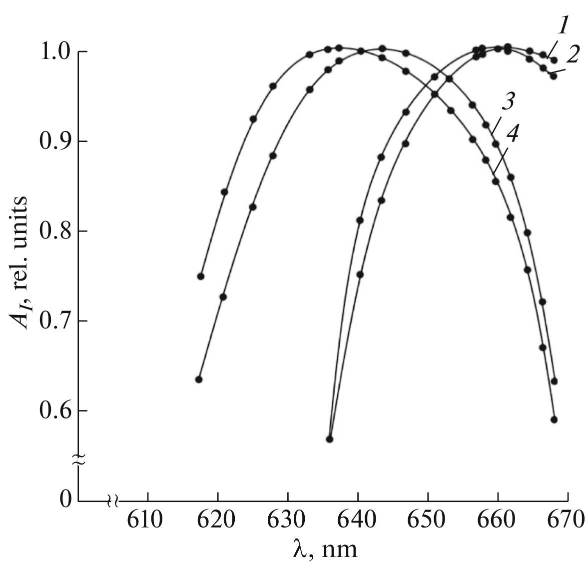

Fig. 3. Dependence of the change in pulse wave amplitude A_I on laser radiation wavelength λ for different people. Graphs *1*, *2*, *3*, and *4* correspond to patients of different genders and ages: 56-year-old male, 21-year-old female, 47-year-old female, and 54-year-old female.

Figure 3 shows as an example the results of a study of the change in the amplitude of a pulse wave with wavelength λ in the visible range of the spectrum for different people. In the experiments, a standard pulse oximeter sensor was used that housed semiconductor laser diodes with different wavelengths in the visible range with a radiation power of $P = 0.2$ mV with a flat angle of the radiation pattern from 10° to 12°.

It was found that in most people, the maximum amplitude of a pulse wave is shifted to smaller wavelengths in the red laser radiation range. The design of modern pulse oximeters provides for the possibility of automatic adjustment of the photodetector by the signal of absorption of laser radiation. In some cases, the pulse oximeter will be adjusted according to the fall of the recorded signal (for example, graph *4* for $\lambda_1 = 660.2$ nm) at a point where the amplitude is less than 30% of the maximum. In the case of using a CCD, this will reduce the resolution of the device in the formation of steps in the pulse wave signal—either part of small-amplitude steps will not be formed or they will be combined with the neighboring one leading to large errors of measuring the values of Δt_n. If photodetectors are used to record a pulse wave, then this will lead to the occurrence of artifacts in the recorded signal. The measurements will be unreliable.

Similar studies have been conducted for near-infrared laser radiation (in the wavelength range from 842.4 to 986.2 nm). It has been found that most people have a shift of the maximum amplitude of the absorption signal towards smaller wavelengths relative to $\lambda_2 = 940.2$ nm.

The experimental data obtained showed that the choice of optimal wavelengths for each person in terms of the processes of oxidation and reduction of hemoglobin forms in the blood allows increasing the signal-to-noise ratio by at least 2 times. This will reduce the error of measuring peak amplitudes and time intervals Δt_n.

To improve the accuracy of measuring the time intervals of pulse wave fronts Δt_n, we have developed the following technique. Figure 2 shows a pulse wave signal formed in the form of steps. The timeline scale is defined as follows. The counter counts N_m the number of peaks per minute (maxima in amplitude A_2, see Fig. 1a). Next, a timeline is constructed with a scale of $60/N_m$ in seconds. The timestamps on the scale correspond to the systolic maxima of the pulse wave signal. There is already an error in this approach, since very often, an integer number of maxima does not fit into a time interval of 60 s. Therefore, we propose to introduce a correction factor $\Delta T_1 = 30/(N_m)^2$ to determine the scale. In this case, the following formula should be used to determine the distance between peaks T:

$$T = 60/N_m + 30/(N_m)^2. \tag{2}$$

In the case of small pulse values (less than 70 beats per minute), it is proposed to introduce additional coefficients in (2),

$$T = 60(1/N_m + 2/(N_m)^2 + \ldots + n/(N_m)^n), \tag{3}$$

where n can vary from 3 to 10 or more.

With an increase in n, the accuracy of determining T increases. Taking into account the specific features of recording absorption signals using a CCD matrix, we propose to determine the position of points t_1, t_2, t_3, t_4, t_5, t_6, and t_7 on the timeline (see Fig. 1a), separating the processes of formation of rising and falling fronts in a pulse wave, since the physics of these phenomena differs; this was not taken into account earlier by other scientists. The position of points t_2, t_4, and t_6 on the timeline will be determined as follows. Consider, for example, determining the time t_2 in accordance with the waveform in Fig. 2. The moment of the end of step formation (register charge) is determined—this is time τ_5. At time τ_6, a step that corresponds to the pulse wave front fall ends to be formed. In this case, the time value t_2 is located between τ_4 and τ_5. Since the formation did not begin with an increase in another step, but there a decrease in the amplitude ΔA_6 occurred (see Fig. 2) that is lower in amplitude than ΔA_5, it can be argued that t_2 is located in the interval between τ_4 and τ_5, $(\tau_5 - \tau_4)/2$. In this case, the value $t_2 = \tau_4 + (\tau_5 - \tau_4)/4$ is selected. The error in determining t_2 in this case is at least two times smaller than when using photodiodes to record absorption signals. Similarly, the values of t_4 and t_6 are determined using a comparison of step amplitudes ΔA_n.

The positions of points t_1, t_3, t_5, and t_7 will be determined as follows. Let us take this as an example of determining the value of t_3. The moment of the end of the step formation (register charge) is also determined—this is the time τ_{n-k}. At the moment of time τ_{n-k+1}, a step corresponding to the pulse wave rise front ends to be formed. In this case, the time t_3 is located between τ_{n-k} and τ_{n-k+1}.

Since the formation of another step did not begin with a decrease in the amplitude, but an increase in the amplitude ΔA_{n-k+1} occurred (see Fig. 2) that is smaller in amplitude ΔA_{n-k}, it can be argued that t_3 is located in the interval between τ_{n-k} and $\tau_{n-k+1} - (\tau_{n-k+1} - \tau_{n-k})/2$. In this case, the value of $t_3 = \tau_{n-k} + (\tau_{n-k+1} - \tau_{n-k})/4$ is selected. In this case, the error in determining t_3 is at least two times smaller than when using photodiodes to record absorption signals. Similarly, the values of t_3, t_5, t_7, and t_1 are determined using a comparison of step amplitudes ΔA_n.

Experimental studies of pulse waves of different people and their comparison with cardiograms and DPPG that were taken synchronously with the pulse wave obtained using an optical sensor with a CCD

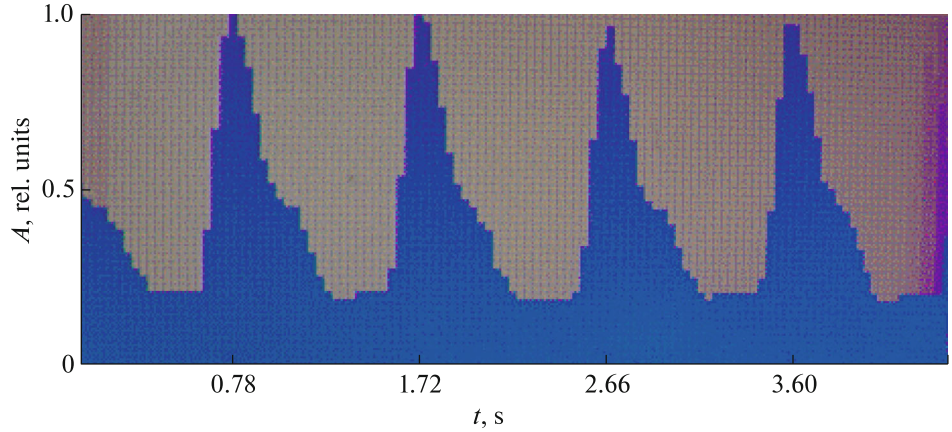

Fig. 4. Shape of absorption signal recorded by CCD matrix in transmission pulse oximetry. The patient is male (age 50).

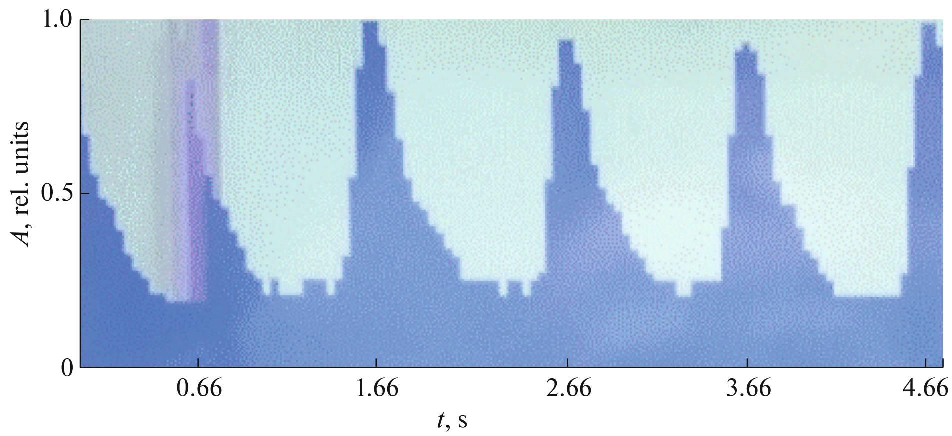

Fig. 5. Shape of absorption signal recorded by CCD matrix in transmission pulse oximetry. The patient is male (age 55).

showed that the use of formula (3) reduces the error in determining τ_n to determine the values of Δt_n several times.

RESULTS AND DISCUSSION

As an example, Figs. 4–7 show recorded pulse wave signals from people with various health conditions that were established using other medical equipment. A visual analysis of pulse waves presented in Figs. 4–7 makes it possible to notice minor deviations in their forms that are quite difficult to associate with changes in the elasticity of arteries and veins. Therefore, we determined the times t_1, t_2, t_3, t_4, t_5, t_6, and t_7 using the developed methods and relations (2) and (3). Using these values, we determined the time intervals Δt_1, Δt_2, Δt_3, Δt_4, Δt_5, and Δt_6. Additionally, we determined the maximum amplitude ratios $\Delta A_1 = A_2/A_4$, $\Delta A_2 = A_2/A_6$, and $\Delta A_3 = A_4/A_6$. These data, as well as the pulse values and percentage oxygen saturation of hemoglobin in the blood, measured at the time of pulse wave recording, are presented in Table 1.

It should be noted that there is no second maximum (time t_6) in the two recorded pulse waves (see Figs. 4 and 5). The presence of this maximum in most cases is associated either with incomplete closure of the semilunar valves of the left ventricle (for various reasons) or with wear of veins and arteries in the human circulatory system. Some data are therefore missing in Table 1.

The conducted studies have also shown that some factors (high degree of fatigue, nervous stress, uncomfortable body position, incorrect sensor positioning, etc.) can lead to the appearance of a second maximum in a pulse wave. Therefore, before using the obtained values of Δt_5 and Δt_6 to determine E, it is

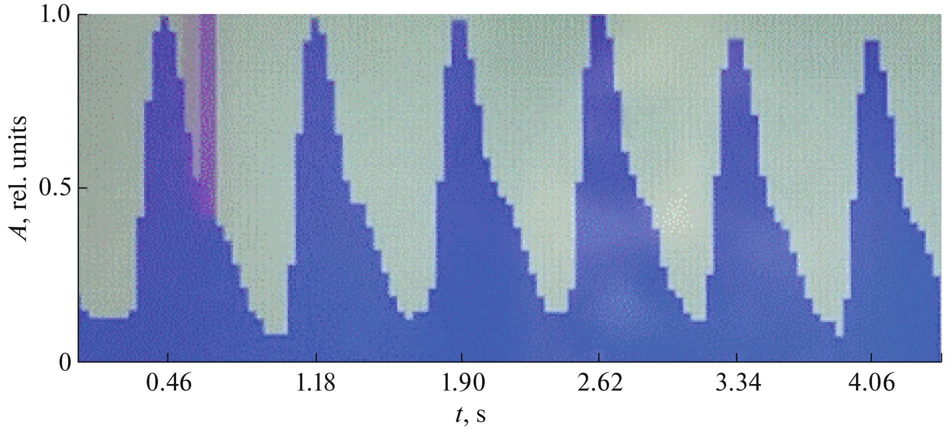

Fig. 6. Shape of absorption signal recorded by CCD matrix in transmission pulse oximetry. The patient is a female (age 56).

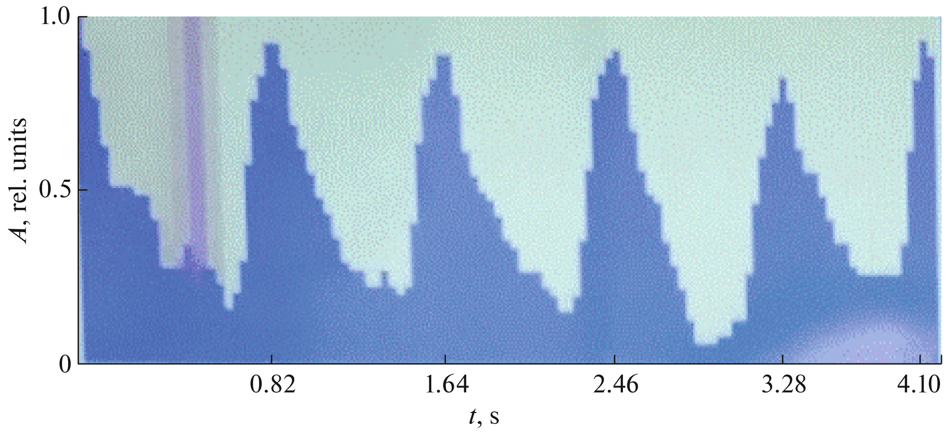

Fig. 7. Shape of absorption signal recorded by CCD matrix in transmission pulse oximetry. The patient is a female (age 47).

necessary to make sure that this third peak is a repeating factor when recording a pulse wave. This allows one to exclude an error that may be more than 20% of the true result.

The following was done to verify the adequacy of results of measuring the elasticity of veins. A catheter was inserted into the median subcutaneous vein of the arm with a pressure sensor connected to the catheter outlet. According to the measured values of change in pressure dP and volume dV, characterizing the filling of the catheter with blood, the value E_c was determined. At this moment, a pulse wave was recorded in the far peripheral zone. Using the methodology developed by the present authors, the values of Δt_n and A_n were determined. Then the calibration was performed (the values of Δt_n and A_n were references to the value of E_c taking into account the human temperature). Without this preliminary grading, the use of the proposed method is ineffective. Determining the time intervals Δt_5 and Δt_6 (the third peak in a pulse wave) with a lower measurement error allows obtaining two additional calibration dependences for a more reliable determination of the value of $E_{\Delta t}$. On the other hand, the form of this peak is unstable in the presence of a disease. This affects the value of other Δt_n and introduces an additional error in determining the elasticity value from the calibration curves. Therefore, if a pulse wave form contains a third peak (even if small in amplitude), it is necessary to calibrate the elasticity value from all values Δt_n and then determine the change in the value of $E_{\Delta t}$ using all values Δt_n.

Next, we conducted the following studies. A 55-year-old female with high blood pressure was undergoing treatment (taking medications to improve the elasticity of veins, arteries, and blood vessels). During treatment, changes in the elasticity value of the median subcutaneous vein were monitored four times with

Table 1. Time position of maxima and minima, rise and fall fronts, the ratio of the maxima of pulse wave peaks, the values of pulse P and the percentage of saturation K of hemoglobin in the blood with oxygen in various people

Human characteristics	P, beats per minute	K, %	Δt_1, ms	Δt_2, ms	Δt_3, ms	Δt_4, ms	Δt_5, ms	Δt_6, ms	ΔA_1, rel. units	ΔA_2, rel. units	ΔA_3, rel. units
Male (age 50, actively engaged in sports). No bad habits	64	99	344.1	155.9	52.2	385.3	–	–	2.854	–	–
Male (age 55, actively engaged in sports). No bad habits	60	97	353.7	163.8	55.7	426.8	–	–	3.072	–	–
Female (age 56, actively engaged in sports). High blood pressure	83	95	291.4	121.4	41.3	72.5	31.8	164.5	2.616	6.366	2.433
Female (age 47, not engaged in sports). Previously smoked	73	96	299.1	146.7	40.3	112.9	44.4	178.5	2.296	6.106	2.659

Table 2. Time position of maxima and minima, rise and fall fronts, the ratio of peak pulse wave maxima, the values of pulse P and the percentage of saturation K of hemoglobin in the blood, as well as indices RI and SI in a person during the course of treatment

Measurement date	P, beats per minute	K, %	Δt_1, ms	Δt_2, ms	Δt_3, ms	Δt_4, ms	Δt_5, ms	Δt_6, ms	ΔA_1, rel. units	ΔA_2, rel. units	ΔA_3, rel. units	RI, %	SI, cm/ms
06.05.2022	83	95	291.4	121.4	41.3	72.5	31.8	164.5	2.616	5.931	2.434	38.21	1.456
11.05.2022	82	96	304.7	114.7	48.1	67.3	35.8	161.1	2.436	5.931	2.434	41.04	1.538
16.05.2022	82	97	312.8	108.2	53.5	63.7	36.8	156.7	2.268	5.552	2.447	44.08	1.630
21.05.2022	80	97	332.1	104.7	61.3	61.1	38.4	152.4	2.180	5.276	2.420	45.86	1.684

Table 3. Comparison of the results of determining the elasticity of subcutaneous vein of the human arm by various methods during the course of their treatment, MPa/mm^3

Date of measurement	E_c	E_i	$E_{\Delta t}$
06.05.2022	1.29	3.74	1.48
11.05.2022	1.30	4.16	1.50
16.05.2022	1.31	4.45	1.52
21.05.2022	1.32	4.62	1.55

an interval of five days using a catheter and the developed method. The results of measurements of E are presented in Table 2. Additionally, according to the data obtained, the indices RI and SI were calculated (woman's height 176.4 cm). Using this data, the values of E_i were determined by calibration dependences for a statistically average person [13, 14]. Table 3 presents a comparison of the results obtained.

The analysis of the results shows that the data obtained using the indices RI and SI differ from the measurement data using a catheter by more than 3–4 times; this makes it possible to monitor the process of change in E only qualitatively. Minor changes in the value of E cannot be determined. When using the developed methodology, the deviation of the value of $E_{\Delta t}$ obtained with its use from measurements by invasive methods (E_c) is approximately 13–14% (see Table 3); this allows one to more adequately monitor the change in E in contrast to the previously used contactless methods. Using the data obtained by the present authors on time intervals Δt_n and A_n, it is possible to monitor the minimum change in the value of E by 0.02 MPa/mm^3. This suffices to monitor the process of treating diseases of veins and arteries, as well as the negative impact of various medications, medical procedures, and other negative factors on their elasticity.

CONCLUSIONS

The results of the research have shown that the use of the developed method allows one to adequately monitor the state of elasticity of veins and arteries without causing damage to their structure. A person can carry out this testing procedure independently without restrictions on the number of measurements at the time one needs. This creates prerequisites for obtaining positive results without damage to the circulatory system (each insertion of a catheter into a vein is a kind of risk).

The use of the proposed novel technique allows reducing the error in determining time intervals Δt_n that, in addition to monitoring the value of E, are also necessary to obtain additional information about the state of human health (for example, the nature of the work of the semilunar valves of the left ventricle, etc.)

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