Improved Method for Measuring the Pulse Wave Propagation Velocity for Palpable Arteries

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Abstract: The results, regarding the development and testing of a quantitative method for diagnosing the condition of the arterial walls, based on the application of a three-channel pneumatic sensor of an original design, are presented. The possibility of using the data obtained by this device in combination with synchronous ECG measurement to determine the velocity of pulse wave propagation between two cross sections of the selected artery has been demonstrated. One of the key points of this technology is the selection of a specific pulse wave element as a reference mark for tracking the signal transit time relative to the R-peak of the synchronous ECG. After collecting measurement statistics, the average values of the wave propagation time between the selected points of the artery, considering the variability of the front delay values, are used to directly calculate the propagation velocity of the pulse wave in the investigated area of the artery. The value of the pulse wave propagation velocity in return is an objective parameter that characterizes the degree of elasticity (or stiffness) of the artery walls and their behaviour in different physiological situations.

1 INTRODUCTION

In previous articles (Mansurov, 2019) (Antsiperov, 2020), the authors have developed and experimentally investigated a line of unique sensors for recording the blood pressure continuous dynamics.

Good results concerning the recording data quality have been achieved with the development of a three-channel pneumatic sensor for continuous non-invasive blood pressure monitoring (Mansurov, 2019).

Thanks to the addition to the sensor of a parallel channel for synchronous recording of the ECG signal, it became possible to for the combined device to measure the absolute timing characteristics of the pressure pulse waves, the main of which is the pulse wave velocity (PWV). On the basis of the latter, it is possible to assess the stiffness of the arterial walls, which is directly related to the early manifestations of symptoms of atherosclerosis.

Undoubtedly, the most important directions in the fight against atherosclerosis are its prevention and early diagnosis. In the early stages, even before the appearance of obvious clinical signs, atherosclerosis is characterized by an outwardly weak loss of the main functions of the blood vessels. First, this is manifested in the loss of elasticity by the vascular walls. The process of increasing the stiffness of the arterial walls leads to an increase in blood pressure (BP), narrowing of the lumen of the arteries and a deterioration in blood circulation in general. It has long been established that one of the most adequate methods for assessing arterial wall stiffness (gold standard) is the measurement of pulse wave velocity (PWV) value. Physically, PWV is the group velocity of a pressure wave propagating along the elastic walls of an artery because of the ejection of a mass of blood from the left ventricle of the heart during systole. Within the framework of the first order, linear approximation, the theory of hydroelasticity gives the following value for the velocity V of an elastic wave (Korteweg, 1878):

$$V = \sqrt{\frac{Eh}{\rho D}}, \tag{1}$$

where E denotes the effective (tangential) Young's modulus, parameters h and D are the wall thickness and vessel diameter in rest, respectively, and ρ is the blood density in the vessel. It follows from (1), which is known as Moens-Korteweg equation, that a growth

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of the elasticity E of the vessel wall and a decrease in its diameter D lead to a increase in the value V of PWV.

The natural method of measuring the pulse wave travel time (PWTT) between a pair of given points located proximally on the arterial wall is a direct method of measuring PWV. Direct measurement of PWV requires a pair of sphygmometric sensors located proximally above the superficial vessels (arteries) and distal to the heart. Any of the pairs of points located above the carotid, femoral, radial, and other palpable arteries are suitable for this role. If the distance d between such pair of points is known and the delay Δt of pulse wave transition time between these points is measured, then the PWV value is obviously the ratio of the first of them to the second:

$$V = \frac{d}{\Delta t} = \frac{d}{t_2 - t_1}, \quad (2)$$

where t_2 and t_1 are the time moments when the pulse wave passes through the locations of the selected points. How these moments are measured is not an easy question. They can be measured by the event of passing of a specific waveform-related label, which can be the foot of the waveform, its maximum, the maximum slope of the wavefront, etc. (Katsuura, T. et al, 2017).

2 PWV ESTIMATION BASED ON CONTINUOUS BLOOD PRESSURE REGISTRATION

Earlier, we developed and tested a new three-channel pneumatic sensor for continuous non-invasive blood pressure monitoring (Antsiperov, 2020). An obvious advantage of the developed device is the possibility of a continuous measurement of the dynamics of blood pressure, which allows not only to determine the current systolic / diastolic pressure, but also to track the dynamics of blood pressure, both within the cycle and at significant time intervals. At the same time, it is not always possible to correctly calibrate the measured value in pressure units. For correct measurement of blood pressure in absolute units, a certain position of the sensor above the artery is required, as well as a rigid base below it, such as the radial bone for the artery of the same name (Antsiperov, 2020). Only under the properly positioned sensor the pressure in the working chamber of the sensor could be considered equal to the blood pressure in the artery (Figure. 1)

It was found experimentally that for arteries whose location does not satisfy the above conditions, it is possible to observe a pressure pulse wave signal, the level and amplitude of which is noticeably distorted by viscoelastic tissues lying both between the sensor and the artery and beneath artery.

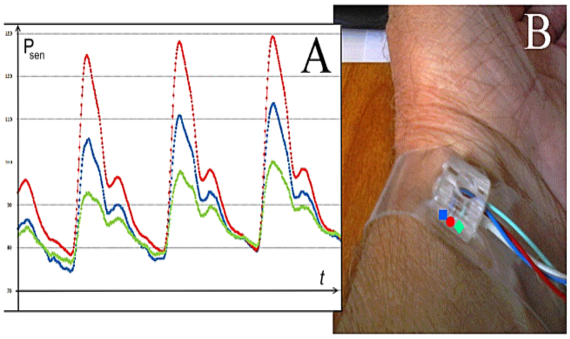

Figure 1: Three-channel pneumatic sensor for continuous non-invasive arterial blood pressure monitoring (B). The difference is the shape of the pulse wave signal from the sensor (A), depending on the position of the sensor pads: ● – pad is directly over the artery, ■. ♦ – pad is shifted to the left and to the right respectively from the central projection of the radial artery.

However, the general structure and corresponding temporal characteristics of the signal are retained in this case as well. This effect can be used to measure the delay of signal front relative to the ECG reference element. To solve the problem of the assessment of pulse wave propagation velocity a unique way to use pneumatic sensors was developed. The idea was the following: if it is possible to take measurements for a pair of points on the artery for a finite time with the patient's condition unchanged, then you can try to do it with the only one sensor. Evidently, the position of the R-wave on the ECG can be used as a periodically repeating reference "zero". For this purpose, an additional channel for synchronous measurement of the electrocardiographic (ECG) signal was integrated into the pressure sensor configuration. In this case, speaking in the language of radio engineering, this channel acts as a kind of reference signal. The ECG amplifier circuit was developed with the expectation of using dry electrodes without a conductive gel and without a neutral electrode, that required the application of both analogue circuitry and digital filters. The simplified scheme was selected to minimize the inconvenience when applying and removing the electrodes and is used so far at the development stage only.

3 PWV BASED ON ABP/ECG JOINT REGISTRATION

To solve this problem, we, together with NEUROCOM LLC, have developed a new design configuration of a portable sensor, which does not require rigid attachment to the patient's body and allows, with a certain skill, to take measurements in many important positions (Figure 2). The three-channel pneumatic sensor of a new configuration (type) allows continuous measurements for several minutes, even in some hard-to-reach places.

Figure 2: The version of the pressure sensor for measuring short patterns of blood pressure with a built-in ECG channel (two coloured button connectors).

Existing methods for assessing the stiffness of the arterial walls by the pulse wave propagation velocity are based, as a rule, on measuring the time of propagation of the pulse wave along the radial artery by the delay in the appearance of the wave on the wrist relatively to the preceding R-peak of the electrocardiogram. It was demonstrated (Kortekaas, 2018) that the moment when the pulse wave appears in the aorta does not coincide with the R-peak of the ECG but has a certain time lag. The magnitude of this lag differs for certain age groups and diseases, and it varies relatively weakly within these categories. Even this small variability has a significant impact on the accuracy of pulse wave velocity (PWV) measurements. The error in measuring the PWV value mainly depends on the accuracy of measuring the signal delay time, since the length of the section of the artery under study can be measured quite precisely. The distance from the aorta to the radial artery in the area of the wrist joint in an adult man is about 0.8 m, and the pulse wave propagation velocity is about 7-14 m/s. Therefore, the signal delay can be estimated to be on the order of 100 milliseconds. For example, for a given measurement accuracy of 5%, the error in measuring the delay should not be greater than 5 ms.

It is not difficult to determine the moment of "dynamic zero" from the "rather narrow" R-peak of the ECG with such accuracy, but it is not so for a pulse wave impulse. Quite often, many researchers, being biologists or physicians by education and scientific background, tend to use the position of the minimum or, conversely, the maximum of the pulse wave impulse as the required timestamps (Kortekaas, 2018). However, it is well known from radiophysics that various transient processes in the region of the establishment of the pulse top or in the region of the minimum significantly distort the main process and interfere with fixing timestamps with the required accuracy. As a solution, pulse technology uses tracking of timestamps on the leading edge of the pulse at the level of 0.1 and 0.9 of its amplitude defining start and rise time. Taking into account these considerations and having a portable pulse wave sensor, a PWV measurement method was developed based on determining the average difference in the delay values from the R-peak on the ECG to the sequential arrival of the pressure pulse front at selected points on the artery under study.

4 EXPERIMENTAL RESULTS OF PWV MEASURMENTS

In our experiments, the results of which are presented in this section, a pulse wave was recorded at five different points along the artery (Figure.3): on the subclavian artery (1), on the brachial artery (2) by 0.35 m lower (just above the ulnar bend), on the radial artery (3) by another 0.27 m lower, at the artery at the base of the middle finger (4) lower by another 0.17 m and at the artery in the pad of the same finger (5) lower by another 0.06 m.

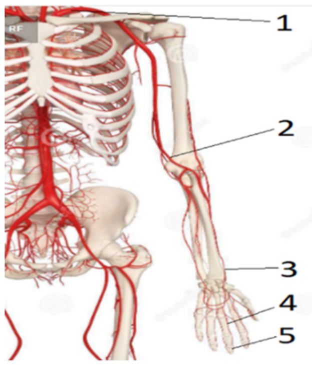

Figure 3: Locations of the points to measure PWV.

Figure 4 demonstrates the joint recording of the pulse wave signal and the ECG, the numbers on the graph correspond to the measurement points on the patient's body (see Figure 3).

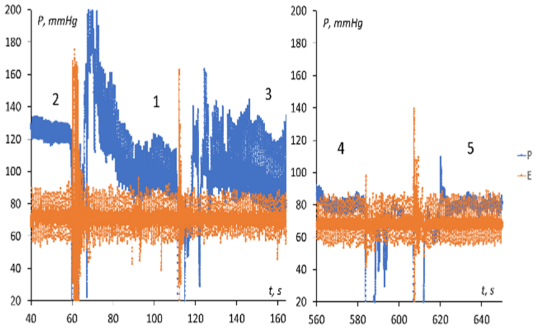

Figure 4: Five-point record of pulse wave and ECG signals.

When deciphering the readings, it should be remembered that only for the radial artery, the pulse wave parameters correspond to the actual blood pressure. In other cases, there are various distortions caused, for example, by the damping properties of the surrounding soft tissues. Therefore, the amplitude and constant component of the pulse wave in these areas may differ rather significantly from the actual pressure in the artery.

Since we are free of the tabular reference values for the “pre-ejection period” (Kortekaas, 2018), we develop a criterion for the stabilization of the wave propagation time, which is more reliable for pressure wave signals. We will consider the beginning and end of the leading edge of the signal, which we determine, as it is customary in pulse technology, at levels 0.1 and 0.9 of the signal amplitudes (timestamps f01 and f09, respectively). In our case, the levels of the thresholds of the pulse wave fronts for each cycle are updated, as demonstrated in Figure 5. Each positive edge of the signal has "personal" marks of the beginning and end of the edge. Of course, the beginning of the front should be taken as the marks of the passage of the wave, since the shape of the pulse wave noticeably changes during the motion, as does the duration of the front.

This algorithm was used first to indicate digital value of R-f01 online. Value of R-f01 turned out to be slightly varying from pulse to pulse and should be averaged to use practically. Below, Figures 6-10 demonstrate the graphs of record fragments of pulse wave and ECG signals at each of the five measurement points and graphs of time parameters for each successive pulse wave front.

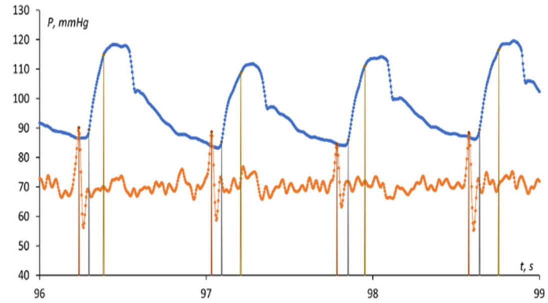

Figure 5: Timestamps at the level of 0.1 and 0.9 of the signal amplitudes for each wave front.

Legend: f01-09 – pulse wave rise time, R-R - time intervals between successive R-peaks, f01-f01 - intervals between successive edges of pulse waves, R-f01 - delay time of pulse wave front relative to R-peak.

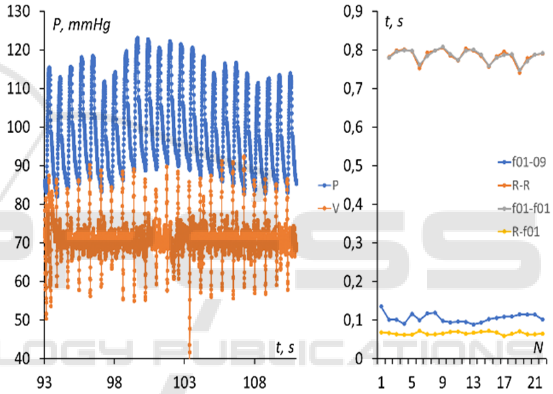

Figure 6: Pulse wave and ECG recording and their temporal characteristics for the subclavian artery (point 1 in Figure 3).

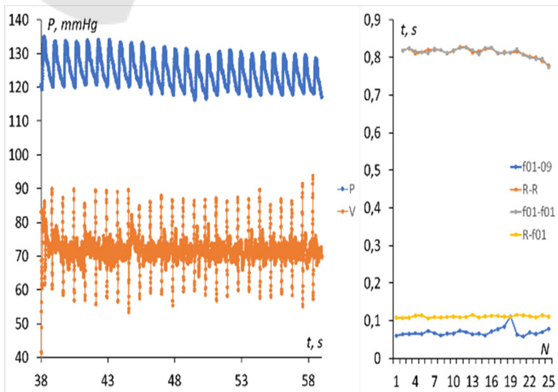

Figure 7: Pulse wave and ECG recording and their temporal characteristics for the brachial artery (point 2).

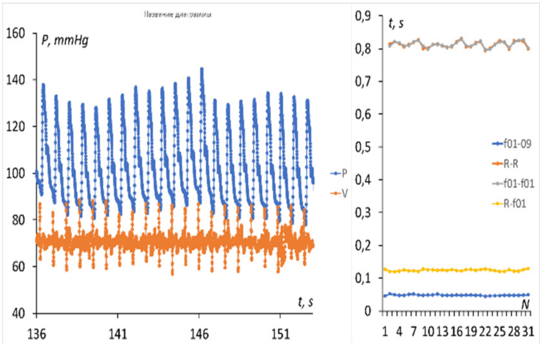

Figure 8: Pulse wave and ECG recording and their temporal characteristics for the radial artery (point 3).

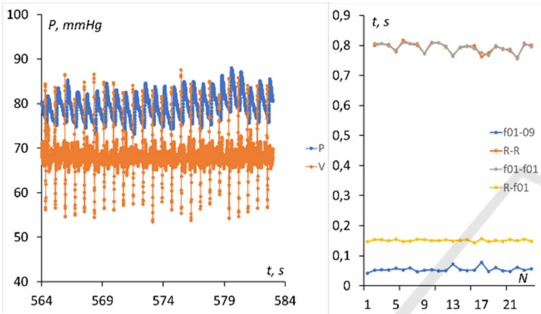

Figure 9: Pulse wave and ECG recording and their temporal characteristics for the digital artery (point 4).

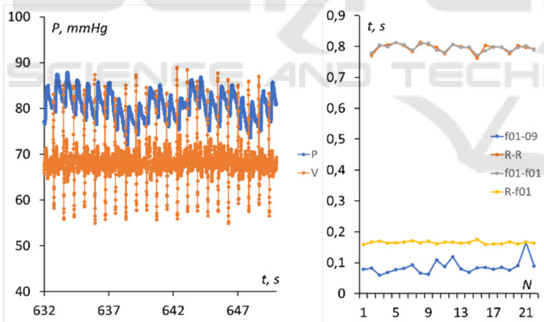

Figure 10: Pulse wave and ECG recording and their temporal characteristics for the digital artery (point 5 in Figure 3).

It should be noted that these data were obtained when measuring the indicators of cardiovascular activity in a specific measuring environment and for certain physiological conditions of the patient. But, even under controlled experimental conditions, the data may change due to the dependence of the patient's reactions in a particular measurement session on many uncontrolled factors. It is obvious that even with a given physiological condition, all time intervals are variable, and averaging is required to correctly estimate the signal delay time. In this case, the standard deviation did not exceed 1 *ms*.

The average delay time of the wave front relative to the R-peak of the ECG was 60.5 milliseconds (*ms*) for the subclavian artery, 109.5 *ms* for the brachial artery, 130 *ms* for the radial artery, 140 *ms* at the base of the finger and 177 *ms* at the fingertip for the digital artery. Accordingly, the assessment of the propagation velocity of the pulse wave in the section 1-2 from the subclavian to the brachial artery was about 7.4 *m/s*, in the section 2-3 - 13.5 *m/s*, in the section 3-4 - 17 *m/s* and in the section 4-5 about 2 *m/s*. The increase in velocity in sites 1 - 4 can be explained by a decrease in the diameter of the branching arteries according Moens-Korteweg equation (1). The decrease in speed in the last section, apparently, is due to more complex reasons. Note that the duration of the pulse wave front is not the same in different points of the measurement sites: 94 *ms* on the subclavian, 65 *ms* on the brachial, 58 *ms* on the radial, 70 and 78 *ms* on the digital arteries, respectively.

5 CONCLUSIONS

Thus, the variant of a three-channel pneumatic sensor described in the paper, designed for measuring and recording short-term blood pressure files, can be used as a pulse wave sensor on all palpable arteries. Experiments on measuring pressure in various superficial arteries show that in the case of arteries located above hard tissues (bones), both the shape of the pulse wave and the actual value of blood pressure can be recorded. This is true, for example, in the case of the radial and femoral arteries. The presence in our device of an additional ECG channel makes it possible to uniformly measure the delay time of the pulse wave at spaced points of the artery, that makes it very easy to calculate the pulse wave propagation velocity along the walls of the artery.

Obtaining and subsequent interpretation of data on the dependence of the distribution of the pulse wave velocity in the arterial system on various internal and external factors is extremely interesting, both for a general understanding of the processes taking place in the human body and for the practical diagnosis of the different diseases. It is important to note that the claimed technology can significantly improve the accuracy of measurements of the pulse wave propagation velocity due to small size of sensing pad.

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