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NON-INVASIVE ULTRASONIC EXAMINATION OF THE LOCAL PULSE WAVE VELOCITY IN THE COMMON CAROTID ARTERY

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Local pulse wave velocity (PWV) is one of important indicators of artery wall's elastic properties. This paper describes a non-invasive method of examinations of local PWV on the basis of ultrasonic blood flow velocity measurements at two points, several centimetres along the common carotid artery (CCA). This velocity was compared with values of PWV calculated on the basis of measurement of distensibility coefficient DC. The examinations were carried out on the group of 12 healthy volunteers ranging in age from 22 to 68 years (mean 39 ± 16 years). Mean value of local PWV for the examined group was 5.1 ± 1.09 m/s. This velocity was linearly correlated ($r = 0.9807$) with the PWV examined on the basis of distensibility coefficient DC, the mean value of which was 6.14 ± 1.29 m/s.

Key words: local pulse wave velocity, common carotid artery, ultrasound.

1. Introduction

The velocity of propagation of pressure and blood flow waves in a human vascular system, called pulse wave velocity (PWV) [1, 2] is, in the medical diagnosis, a very popular indicator of elasticity of the arterial vessel wall [3]. This velocity depends on the artery vessel walls Young's modulus, according to the Moens–Korteweg equation:

$$\mathrm{PWV} = \sqrt{\frac{Eh}{2\rho\, R}} \quad [\mathrm{m/s}], \tag{1}$$

where E is the Young's modulus, h is the vessel wall thickness, R is the radius of the artery and ρ is the blood density.

In the clinical practice the examinations of the PWV are usually performed by means of two transducers located far away from each other on the vessel system, i.e.

between the carotid and femoral arteries [4–6]. In this method, the basis for evaluating the PWV is the measurement of the time delay of the pressure or blood flow waves over a known distance between two measurement points. The measurement of PWV on a long distance of the vessel tree allows only for a global assessment of the vessel system elasticity. Contemporary ultrasonic methods applied for arterial wall elasticity examinations are directed towards selective examination of elasticity in the particular arterial cross-section. The ultrasonic examinations concern the measurements of maximal and minimal artery diameters corresponding to the systolic and diastolic blood pressure measurements respectively provided by the sphygmomanometer. On this basis, series of indicators for assession of the arterial elastic properties i.e. distensibility DC [7], compliance CC [7], elastic modulus E_p [8], stiffness coefficient α [9] and β [10] are determined. A serious limitation of those methods is the fact that the measurements of the systolic and diastolic pressure are made in the different times and places than the measurement of arterial diameter. For example, during the examination of the common carotid wall elasticity, the measurements of blood pressure were carried out in the brachial artery. Therefore it is necessary to work out a method for assessment of the local elastic arterial wall properties which does not require the measurement of the blood pressure. A proper method could be the measurement of the local PWV on a short segment of the artery. Results of examinations of the local PWV were published, among other authors, for example by CHUBACHI *et al.* [11], ERIKSSON *et al.* [12] and by MEINDERS *et al.* [13]. In all these papers the basis for evaluation of the local PWV is the measurement of the radial displacement or the velocity of displacement of the artery wall under influence of the pulse pressure.

In this study, the local PWV in the CCA is evaluated by the authors on the basis of the ultrasonic blood flow velocity measurement. The local PWV is compared with the PWV calculated from the Bramwell-Hill relation [2].

2. Method and equipment

The experimental setup (Fig. 1) consists of two VED (Vascular Echo Doppler) apparatuses [14]. The examination of the blood flow velocity was performed by two Doppler continuous wave flowmeters.

The frequency of the ultrasonic wave transmitted by probes P1 and P2 for both flowmeters was 4.5 MHz. The waves were focused on the depth of 1–3 cm below the body surface. The width of the ultrasonic beam in focus (for -6 dB) was, respectively, 0.66 mm along and 1.7 mm across the vessel axis. The ultrasonic beams of P1 and P2 were located in parallel at the distance $l = 2.3$ cm. The distance d of the artery segment between two measurement points was calculated (according to Fig. 1) from the relation:

$$d = \sqrt{\left[\sqrt{k^2 - l^2} - \frac{w}{2}\left(t_1 - t_2\right)\right]^2 + l^2}. \tag{2}$$

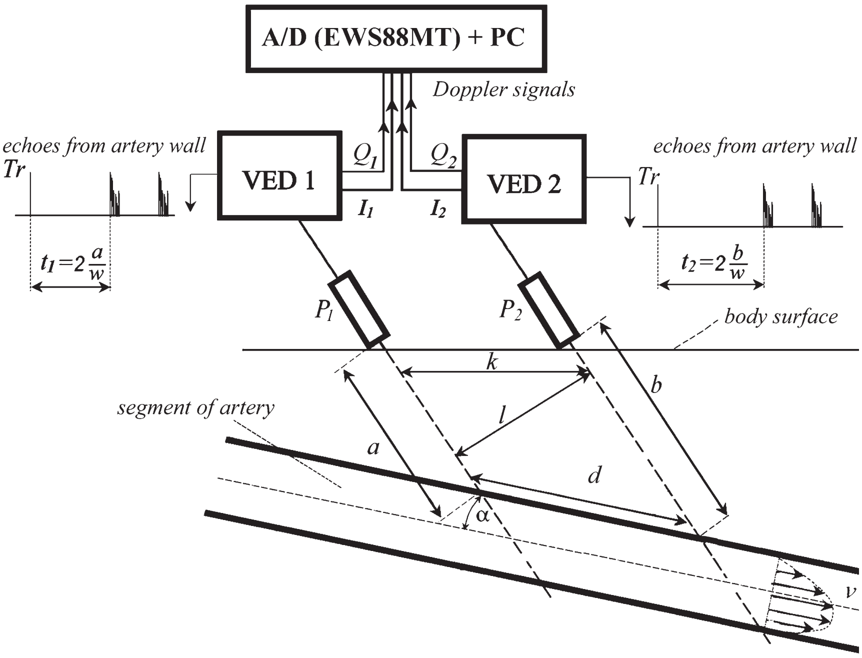

Fig. 1. Experimental setup: k, l – distances between the axes of ultrasonic beams of probes P1 and P2, w – velocity of ultrasonic waves, $t_1/2$, $t_2/2$ – times of the ultrasonic pulse propagation between the probes P1, P2 and the anterior surface of the carotid wall, α – angle between the ultrasonic beams and the axis of the artery ($0° < \alpha < 90°$).

For measuring the propagation times t_1 and t_2, both of VED apparatuses were switched for the pulse mode.

During the blood flow velocity measurement, achievement of the maximum amplitude of Doppler signal spectral components as indicators of properly fixed ultrasonic probes P1 and P2 along the arterial axis was used. These spectral components were obtained in VED by means of 128 points FFT and were presented during examination on the screens of both apparatuses. Doppler quadrature signals (I_1, Q_1, I_2, Q_2) were synchronically recorded by the sound PC card type EWS 88MT (TERRATEC). The frequency of signals acquisition was 44 100 Hz with dynamics of 16 bits. The time of acquisition session was 8 seconds. The separation of the blood flow directions was obtained by application of a software Hilbert transform to the digital values of Doppler quadrature signals. As a post-processing, the spectral density and its mean frequency of Doppler signals on the basis of 1024 points FFT were estimated. For one second, 11008 successive spectra of the Doppler signal were estimated. As a result, the evaluation of mean Doppler frequency was carried out with sampling rate of 11008 Hz. The blood flow waves evaluated in that way for a few successive hart cycles were averaged in time on the basis of the beginning of the systolic phase.

The blood pressure and blood flow waves propagating along the vascular tree are reflected [14, 15]. As a result, the shape of both the waves are changed along the

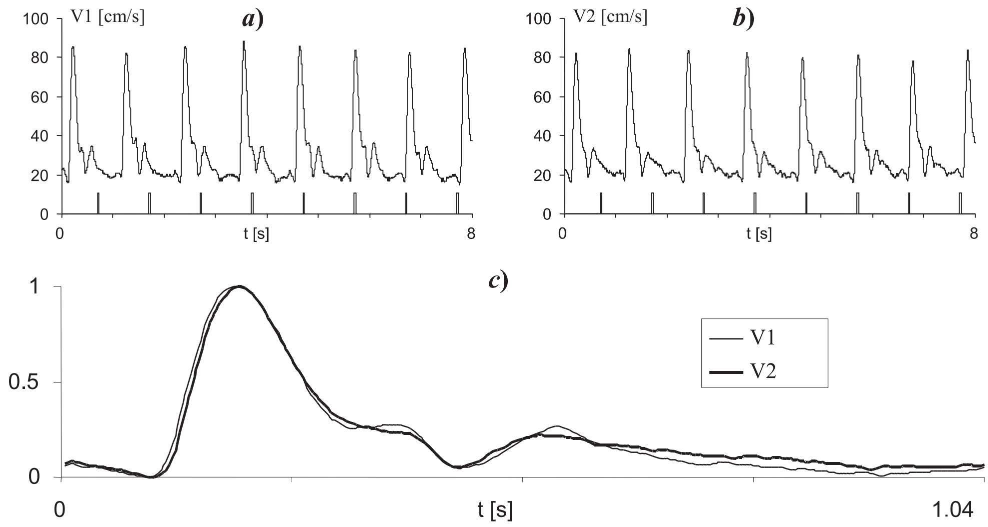

Fig. 2. Tracing of blood flow velocity recorded in the CCA of a man 27 years old: a) far from bifurcation of the CCA, b) near the bifurcation of the CCA, c) relative change of blood flow velocity in both measurement points for an averaged hart cycle.

artery segment between two measurement points. For determination of the local PWV the "foot to foot" method is usually used [2]. In this paper, the authors evaluated the local PWV on the basis of measurement of the time delay for the initial part of the systolic phase of averaged blood flow wave between two measurement points. Only the parts of rising slopes with similar gradients of blood velocity in two measurement points (Fig. 3) were taken into account.

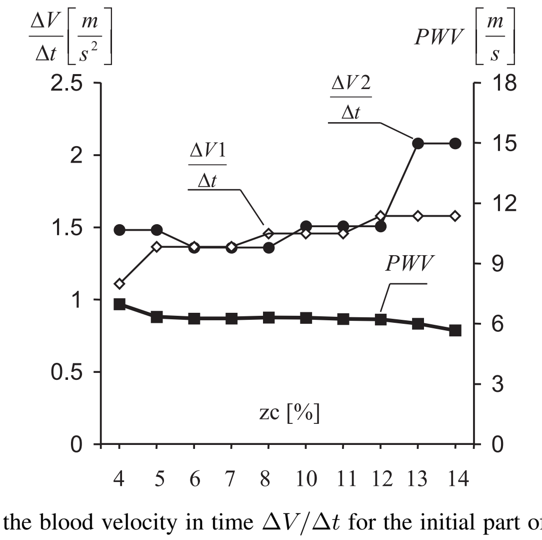

Fig. 3. The increase of the blood velocity in time $\Delta V/\Delta t$ for the initial part of systolic phase for measurement points 1 and 2 and local PWV for different increases of blood flow amplitude levels zc. Results concern a man 58 years old. PWV $= 6.6$ m/s for $\Delta V1/\Delta t \approx \Delta V2/\Delta t$.

The local PWV evaluated on that basis was referred to the PWV calculated on the basis of measurements of maximal diameter $D_{\max}$ and minimal diameter $D_{\min}$ of the CCA, and systolic P_s and diastolic P_d blood pressures in the brachial artery (BA) and blood density ρ, according to the modified Bramwell–Hill relation [2]:

$$\mathrm{PWV} = \sqrt{\frac{(P_s - P_d)\, D_{\min}^2}{\rho\left(D_{\max}^2 - D_{\min}^2\right)}} = \sqrt{\frac{1}{\rho\,\mathrm{DC}}} \quad [\mathrm{m/s}]. \tag{3}$$

The measurements of the maximum and minimum CCA diameters were performed using the ultrasonic VED apparatus. For this purpose, an ultrasonic pulse system tracking the displacement of vascular wall with accuracy equal to $7 \cdot 10^{-6}$ m was used. The inner diameter was determined by means of digital time measurement between the selected echoes (RF signal) received from the inner vascular wall layer. The frequency of transmitted ultrasound was 6.75 MHz. The ultrasonic wave was focused at 1 to 3 cm below the skin surface. The measured data were stored in the computer memory of the VED system. Apart from the data obtained from ultrasonic measurements, also the values of systolic P_s and diastolic P_d blood pressure were transmitted to the computer memory. The blood pressures P_s and P_d were measured by sphygmomanometer in the BA before and after ultrasonic examinations in the CCA. For determination of the PWV from the relation (3) were used the average values of pressures P_s and P_d and the average values of CCA diameters $D_{\max}$ and $D_{\min}$ evaluated for 10 successive cycles of hart beats.

3. Results

The study is performed on 12 healthy volunteers ranging in age from 22 to 68 years, without any atherosclerotic symptoms and without atherosclerotic risk factors. Examinations were carried out in supine position after a 15 min. rest. Ultrasonic examinations of the local PWV were carried out in the right CCA a few centimetres before bifurcation. For a group of examined persons, averaged values of the evaluated parameters are presented in Table 1. In Fig. 4 the dependence of the PWV as a function of age and the relation between the PWV measured on the short artery segment and calculated from formula (3) are presented.

Table 1. Averaged values of evaluated parameters for the examined group.

age [years]	P_d [mmHg]	P_s [mmHg]	$D_{\min}$ [mm]	$D_{\max}$ [mm]	PWV1 [m/s]	PWV2 [m/s]	heart cycle [s]
39 ± 16	76.3 ± 9.7	125.7 ± 9.9	6.86 ± 1.33	7.42 ± 1.27	6.14 ± 1.29	5.1 ± 1.09	1.011 ± 0.243

The results of PWV examinations obtained on the basis of measurement of the time of blood flow wave propagation on a short CCA segment indicate that PWV evaluated in this manner has a ca. dozen or so percent lower value than the pulse wave velocity evaluated on the basis of measurement of the changes of the CCA diameter and blood pressure in the BA measurement. These results indicate a strong linear dependence between the values of PWV evaluated by both methods ($r = 0.9807$).

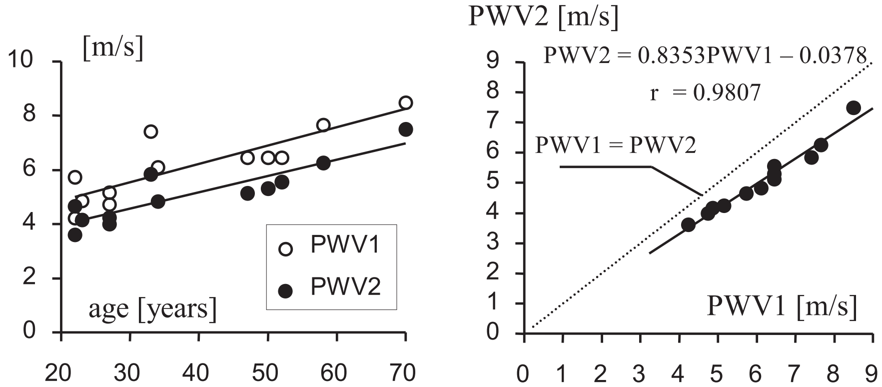

Fig. 4. Pulse wave velocity (PWV) in the CCA for the examined group: PWV1 – calculated from formula (3), PWV2 – measured on the short artery segment.

4. Discussion

The local PWV is an important indicator of elastic properties of the arterial wall. Its values increase with growing stiffness of the arterial wall (Eq. (1)). In the Fig. 5 are presented the values of PWV evaluated by the authors in the CCA on the basis of the measurements of distensibility coefficient DC by using the VED apparatus. Examinations were carried out on the group of volunteers without atherosclerosis (192 arteries)

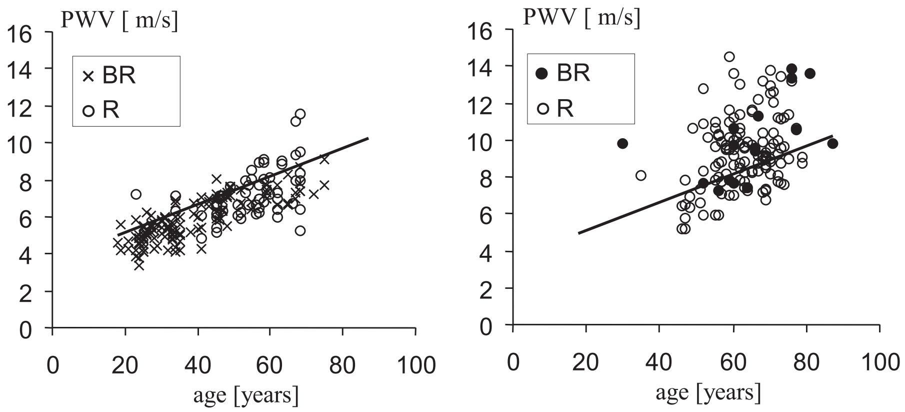

Fig. 5. Pulse wave velocity (PWV) evaluated in the common carotid artery on persons without symptomatic arterial atherosclerosis (a) and on patients with atherosclerosis of carotid arteries (b). BR – persons without atherosclerotic risk factors, R – persons with atherosclerotic risk factors.

and on the group of patients with atherosclerosis of carotid arteries (160 arteries) [16]. Both groups consisted of persons without and persons with risk factors of atherosclerosis, i.e. hypercholesterolemia, arterial hypertension and diabetes mellitus. The results obtained indicate the influence of the age and atherosclerosis on the values of PWV. In the Fig. 5, the continuous line (PWV $= 0.076$ [age] $+ 3.6062$ m/s) represents the critical level of the PWV. Above this line, the value of PWV could be recognized as an indicator of risk for atherosclerosis. The principle of construction of this criterion was a line of linear regression between the PWV and the age of persons obtained for healthy persons without risk factors for atherosclerosis, increased by the value representing standard deviation of linear regression of variability of PWV for healthy people.

The local PWV measured on a short distance of the CCA was the aim of examination of ERICSSON *et al.* [12] and MEINDERS *et al.* [13]. Ericsson *et al.* used the tissue Doppler imaging (TDI) technique. The PWV was evaluated using 16 transducers (18 mm) of the 32-element ultrasonic linear array probe. The examinations was carried out on a healthy man 30 years old with the result of PWV = 8.29 $\pm$ 0.64 m/s. This value is much higher than the values obtained for persons of the same age by authors of the present paper (Fig. 4). Meinders *et al.* evaluated PWV on the basis of change of the arterial diameter in time and along a short segment of the artery. The PWV was examined by means of 16 transducers (15.8 mm) of the 128 element ultrasonic linear array probe. Examinations were carried out on a group of presumably healthy volunteers ranging in age from 21 to 51 years old (mean value 38 $\pm$ 9 y.o.) and they have obtained the PWV = 5.5 $\pm$ 1.6 m/s. This value was compared with the value of PWV calculated from the Bramwell-Hill dependence (Eq. (3)). This velocity was equal to 6.7 $\pm$ 1.2 m/s. These results are very similar to the results presented in this paper obtained for the group of person with a similar range of age (Table 1). Both Meinders *et al.* and authors of this paper obtained the values of PWV from the Bramwell-Hill dependence (Eq. (3)) about 20% higher than those obtained by measurement on the short segment of the artery. One of the reasons of discrepancy of the above results is the possibility of a difference between the values of blood pressures in the CCA and in the brachial artery (BA). According to MEINDERS *et al.* [17], there exists a difference of the systolic blood pressure between the CCA and the BA. Assuming that the diastolic and mean blood pressure do not differ significantly between the CCA and the BA [17], the hypothetical value of systolic blood pressure $P^*_{s\mathrm{CCA}}$ in the CCA could be evaluated on the basis of the instant change of the CCA diameter $D(t)$ [9, 14] according to the relation (4) (see Fig. 6):

$$\int_0^T P^*(t)dt = \int_0^T P_{d\mathrm{BA}} \exp\left[\frac{D^2(t) - D^2_{\min}}{D^2_{\max} - D^2_{\min}} \ln\left(\frac{P^*_{s\mathrm{CCA}}}{P_{d\mathrm{BA}}}\right)\right] dt$$
$$= P_{m\mathrm{BA}} = P_{d\mathrm{BA}} + \frac{(P_{s\mathrm{BA}} - P_{d\mathrm{BA}})}{3}, \qquad (4)$$

where $D_{\min}$, $D_{\max}$ – minimal diameter and maximal diameter of the CCA, $P_{d\mathrm{BA}}$, $P_{s\mathrm{BA}}$, $P_{m\mathrm{BA}}$ – diastolic, systolic and mean blood pressure in the BA, T – time of cardiac cycle.

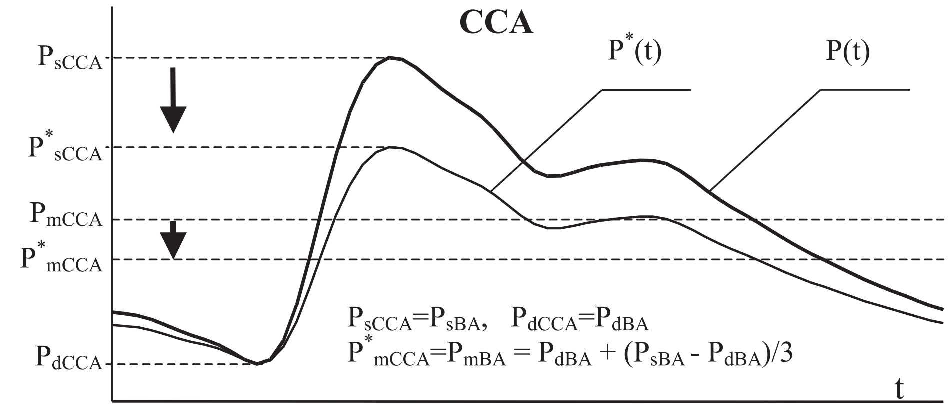

Fig. 6. Procedure of determination of the systolic blood pressure in the CCA: $P(t)$, $P^*(t)$ – blood pressure waves in the CCA determined on the basis of variation of the artery diameter, $P_{s\mathrm{CCA}}$, $P_{d\mathrm{CCA}}$ – systolic and diastolic blood pressure in the CCA determined on the basis of systolic $P_{s\mathrm{BA}}$ and diastolic $P_{d\mathrm{BA}}$ blood pressure measured in the BA, $P^*_{s\mathrm{CCA}}$ – systolic blood pressure determined in the CCA under assumption that mean $P^*_{m\mathrm{CCA}}$ and diastolic $P^*_{d\mathrm{CCA}}$ blood pressures in the CCA are equal to mean $P_{m\mathrm{BA}}$ and diastolic $P_{d\mathrm{BA}}$ blood pressures in the BA.

Figure 7 presents the ratio of the pulse pressure $\Delta P^*_{\mathrm{CCA}} = P^*_{s\,\mathrm{CCA}} - P_{d\mathrm{CCA}}$ in the CCA to the pulse pressure $\Delta P_{\mathrm{BA}} = P_{s\mathrm{BA}} - P_{d\mathrm{BA}}$ in the BA, determined on the basis of the procedure described above (Eq. (4), Fig. 6). Mean value of this ratio equals 0.71 ± 0.04 and is similar to the value given by MEINDERS *et al.* [17]. Substituting in Eq. (3) the new value of the systolic blood pressure $P^*_{s\,\mathrm{CCA}}$ a hypothetical value of the PWV1* (Fig. 8) was obtained, which is very similar to the PWV2 measured on a short segment of the CCA. Mean value of the velocity PWV1* for the group of people examined was equal to 5.23 ± 1.14 m/s.

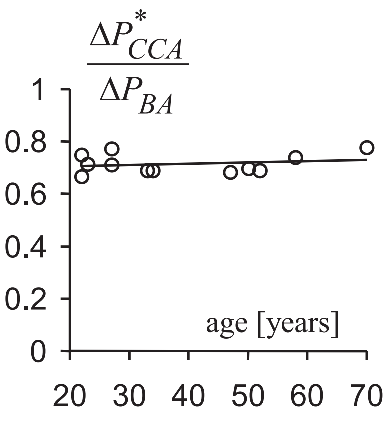

Fig. 7. Hypothetical value of the ratio of pulse pressure $\Delta P^*_{\mathrm{CCA}}$ in the CCA to the pulse pressure ΔP_{BA} in the BA for the group of the people examined (see Fig. 4).

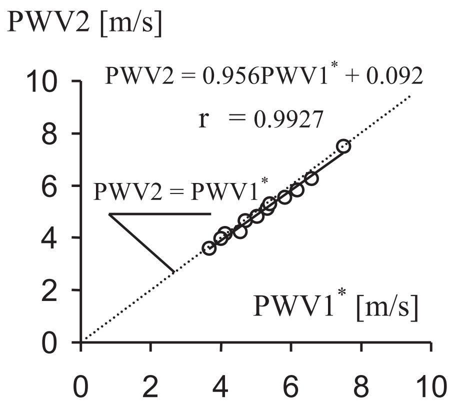

Fig. 8. Pulse wave velocity (PWV) in the CCA for the group of people examined: PWV1 * – determined on the basis of measurement of variation of the diameter CCA and blood pressure in the BA (Eq. (3)) under the assumption that the diastolic and mean blood pressure in the CCA are the same as in the BA; PWV2 – determined on the basis of the blood flow velocity measured at two points.

5. Conclusions

The preliminary results of PWV examinations obtained on the basis of ultrasonic blood flow velocity measurements at two points, several centimetres along the CCA, indicate that the local PWV is a ca. dozen percent smaller that the value determined on the basis of measurement of variation of CCA diameter and blood pressure variation in the BA (Eq. (3)). The reason of this effect my be the difference of blood pressures between the CCA and the BA, especially concerning the values of systolic blood pressures. The results obtained indicate a linear relation between the pulse wave velocities determined of both the methods ($r = 0.9807$). The method of examination of the local PWV presented in this paper requires further verification on a larger group of patients.

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