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Pulse pressure velocity measurement – A wearable sensor

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Abstract—Pulse pressure velocity measurements (PPV) may be a source of useful information on artery state. A 2007 guideline of the European Society of Hypertension recommends assessing arterial stiffness in patients with arterial hypertension, by measuring the PPV. Mechanical changes in the cardiovascular tree involved in the blood ejection have been measured at the thorax level and on the wrist using impedance techniques, in combination with a one channel electrocardiographic signal. Performing impedance measurements on the wrist is a very challenging task because of the very low conductivity changes in combination with the relatively high value of the basal impedance. This is especially a problem, when the dimensions of the measuring probe are required to be adequate for integration into a whole day wearable sensor. In this study, it has been shown that such measurements are indeed possible, and that they may deliver very useful information on pulse pressure velocity when compared to the classical approach based on the measurement of the PPV delay in relation to the ECG signal. However, it has been found that a significant discrepancy may exists between results obtained using the classical approach and the impedance measurement technique proposed here.

***Keywords*: Pressure pulse velocity, impedance technique, body sensors**

I. Introduction

PULSE pressure wave velocity (PWV) measurements allow evaluation of arterial stiffness. The pressure wave following the ejection of blood by the heart is gradually conveyed to the periphery. Close to the heart the wave velocity is of the order of 5 m/s and gradually increases towards the periphery and then again decreases in very small arteries. However, only the mean value of the wave velocity over a given segment is considered, while the vessel wall characteristics are highly site dependent. On the other hand, pulse wave velocity assessment has the advantage that no blood pressure information (measured invasively) is required. Several studies have shown that wave velocity measurements allow assessment of the haemodynamic profile of a patient as a function of age and pathophysiological conditions [1], [2]. The European Society of Hypertension recommends in its guidelines evaluation of this parameter in patients with arterial hypertension. It is also suggested that carotid-femoral pulse wave velocities greater than 12 m/s are an indicator of organ damage [1].

In general, apart from geometric non-uniformity, the changing elastic properties and viscous components make the structure of the artery much more complicated. E.g. for the aorta, stiffness is increasing with the distance away from the heart, which is accompanied by increased damping properties due to differential compositions of the wall.

Different approaches for PWV measurement may be applied [1], [6], while in most of them the pulse pressure is measured at two distant points separated by a known distance, and the propagation delay between the two signals is estimated. This time delay is determined between two "corresponding" points on the pressure wave. Measurements should be performed at stable conditions, as the propagation of the pressure wave depends on many factors, including the pressure itself. The difference between results obtained by applying two different approaches in estimation of the PPV is examined in this paper. A first approach is based on the time delay between the R wave and a selected point of the pulse, while the second one is based on the measurement of the time delay between two impedance signals, one measured on the thorax and another at the wrist.

II. Methods

Impedance measurements have been already applied in evaluation of blood flow in artery and venous system. However, the most widespread method, called impedance plethysmography, is based on occlusion of venous return in selected part of a limb (a segment) and measurement of impedance change following the change of segment volume [11]. Pairs of measurement electrodes are localized distantly when comparing to limb diameter in this application. It is in contradiction to the presented application. On the other hand, it is known phenomenon that configuration of electrode matrix (a probe) influences a shape of the recorded signal [11].

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A. Theoretical backgrounds

Sensitivity of impedance measurements to conductivity changes localized in examined segment can be evaluated by relation $S = \nabla\phi_n \cdot \nabla\varphi_n$, where ϕ_n and φ_n are normalized potential distributions associated with unit current flowing respectively between "current" and "voltage" electrodes. The latter one is a hypothetical one. The impedance change involved by conductivity changes is described by the following relationship [3]

$$\Delta Z = -\int_V \Delta\sigma(x,y,z)\nabla\phi_n \cdot \nabla\varphi_n dv \qquad (1)$$

where $\Delta\sigma(x,y,z)$ - spatially distributed change of conductivity. The relationship (1) allows calculation of impedance change (ΔZ) associated with conductivity change ($\Delta\sigma$) undergone in the subject. A geometry, conductivity distribution and the electrode matrix configuration are reflected in spatial distribution of potentials (ϕ_n, φ_n).

According to the relationship (1) a value of the measured impedance change depends on a volume of the conductivity change. Thus, it is important to estimate how volume changes of an artery reflect pulse pressure propagation along it.

The wave propagation velocity, based on theoretical model in [4], [5], is described by the relationship

$$c^2 = \frac{Eh}{2\rho(1-\mu^2)r_0}\left[1 - \frac{2}{ar_0}\frac{J_1(ar_0)}{J_0(ar_0)}\right] \qquad (2)$$

where c is the pulse pressure velocity, E the modulus of elasticity, μ the Poisson's ratio, $J_0(ar_0)$ and $J_1(ar_0)$ Bessel functions of the first kind, and $a^2 = -j\omega\rho/\eta$, where ρ, ω and η denote the blood density, circular frequency, and blood viscosity, respectively.

For inviscid fluids, $\eta = 0$, $a \to \infty$ the second term in square brackets in eq. 1 becomes zero

$$\frac{2}{ar_0}\frac{J_1(ar_0)}{J_0(ar_0)} = 0. \qquad (3)$$

As a result the following relationship is obtained:

$$c^2 = \frac{Eh}{2\rho(1-\mu^2)r_0}. \qquad (4)$$

Setting $\mu = 0$, (3) reduces to the well-known Moens-Korteweg equation:

$$c_0^2 = \frac{Eh}{2\rho r_0}. \qquad (5)$$

The above relationship has been obtained assuming that [4]:

1. The wall material is homogeneous, elastic, isotropic, and follows Hooke's law.
2. The relative variations of the dimensions are small.
3. Variation of length is not possible.
4. There exists only a radial motion; so there is no rotation (in other words rotational symmetry is assumed).
5. The thickness h of the wall tube is small, as compared with the inner radius r_0, at the mean pressure in the tube.

However, in the actual measurements the above assumptions are not fulfilled exactly. In spite of this, it is generally accepted that pulse velocity measurement is essential in evaluation of cardiovascular state [6]. The relationship (2) can be rewritten in the following form [7]

$$c = \frac{c_0}{(X - jY)(1 - j\omega W)} \qquad (6)$$

where viscous effects of the blood on phase velocity, (c), are taken into account by $X - jY$ term and the second term in the denominator of equation (6) takes into account the effects of wall viscosity as well as a complex Poisson ratio.

B. Measurement technique

The pressure wave or its function has been measured at the thorax and wrist using two different sensors, both being based on the impedance technique. The thorax sensor is described elsewhere [8]. Fig. 1 shows the electronic circuit of the wrist sensor, which consists of a current source built around two operational amplifiers, that delivers a current of 0.1 mA_{pp} at 20 kHz. The high output impedance of the current source ensures a constant current for changes of load impedance up to 10 kΩ. These specifications have been realized by means of an active feedback design.

The voltage arising from the current flowing between electrodes I_1 and I_2 is measured by means of voltage electrodes V_1 and V_2, and amplified 50X. In the next stage, a programmable amplifier allows to select four levels of amplification, such as to ensure a maximum SNR ratio.

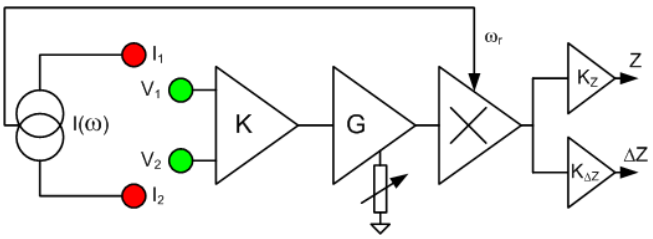

Fig. 1. Schematic diagram of measurement circuit, I(ω) – current source, K, G, K_Z $K_{\Delta Z}$ - gain of respective stage of bandpass amplifier, mark × stands for synchronous demodulator

The amplified signal is then passed to a synchronous demodulator, built around another operational amplifier. The maximum input amplitude is determined by the linear range of the synchronous demodulator.

After demodulation the signal is split between two channels, acting as low-pass and pass-band filters, respectively. The former delivers a signal proportional to the value of the basal impedance Z, depending on the average electrical and the geometrical properties of the wrist. The amplifier marked K_Z in Fig. 1, is a low pass filter with a corner frequency equal to 2 Hz. The time-dependent signal, ΔZ, reflects the changes of blood volume in the measured segment of the body, and in

general is synchronous with the blood induced conductivity changes in the wrist segment of the hand. The amplifier $K_{\Delta Z}$, instead acts as a pass-band filter with middle frequency equal to 11 Hz, a low corner frequency below 0.3 Hz, and high corner frequency equal to 20 Hz. For the middle frequency, this two-stage amplifier has a constant gain equal to 200.

C. Model and realization of the sensor's probe

It was assumed that the sensor should be in the form of a watch with the four electrodes mounted on the watch strap. To allow calculation of the sensitivity function for different sensor configurations, a FEM model of the limb segment was built.

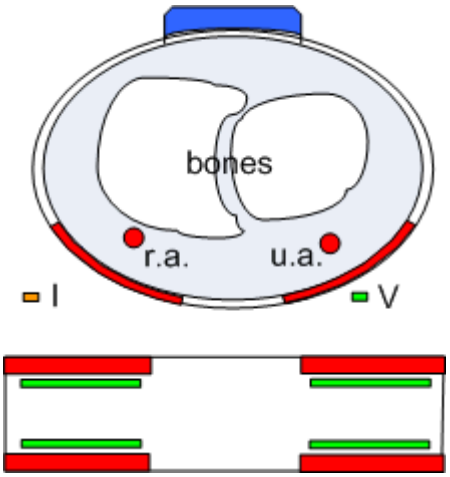

Fig. 2. Schematic representation of a cross-section of the wrist. Construction of the four electrode sensor (current injecting electrodes marked red, voltage measuring electrodes marked green); r.a. indicates the radial artery, while u.a. stands for ulnar artery.

This model was in the form of a cylinder of finite length with four circumferential electrodes. Two of them (marked by letter I) were used for current injection, while the other two (marked V) serve as voltage measurement electrodes. Several variants, differing in geometry, were examined both theoretically and experimentally.

First, a sensor consisting of classical circumferential electrodes was examined for equally spaced electrodes. Then, the voltage electrodes were moved closer to the current electrodes, so that these are no longer equally spaced, but with the external dimensions preserved. This latter sensor configuration is shown in Fig. 2.

D. Signal analysis

The recorded signals were analyzed manually after being collected and imported into Microsoft Excel. Time relations between the impedance signals were calculated and assessed according to the definitions specified in Fig. 3. Although only one time course of the impedance signal is illustrated in Fig. 3, the same parameters were calculated for the impedance signals recorded on the upper thorax and the wrist. The effect of the arm position is known to influence both the arterial and venous pressure [9], [10], and was closer examined in our experiments.

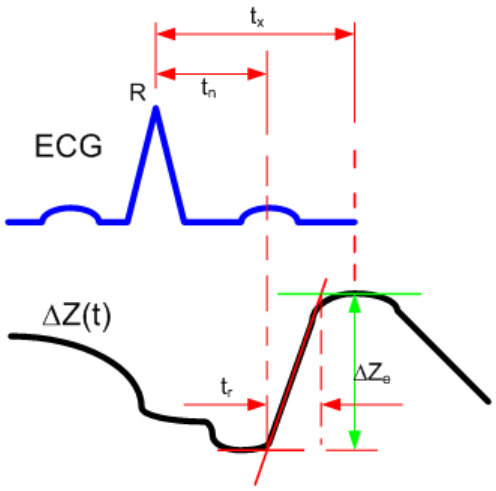

Fig. 3. Definition of time parameters and amplitude of the signal: t_n, the time delay between R wave in the ECG and the beginning of impedance signal ΔZ_e, t_x, the time delay between the R wave and the maximum of impedance signal ΔZ_e, and t_r, the rise time of the signal.

Pressure waves were recorded for the arm hanging freely down the body, kept at the heart level, and raised straight above the head.

III. Results

It can be assumed, from relationship (2) that the pulse velocity is larger than 5m/s for the considered arteries (starting from the ascending aorta and finishing at the ulnar or the radial arteries). Moreover, assuming that heart rate is 1 Hz (in fact it is a little bit higher) a wavelength of generated pressure wave of frequency 1 Hz is around 5 m. Higher harmonics are of course respectively shorter.

Sensitivity values were calculated for a circumferential electrode configuration applied to a cylinder of radius R and height ±10 R.

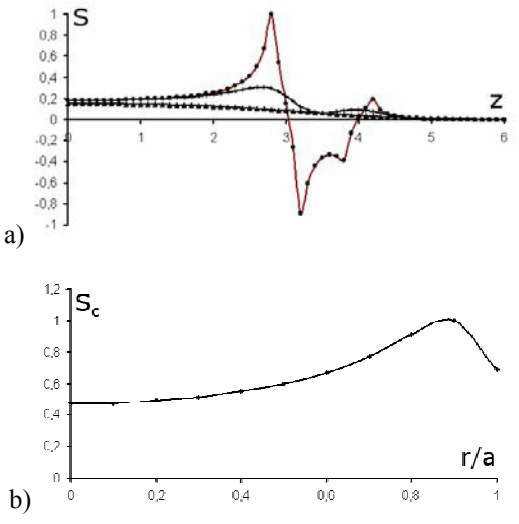

Fig. 4. Local sensitivity function (a) for conductivity changes at 0.5R (filled triangles), 0.9R (crosses) and 0.99R (filled squares), where R is the radius of the cylinder and b) total sensitivity, S_c, calculated as the sum (integral) of sensitivity S along coordinate "z" for different value of r. An anisotropic model, with anisotropy ratio equal to 2, was assumed.

Anisotropy of the conductivity was assumed, i.e. the ration σ_l/σ_t =2, 4, etc., with σ_l and σ_t the longitudinal and transfer conductivity, respectively. The voltage electrodes were placed at ±R, with the current electrodes positioned at ±1.3 R. Sensitivity values for a conductivity changes localized at 0.5 R, 0.9 R and R respectively are shown in Fig. 4a. The integral sensitivity (calculated as a sum of sensitivity for a certain depth) is also dependant on artery localization in the segment (Fig. 4b).

The developed sensor was found to allow measurement of impedance changes less than 15 mΩ in magnitude, with the channel bandwidth of the impedance measurements limited to 20 Hz, while the basal impedance value ranged from 10 Ω up to 60 Ω. However, the impedance measured between "current" electrodes belonged to range (200 ÷20 000) Ω and strongly depended on types of electrodes used.

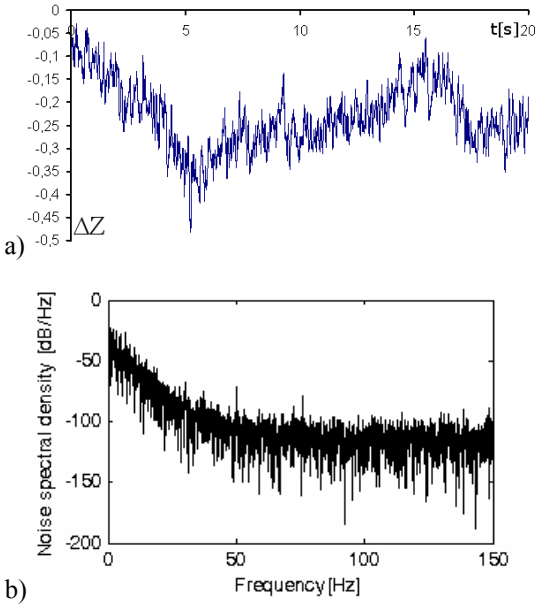

Fig. 5.(a) Noise of the ΔZ channel recorded on a resistance of 1 kΩ, (b) resulting power density.

The noise level of the sensor was estimated by means of measurements recorded on a 1 kΩ resistor (Fig. 4). The signals recorded using the thorax and wrist sensors are presented in Fig. 6 and 7, respectively. All signals have been acquired for standing person with the hand hanging freely down the body.

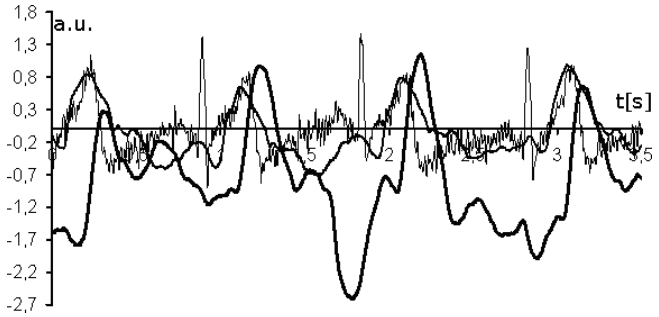

Fig. 6. Recorded signals of electrical heart activity and impedance changes on the upper part of the thorax and the wrist. The signals are marked in more detail in the following figures

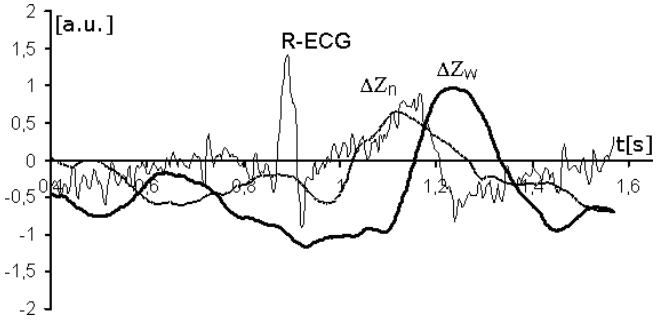

Fig. 7. One cycle of the signals presented in Fig. 6. ΔZ_n denotes the impedance change recorded on the upper part of thorax, while ΔZ_w is the impedance change recorded at the wrist. The hand was kept hanging down.

All signals presented in Fig. 8 have been obtained with the wrist kept at the level of the heart. Parameters of the signal measured on the thorax are changed slightly in comparison to previous one.

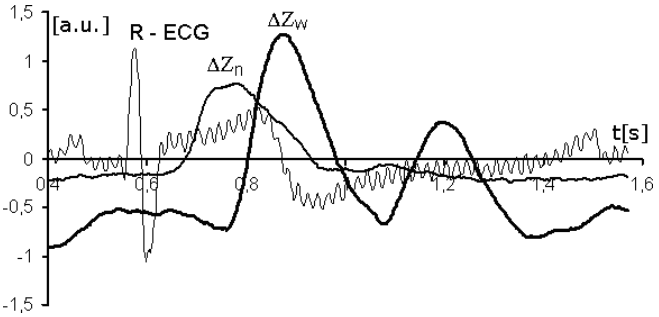

Fig. 8. One cycle of recorded signals with the hand kept at the level of the heart, ΔZ_n denotes the impedance change recorded on the upper part of thorax, while ΔZ_w is the impedance change recorded at the wrist.

The most significant and noticeable changes are observed for the signal recorded at the "wrist," rather than for the "cardiac" signal recorded on the thorax. However, when raising the hand above the head, both impedance signals show changes (Fig. 9). An enhanced wave coinciding with the QRS complex of the ECG has appeared, probably reflecting the activity of the atria.

Differences in time delay were evaluated for different positions of the arm, i.e. freely hanging along the body, kept at the level of heart, and held straight above the head. Time delays were calculated between the ECG and impedance signals measured on the wrist (Fig. 10), as well as between the impedance signals measured on the thorax and at the wrist (Fig. 11).

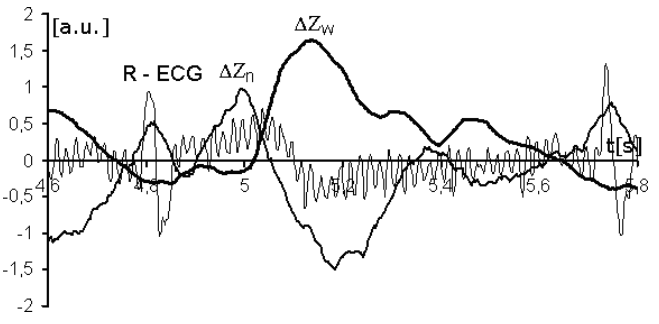

Fig. 9. One cycle of recorded signals with the hand kept above the head, ΔZ_n denotes the impedance change recorded on the upper part of thorax, while ΔZ_w is the impedance change recorded at the wrist.

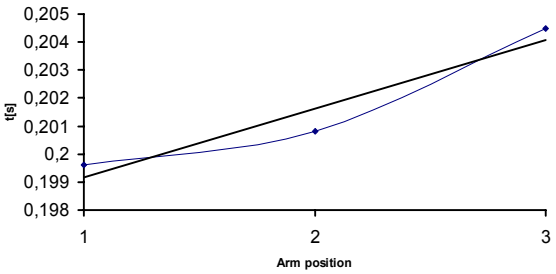

Fig. 10. Time delay between the R wave of the recorded ECG starting point of ΔZ_w signal, for three positions of the arm: (1) arm hanging freely down the body, (2) arm kept at the level of the heart, and (3) arm hold straight above head. Indicated in bold is the trend line.

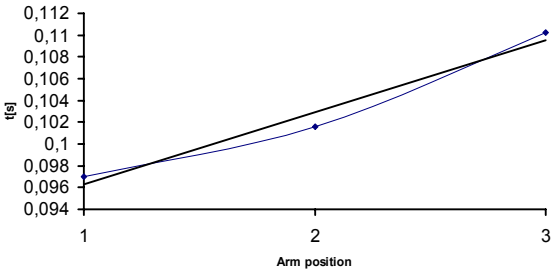

Fig. 11. Time delay between the starting the point of the impedance signal measured on the thorax, and the starting point of ΔZ_w as measured at the wrist, for three positions of the arm: (1) arm hanging freely down the body, (2) arm kept at the level of the heart, and (3) arm hold straight above head.

The distance between the heart and the wrist is relatively short so that different measurement methods may yield large discrepancies. The differences observed between the two methods were indeed quite large, often separated by almost a factor 2.

IV. Discussion

Recently, the measurement of large artery stiffness, as a factor indicating the development of cardiovascular complications, has become one of the most essential parameters in patients with cardiovascular diseases such as arterial hypertension [1]. It is assumed that the measurement of pulse pressure velocity can contribute useful information on the state of the cardiovascular system.

Typically, impedance plethysmography has been applied to relatively big segments, e.g. lower or upper limb, with the electrode separation assumed bigger than the radius of the segment being measured. In the current application, however, the distance between the electrodes is less than or comparable with the radius of the segment (the distance between current electrodes was equal to 4 cm), which results in significantly higher sensitivities for superficial than deep conductivity changes. With both arteries taken into account, the radial and the ulnar being superficial, this could be considered as rather advantageous effect. However, the sensitivity function of the probe becomes bipolar, especially for regions close to the surface, as seen in Fig. 4a. This effect can decrease the total sensitivity for superficially located arteries (Fig. 4b). In order to minimize the region of negative sensitivity, voltage and current electrodes were brought close together. The effect can be eliminated completely when a two-electrode technique is utilized. Unfortunately, as two-electrode techniques are known to be very susceptible to motion artefacts they cannot be used in the current application. The proposed measurement probe consists of pairs of electrodes, each containing of a voltage and current electrode positioned closely to each to other. The separation between electrode pairs instead should be as large as possible, such as to ensure that larger magnitude impedance signals are recorded.

Further studies are needed to investigate the origin of the recorded signals when utilizing electrical impedance measurements. As the pulse is travelling along the artery, the shape of the recorded signal is influenced by many factors, also by measurement procedures and equipment used. However, assuming the pressure wave velocity of 5 m/s, and a distance between the voltage electrodes equal to 6-8 cm, the time pulse propagation between electrodes is less than 8 milliseconds. So, in a first approximation this effect may be neglected, and the signal source can be assumed arising from a uniform change of volume within the artery. This uniform radial expansion of artery results from response to transmural pressure (difference between pressure inside and outsight the artery). It follows from a comparison of the maximal distance between electrodes, equal to ~8 cm, with wavelength of the dominating harmonics of pulse wave in vascular tree. In fact, instead of distance between electrodes the length of the artery participating in measurement should be considered. This length can be evaluated using the data presented in Fig. 4. As it is shown this length depends on the artery localization (depth) in the segment. In spite of this, it can be assumed that the impedance measurements reflect changes of the local artery diameter.

According to the relationships (2) and (6) pulse wave velocity is determined both, by the properties of fluid (blood) and vessels. Analysis of these relationships allows the following conclusions: for a large artery, and consequently a large value of the term $|a|r_0$, the phase velocity, c, approaches the phase velocity in an inviscid fluid. On the other hand, for very small values of $|a|r_0$ the phase velocity becomes very small. Thus, when estimating the artery's properties from the phase velocity measurements certain precautions must be taken. Moreover, some properties of artery's wall depend also on pressure. Thus, its value should be included in estimation procedure.

The recorded signals are also affected by conditions of measurement (Figs. 7-9). Position of the hand, in relation to the heart level, influences essentially shape of impedance signals, both recorded on the thorax and the wrist. However, this does not change significantly time relations between recorded signals. However, noticed a change in time delay between R wave and ΔZ_n also can be considered as a "natural" response of cardiovascular system. Note, that slopes of regression lines presented in Figs. 10 and 11 are different. It seems that further experiments should also include other measurement method allowing precise determination of ejection time from left ventricle.

The proposed measurement method generally is known as impedance plethysmography, and is sensitive to conductivity changes undergoing within the electrical field created by the current flowing between the current electrodes, as well as to the geometrical relation between the voltage and current electrodes, and the shape of the examined body segment [11].

It must be underlined that this type of measurements also is susceptible to artefacts arising from wrist movements. This was also the reason to divide the probe into two parts, each located closely to the corresponding artery. Unfortunately, muscles and tendons are also in close proximity to the arteries.

It is observed from the results that the difference between the velocity estimated from measuring the time delay between the R wave of the ECG signal and the pulse pressure at the wrist, and that estimated from the time delay between the two impedance signals, is almost 100 %. In the former method, the time of isovolumetric contraction of the chambers is included. Taking into account that this time is comparable with the time of pulse propagation along the arm, this explains largely the measurement discrepancy. This effect may be amplified in persons having vessels affected by arteriosclerosis, as such vessels demonstrate a higher pulse propagation speed. Heart insufficiency may further increase the time of isovolumetriuc contraction, and also cause larger measurement variances.

Taking into account the effect of different arm positions on the recorded signal, it seems that supplementary information obtained by means of an accelerometer would be useful.

Despite providing a more precise estimation of the pulse velocity propagation along the arteries in arm, the proposed method also has its drawbacks. As the magnitudes of the conductivity changes are very low, the very high gain required to recover a reasonable and interpretable signal, makes the measurement system susceptible to interferences and artefacts. The total gain used in the proposed system such as to obtain the presented signals, ranged from 2000 up to 5000. Additional artefacts may also arise from changes in geometry of the wrist and tissues involved in operating and moving hand.

V. CONCLUSIONS

It has been shown that measuring the propagation of pulse pressure along the arm is possible using an impedance technique. Moreover, it has been shown that the conventional method, based on the simultaneous measurement of ECG and pulse, contains a large time delay component arising from the isovolumetric ventricle contraction. However, it should be underlined that the presented impedance technique may also be susceptible to measurement artefacts as a consequence of the high amplification required to obtain appropriate signal levels that allow proper parameterization and interpretation.

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