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Quantitative Determination of Arterial Pulse Wave Velocity by Non-interfering Bioimpedance Sensing

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***Abstract*—Measurement of pulse wave velocity (PWV) in arteries is demonstrated with a novel measurement technique. Localized measurements are facilitated over arterial lengths below 10 cm on both the bicep and the wrist. Measurements do not interfere with the physiological state of the person or patient on which the measurement is performed. The scheme is based on impedance sensing utilizing novel conjugate impedances that eliminate destructive cross coupling between sets of electrodes. Initial measurements show that the test subject has a PWV of 8.0 +/- 0.1 m/s and 7.1 +/-0.2 on the wrist and bicep, respectively. The measurement distances were 8 cm and 5 cm, respectively.**

I. INTRODUCTION

We report on what to our knowledge is the first measurement of pulse wave velocity (PWV) in arteries over very short stretches (about 10 cm) and without any noticeable interference to the arterial system as well as the test subject in general.

Pulse pressure wave propagation in arteries is a well-known phenomenon. The contraction and subsequent relaxation of the heart generates a pulse, which propagates along the arterial system at a much higher velocity than the flow velocity: pulse propagation is essentially an acoustical phenomenon. The velocity depends on the properties of the cardiovascular system, which changes with the distance from the heart and with distension. It is therefore a nonlinear phenomenon. The PWV have shown to have a high diagnostic relevance to the medical community [1]. Arterial stiffness and even blood pressure may be assessed on the basis of measured PWV [2].

Current measurements are generally performed along a relatively long arterial stretch, such as from the heart to the wrist or from the heart to the thigh. Quantitative measurements on long stretches in the human body are, however, hampered by several drawbacks: The exact measurement path is cumbersome and almost impossible to estimate and it varies from person to person; the propagation velocity depends on the arterial dimensions, which generally become smaller as the distance from the heart increases, and bifurcations will cause reflections. This in addition to the nonlinear properties of the arterial response may change the pulse shape along the measurement path significantly and complicates estimation of PWV as an indicator of local arterial stiffness. This indeed implies that the long path measurements reveal an inaccurate PWV for a local body region. In addition to these facts current methods do not allow for localized diagnostics.

For these reasons and because we need a quantity related to the arterial stiffness to fulfill the ambition of performing non-invasive blood pressure measurements, we have embarked on developing methods for measuring the PWV over very short stretches (to below 10 cm) and this without any noticeable interference to the vascular system. This we are convinced that can be transformed into the blood pressure which is investigated in a later work. It is noted that essentially all currently applied blood pressure sensing devices apply a counter pressure to the vessel; however, this unarguably compromises the exactness of the estimation and is invasive to the patient's state.

Our approach for finding the local PWV is based on impedance measurements at relatively high excitation frequencies (100 kHz – several MHz) as well as a highly sensitive measurement scheme where two pairs of standard ECG electrodes are driven with different excitation frequencies. Each pair's sensing electronic is made blind to the other pair's excitation frequency. The mutual spacing of the electrodes in the direction of a selected artery is about 10 cm and each pair is placed normal to the artery's propagation direction. The spacing between electrodes in each pair should ensure penetration of the artery by field lines arising during electrical excitation.

II. PULSE PROPAGATION IN ARTERIES

Pulse wave waveforms and propagation in arteries have been investigated for decades. It is well established that the pulse pressure amplitude as well as pulse velocity increase with increasing distance from the heart until dissipation in the vascular capillaries [3]. This fact can be seen from the Moens-Korteweg equation, which in a modified form that takes Poisson's ratio into account, reads as follows:

$$v = \sqrt{Eh/2\rho r(1-\sigma^2)}, \qquad (1)$$

where E is the elastic modulus of the vessel wall, h is the wall thickness, ρ is the blood density, σ is the Poisson ratio, and r is the radius of the vessel. It is often assumed that σ = 0.5 and that the wall thickness is typical less than one tenth of the diameter. Both E and r change with increasing distance from the heart, but the dominating effect appears to be the decrease in radius.

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The pulse velocity can be expressed by the Bramwell-Hill equation assuming that the longitudinal expansion of the vessel is negligible

$$v = \sqrt{\frac{1}{2}\frac{\partial P}{\partial r}\frac{r}{\rho}}. \quad (2)$$

The pulse velocity depends on the radius of the artery as can be seen from (1). For relative distensions much smaller than unity the actual radius can be substituted by the time averaged radius at the location of the measurement. For larger variations nonlinear effects must be considered. The effect is a pulse broadening because the velocity decreases with increasing radius. This effect is contrary to the effect that the tapering of the arteries implies a pulse sharpening.

The pulse shape depends on both heart and arteries. Characteristic pulse shapes are shown in Fig. 1 recorded at the lower forearm with custom made electronics (measurements have also been performed with a commercial impedance analyzer, Hioki IM 3570). The pulse becomes narrower with increasing distance from the heart and the shape is additionally affected by the specifics of the cardiac cycle as well as by reflections primarily caused by arterial bifurcations but also by changes in arterial stiffness. For accurate velocity measurements it is important to select the part of the pulse that represents the forward moving pressure pulse. We have applied the following expression for the pulse at the lower forearm:

$$d(t) = a\left[\sin(2\pi t/t_1)\exp\left(-(t/t_2)^2\right)+(1-t)\left(1-\exp(-t/t_3)\right)\right]\times \left[\text{unitstep}(t)-\text{unitstep}(t-1)\right], \quad (3)$$

which provides for a good representation of measured pulses at (Fig. 1). Typical pulse velocities are in the 5 – 10 m/s region [4], which implies that the spatial extension of a pulse is a little less than one meter, which again implies that the relative displacements of the two sets of curves will be much smaller than the width of the pulses.

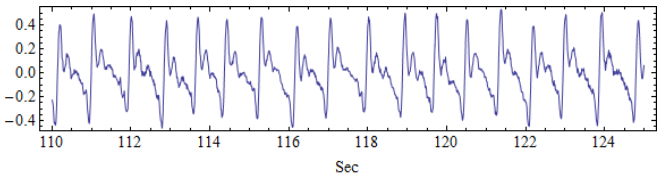

Fig. 1. A measured sequence of pulses (abs. impedance variations in ohms) from the upper arm of a healthy 29 year old male. The excitation frequency was 1 MHz.

III. Impedance Sensing

The conductivities and permittivities of biological tissues depend on the specific type of tissue and on the excitation frequency. Furthermore, tissues are often anisotropic, e.g. muscles conductivities in the direction of the fibers may be different from conductivities perpendicular to the fibers [5]. The response may be nonlinear, depending both on bias voltage and excitation amplitude.

Diagnostics based on impedance sensing has the advantage of being able to be performed without affecting the state of the object. However, applications are often hampered by poor resolution and complicated responses [6].

The electric field distribution in an organ subjected to electrodes and excited at a specified frequency can be evaluated from the Poisson Equation using a finite element method. A simpler equivalent circuit can be established based on combinations of resistors and capacitors. In the extreme of having a very large number of components results must be equivalent to finite element calculations. We have used both approaches to verify our method. The distribution of field lines will roughly expand to a width given by the diameter of the tissues with a high conductivity (predominantly muscles). This is true if it is assumed that the electrodes have dimensions much smaller than the characteristic organ dimension. Fat layers right below the skin contributes very little to the spreading of field lines but have a significant impact on the magnitude of the total impedance. Arteries embedded in fat are more difficult to observe than arteries embedded in muscles.

Our first approach was to use impedance sensing with two sets of electrodes. Each set consists of two electrodes placed normal to an artery's propagation direction and with a separation distance that ensures the field lines' penetration of the artery of interest. The electrode sets themselves are separated along the artery in a well-defined distance. The brachial artery is here a preferred choice because of its straightness and accessibility; however, initial tests have shown that almost any straight artery can be measured on even in a finger.

The electronic scheme for the first version in this study was based on an auto balancing impedance bridge having a response time longer than the characteristic temporal distance between pulses. The short term bridge imbalance is used as a measure of the arterial distension. This is illustrated in Fig. 2. A disadvantage of the bridge is that with impedances where both the real and imaginary parts can vary considerably balancing may be very delicate and undesirable phase shifts may be encountered. Alternatively an impedance detector based on a simple op-amp with known feedback impedance and the unknown impedance as the input impedance may be applied (also known as *auto balancing* [7]). The constraint here, however, may be the dynamic range or noise of the op-amp.

A major problem in using two closely spaced sets of electrodes is that one set affects the impedance observed by the other set. Without special precautions for electrically isolating the two sets simultaneous measurements of distension and thus time lag are essentially impossible. The problem is illustrated in Fig. 3. We note that even a complete galvanic separation does not solve the problem.

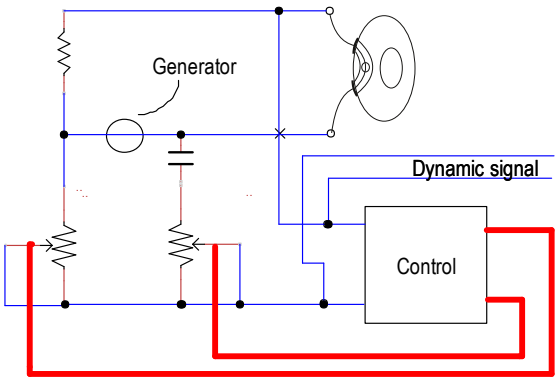

Fig. 2. The measuring bridge with complex impedances. C*ontrol* ensures a bridge balance on a time scale larger than the pulse spacing. The dynamic signal is obtained as the short term unbalance of the bridge.

Our solution implies two essential measures: (1) Two slightly different excitation frequencies are applied, one for each set of electrode, in conjunction with coherent quadrature detection, and (2) the use of dedicated conjugate impedances that essentially makes one set of electrodes invisible to the other set and vice versa.

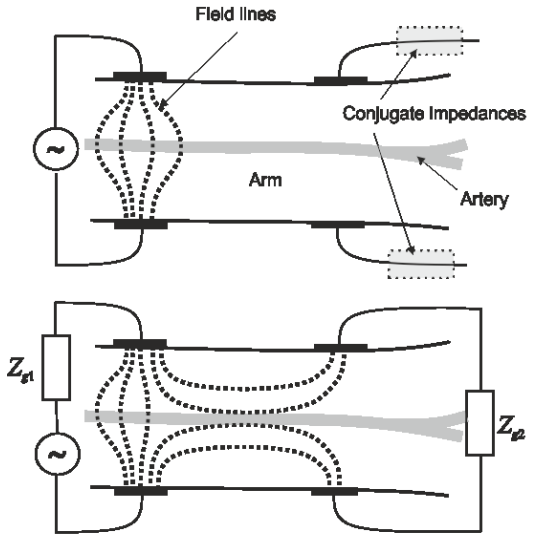

Fig. 3. Field lines with two sets of electrodes. (Top): The first set of electrodes is driven by a low impedance generator; the second set is connected with infinitely large load impedance (conjugate impedances). The field lines of the first electrode set are hereby unaffected by the second set. (Bottom): The second set of electrodes is here connected to a low impedance load, which drastically alters the spatial distribution of field lines. (Generator 2 and the associated field lines are not shown.) Inserting conjugate impedances will eliminate the effect of the load Z_{g2} and vice versa (not shown for the left electrode set).

The absolute transfer responses for the conjugate impedances are illustrated in Fig. 4 where the notch of the first impedance corresponds to the peak of the second impedance and vice versa. We note that filter adjustments are very delicate since just a small difference in phase delay can have a significant impact on the delay estimation.

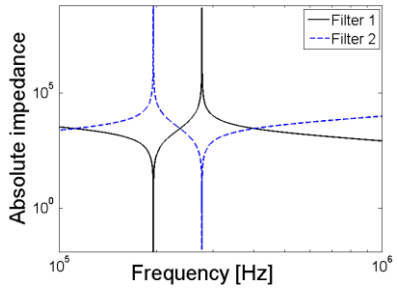

Fig. 4. Conjugate impedances.

A subsequent solution is based on six electrodes: one set for excitation by current injection and two sets for sensing. This method turned out to be much more robust than the first method. The effective path length is equal to half the spacing between the two sets of sensing electrodes if the electrodes of each set has the same separation perpendicular to the artery and a local isotropy in the direction of the artery can be assumed.

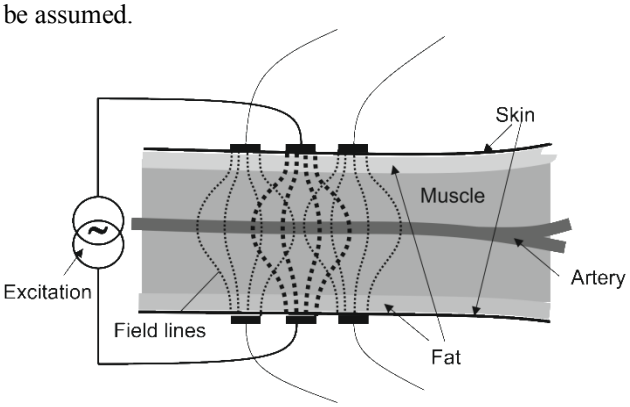

Fig. 5. A six electrode configuration. The middle set is for excitation. The two outer sets are for detection.

IV. Delay Estimation

The delay estimator must be devised considering that the primary unwanted signal perturbations are not caused by broad-band thermal noise but by body movements uncorrelated with the heartbeat. These perturbations can have roughly the same bandwidth as the desired signal, but have neither the same shape nor the same quasi-repetitive nature as pulses caused by heartbeats. The perturbations are typically not statistically stationary over the timescales considered. Estimation may be performed by direct timing of validated pulses from each electrode set. The steep pulse front provide the best part of the pulse for temporal location if broad band noise is negligible and the front does also appear to provide the most well-defined part of the propagating pulse [8]. Our approach is to select sequences of at about ten pulses where quasi-repetitive signals are identified, calculate the cross-correlation function, and perform a parabolic fit to the peak. This has been performed using the *Mathematica*™ program environment.

V. Results

Measurements of pulse velocity have been performed on the lower forearm and on the biceps (where cuffs for blood pressure measurements typically are placed). An example of pulses taken at the middle upper arm is shown in Fig. 1 and a cross correlation of pulses from two displaced detection electrodes (Fig. 5) is shown in Fig. 6.

Results can be summarized as given in Table 1. The *Distance* is the spacing between the two sets of electrodes. The uncertainty is evaluated from the variance of velocity estimates obtained from independent sets of data.

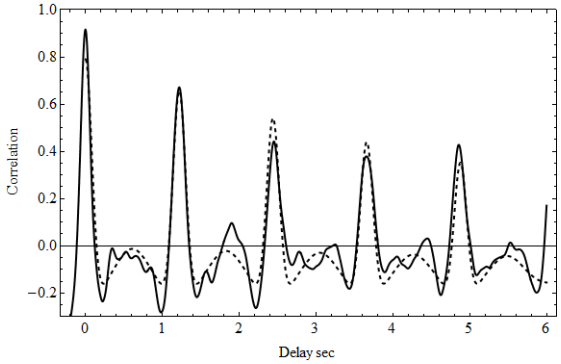

Fig. 6. Cross correlation of detector signals obtained at the upper arm using the configuration of Fig. 5 and a fitted reference signal (dashed). Electrodes were 2 cm x 2 cm with a spacing of 5 cm in the direction of the artery. Excitation frequency: 200 kHz.

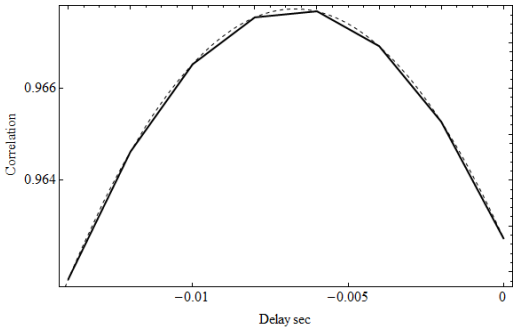

Fig. 7. Normalized cross correlation as in Fig. 6 except for a much smaller delay range, which reveals the displacement of the peak. The dashed curve is a parabolic fit.

Table 1. Pulse velocities.

Location	Pulse velocity m/s	Uncertainty m/s	Distance m
Lower forearm, radialis, Fig. 3	8.0	± 0.1	0.08
Upper arm, brachalis, Fig. 5	7.1	± 0.2	0.05

VI. Discussion

Clearly an increase in PWV from biceps to lower forearm is identified in Table 1. This complies with the literature and we are therefore convinced that the results are representative for the actual pulse propagation velocities. Uncertainties in the measurements may be caused by an uncertainty of the distance between electrodes, which can vary if the electrode sets are not placed perpendicular to the artery. Disturbances from neighboring arteries may skew the measured pulse and introduce misleading pulse flanks and external disturbances caused by body movements are always present. A large dataset and selective filtering and validation are therefore mandatory in order to attain the desired accuracy. The conjugate impedances must be matched very closely; otherwise, they may introduce unacceptable large phase shift, which can be damaging to the estimation.

Validation with a complementary method is desirable. A specially implemented high-frequency ultrasound system appears to be the only possibility for direct validation. An indirect validation has been performed by measuring the blood pressure, the vascular dimension from an MR image, and then calculating the pulse velocity from Eq. (2) with a good agreement. However, it is estimated that the direct measurement of pulse velocity implies a smaller uncertainty than the indirect method.

VII. Conclusion and Outlook

Localized measurement of PWV without interfering with the physiological state of the test person has been demonstrated to our knowledge for the first time. Two methods have been applied: one with two sets of electrodes and conjugate impedances, and one with two sets of electrodes for sensing and one set of electrodes for current injection. The latter method is the more robust of the two.

The possibility of localized measurements may facilitate improved diagnostics for arterial deficiencies. It may also facilitate noninterfering measurement of calibrated blood pressure.

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