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Implantable Acceleration Plethysmography for Blood Pressure Determination

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***Abstract*— This paper presents an implantable accelerometer which detects plethysmograms directly at an artery. The sensor provides a new method for continuous blood pressure monitoring. *In vivo* measurements indicate that the accelerometer is well suited for determining the Pulse Transit Time (PTT) and the Reflected Wave Transit Time (RWTT). Both parameters show a high correlation with the systolic blood pressure. By varying the blood pressure, it was seen that RWTT more closely agrees with theory than PTT. Through several blood pressure sweeps the RWTT, as detected by the accelerometer, coincided very well with the systolic blood pressure, with a correlation coefficient of 0.96 and mean deviation of 4.3 % for 1800 pulses.**

I. Introduction

In a world with an ageing population, long-term monitoring of cardiovascular parameters for optimized therapy becomes increasingly important. A monitoring device needs to have a low risk and should not limit the patients in their day-to-day life. Standard methods for blood pressure monitoring are arm- or wrist-cuff systems, which measure the pressure by inflation; this is uncomfortable for continuous monitoring over several days.

In response to this issue, pulse wave velocity (PWV), which is strongly correlated with blood pressure, has become of considerable medical interest. The most common method for PWV calculation is to use an electrocardiogram (ECG) and a photoplethysmogram (PPG) to determine the pulse transit time (PTT) [1], from which blood pressure may be determined. We present here a new approach using a sensor to directly detect the arterial acceleration. The sensor is implanted, but without vessel invasion, and therefore with low risk for the patient. The measured signal is the acceleration plethysmogram (APG) which allows more accurate detection of the PTT than the PPG [2]. Additionally, the sensor is able to detect the reflected wave transit time which also correlates with the PWV and the blood pressure. The current consumption of this new sensor is at least one magnitude smaller than that of standard PPG sensors.

II. Theory

A. Acceleration Plethysmography

Each heart contraction leads to a pressure change in the blood vessels. Blood pressure is coupled to vessel extension by the properties of the vessel wall. Due to the viscoelastic properties of large arteries, the relationship is non-linear and shows a hysteresis. Therefore, it is not possible to directly calculate blood pressure by simply measuring the vessel expansion. Despite this hysteresis, the blood pressure and vessel expansion curve have a similar appearance [3].

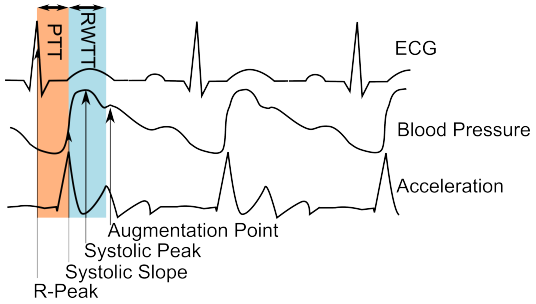

Fig. 1. *Pulse transit time (PTT) is the time delay between the R-Peak in the ECG signal and the systolic slope in the pressure signal, which coincides with the maximum in the acceleration. The reflected wave transit time (RWTT) is the delay between systolic slope and arrival of the reflected wave at the augmentation point.*

The measurement of vessel expansion over time is called plethysmogram. The most common method for its measurement is optically, using a pulse oximeter, which is therefore referred to as a photoplethysmogram (PPG). Since the cardiac signals are periodic, the PPG can be described as a Fourier series, consisting of the heart frequency and its overtones.

For further signal processing, many groups use the second derivative of this signal, which is the acceleration plethysmogram (APG)[1] [2], [4]. The plethysmogram and its second derivative can be expressed as

$$|s(t)| = A_0 + \sum_{k=1}^{\infty} A_k \cdot \sin(k \cdot \omega_0 \cdot t + \varphi_k) \quad (1)$$

$$|a(t)| = |\ddot{s}(t)| = \sum_{k=1}^{\infty} k^2 \omega_0^2 \cdot A_k \cdot \sin(k \cdot \omega_0 \cdot t + \varphi_k) \quad , \quad (2)$$

where s is the vessel extension, a the vessel acceleration, k the overtone-number, A_k the amplitude, ω_0 the heart rate and φ_k the phase.

The two signals differ by the factor $k^2\omega_0^2$ which is the square of the overtone frequency. Thus, the APG power spectrum is amplified at higher frequencies, which allows a better detection of fast signal changes.

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[1]This is sometimes also referred to as Second Derivative Photoplethysmogram (SDPPG)

B. Blood Pressure Determination

It has been shown that the pulse wave velocity is strongly dependent on the arterial blood pressure [3], [5]. Increased arterial stiffness, as well as high blood pressure, lead to faster PWV. As PWV is not directly measurable, the pulse transit time (PTT) is determined. PTT is the time the pulse wave travels from the heart to the sensor position, which is inversely proportional to PWV. We present here an approach to detect PTT by cross-correlating an ECG signal with the APG signal measured at the artery. This approach also includes the pre-ejection period (PEP), which is the time between the ECG peak and the heart ejection, and which also varies with blood pressure.[2]

Pulse waves travel from the heart to the peripheral arteries. Within the vascular system, reflections occur, causing part of the pulse wave to travel backward, again with the pulse wave velocity. The reflected wave transit time (RWTT) is the temporal difference between incident and reflected wave at the same position. RWTT was measured by only one acceleration sensor and also used as a blood pressure estimator, since it is inversely proportional to PWV. PTT as well as to RWTT are illustrated in Fig. 1.

PWV can be calculated by the Moens-Korteweg equation.[3] Numerous measurements have shown that Young's modulus E of large arteries increases almost linearly with blood pressure (BP) within the physiologically relevant pressure range [6]. Assuming a linear relationship with slope $m_E = \frac{\Delta E}{\Delta BP}$ and offset E_0, the Moens-Korteweg equation can be modified to

$$BP = \frac{2\rho \cdot a_i}{m_E \cdot h} \cdot PWV^2 - \frac{E_0}{m_E} := m_{BP} \cdot PWV^2 + c_{BP}, \quad (3)$$

with constant length s of the arterial path. Supposed that the blood density ρ as well as the ratio of the inner vessel diameter a_i and the vessel wall thickness h are constant, the equation can be simplified as shown on the right, by introducing an empirical slope m_{BP} and offset c_{BP}. Both parameters depend on the individual characteristics of the patient and require calibration measurements.

The aortic pulse wave velocity of a healthy rabbit at rest is between 4 and 8 m/s [7].

C. Sensor Concept

To detect the pulse wave, an Analog Devices ADXL327 accelerometer was mounted directly on a blood vessel. The MEMS sensor with dimensions of $4 \times 4\,\mathrm{mm}^2$ draws only 350 µA. Silicone strips are used to mount two accelerometers on opposite sides of one artery as shown in Fig. 2. It has been shown that the utilized highly elastic silicone rings are stable over long-term periods and do not cause any vessel constriction [8].

Using two sensors makes it possible to cancel artifacts from the pulse signal caused by the patient's movement. The acceleration, seen by each sensor, can be described as $\vec{a}_{\text{total}} = \vec{a}_{\text{vessel}} + \vec{a}_{\text{body-movement}}$.

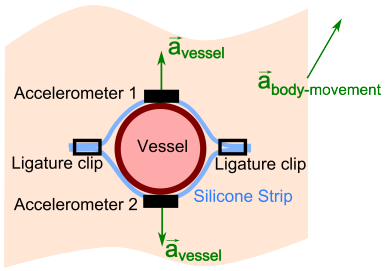

Fig. 2. *Two accelerometers positioned diametrically opposed on an artery. Highly elastic silicone strips and ligature clips are used for perivascular mounting.*

Since the sensors are placed on opposite sides of the artery, the vectors $\vec{a}_{\text{vessel}}$ point in opposite directions, whereas the movement artifact $\vec{a}_{\text{body-movement}}$ is unchanged. The difference of the two sensor signals equals $2 \cdot \vec{a}_{\text{vessel}}$ and is independent of movement artifacts.

III. In Vivo Measurements

A. Measurement Description

The sensor was evaluated *in vivo* inside an 8 month old male New Zealand White rabbit under narcosis with body weight of 3.5 kg. The experiments were approved by the local ethics committee and performed in compliance with the rules and regulations of the German animal protection law.

Blood pressure variations were induced by intravenous injection of dopamine (2.5 mg), which results in a quick rise followed by a slow decrease of the arterial blood pressure.

The reference pressure was measured by an intra-arterial tip-catheter (Millar Instruments). ECG was acquired by extracorporeal electrodes on the left and right side of the thorax.

The accelerometer was placed around the aorta using elastic silicone strips, as shown in Fig. 3. The rabbit aorta has a diameter of approx. 6 mm, which is comparable to the diameter of the carotid or femoral artery of a human. After preparing the vessel, the sensor was mounted in less than five minutes, without any complications.

Fig. 4 illustrates the position of the accelerometer about 30 cm from the heart and 5 cm from the aortic bifurcation. The intra-arterial reference sensor was placed in the descending aorta, 5 cm proximal. The aortic bifurcation is the point where the descending aorta splits into the left and right iliac arteries and acts as the main source of pulse wave reflection [7].

By injection of dopamine, the systolic blood pressure was repeatedly varied in the range between 80 and 160 mmHg. ECG was measured extracorporeally for PTT calculation.

B. Acceleration Plethysmogram

Figure 5 shows the acceleration plethysmogram, the blood pressure and its second derivative. The APG signal displays a clear peak at the systolic rise of the blood pressure. The shape is similar, but not identical to the second derivative of the blood pressure. The deviation can be explained by the

[2]This is sometimes also referred to as pulse arrival time (PAT). PAT = PTT + PEP .

[3]The Moens-Korteweg equation models the relationship between PWV and Young's modulus of arteries. It is often referred to in biomechanics.[3]

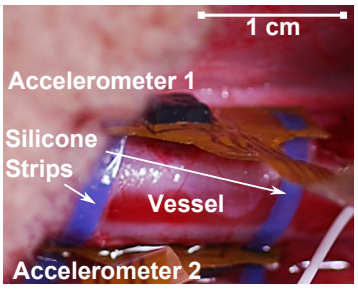

Fig. 3. *Acceleration sensor mounted at the aorta of a rabbit.*

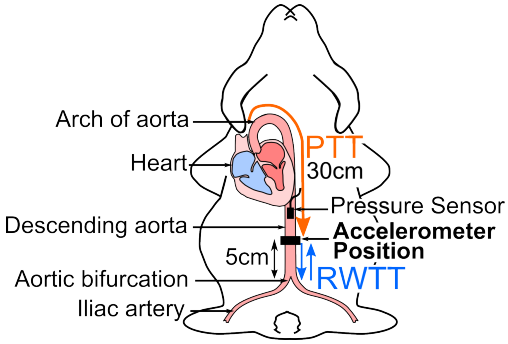

Fig. 4. *Acceleration sensor position inside the rabbit.*

viscoelastic properties of the aorta and the distance between the sensor positions.

By taking the derivative of a signal, higher frequencies are amplified as seen in Equation 2. Since the acceleration is the 2nd derivative of the vessel expansion, caused by blood pressure change, the power spectral density (PSD) of the APG shown in Fig. 6 displays signal components at higher frequencies than the blood pressure. The existence of higher frequency components allows a more precise temporal detection of significant points, such as the systolic rise or the first pressure wave reflection.

C. Blood Pressure Estimation

The blood pressure was varied three times by drug injection. The temporal changes can be seen in Fig. 7.

For all 1800 pulses, PTT, RWTT and intra-arterial systolic blood pressure were detected. Fig. 8 shows one measured ECG and APG pulse as an example. The PTT is the delay between the maxima in ECG and APG, determined by the cross-correlation of both signals. The RWTT is determined only from the APG-signal as the delay between the first and second peak.

Fig. 9 shows the relation between respiration-filtered transit times and blood pressure. Eq. 3 holds with a correlation coefficient of 0.94 for PTT and 0.96 for RWTT. The standard (mean) deviation from the theoretic linear function is 9.4 % (7.5 %) for PTT and 5.6 % (4.3 %) for RWTT.

A quadratic fit is more accurate for the PTT curve with a correlation of $r = 0.96$ and a standard and mean deviation of 4.0 % and 3.0 %, respectively. This can be explained by the pre-ejection period (PEP) of the heart, which strongly

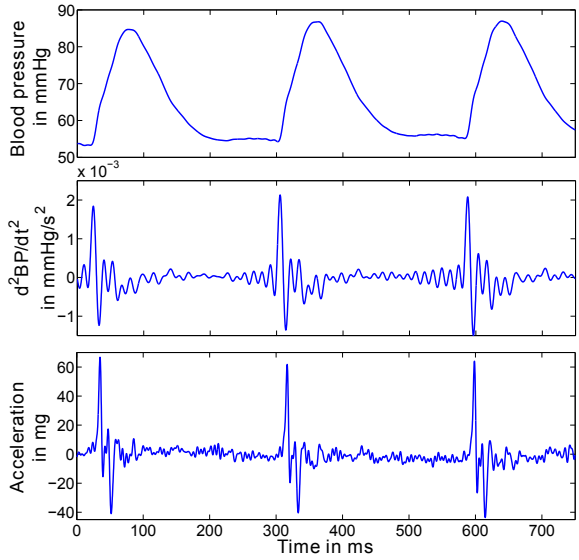

Fig. 5. *In vivo signals over 0.75 s: Intra-arterial blood pressure, its second derivative and APG, i.e. the measured acceleration in radial direction.*

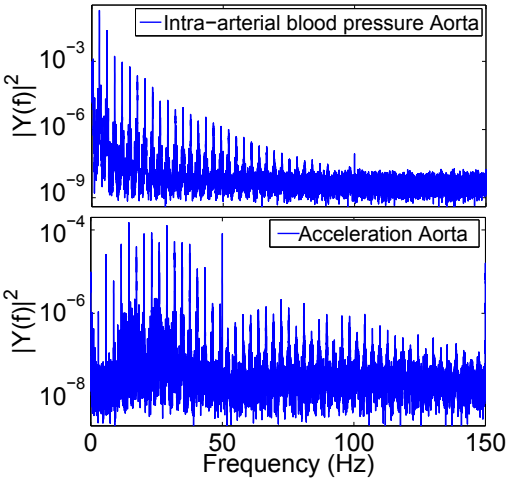

Fig. 6. *PSD of the blood pressure, measured with a tip-catheter inside the aorta (upper plot), and PSD of the acceleration signal from the sensor outside the aorta (lower plot).*

influences PTT over the short distance but does not affect RWTT.

Both curves – PTT more clearly than RWTT – show a hysteresis. The fast blood pressure increase, due to the injection of dopamine, leads to higher values than the slow blood pressure decrease.

D. Discussion

The APG is a fast signal that allows a highly reliable detection of the systolic slope as well as of the wave reflection. Table I summarizes the correlation between the blood pressure and the two transit times. Using the cross-correlation between ECG and APG, the detection of pulse transit time turns out to be very reproducible. Since the PTT is influenced by the PEP, the theory from Eq. 3 does not hold exactly. RWTT is not affected by PEP and therefore better

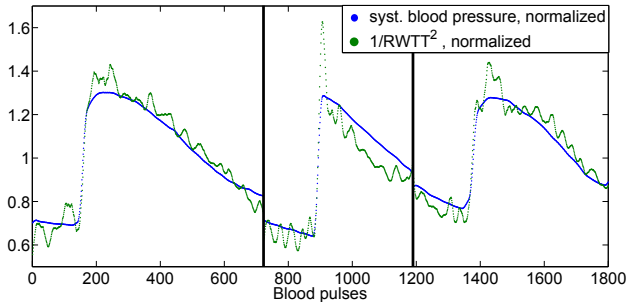

Fig. 7. *The blood pressure was varied three times by injection of a dopamine bolus. Each bolus causes a quick pressure rise from ca. 80 to 150 mmHg. Each measurement takes 2 – 4 minutes. The heart rate is 170/min. The quantity 1/RWTT^2 clearly follows the reference pressure.*

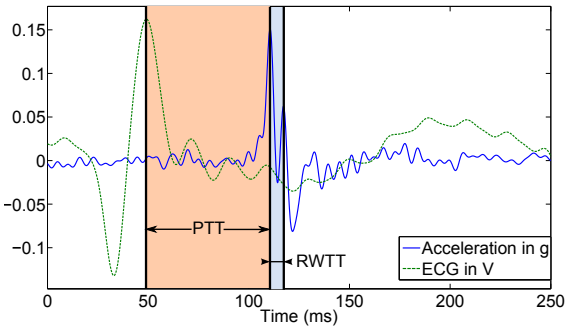

Fig. 8. *Measured ECG and APG. The vertical lines define the PTT and RWTT.*

agrees with theory, making it well suited to estimate systolic blood pressure. Another benefit of using RWTT is that only one sensor is necessary, and that no further ECG electrodes are needed. The fixation of the sensor around the rabbit aorta was unproblematic. For use in humans, the sensor could be applied to an artery of similar size as the carotid or femoral artery.

TABLE I
CORRELATION OF BLOOD PRESSURE WITH PTT AND RWTT

parameter compared to blood pressure	correlation coeff.	standard deviation	mean abs. deviation
1/RWTT^2	0.96	5.6 %	4.3 %
1/PTT^2	0.94	9.4 %	7.5 %
1/PTT	0.96	4.0 %	3.0 %

IV. CONCLUSION

It has been shown that an acceleration plethysmogram can be measured directly at the blood vessel with high quality. Pulse transit time and reflected wave transit time detected by the acceleration sensor show a very high correlation to the systolic blood pressure, making this sensor a promising, new, implantable, low-power approach for continuous long-term monitoring of blood pressure. Blood pressure from reflected wave transit time agrees better with theory and simplifies the approach such that only one sensor has to be employed.

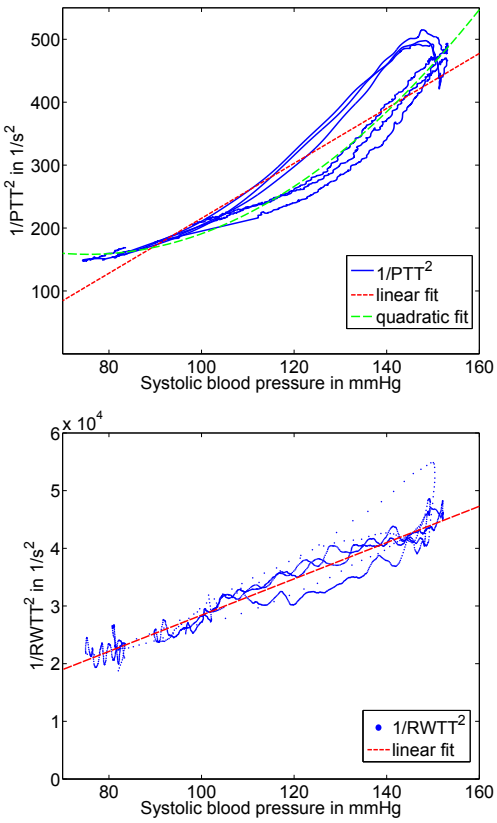

Fig. 9. *Inverse squared transit times as a function of the systolic blood pressure for 1800 pulses measured at the aorta.*
Upper plot: pulse transit time from the interval between the R-peak of the ECG signal and the peak of the acceleration plethysmogram.
Lower plot: reflected wave transit time.

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