Sensors and Actuators A 206 (2014) 151–158

Contents lists available at ScienceDirect

Sensors and Actuators A: Physical

journal homepage: www.elsevier.com/locate/sna

Implantable accelerometer system for the determination of blood pressure using reflected wave transit time☆

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ARTICLE INFO

Article history:
Received 20 September 2013
Received in revised form 2 December 2013
Accepted 3 December 2013
Available online 14 December 2013

Keywords:
Accelerometer
Blood pressure measurement
Pulse wave velocity
Reflected wave transit time
Medical implant

ABSTRACT

A sensor system for continuous monitoring of blood pressure using an acceleration sensor implanted on an artery using minimally invasive techniques is described. The sensor relies on measurement of the reflected wave transit time (RWTT). This implantable system is fabricated on a flexible substrate using 2 mm × 2 mm acceleration sensors and a telemetric unit for transmission of the data. *In vivo* experiments show that the RWTT can be reliably determined from arterial acceleration signals. RWTT and systolic blood pressure are shown to be strongly coupled, with a correlation coefficient of 0.96, as determined from measurement of 1800 pulses with a mean deviation of the blood pressure of only 4.3%. The system was implanted in an animal and was able to telemetrically transmit acceleration plethysmographs with high quality out of the awake animal.

1. Introduction

Even though blood pressure is one of the most important physiological parameters, there is at present no means available for continuously monitoring this quantity without limiting a patient's mobility. Continuous long-term monitoring of blood pressure over longer periods, such as a day, would provide considerably more physiological insights than isolated clinical measurements [1], but the means to do so are presently unsatisfactory.

Several promising approaches for continuous blood pressure monitoring have been presented, most notably those using arterial pulse wave velocity (PWV), determined by measuring the pulse transit time between an electrocardiogram and a photoplethysmogram [2]. Pulse wave velocity and blood pressure have been shown to be clearly correlated [3,4] with a stable long-term relationship [5]. However, a disadvantage of measuring pulse transit time is that two sensors at a certain spacing are necessary to determine PWV.

Since pulse waves are subject to reflection at bifurcations in the circulatory system, it is possible to detect both the incident and the reflected pulse wave, using a single sensor. The temporal delay between the two – the so-called reflected wave transit time (RWTT) – has been proposed as indicator of blood pressure. However, this measurement approach has not been pursued, since the detection of RWTT using extracorporeal devices has shown poor reproducibility. [6]

We present here an implantable system which uses an accelerometer to reliably determine RWTT and to estimate the blood pressure. Since only one measurement position is needed, the implant size can be minimized and thus make it possible to monitor a patient's vital signs without impairing mobility.

2. Theory

2.1. Reflected wave transit time

Each heart contraction generates a pressure wave, which passes through the arteries at the pulse wave velocity (PWV). PWV is not directly measurable, but the time delay of the traveling pulse wave between two points of the cardiovascular system may be detected. This measurement can also be realized using only one sensor which detects both the incident pulse wave and the same wave reflected at the end of the arterial segment. Reflections are caused at several parts of the arterial system [7,8].

The main reflections occur at the aortic bifurcation, where the descending aorta splits into the left and right iliac arteries [9]. Due to its smaller amplitude, it is difficult to exactly determine the reflected wave in a standard plethysmogram.

☆ Selected Paper based on the paper presented at The 17th International Conference on Solid-State Sensors, Actuators and Microsystems, June 16–20, 2013, Barcelona. Spain

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http://dx.doi.org/10.1016/j.sna.2013.12.006

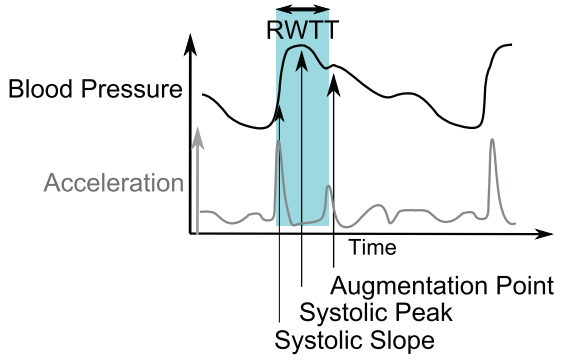

Fig. 1. The acceleration signal is the second derivative of the vessel expansion caused by the blood pressure pulses. Reflected wave transit time (RWTT) corresponds to the delay between systolic slope and arrival of the reflected wave at the augmentation point.

2.2. Arterial acceleration

Acceleration is the second derivative of blood vessel expansion. The corresponding signal is called acceleration plethysmogram and has been acquired indirectly by differentiating a photoplethysmogram two times. These acceleration plethysmograms have proven to be useful for pulse contour analysis, since rapid changes in the signal can be detected more reliably [10,11].

The maximum acceleration within each pulse is the largest change in slope of the arterial expansion which occurs at the beginning of the systole. The next local maximum in the acceleration signal is caused by the reflected pulse wave. These two maxima yield RWTT, depicted in Fig. 1, which is inversely proportional to PWV [12].

By using this indirect method of determining the acceleration, it is necessary that the measured plethysmogram is clearly above the noise threshold, since noise is amplified when the signal is differentiated two times. Hence, direct access to the arterial acceleration will definitively yield an improved signal-to-noise ratio and a more reliable detection of the significant points needed for RWTT determination. The arterial expansion and its second derivative can be expressed as

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

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

where s is the vessel extension, a is the vessel acceleration, k is the overtone-number, A_k is the amplitude, ω_0 is the heart rate and φ_k is the phase. $s(t)$ and $a(t)$ differ by the square of the overtone frequency $k^2\omega_0^2$. Thus, the power spectrum of the acceleration signal is amplified at higher frequencies and improves the detection of fast signal changes [4].

2.3. Blood pressure determination

It has been shown that arterial PWV strongly depends on the arterial blood pressure according to the Moens–Korteweg equation [13,14]. Numerous measurements have shown that Young's modulus E of large arteries increases almost linearly with blood pressure (BP) [15].

Assuming this linear correlation with slope $m_E = \Delta E/\Delta \mathrm{BP}$ and offset E_0, the Moens–Korteweg equation is rearranged to yield

$$BP = \frac{2\rho \cdot a_i}{m_E \cdot h} \cdot \frac{s^2}{RWTT^2} - \frac{E_0}{m_E} := m_{BP} \cdot \frac{1}{RWTT^2} + c_{\mathrm{BP}}, \quad (3)$$

with constant arterial path s between incident and reflected wave.

The blood density ρ is mainly dependent on the patient's hematocrit level, which varies little over a period of weeks. Inner blood vessel diameter a_i and vessel wall thickness h are dependent on the blood pressure. However, *ex vivo* measurements have shown that a_i of a young pig's carotid artery increases by only 0.06%/mmHg within the physiologically relevant range. This causes a decrease of h by −0.03%/mmHg. For the blood pressure estimation using *RWTT*, the simplification that the ratio of a_i and h remain constant leads to a calculated mean error of less than 1% over the entire physiological blood pressure range.

Assuming that the blood density ρ as well as the ratio of a_i and h are constant, Eq. (3) simplifies to that on the right-hand side by introducing the empirical terms corresponding to slope m_{BP} and offset c_{BP}. The system must be initially calibrated individually and after a time recalibrated to determine m_{BP} and c_{BP} for each patient [3].

3. Sensor

3.1. Measurement principle

The arterial movement is acquired by acceleration sensors which are sensitive to all movements including body motion. Therefore, the system consists of at least two sensors to operate in a differential mode. Fig. 2 shows two sensor configurations and the most important acceleration vectors. The diametral concept (Fig. 2A) features two accelerometers on opposite sides of an artery. They are connected by a highly elastic perivascular silicone strip. Both sensors see all body accelerations $\rightarrow a_{\text{body-motion}}$ and the arterial acceleration $\rightarrow a_{\text{artery}}$, but the latter one in opposite direction. Therefore, the difference signal

$$\begin{aligned} \rightarrow a_{\text{total}} &= \rightarrow a_1 - \rightarrow a_2 = (\rightarrow a_{\text{artery}} + \rightarrow a_{\text{body-motion}}) \\ &\quad - (-\rightarrow a_{\text{artery}} + \rightarrow a_{\text{body-motion}}) = 2 \cdot \rightarrow a_{\text{artery}} \end{aligned}$$

is not dependent on body motion. In a parallel configuration (Fig. 2, B), no direct contact with the artery is required. Both sensors are placed subcutaneously on a thin layer of connective tissue, joined by a flexible polyimide (PI) substrate. Only one of the sensors is placed directly above the artery, thereby facing $\rightarrow a_{\text{artery}}$. By using the differential signal

$$\begin{aligned} \rightarrow a_{\text{total}} &= \rightarrow a_1 - \rightarrow a_2 = (\rightarrow a_{\text{artery}} + \rightarrow a_{\text{body-motion}}) \\ &\quad - (\rightarrow a_{\text{body-motion}}) = \rightarrow a_{\text{artery}} \end{aligned}$$

we again eliminate the movement artifacts.

3.2. Implantable system

For continuous wireless monitoring, the autonomous sensor system, shown in Fig. 3 has been built. The block diagram in Fig. 4 symbolizes the configuration of the circuit. Three digital accelerometers (Bosch BMA280) are mounted on flexible PI foils. They connect the sensors to an RF microcontroller (Texas Instruments CC430), a balun network and a chip antenna (Yageo CAN4311). The complete system is powered by a battery. Sensor data is acquired, using a serial peripheral interface (SPI), and transcutaneously transmitted in the Medical Implant Communication Service (MICS) frequency band [16] at 403 MHz. A