

A System for In-Ear Pulse Wave Measurements

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Abstract—The measurement of the pulse wave has proven to be a vital tool in medical diagnosis. Whereby most pulse wave measurements are carried out at extremities, this work proposes a system for measuring the pulse wave and the Pulse Arrival Time (PAT) in the interior of the ear. The developed measurement device is based on a battery powered microcontroller system. The measurement device can simultaneously acquire a single channel Electrocardiogram (ECG), a dual wavelengths Photoplethysmogram (PPG), the pressure in both ears, the body core temperature, as well the subjects motion. The acquired measurement data can either be saved on a micro SD-card or can be transmitted wireless via Bluetooth or wired via USB to a host PC for further analysis. In Bluetooth communication mode the device can operate battery powered up to 8 hours. In addition to the system description, first measurements carried out with the system will be presented.

I. INTRODUCTION

A key to future personal monitoring is portability and adaptability. For that purpose it is desirable to record vital signs non-invasively and comfortably for a long period of time.

Pulse Arrival Time (PAT) and the morphology of the pulse wave are considered as indicators of arterial stiffness and are also known as (prognostic) markers for cardiac and vascular diseases [6], [7]. In the past various methods and sites have been demonstrated for pulse wave measurements, mostly on carotid and femoral artery through invasive and non-invasive methods.

This work proposes a system for detecting the pulse wave inside the auditory canal for PAT and Pulse Wave Velocity (PWV) measurements, as proposed by [3]. The developed measurement device is based on a battery powered microcontroller system and can simultaneously acquire a single channel Electrocardiogram (ECG), a dual wavelengths Photoplethysmogram (PPG), the pressure in both ears, the body core temperature, as well the subjects motion. After data acquisition the

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measurement data can be transmitted wireless via Bluetooth or wired via USB to a host PC for further analysis. Additionally it is also possible to save the measurement data on a micro SD-card.

II. MEASUREMENT METHODS

A. Electrocardiogram (ECG)

Via the recording of the Electrocardiogram (ECG) it is possible to detect the electrical activity of the heart, which is closely correlated with the contractions of the heart atria and ventricles and can therefore provide a reliable time reference for the PPG and pressure measurements.

B. Photoplethysmography (PPG), Pulse Arrival Time (PAT) and Pulse Wave Velocity (PWV)

Caused by the rhythmic contraction of the heart, blood is pumped in a pulsation through the arterial system. The blood pulsation is time and site depended and can be measured in terms of flow, pressure and volume. Via the PPG this volume changes can be measured non-invasively by exploiting the different optical properties of oxygenated and non-oxygenated blood. The morphology of the obtained waveform, as well as the PAT are very significant and direct prognostic markers for the stiffness of the arteries. Hence providing information regarding the overall condition of the cardiovascular system. PAT values can also be used for the calculation of the PWV, which is an additional important marker for arterial stiffness [4], [6].

The PAT is defined as the time span between the R-peak of the ECG and the arrival of the pulse wave in the PPG signal. The PAT is commonly defined in one of three possibility according to Fig. 1.

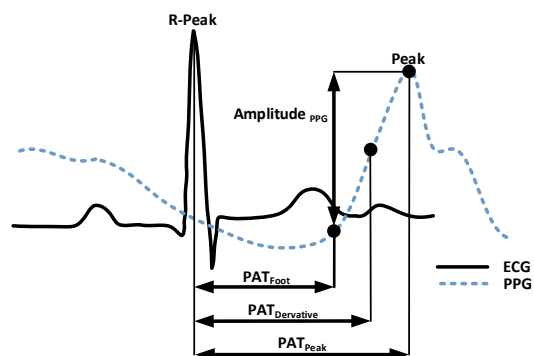


Fig. 1. Calculation principle for the Pulse Arrival Time (PAT).

The PWV (1) is given by the ratio between the distance of the two measurement sites (Δx) and the Pulse Transient Time

(PTT) [5].

$$PWV = \frac{\Delta x}{PTT} \quad (1)$$

The PTT is the time required by the pulse wave to travel through the whole segment. The PTT can also be expressed as relative measurement through the PAT and the Pre-Ejection Period (PEP) according to (2). Whereby the PEP is the time between the R-peak of the ECG and the opening of the aortic valve.

$$PTT = PAT - PEP \quad (2)$$

C. In-Ear Pressure

To be able to measure the internal pressure variation inside the auditory canals, the canals have to be sealed. Based on the ideal gas law $PV = nRT$ (with the pressure P , the volume V , amount of the substance n , the gas constant R and the absolute temperature T) a change in volume in a sealed cavity under the assumption of a constant temperature leads to a change in pressure. Fig. 2 shows a principle drawing of the in-ear pressure model of the auditory canal. The basic idea is that the

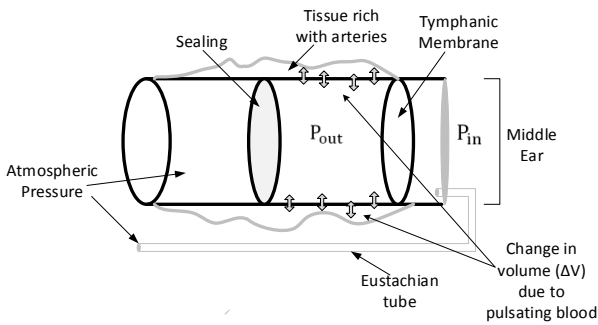


Fig. 2. Basic model of the pressure changes in the auditory canal.

arteries around the auditory canal expand based on the heart activity leading to a blood volume change. This change of volume can then be measured with a pressure sensor applied to the sealing. Fig. 3 shows a photograph of the developed custom in-ear sensor. The sensor buildup achieved the required

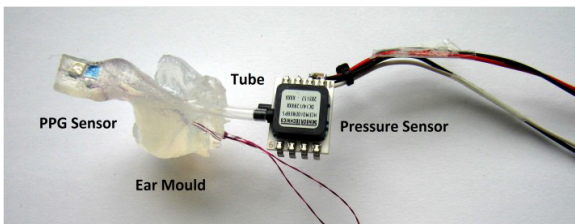


Fig. 3. Photograph of the preliminary in-ear sensor comprising of the PPG and a pressure sensor.

sealing through the mould, while allowing the measurement of a PPG, as well as the pressure inside the ear.

D. Acceleration and Temperature

For the assessment of the subjects position and physical activity an acceleration sensor is implemented in the measurement system. Additionally the core temperature can be measured via a NTC based temperature sensor.

III. IMPLEMENTATION

A. Block diagram

Fig. 4 shows a block diagram of the developed system. The microcontroller based embedded system acquires all measurement data from the subject and transmit the measured data to the host PC for further analysis and display. The data can be transmitted via Bluetooth / USB or can be stored on the micro SD-card. To allow a higher degree of flexibility the system can be battery powered, too.

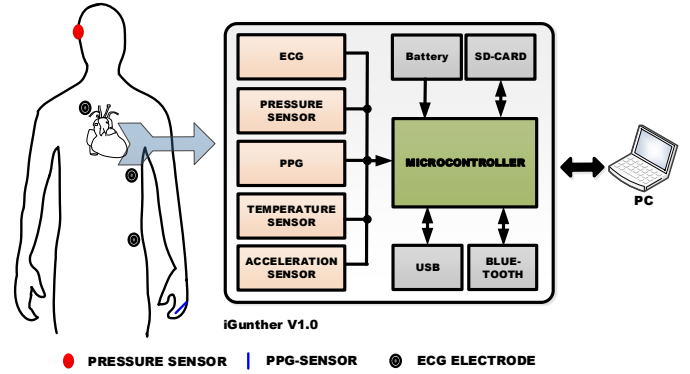


Fig. 4. Block diagram of the microcontroller based measurement system.

B. ECG Module

The developed ECG module consists of an Instrumentation Amplifier (INA) with provisions for a Driven Right Leg (DRL) and a driven shield circuit, a base line wandering rejection circuit, as well as an optional 50 Hz notch filter to mitigate power-line noise. The ECG module is able to measure the ECG in a frequency range of 50 mHz to 150 Hz. The analog ECG signal is digitalized via the internal 12 bit Analog to Digital Converter (ADC) of the microcontroller.

C. Photoplethysmography (PPG) Module

The implemented PPG circuit is designed for reflective mode and is based on two LED with wavelengths of 660 nm and 940 nm. The LED are wired in a way, that only one LED can be switched on at a time. The reflected light will be sensed by a photo diode, which is connected to a transimpedance amplifier (OPA2380 from Texas Instruments). Based on the voltage present at the output of the amplifier, digitalized via the ADC of the microcontroller, the LED current is regulated via the microcontroller's Digital to Analog Converter (DAC) to keep the output voltage of the transimpedance amplifier constant. Afterwards the analog PPG signal is high-pass filtered, to remove the DC-offset due to the dark current of the photo diode, with a servo feedback high-pass filter with a cutoff frequency of ≈ 15 mHz. A subsequent 5th-order Sallen-Key low-pass filter with a cutoff frequency of ≈ 30 Hz is used to mitigate 100 Hz flickering of the ambient light. The cutoff frequencies are chosen to allow a distortion free investigation of the expected PPG waveform, with its spectral components ranged mainly between a few mHz up to ≈ 20 Hz. The band-pass filtered analog PPG signal is afterwards digitalized via another channel of the microcontrollers ADC.

D. Pressure Measurement

The pressure sensors (HCE-M010DBE8P3 from Fist Sensor AG) have a calibrated and compensated pressure measurement range of $\pm 1000 \text{ Pa}$ ($\pm 10 \text{ mbar}$). The pressure sensor supports up to 1 kSPS with 14 bit precision, which leads to a theoretic resolution of 0.122 Pa . The temperature sensors are read out via an Serial Peripheral Interface (SPI) which is connected to the microcontroller.

E. Acceleration Measurement

The acceleration measurement is based on a LIS3DH (ST Microelectronics) acceleration sensor. The sensor can measure in three directions with full scale ranges from $\pm 2 \text{ g}$ up to $\pm 16 \text{ g}$ with 16 bit precision per direction at a sample rate from 1 Hz to 5 kHz . Furthermore the sensor has a build-in functionality for free-fall detection and is connected to the microcontroller via the SPI interface.

F. Temperature Measurement

The temperature measurement is based on a NTC sensor with dimensions of $1.0 \text{ mm} \times 0.5 \text{ mm}$. The sensor is therefore small enough to be located inside the ear. The NTC resistance is evaluated with a quarter Wheatstone bridge and will be calibrated to $37 \text{ }^\circ\text{C}$ for optimal sensitivity. The output signal of the bridge is amplified and digitalized with the internal ADC of the microcontroller.

G. Housing and Battery

The housing for the measurement system was designed in SolidWorks (Dassault Systems) and manufactured with a 3d-printer (MakerBot Replicator 2X). The housing consists of ABS plastic and has a size of $71.5 \text{ mm} \times 71.5 \text{ mm} \times 38 \text{ mm}$. Fig. 5 shows a exploded assembly drawing of the housing incl. Printed Circuit Board (PCB) and the used battery. The li-ion battery has 3.7 V by 1.25 Ah with dimensions of about $53 \text{ mm} \times 34 \text{ mm} \times 5.5 \text{ mm}$ and a weight of about 23 g . The expected system run-time with the battery is about 8 hours.

H. Microcontroller System and Communication Interfaces

The microcontroller (ATxMega128A4U from Atmel) acquires the measurement data from the different sensors and is also responsible for the data transmission to the host PC via Bluetooth or USB and for the optional storage of the measurement data on the micro SD-card. While USB communication is realized via the internal USB stack of the microcontroller, Bluetooth is implemented with a commercially available module (RN42-I/RM from Roving Networks). The Bluetooth module is certified according to Bluetooth V2.1 incl. Enhanced Data Rate (EDR) mode and supports master and slave mode with up to 300 kbps . The interfacing to the microcontroller is implemented as asynchronous serial connection.

The system is externally supplied either via USB or via the system battery. Whereas the digital components operate with $+3.3 \text{ V}$ only, the analog components are supplied with $\pm 3.3 \text{ V}$. For electrical safety considerations it is recommended to operate the system in wireless or SD-card storage mode while running from the system battery.

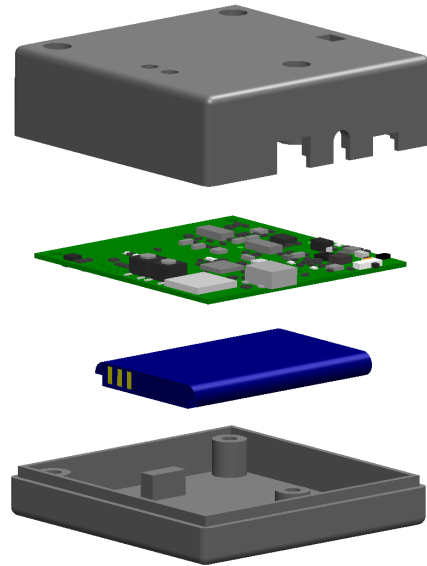


Fig. 5. Exploded assembly drawing of the housing of the measurement system inclusive the Printed Circuit Board (PCB) and the used battery.

I. Manufactured System

Fig. 6 shows a photograph of the manufactured and populated Printed Circuit Board (PCB) of the developed measurement system. The PCB has a size of about $60 \text{ mm} \times 60 \text{ mm}$ and contains about 200 components.

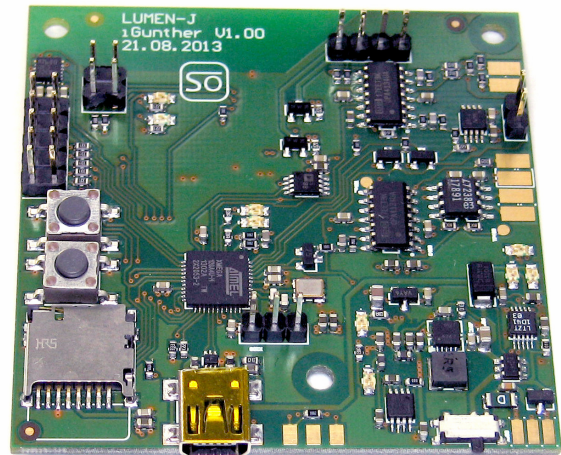


Fig. 6. Manufactured and populated Printed Circuit Board (PCB) of the developed measurement system. The PCB has dimensions of $60 \text{ mm} \times 60 \text{ mm}$ and contains about 200 components.

J. Software

The firmware of the developed measurement system is based on the Atmel Software Framework (ASF) and is written in C language.

The PC interface software is written in C# language and is based on the .NET framework from Microsoft. It is able to display the measured waveforms in real-time with a latency of about 50 ms and can be used to configure the embedded measurement systems in terms of active channels and sample

rates. With the interface software, waveforms can be recorded and exported to MathWorks MATLAB or Excel. Fig. 7 shows an image of the interface software. Visible is additional to the Graphical User Interface (GUI) the pressure inside the ear, the ECG, as well as a PPG, acquired with the developed system. It can be seen, that the waveforms have the expected typical shapes. Based on the recorded waveforms calculations of the PAT and PWV values based on PPG, ECG and the pressure signal are possible.

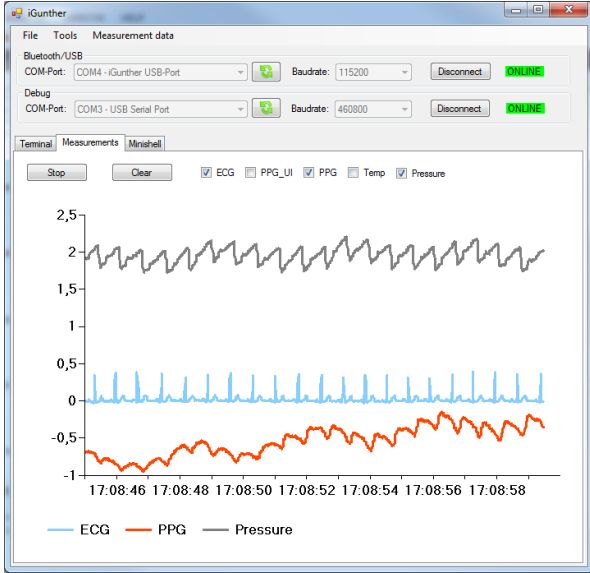


Fig. 7. Screenshot of the developed C# based PC interface software.

IV. RESULTS AND DISCUSSION

Table I shows preliminary measurement results acquired over a period of 180 seconds on three healthy subjects. For simplicity reasons only measurements on the left ear were taken and the PWV was calculated while assuming a PEP of 0 s. The PPG sensor was placed around incisura and tragus targeting the arteria temporalis superficialis and arteria auriculares anteriores.

The acquired measurements have a sample rate of 1 kHz and are filtered via wavelet decomposition in MathWorks MATLAB. The filtered signals are fed to an automated peak detection and are referenced to the ECG for evaluating the PAT. Together with the measured distance (Δx) the PWV is calculated according to (1).

The PAT derived from the pressure measurements are in a range of 100 ms to 200 ms and the PAT derived from the PPG measurements are in a range of 200 ms to 240 ms. The results are reasonable taking into account that usually reported PAT values on extremities are in the range of > 240 ms [8] and that [3] reported PAT values for in-ear pressure measurements around 100 ms.

V. CONCLUSION

A prototype of an in-ear pulse wave measurement system was developed and tested. The results are very promising and showing a good performance. In future the prototype has to

TABLE I
MEASUREMENTS

	Subject 1	Subject 2	Subject 3
Age	25	29	33
Sex	male	male	female
Height	170 cm	188 cm	158 cm
PAT length (Δx)	27 cm	30 cm	26 cm
Weight	68 kg	134 kg	52 kg
Heart Rate	68 BPM	72 BPM	70 BPM
Mean $PAT_{peak, PPG}$	207 ms	212 ms	235 ms
Mean $PAT_{peak, Pressure}$	114 ms	171 ms	123 ms
Mean PWV_{PPG}	1.3 m/s	1.4 m/s	1.1 m/s
Mean $PWV_{Pressure}$	2.4 m/s	1.8 m/s	2.1 m/s

be miniaturized and the interface software must be enhanced by the possibilities to measure PAT and PWV, as well as the heart rate directly. Furthermore a second PPG channel for measurements on the extremities should be added to have an additional reference measurement side for comparisons.

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REFERENCES

- [1] M. Gekle, E. Wischmeyer, S. Gründer, M. Petersen, A. Schwab, Taschenlehrbuch Physiologie, Thieme, S. 174, 2010
- [2] M. Middeke, Pulswellenanalyse, Renaissance einer alten Methode als moderner Biomarker der Gefäßsteifigkeit, 2010
- [3] S. Kaufmann, G. Ardelt, A. Malhotra, and M. Ryschka. In-ear pulse wave measurements: A pilot study. In Proceedings BMT (Biomedizinische Technik), volume 58, 2013
- [4] S. Laurent, J. Cockcroft, L. V. Bortel, and P. Boutouyrie. Expert consensus document on arterial stiffness: methodological issues and clinical applications. Eur Heart J, 27:2588-605, 2006.
- [5] S. Federico, Cattivelli and Harinath Garudadri. Noninvasive Cuffless Estimation of Blood Pressure from Pulse Arrival Time and Heart Rate with Adaptive Calibration. IEEE computer society, 2009.
- [6] S. Laurent, P. Boutouyrie, R. Asmar, et al: Aortic stiffness is an independent predictor of all-cause and cardi-vascular mortality in hypertensive patients. Hypertension 2001; 37: p. 12361241
- [7] A. P. Guerin, J. Blacher, B. Pannier, S. J. Marchais, M. E. Safar, G. M. London: Impact of aortic stiffness attenuation on survival of patients in end-stage renal failure. Circulation 2001; 103: p. 987992
- [8] M. C. Kortekaas, S. P. Niehof, M. H. N. van Velzen, E. M. Galvin, R. J. Stolker, F. J. O. M. Huygen: Comparison of bilateral pulse arrival time before and after induced vasodilation by axillary block. Physiol. Meas. 2012, 33, p.1993-2002
- [9] J. Baulmann, J. Nrnberger, J. Slany, R. Schmieder, A. Schmidt-Trucks, D. Baumgart, P. Cremerius, O. Hess, K. Mortensen, T. Weber, Arterielle Gefäßsteifigkeit und Pulswellenanalyse - Positionspapier zu Grundlagen, Methodik, Beeinflussbarkeit und Ergebnisinterpretation, Deutsche Medizinische Wochenschrift 135, S4-S14. 2010