Evaluation of design parameters for a reflection based long-term pulse oximetry sensor

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***Abstract*—Pulse oximetry is a standard parameter for many years in clinical patient monitoring. There are also sensors that can be used at home. But all these sensors use the transmission based method to measure the oxygen saturation which require finger or ear clips that are uncomfortable and confining and therefore not fit for long-term monitoring. Sleep-related breathing disorders or breathing thin air in high altitudes can lead to insufficient oxygen intake. Insufficient oxygen supply can cause permanent damage to the tissue and in some cases even death. For early ambulatory diagnosis, a reflection based long-term pulse oximetry sensor would be the best solution. There are no available devices on the market, even so many studies showed promising results. That why, we evaluated design parameters for a reflection based long-term pulse oximetry sensor, located at the wrist. The prototype we developed consists of two LEDs (one red and one infrared), a photodiode and the evaluation board SLAU480B from Texas Instruments. We tested three sensor layouts and discovered that the distance between the LEDs and the photodiode, the contact pressure and motion artefacts were the most important signal influencing parameters. After the Signal processing we obtained a signal to noise ratio of 22 dB (red) and 30 dB (infrared) and a AC/DC ratio of 1-3 %, which should be more than enough to calculate the SpO_2 value. Also motion artifacts were tested and seem to affect the measurement at the lower wrist strongly. In conclusion we found some reasons why there is no such device on the market yet.**

Keywords—Pulse oximetry; long-term monitoring; reflexion; hypoxia; sensor; SpO_2; PPG

I. Introduction

Pulse oximetry is a standard parameter for many years in clinical patient monitoring. There are also sensors that can be used at home. But all these sensors use the transmission based method to measure the oxygen saturation which require finger or ear clips that are uncomfortable and confining and therefore not fit for long-term monitoring. Sleep-related breathing disorders or breathing thin air in high altitudes can lead to insufficient oxygen intake. Insufficient oxygen supply can cause permanent damage to the tissue and in some cases even death.

Sleep-disordered breathing can lead hypoxia is a condition where the oxygen supply to the tissues is insufficient [1]. This can happen e.g. during sleep when no air is transported to the lung because sleep apnea (obstructive or central) [2]. The estimations of effected middle-aged adults are 9 percent among women and 24 percent among men. 17.4 percent of the affected have hypersomnia [3]. The main syndrome of it is an excessive daytime sleepiness that affects the life of people suffering from it dramatically.

Another cause of hypoxia can be breathing the thin air in high altitudes. For this reason mountain climbers or glider pilots need to pay special attention to hypoxia. The WHO estimates that that there are around 40 million tourists seeking high or extreme altitudes every year. This people are exposed to the risks of hypoxia [4].

For an early diagnosis and treatment the blood-oxygen level has to be monitored during the period of increased hypoxia-risk. Therefore a sensor, which is capable of monitoring the blood-oxygen level over several days, not interfering with the subject's activity, would be ideal. Pulse oximetry is the simplest and most common method to measure the peripheral oxygen saturation (SpO_2) non-invasively. Our market research fond no sensors available on the market that are using the transmission based method. Although wearing a finger or ear clip during sports or sleep is uncomfortable and confining. Therefore we evaluated the design parameters for a reflection based long-term pulse oximetry sensor, located at the wrist.

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II. MATERIALS AND METHODS

The components for a reflection based and a transmission based sensor are the same. Two LEDs one red and one infrared and a photodiode to measure the reflected or transmitted light are needed. Therefore we could use the existing evaluation board, the SLAU480B from Texas Instruments (TI) to drive the LEDs and read the photodiode. This board uses the integrated circuit AFE4490 from TI, with additional periphery. TI provides also the software to connect the evaluation board via USB to the PC and capture the data. The main focus of this paper lies therefore on the arrangement of the LEDs and the Photodiode, as well as the ambient conditions under which the measurement is done.

A. Selection of components

As test environment we used the evaluation board SLAU480B from TI. It consists of the AFE4490 as an integrated circuit (IC) that is designed for pulse oximetry measurements. This IC has a LED driver circuit and a measurement unit where a photodiode can be attached. The driver circuit has an 8 bit current control for both LED channels. The measurement unit processes the signal received from the photodiode. The current received from the photodiode is converted by a transimpedance amplifier into a Voltage. The gain can be adjusted by software. The next stage can subtract a variable DC value from the signal. This can be used by a control loop to compensate ambient light analogously. Afterwards there is a low pass filter stage with 500 or 1000 Hz cutoff frequency. The resulting signal gets buffered and fed to a 22 bit analog digital converter (ADC). Besides values for both wavelength and ambient light a digitally compensated value is provided. This compensated value is the subtraction of signal and ambient light. For the measurements preformed for evaluation tests we used the digitally compensated values.

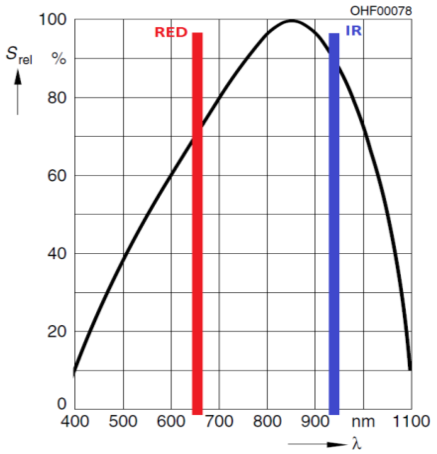

Fig. 1. Characteristic of BP 104 S photodiode [4] with peak wavelengths of the used LEDs

Photodiodes can detect only a small bandwidth of wavelengths. Out of this area the sensitivity decreases rapidly. For pulse oximetry we need to detect 2 different wavelengths with one photodiode. Hence the diode should be sensitive to both wavelengths. For the sensor frontends we used the "BP 104 S" (OSRAM). Fig. 1 shows its sensitivity over the wavelengths and also displays the red and infrared wavelengths needed for the SpO_2 measurement. The sensitivity for the infrared wavelength is slightly higher than the one for the red wavelength. Therefore the signal intensity of the red signal is expected to be lower than the one of the infrared signal.

We used standard 0805-SMD packages for the LEDs and paid special attention that they were of the same height. This ensures that the distance between the tissue and photodiode is the same for both LEDs. In TABLE I. the most important characteristics for both LEDs are listed.

TABLE I. LED CHARACTERISTICS

LED	**KP-2012SRC-PRV**	**KP-2012F3C**
Color	red	infrared
Manufacturer	Kingbright	Kingbright
Peak wave-length	660 nm	940 nm
Typ. forward voltage	1.85V (IF=20 mA)	1.2V (IF=20 mA)
Max DC current	30mA	50mA
Max peak current	155mA	1.2A
Height	1.1 mm	1.1 mm

B. Sensor design

For the sensor design there are several parameters to pay attention to. The most important is the distance between LEDs and photodiode. The dimensions are limited to the size of the components and to the size of the area of the wrist where the sensor can be placed. Because the sensor is built on a FR4 printed circuit board (PCB), the substrate is rigid and should be placed on an approximately flat area.

In [5] they found out that the photon-visit depth is dependent on the skin surface and color. They suggested sensor detector distances from 6 to 12.5 mm. In [6] it was found that the ideal distance for measuring at the forehead or cheek is between 5 und 7 mm. The diffusely reflected photons pierce the tissue on a curved course. Out of the information they found in [5] we can derive the following relations between the distance of LED and photodiode and the signal:

- The larger the distance, the larger is the penetration depth of the photons
- The larger the distance, the weaker is the intensity of the detected signal

The main challenge in the sensor frontend design is to find the best layout for the LEDs and photodiode. The distance should be large enough for the photons to penetrate the tissue until pulsating blood vessels. On the other hand the distance should be as close as possible to get the best signal intensity possible.

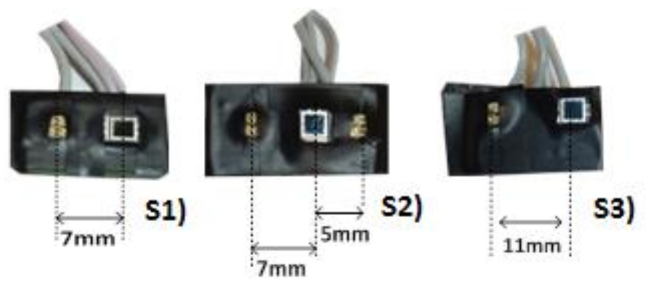

Fig. 2. Sensor prototypes used for examination

In our tests we examined 3 different sensor prototypes (Fig. 2). The prototypes (S1) and (S3) have the same structure, with the LEDs on one side and the photodiode on the other side. The only difference is the distance between the sensor and detector elements. The prototype (S2) on the other hand has a different structure. Here the photodiode is in between two LED pairs. On each side there are one red and one Infrared LED. The sensors were placed at the lower wrist and covert by a sleeve to reduce the influences of ambient light.

III. Results

TABLE II. SENSOR PROTOTYPE TEST RESULTS

	Sensor (S1)	Sensor (S2)	Sensor (S3)
I_R [mA]	15	10	25
I_{IR} [mA]	10	6	20
$\hat{U}_{AC\,R}$ [mV] (15 mmHg)	2.7	1.5	1.1
$\hat{U}_{AC\,IR}$ [mV] (15 mmHg)	2.8	1.7	1.4
$\hat{U}_{AC\,R}$ [mV] (30 mmHg)	5.2	3.8	3.7
$\hat{U}_{AC\,IR}$ [mV] (30 mmHg)	6.0	2.0	5.0
$\hat{U}_{AC\,R}$ [mV] (45 mmHg)	7.5	5.5	12.8
$\hat{U}_{AC\,IR}$ [mV] (45 mmHg)	7	3.4	15.2
$\hat{U}_{AC\,R}$ [mV] (60 mmHg)	15.6	10.5	29.5
$\hat{U}_{AC\,IR}$ [mV] (60 mmHg)	11.0	6.6	46.5

The results show that by increasing the pressure the signal strength increases also. For 15 mmHg all prototypes showed the weakest signal. Also the signal shape was not clear defined and because there was no indication of the reflected pulse wave and only the fundamental pulse frequency was shown. At 30 mmHg pressure the signal shape was good for all sensors and the reflected wave could be seen. Sensor 1 showed the highest signal amplitude at this pressure. At 45 and 60 mmHg Sensor 3 showed the highest signal amplitude. The shape of the wave form did not change by altering the cuff pressure between 30 and 60 mmHg. The wearing comfort decreased with increasing pressure. We found out that 45 mmHg was still comfortable during a half hour measurement whereas 60 mmHg was already very constraining. Hence suggest to use a contact pressure of 45 mmHg or below for long-term measurements.

Signal processing and analysis

To design the optimal filter range and imported frequency components, we transformed the signal with a fast Fourier transformation (FFT).

In Fig. 3 the FFT of the unfiltered AC signal is shown. The 50 Hz mains hum shows significantly as a peak. Furthermore the signal amplitude is highest in the range from 0 to 1.2 Hz. The frequency components between 0 and 0.5 Hz are likely associated with respiration and motion artefacts. The frequencies around 1 Hz are likely related to the pulsation which we want to detect.

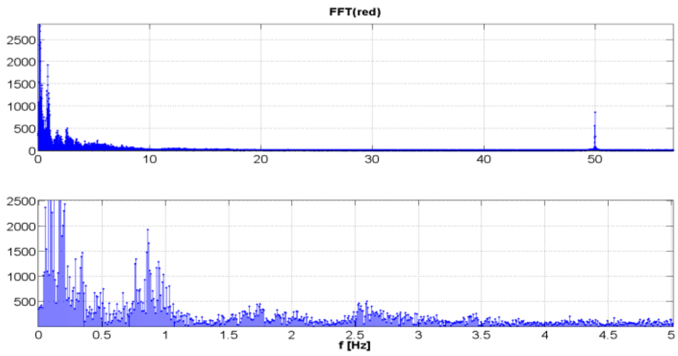

Fig. 3. Fast fourier transformation of the DC free Signal

The main challenge for the filter design is to distinguish between the PPG signal and other signal components. The frequency components of our wanted signal are likely around the pulsation frequency and its harmonics. The amplitudes of the FFT signal are decreasing rapidly after the second harmonic. The cut off frequency of the high pass filter should therefore be around three times the maximum pulsation frequency. We choose 10 Hz calculating a maximum pulse of 200 beats per minute. The cut of frequency for the low pass should be around 0.5 Hz to eliminate the unwanted signal components mentioned before.

To eliminate influences of noise and other unwanted signal parts we processed the signal. First the signal was filtered and digitalized with the analog digital converter of AFE4490. The AFE4490 cuts the signal before digitalizing at 500 Hz to avoid aliasing. Afterwards we applied a notch filter to cut off the 50 Hz mains hum. Next a band pass filter with a pass band between 0.5 and 10 Hz was applied to purge spectral power out of the spectrum of interest. Finally the signal was inverted to get the typical PPG waveform. Fig. 4 shows the outcome.

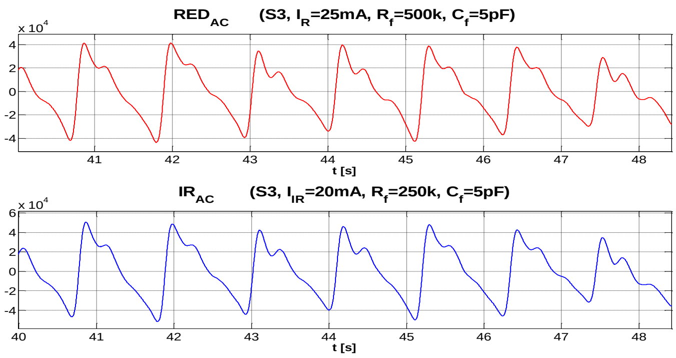

Fig. 4. Signal after all filter stages at 45 mmHg cuff preasure

In terms of signal quality the most important parameters are AC/DC ratio for both wavelength and the signal to noise ratio (SNR). For transmission based methods the AC/DC ratio is between 2 and 5 %. Reflectance methods can have up to 10 times weaker AC signals [7]. For our measurements we had an AC/DC ratio of 1-3 %. The SNR was calculated with the following equation:

$$SNR = 10 \log_{10}\left(\frac{\sum_{n=0}^{k} u_{n\,AC\,signal}^2}{\sum_{n=0}^{k} u_{n\,AC\,noise}^2}\right)$$

As signal component we used signal processed as shown in Fig. 4. As noise component we used two different setups. One was with a pressure cuff at the upper arm to stop the pulse wave before reaching the sensor to eliminate the Signal component. For the other we used the ambient light signal as noise. For the first method we calculated a SNR of 16.4 dB (red) and 18.6 dB (infrared). Using the ambient light as noise source we got 22 dB for the red and 30 dB for the infrared signal. The pressure cuff couldn't erase the signal completely and at 150 mmHg there were still minor pulsations at around 1 Hz visible. This indicates that the second value is likely the more accurate one. The obtained SNR and the AC/DC ratios indicate that the SpO_2 calculation is possible.

A. Influence of motion artifacts

For continuous measurements over a long time during everyday activities the sensor should also be tested under non static conditions. We tested the influence of motion artifacts to the signal by capturing new data, which is shown in Fig. 5. During the measurement the subject did three predefined movements. Movement a) was snapping the finger. Movement b) was fist clinching and c) was bending the arm. During the movement periods you can see artifacts dominating the AC signal.

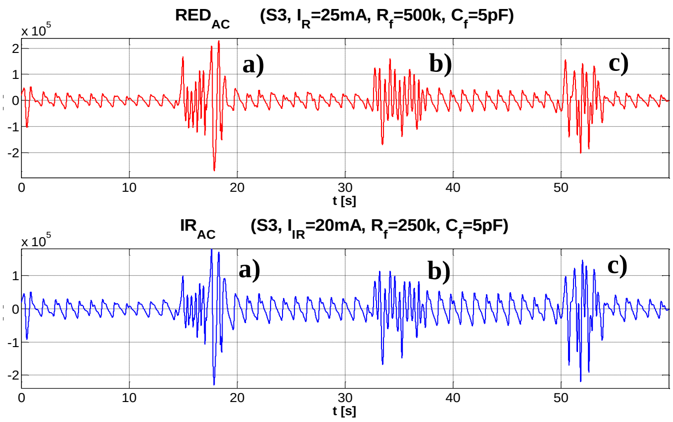

Fig. 5. Motion artifacts within the signal; a) finger snaping b) fist clenching c) bending the arm

The AC PPG signal can't be detected during all three movement periods. The kind of motion doesn't seem to make a difference. This indicates that motion in general affects the signal quality at the lower wrist dramatically. On the other hand the measurement shows that the signal recovers quickly after periods of motion.

IV. Conclusion

During this study we found different reasons that could explain why we found no commercial sensors solution, that use the reflection based method to measure the oxygen saturation. E.g. the required contact pressure of at least 15 mmHg and the strong influence of motion artifacts could be hard to handle in a commercial product. The SNR of 16.4/22 dB for the red and 18.6/30 dB for the infrared signal and the AC/DC ratio of 1-3 % at 45 mmHg contact pressure on the other hand indicate that a SpO_2 calculation is possible during phases with no motion. In the future we will calculate the SpO_2 value out of the signal and compare it in a small study to a finger-clip sensor. Furthermore we will investigate the power consumption and also different body positions.

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