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HIGH RESOLUTION HADAMARD TRANSFORM OXIMETRY

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Abstract: The principle of pulse oximetry is based on the different absorption of haemoglobin at various wavelengths in the visible and IR-region, utilizing Lambert-Beers law. With conventional oximeters a part of the human body is penetrated by radiation at two wavelengths and the oxygen saturation is calculated from the detected signals. Intention of the optical set-up presented here is the improvement of the signal-to-noise-ratio of the detected signals by a modified signal generation. Radiation of three different sources penetrates the human skin in a time-multiplex sequence, which is designated by a Hadamard matrix. The signals of the distinct radiation sources are obtained mathematically from the detected sum signals. The optical set-up exemplifies the principle, which can be transferred to other blood ingredients than haemoglobin when the emitting wavelengths of the radiation sources are changed.

Keywords: NIR-Spectroscopy, Pulse Oximetry, Hadamard Transformation

1 INTRODUCTION

In clinical practice, the determination of the oxygen saturation of the human body is routinely performed. The importance of this parameter is generally accepted. Nevertheless the readout of the oxygen saturation of commercial pulse oximeters has an accuracy of ±1%. This certainly can be improved.

The precision of a calculated parameter like the oxygen saturation can only be improved if the signal quality of the photophlethysmograms the oxygen saturation is based on is enhanced. This can be achieved by increasing the signal-to-noise ratio and reducing the influence of motion artifacts [1]. Because our further objective is the measurement of other blood ingredients which are more difficult to detect, the increment of the signal-to-noise ratio is the purpose of our optical set-up. One method could be to raise the intensity of the radiation sources while the detector noise remains unchanged, but the improvement of the intensity has of course physical limits. Another option is the benefit of the so-called “multiplex advantage” by the use of multiplexing wavelengths in different groups and re-calculating the distinct intensities afterwards. This kind of spectroscopic approach and the usage of optimal multiplexing for certain numbers of wavelengths is known as Hadamard Transform Spectroscopy [2].

We present a new concept for a pulse oximeter, where distinctive radiation sources are controlled in a way which is determined by a Hadamard matrix. Transmission signals are measured through the finger using three different wavelengths. The calculation algorithm of the oxygen saturation remains the same while the signal quality is improved.

2 THEORETICAL BACKGROUND

2.1 Pulse Oximetry

Pulse oximetry is a noninvasive spectroscopic method to determine the oxygen saturation of blood. The oxygen saturation describes the percentage of oxyhaemoglobin on total haemoglobin and it is a parameter which is an indicator for the oxygen accommodation of the human body. The different absorption of total haemoglobin and oxyhaemoglobin is utilized [3].

A part of the human body, e.g. a finger or an earlobe, is transmitted by radiation of two distinct spectral ranges, at approx. 660 nm and between 830 nm and 950 nm. These wavelengths are all located in a so-called optical window where radiation can penetrate the human skin easily. At 660 nm the extinction coefficient of total haemoglobin is much higher than one of the oxyhaemoglobin whereas between 830 and 950 nm they are approximately the same.

Two diodes are used as radiation sources which are operated in time-multiplex. The radiation of the two radiation phases is detected by a photodetector located opposite the diodes. Two time-multiplexed photoplethysmograms are obtained, each corresponding to one wavelength. After demultiplexing, the minima and maxima of each pulse wave must be extracted for calculating the oxygen saturation. By subtracting the logarithmic intensities at the minima from the logarithmic intensities at the corresponding peaks, the influence of the tissue on the signals is eliminated [4]. The algorithm is based on the assumption, that the finger consists of three layers: a tissue layer, a layer of constant thickness containing blood with the thickness of the minimal diameter of the arteries in the finger and a layer with changing thickness, describing the part of the arteries that change their diameter due to the pulse wave.

The logarithmic differences of the intensities $I_{max}(\lambda_i)$ and $I_{min}(\lambda_i)$ for the two wavelengths are divided. Defining the factor

$$\Omega = \ln\left(\frac{I_{\max}(l_1)}{I_{\min}(l_1)} / \frac{I_{\max}(l_2)}{I_{\min}(l_2)} \right) \tag{1}$$

the oxygen saturation is calculated by

$$S_{o_2} = \frac{\Omega \cdot e_{Hb}(l_2) - e_{Hb}(l_1)}{\Omega \cdot \left(e_{Hb}(l_2) - e_{HbO_2}(l_2)\right) + e_{HbO_2}(l_1) - e_{Hb}(l_1)} \tag{2}$$

The formula can be traced back to Lambert-Beer's law and can be applied to the calculation of the oxygen saturation down to 70%. Below 70% Lambert-Beer's law is no longer valid [4].

A different technique is utilized with the Masimo Signal Extraction Technology [1], which deploys a so-called Discrete Saturation Transform. In this approach reference signals are created to find adaptively a signal that corresponds to the arterial oxygen saturation. The oximeter consists of two diodes like conventional oximeters. The algorithm utilizes the fact that the venous component in the detected signal is more susceptible to motion artefacts than the arterial component and calculates a noise reference.

2.2 Hadamard Transform Spectroscopy

A pulse oximeter operating with two diodes is the easiest form of a dispersive spectrometer. Mathematically a measurement of a whole spectrum with a dispersive spectrometer can be expressed as

$$\vec{I} = \underline{W} \cdot \vec{o} + \vec{e} \tag{3}$$

where $\vec{I}$ is the vector containing the measured intensities I_{l_i}, $\vec{o}$ is the vector with the unknown real intensities, $\vec{e}$ describes the error that is made with each measurement and $\underline{W}$ is the so-called design-matrix. For a dispersive spectrometer $\underline{W}$ is the unity matrix, at each single wavelength or small wavelength range the intensity measured is assigned to that wavelength [5].

If the measurement is perfect the error is zero and the measured intensities are the real intensities $\vec{o}$:

$$\vec{o} = \underline{W}^{-1} \cdot \vec{I} \ . \tag{4}$$

It is assumed that the e_i in different measurements are random and uncorrelated, that they have a standard deviation of σ^2 and that the detector response is linear. Then it can be said that

$$\left\langle e_i e_j \right\rangle = 0 \qquad i \neq j \ . \tag{5}$$

The aim of every measuring device is to minimize the errors occurring. If $\vec{\hat{o}}$ describes the best estimation of the true intensities, corresponding to formula 3 $\vec{\hat{o}}$ can be calculated by $\vec{\hat{o}} = \underline{W}^{-1} \cdot \vec{I}$, the errors e_i are not known. $\vec{e}$ can be minimized if the difference between the estimated intensities $\vec{\hat{o}}$ and the real intensities $\vec{o}$ are minimized:

$$\vec{\hat{o}} - \vec{o} = \underline{W}^{-1} \cdot \vec{I} - \underline{W}^{-1} \cdot \left(\vec{I} - \vec{e}\right) = \underline{W}^{-1} \cdot \vec{e} \ . \tag{6}$$

The mean squared error of $\vec{p}-\vec{o}$ is $\vec{e}=\frac{s^2}{N}\cdot\sum_{i=1}^{N}\sum_{j=1}^{N}\left(w_{ij}\right)^2$. (7)

This term is minimized if $\underline{W}\cdot\underline{W}^T=N\cdot\underline{E}$, where $\underline{E}$ is the unity matrix of order *N*. Matrices who satisfy this equation are called Hadamard matrices [6]. The matrices consist of the elements $\{-1,1\}$. Using Hadamard matrices instead of unity matrices, the mean square error is reduced by a factor of $1/N_H$, where N_H is the matrix order, and the signal-to-noise ratio is increased by a factor of $\sqrt{N_H}$ [7]. Hadamard matrices exist for the orders of 1, 2 and multiples of 4. Hadamard Transform Spectroscopy utilizes these optimal design matrices. The matrices describe which wavelengths must be detected in the spectrometer simultaneously. The radiation corresponding to each single wavelength is calculated by $\underline{W}^{-1}\cdot\vec{o}$. Hadamard matrices are used for two-detector-designs, the elements including a "1" describe wavelengths to be transmitted and to be detected with one detector, and the "-1"-elements characterize wavelengths to be reflected and detected by a second detector.

From Hadamard matrices so-called S-matrices can be obtained. These matrices consist of the elements $\{0,1\}$ and they are of the orders N_H-1. In comparison to Hadamard matrices the S-matrices are used for single-detector designs where the "0" mark wavelengths to be rejected. Using S-matrices the mean square error is reduced by a factor of $(4\cdot N_S)/(N_S+1)^2$ and the signal-to-noise ratio increases by a factor of. $(N_S+1)/(2\cdot\sqrt{N_S})$ [8].

Hadamard spectrometers show the advantages of wavelength multiplexing spectrometers without the disadvantages of moving parts. Normally they consist of a radiation source, a dispersive element like a grating or a prism, a sample and one or two detectors. Instead of a slit behind the grating or prism, like a dispersive spectrometer, a Hadamard spectrometer utilizes additionally to the other components a so-called Hadamard encoding mask which converts the mathematical description of the matrix into a physical realization. The Hadamard encoding mask is either a rotating mechanical mask [8], realized by crystals who are opaque or transparent corresponding to their temperature [2], or by the transmitting wavelengths of an acousto-optical filter [9].

3 INSTRUMENTAL SET-UP

To improve the signal-to-noise ratio of the photoplethysmograms a Hadamard spectrometer with IR-diodes and a LED is constructed (figure 1).

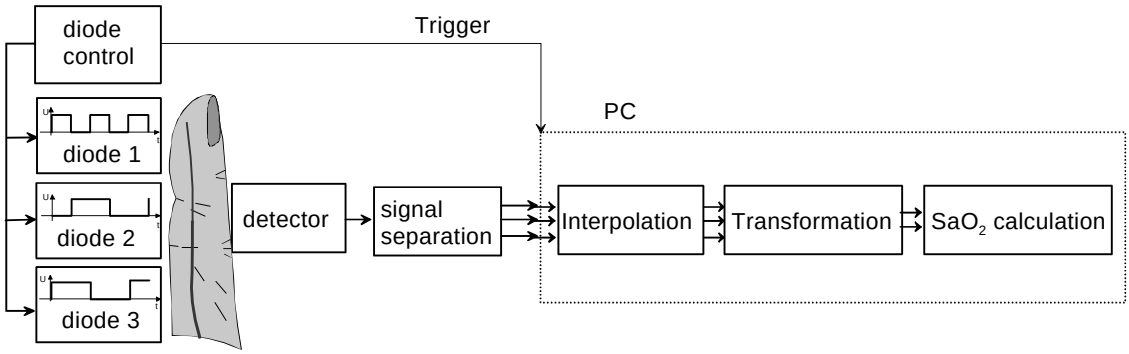

Figure 1. functional block diagram

It is expected that recalculated signals of the three diodes have a better quality, so the error due to error propagation in the calculation of the oxygen saturation decreases. The smallest reasonable S-matrix, order N=3, is utilized. In our optical set-up the encoding mask is formed by the radiation sources themselves. By switching groups of 2 diodes in the way the corresponding S-matrix dictates (eq. 8), the encoding mask is realized.

$$\begin{bmatrix} I_{\text{phase1}} \\ I_{\text{phase2}} \\ I_{\text{phase3}} \end{bmatrix} = \begin{bmatrix} 1 & 0 & 1 \\ 0 & 1 & 1 \\ 1 & 1 & 0 \end{bmatrix} \cdot \begin{bmatrix} I_{\text{Diode1}} \\ I_{\text{Diode2}} \\ I_{\text{Diode3}} \end{bmatrix} \tag{8}$$

In each phase two diodes are radiating at a time. After the three radiation phases a phase with no diode emitting follows. The detected radiation during this phase represents the basic offset. The diodes and the detector are placed on opposite sides of a finger sensor.

4 SIGNAL PROCESSING

The classical Hadamard transformation is applied to stationary signals whereas the photoplethysmograms are time-dependent. The radiation sources are triggered in time-multiplex corresponding to the S-matrix. As a consequence there is a time difference between the three detected signals. To obtain the Hadamard transformation algorithm, for each diode the values at the sampling instants of the other diodes and the non-emitting phase are linearly interpolated. The basic offsets are also interpolated. As a result all signals are reconstructed for the whole measuring time. The basic offsets are subtracted from the other signals and the Hadamard transformation can be applied.

The signals corresponding to the single diodes are calculated by

$$\begin{bmatrix} I_{\text{Diode1}} \\ I_{\text{Diode2}} \\ I_{\text{Diode3}} \end{bmatrix} = \frac{1}{2} \begin{bmatrix} 1 & -1 & 1 \\ -1 & 1 & 1 \\ 1 & 1 & -1 \end{bmatrix} \cdot \begin{bmatrix} I_{\text{phase1}} \\ I_{\text{phase2}} \\ I_{\text{phase3}} \end{bmatrix} \tag{9}$$

After signal separation, the re-calculated intensities of the 660 nm-diode and the 880 nm-diode are used to determine the oxygen saturation with the algorithm from eq. (2). The spectral characteristics of the diodes were measured using an Oriel MIR8000 FT spectrometer, and the extinction coefficients corresponding to the measured wavelengths are used [10]. The third diode in the set-up is necessary to enable the Hadamard transformation. In comparison to the 2-diode-oximeter each diode radiates at a longer time interval because of the modified radiation sequence.

5 RESULTS AND DISCUSSION

The signal-to-noise ratio of photoplethysmograms can only be estimated because the pure signal without any noise is not known. To determine an indicator for the signal-to-noise ratio a measurement was made with the Hadamard set-up (figure 1), positioning an optical filter which attenuates radiation uniformly for all wavelengths in the finger sensor. In turn two diodes were switched off and the signal was taken while one diode was emitting alone. So constant signals were measured, whose standard deviations serve as an indicator for the signal-to-noise ratio. Table 1 shows the computed standard deviations of the recalculated signals and of the corresponding single-diode signals.

Table 1: standard derivations of recalculated and single-diode signals

Wavelength in nm	660 nm	880 nm	935 nm
Standard deviation recalculated signals $ó_R$	0.0186	0.0119	0.0127
Standard deviation single-diode signals $ó_s$	0.0211	0.0223	0.0185
Quotient $ó_R / ó_s$	0.8806	0.5341	0.6842

The standard deviations of the recalculated signals are better for each diode than the standard deviation of the single-diode signals. The decrease in the standard deviation is not the same for the three wavelengths, but the experimentally achieved values correspond approximately to the theoretical value of 0.75.

For verification of the Hadamard set-up the oxygen saturation of a subject was measured with our instrument and simultaneously with two commercial oximeters[1]. The nasal air flow was also recorded, using a thermistor. The Hadamard finger sensor was placed at the middle finger, the oximeter1 [11] sensor at the ring finger and the oximeter2 sensor [12] at the forefinger of the left hand. The test person stopped breathing for 35 seconds and breathed slightly another 80 seconds during the measuring time of four minutes leading to a decrease in the oxygen concentration, fig. 2. Oximeter1 and oximeter2 show this effect 20 seconds after breathing stop, oximeter2 displays the reduction faster.

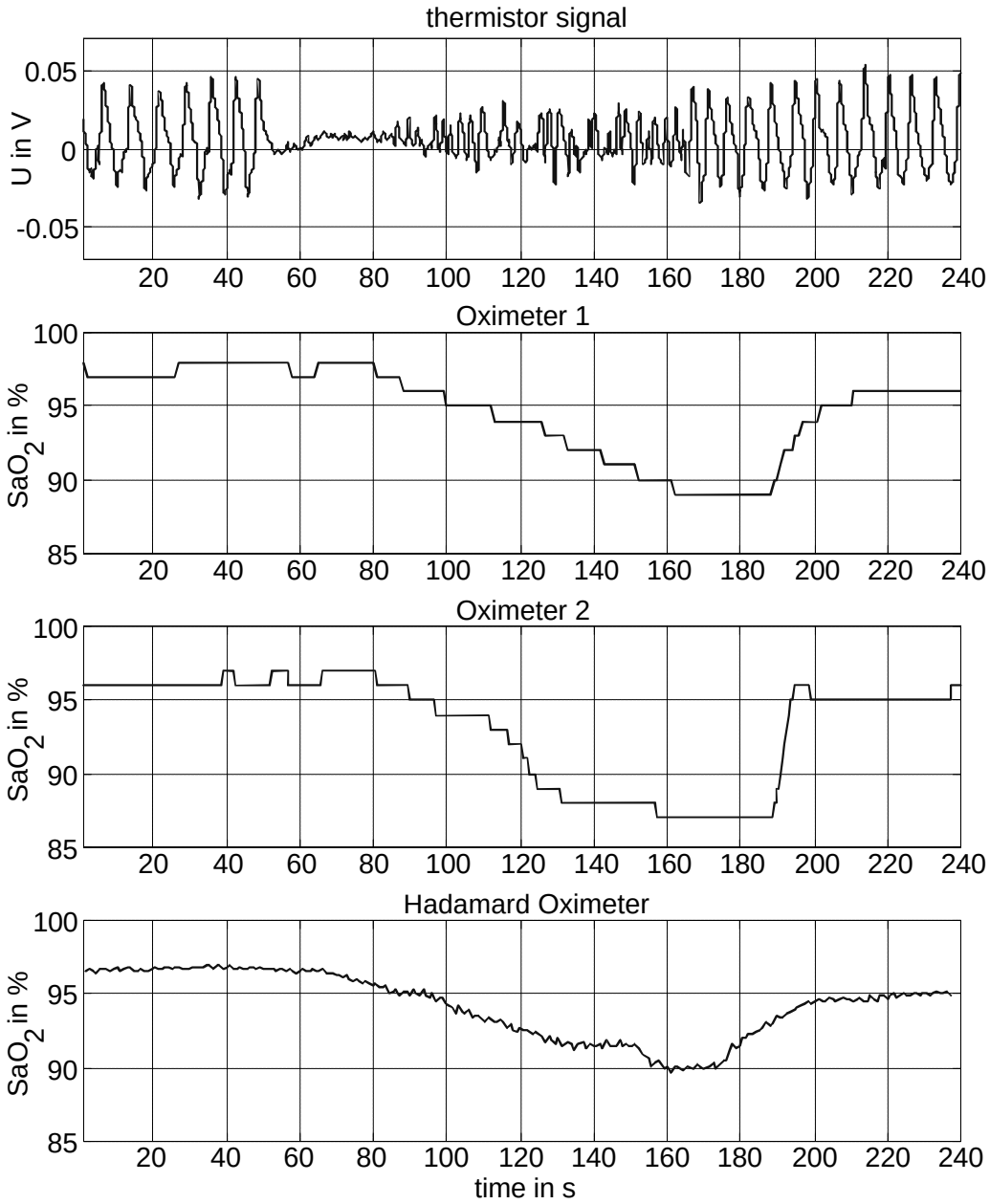

Figure 2: Nasal thermistor signal and outputs of oximeters 1, 2 and Hadamard oximeter at hypoxia experiment

The trend of the oxygen saturation is the same in all oximeters. The Hadamard optical set-up displays the beginning of the decrease in oxygen saturation about 10 seconds earlier than oximeter1 and oximeter2. In our approach the oxygen saturation is calculated level-continuously after every pulse

[1] Hadamard oximeter, Nellcor 200 (oximeter1), MCC PO300 (oimeter2)

wave while the outputs of the commercial instruments are updated every second. We have verified experimentally, that the commercial instruments have time constants of 9 and 7 seconds, respectively, so the different time response seems reasonable.

Regarding the concentration sensitivity, oximeters1 and 2 are specified with an accuracy of 1% and 1.5 % [11,12], respectively. Our set-up was calibrated utilizing the output signal of oximeter1 as a reference. A constant offset of 1.2% had to be subtracted from the computed Hadamard oxygen saturation to match the oxygen concentration of oximeter1 under normal breathing conditions.

The accuracy, the sensitivity, the time response and the artefact behaviour of our oximeter remain to be further analyzed. The application of this Hadamard transform set-up is of course not limited to pulse oximetry. The method was chosen because various blood ingredients have typical extinction peaks in the NIR or IR region where the water absorption dominates in the human body [13].

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