Employing Microtechnology for Non-Invasive Determination of Local Blood Oxygen Saturation Based on Tissue Remission Spectra

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ABSTRACT

A new micro measurement system - according to the principle of the EMPHO-System - has been developed to record tissue remission spectra within the range of 510 to 590 nm. The device hardware basically involves a white LED, optical fibers, and a miniaturized grating spectrometer. In general, exact computation of blood oxygen saturation and relative hemoglobin concentration from these tissue remisson spectra is not possible. The main reason is that - except for oxyhemoglobin and desoxyhemoglobin - many other light-scattering and light-absorbing substances such as dopa-melanin and bilirubin cause measurement errors. A new algorithm for the determination of local blood oxygen saturation and relative hemoglobin concentration from tissue remisson spectra is presented. This method is based on the probability of light photons propagation and reference coefficients. In addition, the algorithm also takes other light-scattering and light-absorbance substances in biological tissue into consideration. The method is compared with the standard procedure of Kubelka and Munk. Among others these two approaches are applied to the tissue remission spectra of ischaemia with subsequent reactive hyperemia. The calculated blood oxygen saturation and relative hemoglobin concentration of both methods are presented and discussed.

Keywords: biological tissue, blood oxygen saturation, hemoglobin concentration, Kubelka-Munk, tissue remission spectra

1. INTRODUCTION

At the institute for measurement and control engineering a micro measuring system for continuous recording of tissue remission spectra in the wavelength range from 500 to 600 nm was developed over the past years. The measurement principle is based on the EMPHO (The Erlangen - micro-lightguide - spectrophotometer).[1,2] The EMPHO uses a xenon arc lamp as a wide-band source of light, which emits light of high intensity within the interested range between 500 and 600 nm, a spectrophotometer consisting of a rotating filter disk with a motor, a position detector, and a photomultiplier. Using this system, 100 tissue remission spectra per second can be recorded. This system has conditional on its components some disadvantages. The system can be used only stationarily because of its large size, it is not robust in relation to vibrations, since it consists of moving parts, and it cannot be battery-operated, since both the xenon arc lamp and also the photomultiplier require high voltage supply.

At the institute for measurement and control engineering a handy sized, robust, and mobile micro measuring system for recording tissue remission spectra could be realized (see fig. 1). Essentially, this micro measuring system consists of a white LED (e.g. Nichia NSPW 300BS), optical fibers (diameters: 50 - 250 μm), a miniaturized grating spectrometer (e.g. the module: CP20 of Jobin Yvon with a line detector from 128 pixels), and a 14-Bit-AD-Converter. The digitized values are transferred over a serial interface (RS232) to a computer. The tissue remission spectra are visualized on the computer and the local oxygen saturation and the relative hemoglobin concentration are computed.

Until today it is difficult to make a secure statement about the local blood oxygen saturation from tissue remission spectra. With the designation of the blood oxygen saturation one differentiates between the functional oxygen saturation and the correct oxygen saturation. The functional oxygen saturation is defined by the ratio of

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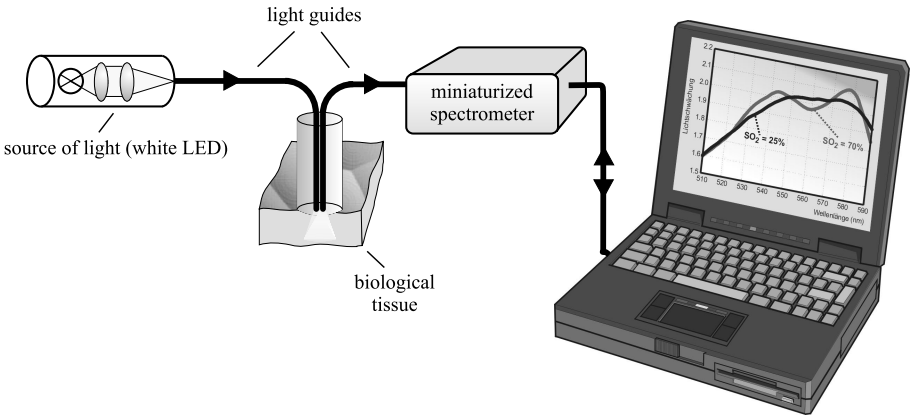

Figure 1. Principal drawing of the micro measuring system for recording tissue remission spectra

the concentrations of oxygenized hemoglobin over the concentration of the sum of desoxygenized and oxygenized hemoglobin:

$$SO_2 = \frac{c_{HbO_2}}{c_{Hb} + c_{HbO_2}} \tag{1}$$

The correct oxygen saturation considers also the concentrations of the other hemoglobin derivatives e.g. carbon monoxide hemoglobin (c_{COHb}) or methemoglobin (c_{MetHb}) and is defined as follows:

$$SO_2(correct) = \frac{c_{HbO_2}}{c_{Hb} + c_{HbO_2} + c_{COHb} + c_{MetHb} + \cdots} \tag{2}$$

Since many other light-absorbing and light-scattering components besides hemoglobin molecules are present in the biological tissue, they will be reflected down in the tissue remission spectrum. The photons, which penetrate into the biological tissue, can be scattered or absorbed by different parts or structures. Except of the light absorption by oxygenized and desoxygenized hemoglobin, there are still different light-absorbing substances, which influence the tissue remission spectrum. Melanin and bilirubin belong to these substances in the wavelength range between 500 and 600 nm. Melanin is a dark (brown or black) coloring material (pigment), those who causes the colouring of skin, and bilirubin is a decomposition product of the haem. In Fig. 2 the extinction spectra of these 4 relevant light-absorbing substances of biological tissue are represented in the wavelength range between 500 and 600 nm.[3]

As the standard method for the calculation of the oxygen saturation from the tissue remission spectra the method of Kubelka and Munk was established. It is used also in the EMPHO for the calculation of the blood oxygen saturation. This method uses very simplified assumptions, which will be specified in this paper. Due to this simplified assumptions a new method is presented, which is based on the probability of light photons propagation, and accordingly the calculation of the oxygen saturation is modified. This new method is called FIR-filter-method, since for the calculation of the oxygen saturation FIR-filter-coefficients are used. In the following, first the theory of the Kubelka-Munk-method and afterwards the theory of the FIR-filter-method are represented.

By using the example of an in-vivo-measurement of an ischaemia with following reactive hyperemia both evaluation methods are used for the determination of the local blood oxygen saturation and the relative hemoglobin concentration. The results are presented and are confronted in a comparative discussion between the Kubelka-Munk-method and the FIR-filter-method.

2. METHODS

2.1. Kubelka-Munk-method

The theory of Kubelka and Munk, also referred to as two-flux-theory, describes the light propagation in the tissue by two diffuse photon fluxes, a diffuse photon flux I_1 into positive z-direction and a diffuse photon flux I_2

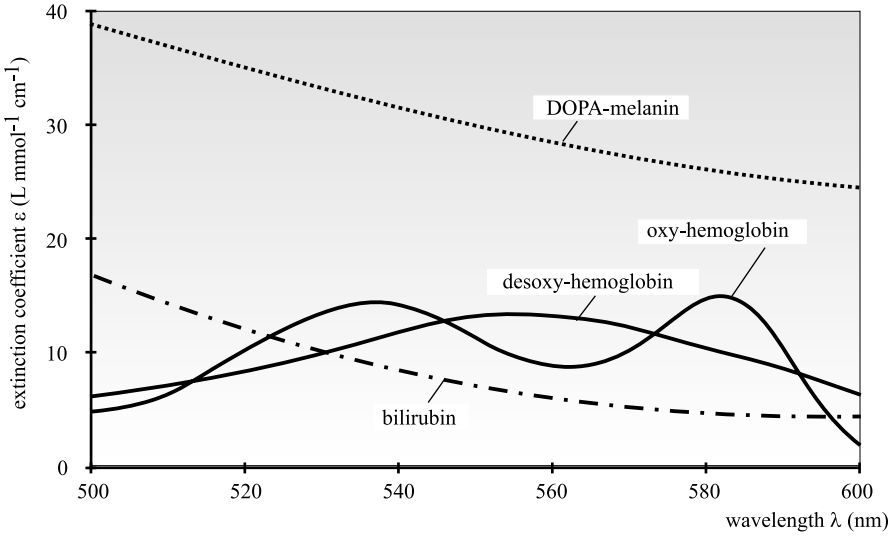

Figure 2. Molecular extinction coefficients of some relevant light-absorbing substances in human skin

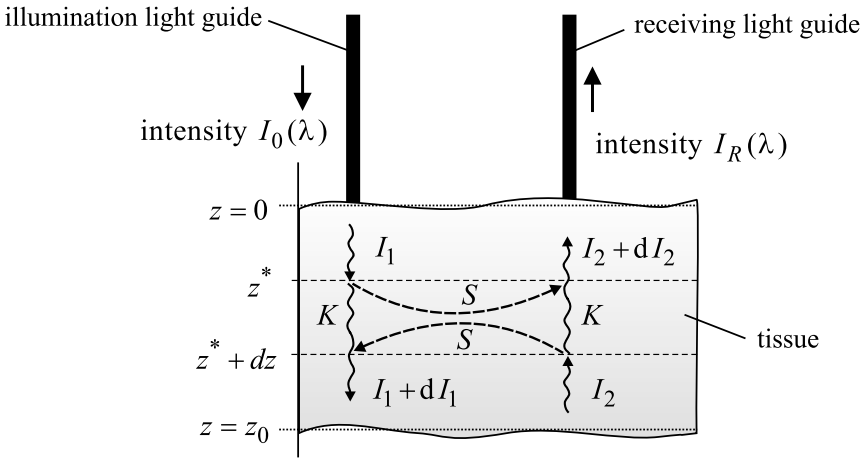

Figure 3. Two-flux-theory of Kubelka and Munk

into negative z-direction. The flux I_1 and/or I_2 can be reduced in each case by absorption (represented by the constant K) or by scattering (represented by the constant S) or be increased in each case by scattering from that different photon flux. A coupled system results, consisting of two differential equations:

$$\frac{d\,I_1(z)}{d\,z} = -(K+S)\cdot I_1(z) + S\cdot I_2(z) \tag{3}$$

$$\frac{d\,I_2(z)}{d\,z} = (K+S)\cdot I_2(z) - S\cdot I_1(z) \tag{4}$$

This set of equations can be solved both for $I_1(z)$ and $I_2(z)$ with an exponential assumption $\exp(\alpha z)$. This leads to the solutions:

$$I_1(z) = C_1\cdot e^{\alpha\cdot z} + C_2\cdot e^{-\alpha\cdot z} \tag{5}$$

$$I_2(z) = C_1\cdot\beta_1\cdot e^{\alpha\cdot z} + (C_2\cdot\beta_2)\cdot e^{-\alpha\cdot z} \tag{6}$$

with

$$\alpha := \sqrt{K(K+2S)} \quad , \quad \beta_1 := \frac{\alpha+K+S}{S}, \quad \text{and} \quad \beta_2 := \frac{-\alpha+K+S}{S}$$

The constants C_1 and C_2 are coefficients, which result from the boundary conditions. Often selected boundary conditions are that at the position $z = 0$ the photon flux into positive z-direction is equal to the incident light intensity I_0 and that at the position $z = z_0$ the photon flux is zero into negative z-direction:

$$I_1(z = 0) = I_0 \qquad \text{and} \qquad I_2(z = z_0) = 0$$

With these boundary conditions one receives for the constants C_1 and C_2:

$$C_1 = I_0 \cdot \frac{\beta_2 \cdot \exp(-\alpha \cdot z_0)}{\Delta} \quad , \quad C_2 = I_0 \cdot \frac{\beta_1 \cdot \exp(\alpha \cdot z_0)}{\Delta}$$

with the abbreviation

$$\Delta := \beta_1 \cdot \exp(\alpha \cdot z_0) - \beta_2 \cdot \exp(-\alpha \cdot z_0)$$

The remission at $z = 0$ as a function of the layer thickness z_0 results then as follows:

$$R_{z_0} = \frac{I_2(0)}{I_0} = C_1 \cdot \beta_1 + C_2 \cdot \beta_2 = \beta_2 \left[\frac{1 + \exp(-2\alpha z_0)}{1 - (\beta_2/\beta_1) \cdot \exp(-2\alpha z_0)} \right] \tag{7}$$

If there is an additional assumption, that the light propagates in a sample with infinite layer thickness ($z_0 \to \infty$), then eq.(7) is simplified to:

$$R_\infty(\lambda) = \beta_2(\lambda) = 1 + \frac{K(\lambda)}{S(\lambda)} - \sqrt{\frac{K(\lambda)}{S(\lambda)} \left(\frac{K(\lambda)}{S(\lambda)} + 2 \right)} \tag{8}$$

According to eq.(8) the remission spectrum depends with infinite layer thickness not explicitly on the Kubelka Munk parameters $K(\lambda)$ and $S(\lambda)$, but only on the quotient $K(\lambda)/S(\lambda)$. The light attenuation in the diffuse medium is given as:

$$A_{Kubelka}(\lambda) = -\ln \left(\frac{I_R(\lambda)}{I_0(\lambda)} \right) = -\ln \left(R_\infty(\lambda) \right) = -\ln \left(1 + \frac{K(\lambda)}{S(\lambda)} - \sqrt{\frac{K(\lambda)}{S(\lambda)} + 2} \right) \tag{9}$$

Eq.(9) can be transformed to $K(\lambda)/S(\lambda)$:

$$\frac{K(\lambda)}{S(\lambda)} = \frac{1}{2} \left(R_\infty(\lambda) + \frac{1}{R_\infty(\lambda)} \right) \tag{10}$$

The absorption $K(\lambda)$ of the light in the tissue is assumed as:

$$K(\lambda) = K_0 + (c_{Hb} \cdot \varepsilon_{Hb}(\lambda) + c_{HbO_2} \cdot \varepsilon_{HbO_2}(\lambda)) \cdot \ln(10) \tag{11}$$

(K_0: constant basic absorption, c_{Hb}: concentration of desoxygenized hemoglobin ,
c_{HbO_2}: concentration of oxygenized hemoglobin, $\varepsilon_{Hb}(\lambda)$: extinction coefficient of desoxygenized hemoglobin,
$\varepsilon_{HbO_2}(\lambda)$: extinction coefficient of oxygenized hemoglobin)

In this model the light absorption in the tissue consists of two parts:

1. a basic absorption K_0, which is caused for example by bilirubin or melanin
2. an absorption dependent on the wavelength due to of oxygenized and/or desoxygenized hemoglobin

For the scattering $S(\lambda)$ it is assumed that it is a linear function in λ in the wavelength range between 500 and 600 nm.[1]

$$S(\lambda) = S_0 + S_1 \cdot \lambda \tag{12}$$

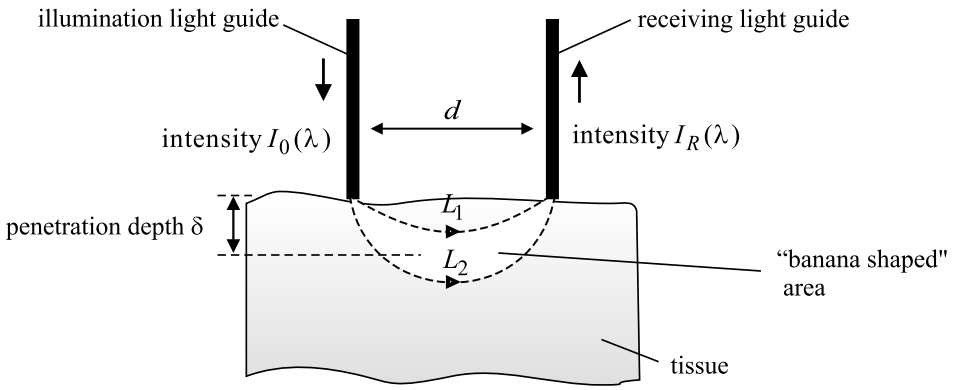

Figure 4. Photon propagation with the FIR-filter-method

Eq.(11) and eq.(12) lead to the following relation with five unknown parameters K_0, S_0, S_1, c_{Hb} und c_{HbO_2}.

$$\frac{K(\lambda)}{S(\lambda)} = \frac{K_0 + (c_{Hb} \cdot \varepsilon_{Hb}(\lambda) + c_{HbO_2} \cdot \varepsilon_{HbO_2}(\lambda)) \cdot \ln(10)}{S_0 + S_1 \cdot \lambda} \tag{13}$$

Since in this equation however only 4 free parameters are present, only the parameters K_0/S_0, c_{Hb}/S_0, c_{HbO_2}/S_0 and S_1/S_0 can be determined by a nonlinear adjustment computation (e. g. by the method of least distance squares):

$$\frac{K(\lambda)}{S(\lambda)} = \frac{\frac{K_0}{S_0} + \frac{c_{HbO_2}}{S_0} \cdot \varepsilon_{HbO_2}(\lambda) \cdot \ln(10) + \frac{c_{Hb}}{S_0} \cdot \varepsilon_{Hb}(\lambda) \cdot \ln(10)}{1 + \frac{S_1}{S_0} \cdot \lambda} \tag{14}$$

From these parameters one receives the searched oxygen saturation SO_2 and the relative hemoglobin concentration rHb using the following equations:

$$SO_2 = \frac{c_{HbO_2}/S_0}{c_{Hb}/S_0 + c_{HbO_2}/S_0} = \frac{c_{HbO_2}}{c_{Hb} + c_{HbO_2}} \tag{15}$$

$$rHb = \frac{1}{S_0}(c_{Hb} + c_{HbO_2}) = \frac{1}{S_0} \cdot tHb \tag{16}$$

2.2. FIR-filter-method

This new represented FIR-filter-method is based on the taylor expansion of a probability function.[6] A photon, which penetrates according fig. 4 into a medium, can be scattered or absorbed there with certain probabilities by particles.

The scattering of the photons is described by the probability function $P_S(\mu_S(\lambda), g, l)$. It is dependent on the scattering coefficient $\mu_S(\lambda)$, the anisotropy factor g and the photon distance l on the trajectory between lighting fibers and receiving fibers. $P_S(\mu_S(\lambda), g, l)\, DS$ indicates the probability that the photon distance lies in absence of absorption in the interval between $[l,\ l + d\, l]$.

The absorption of a photon is described by the absorption coefficient $\mu_A(\lambda)$ of the medium. The higher the absorption coefficient $\mu_A(\lambda)$ is, the more largely is the probability that the photon on the way of the length l is absorbed. The probability P_a, that a photon is absorbed after the distance l, is proportionally to $\exp(-\mu_A(\lambda) \cdot l)$.

The integral of the probability density functions for the scattering P_S and for the absorption P_a over all photon distances l determines the light attenuation $A(\lambda) = -\ln(I(\lambda)/I_0(\lambda))$ of the photon flux between the

lighting light guide $I_0(\lambda)$ and the receiving light guide $I(\lambda)$.

$$\frac{I(\lambda)}{I_0(\lambda)} = \int_0^\infty P_S(\mu_S(\lambda), g, l) \cdot \exp(-\mu_A(\lambda) \cdot l)\ d\,l \tag{17}$$

The distribution function $P_S(\mu_S(\lambda), g, l)$ is not well-known in straying media. For this reason this equation cannot be solved closed. In order to getting at least one approximation for the attenuation of the light in the medium, the eq. (17) is expanded with Taylor at the point $\mu_A(\lambda) = \mu_a^0(\lambda)$. The expansion up to the square term is given as follows:

$$\begin{aligned} A(\lambda) = -\ln \frac{I(\lambda)}{I_0(\lambda)} \quad &\approx \quad A(\mu_a^0(\lambda)) + \left.\frac{\partial A(\lambda)}{\partial \mu_a(\lambda)}\right|_{\mu_a(\lambda)=\mu_a^0(\lambda)} \cdot (\mu_a(\lambda) - \mu_a^0(\lambda)) \\ &+ \frac{1}{2!} \left.\frac{\partial A(\lambda)}{\partial \mu_a(\lambda)}\right|_{\mu_a(\lambda)=\mu_a^0(\lambda)} \cdot (\mu_a(\lambda) - \mu_a^0(\lambda))^2 + \ldots \end{aligned} \tag{18}$$

For the expansion coefficient of the linear term follows:

$$\left.\frac{\partial A(\lambda)}{\partial \mu_a(\lambda)}\right|_{\mu_a(\lambda)=\mu_a^0(\lambda)} = \frac{\int_0^\infty l \cdot P_S(\mu_S(\lambda, g, l)) \cdot \exp(-\mu_a^0(\lambda))\, dl}{\int_0^\infty P_S(\mu_S(\lambda, g, l)) \cdot \exp(-\mu_a^0(\lambda))\, dl} = \langle L \rangle \tag{19}$$

That coefficient of the linear term indicates the medial photon distance $\langle L \rangle$ of the distribution function. For the expansion coefficient of the square term one receives:

$$\frac{1}{2!} \left.\frac{\partial^2 A(\lambda)}{\partial (\mu_a(\lambda))^2}\right|_{\mu_a(\lambda)=\mu_a^0(\lambda)} = \frac{-1}{2} \cdot \left(\langle L^2 \rangle - \langle L \rangle^2\right) = \frac{-1}{2} \cdot \left\langle (L - \langle L \rangle)^2 \right\rangle \tag{20}$$

This coefficient describes the scattering of the medial photon distance $\langle L \rangle$.

With the abbreviation $\Delta\mu_a(\lambda) := \mu_a(\lambda) - \mu_a^0(\lambda)$ the light propagation in the diffuse medium can be described as follows:

$$A(\lambda) = -\ln \frac{I(\lambda)}{I_o(\lambda)} \approx A_0(\lambda) + \langle L \rangle \cdot \Delta\mu_a(\lambda) - \frac{1}{2} \left\langle (L - \langle L \rangle)^2 \right\rangle \cdot (\Delta\mu_a(\lambda))^2 \tag{21}$$

Linear approximation is applicable if the value of the square term is much smaller than of the linear term:

$$\frac{1}{2} \left\langle (L - \langle L \rangle)^2 \right\rangle \cdot \Delta\mu_a(\lambda) \ll \langle L \rangle \tag{22}$$

The condition eq.(22) states that first of all the changes of the absorption coefficient $\Delta\mu_a(\lambda)$ should be small in comparison to the value $\mu_a^0(\lambda)$ and secondly the medial photon distance $\langle L \rangle$ scatter as little as possible. If the requirement for linear approximation eq.(22) is fulfilled in the biological tissue, the attenuation of the light in the tissue can be described in first approximation as follows:

$$A(\lambda) = -\ln \frac{I(\lambda)}{I_0(\lambda)} = A_0(\lambda) + \langle L \rangle \cdot \mu_a(\lambda) \tag{23}$$

with $A_0(\lambda)$: light attenuation in the medium (straying and absorption)

$\langle L \rangle$: medial photon distance between lighting and receiving fibers

$\mu_a(\lambda)$: absorption coefficient

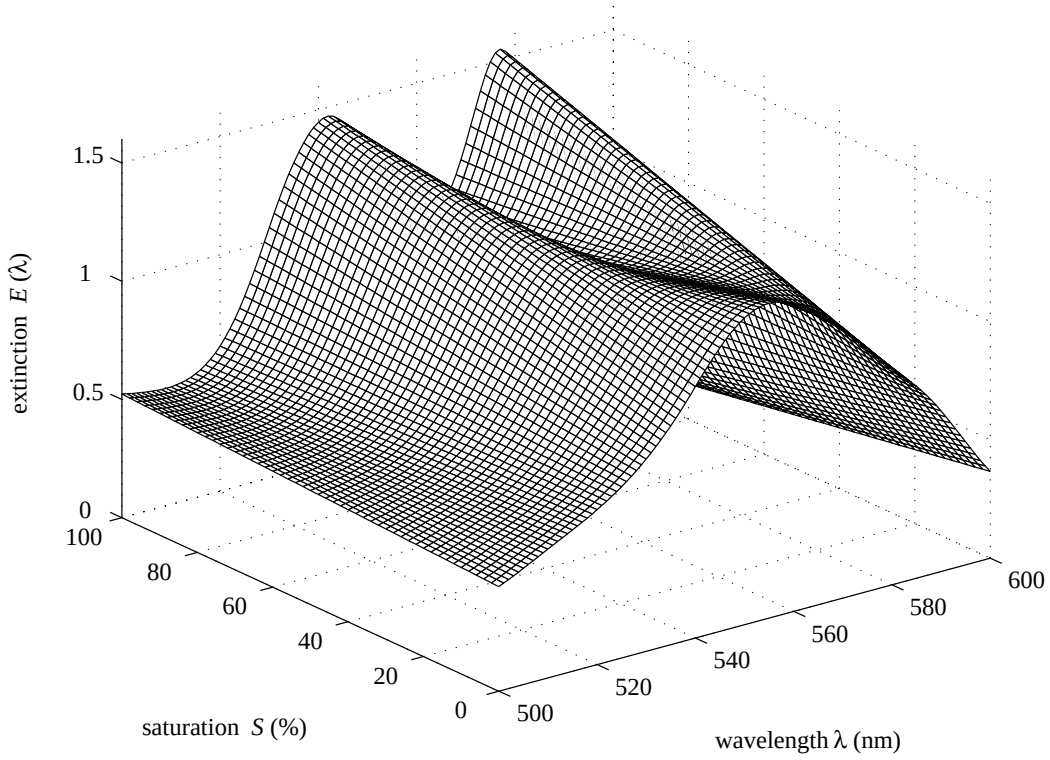

Figure 5. Reference spectra of different oxygen saturations S

For the computation of the local oxygen saturation assumptions for the light attenuation $A_0(\lambda)$ and the absorption coefficient $\mu_A(\lambda)$ in eq.(23) must be made. For the absorption coefficient it is assumed that it consists of the absorption of the oxygenized and desoxygenized hemoglobin:

$$\mu_a(\lambda) := \{c_{Hb} \cdot \varepsilon_{Hb}(\lambda) + c_{HbO_2} + \varepsilon_{Hb}(\lambda)\} \cdot \ln(10) \tag{24}$$

For the light attenuation $A_0(\lambda)$ is assumed that all other light-absorbing and light-scattering components in the tissue within the range between 510 and 590 nm can be described by a linear function in λ:

$$A_0(\lambda) := \alpha_0 + \alpha_1 \cdot \lambda \tag{25}$$

If one inserts eq.(24) and eq.(25) into the eq.(23), then one receives a tissue model function by means of the view of probability, which in the following is called the FIR-filter-method:

$$\begin{aligned} A_{FIR-filter}(\lambda) &:= \alpha_0 + \alpha_1(\lambda) + \langle L \rangle \cdot \{c_{Hb} \cdot \varepsilon_{Hb}(\lambda) + c_{HbO_2} \cdot \varepsilon_{HbO_2}(\lambda)\} \cdot \ln(10) \\ &= \alpha_0 + \alpha_1(\lambda) + \langle L \rangle \cdot E_{ref}^{(S)}(\lambda) \cdot \ln(10) \end{aligned} \tag{26}$$

The parameters α_0, α_1 and $\langle L \rangle$ depend on the biological tissue and describe the influences of the tissue.

$E_{ref}^{(S)}(\lambda)$ describes the extinction spectrum of oxygenized and desoxygenized hemoglobin for a certain oxygen saturation S. One can receive these extinction spectra in good approximation by a linear superposition of oxygenized and desoxygenized hemoglobin. They are represented in fig. 5.

It proved as practicable to describe these values of the reference spectra by an analytic function. With the FIR-filter-functions in such a way specified one receives a max. relative error using a filter order of 36 less than 1 %. Neither by polynomials of n-th degree, nor by rational functions, or trigonometric functions such a good

adjustment can be reached. Therefore in the following the reference spectra are described by the norm of a FIR-filter-function of the order 36[4,5]:

$$E_{ref}^{(S)}(\lambda) = \left| \sum_{\nu=0}^{36} a_{\nu}^{(S)} \cdot \exp\left(-j\frac{\lambda}{\lambda_0} \cdot \nu\right) \right| \tag{27}$$

Each reference spectrum ist given by a set of 37 filter coefficients $a_{\nu}^{(S)}$ $(\nu = 0, 1, 2 \ldots 36)$.

After recording the tissue remission spectrum $I_R(\lambda)$ by means of the micro measuring system in fig. 1 the experimental tissue light attenuation spectrum $A_{exp}(\lambda) = -\ln(I_R(\lambda)/I_0(\lambda))$ is computed. At the beginning of the experiments once the spectrum of the incident light $I_0(\lambda)$ must be recorded. Using the isobestic points the tissue parameters α_0, α_1 and $\langle L \rangle$ are determined. Subsequently, a corrected spectrum can be computed, which only depends on the extinction of the oxygenized and desoxygenized hemoglobin. The experimental FIR-filter coefficients are determined by this corrected spectrum and these filter coefficients are compared with the reference filter coefficients. By minimization of the error squares one receives the searched oxygen saturation SO_2:

$$SO_2 = \frac{c_{HbO_2}}{c_{HbO_2} + c_{Hb}} \tag{28}$$

Additionally with the parameter $\langle L \rangle$ the relative hemoglobin concentration can be calculated:

$$rHb = \langle L \rangle \cdot (c_{Hb} + c_{HbO_2}) \tag{29}$$

2.3. Measurement of a ischaemia with following reactive hyperemia

A good method for realization of different oxygen saturations at living biological tissue is to put a tourniquet at the finger basic joint. With closing this tourniquet the blood supply into the finger and/or to the measuring point is stopped and by the continuous blood oxygen consumption of the tissue the oxygen saturation is lowered. After sufficiently long ischaemia period the tourniquet again is opened and it comes to exceeding influxes of blood into the finger and/or to the measuring point, so that the oxygen saturation increases fast again. Thus, it comes briefly to a in relation to the normal blood circulation condition increased hemoglobin concentration per tissue volume. In this context one speaks of reactive hyperemia.

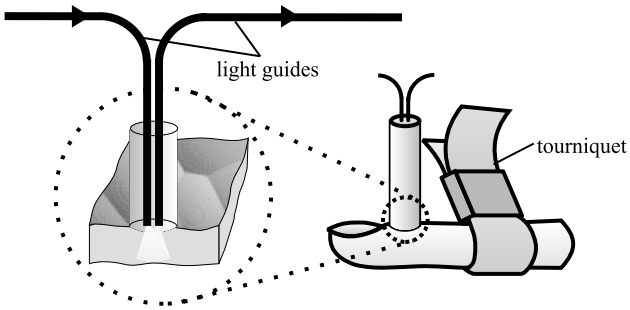

Figure 6. Generating an ischaemia with following reactive hyperemia

The optical fibers are held in a light contact to the skin surface of the finger during the experiment. The remission spectra are recorded with the micro measuring system represented in fig. 1. From the tissue remission spectra the oxygen saturation, the relative hemoglobin concentration and the relative error are computed according to the Kubelka-Munk-method and the FIR-filter-method.

3. RESULTS

In the following a measurement of tissue remission spectra is represented. In the first 60 s the tourniquet is open, afterwards for 300 s the tourniquet is closed, and than for further 240 s it is open again. During this experiment each second a light attenuation spectrum with an average time of 150 ms is recorded. The spectra are represented in fig. 7.

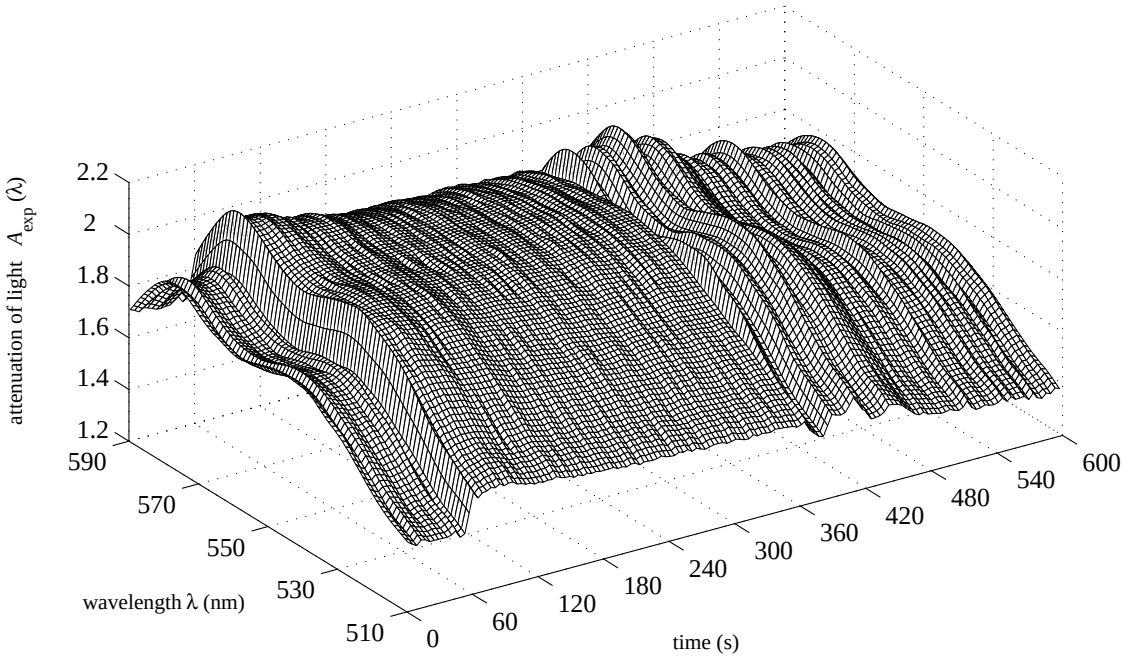

Figure 7. Light attenuation spectra of a ischaemia with following reactive hyperemia

The appropriate computations of oxygen saturations, relative hemoglobin concentrations, and relative errors are represented in the fig. 8, 9 and 10.

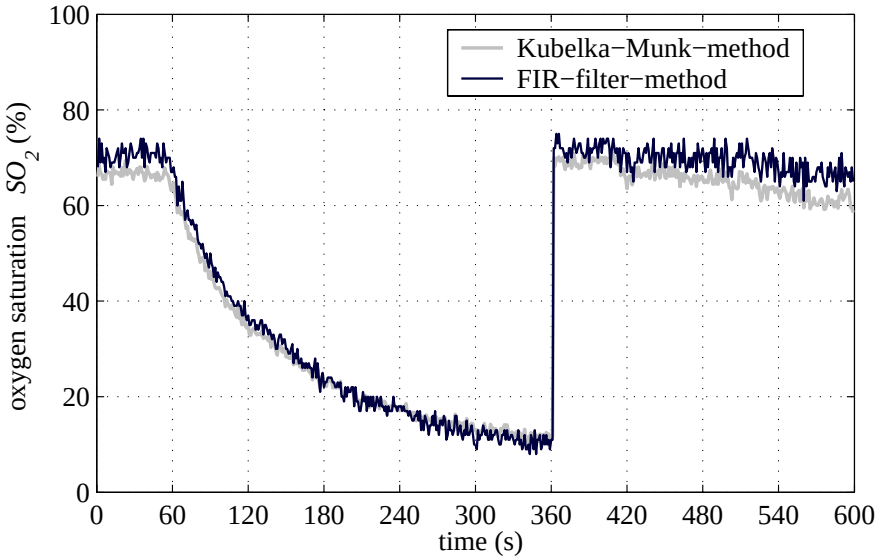

Figure 8. Oxygen saturation of a ischaemia with following reactive hyperemia

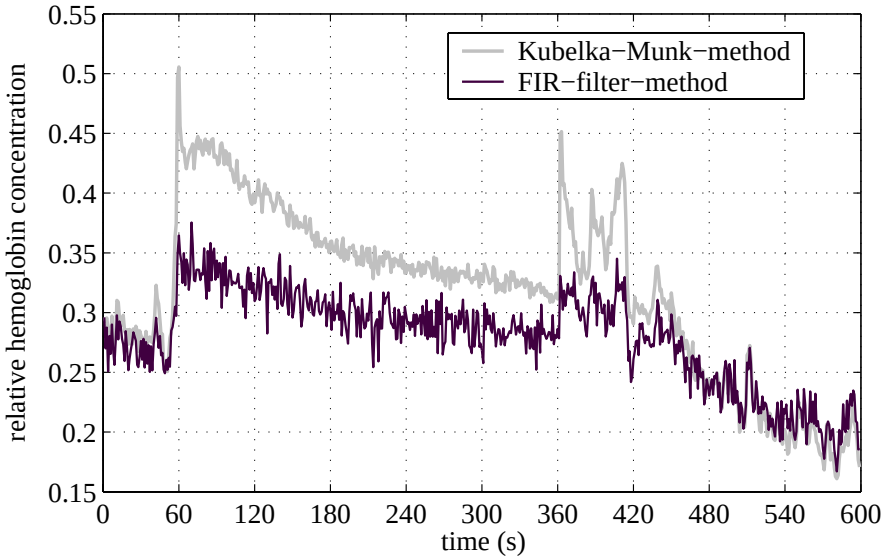

Figure 9. Rel. hemoglobin concentration of a ischaemia with following reactive hyperemia

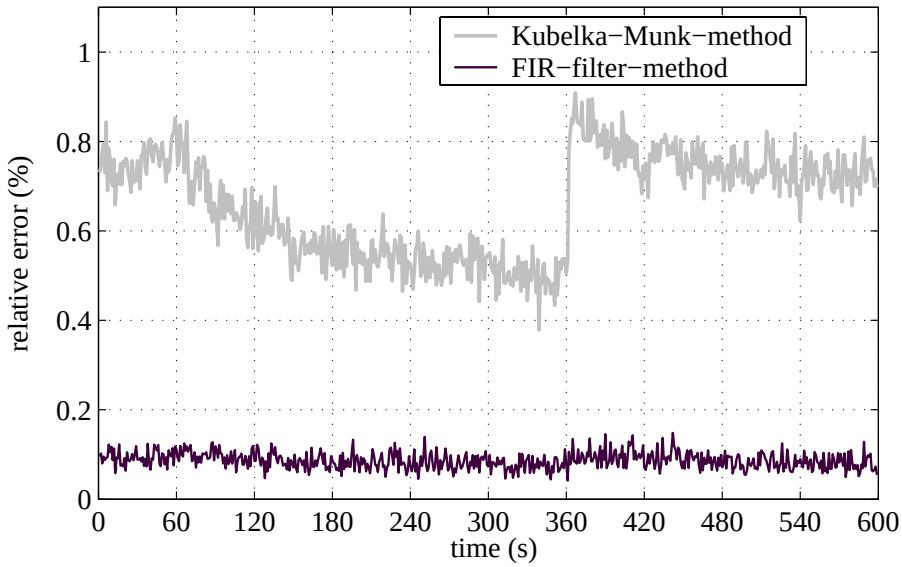

Figure 10. Relative error of a ischaemia with following reactive hyperemia

The average relative error amounts with the Kubelka-Munk-method 0.66 % and with the FIR-filter-method 0,09 %.

4. DISCUSSIONS

The new represented FIR-filter-method has the advantage in relation to the Kubelka-Munk-method that it is based on a more general heuristic assumption. The photon flux with the Kubelka-Munk-method may propagate only in positive or negative z direction. This restriction the FIR-filter-method does not need. Furthermore, the Kubelka-Munk-method needs the condition of infinite layer thickness and that the intensity of the two photon fluxes disappears for $z \to \infty$. The assumption of a constant basic absorption K_0 in eq.(11) does not

correspondent to reality. If one looks at fig. 2, one can see that it is even still a simplified approximation using a linear approximation for melanin and bilirubin within the wavelength range between 510 and 590 nm. For these reasons it is to be expected that the model function of the FIR-filter-method can be adapted better to measurement data than the model function of the Kubelka-Munk-method.

Exemplary this is confirmed by the in-vivo-measurement of a ischaemia with following reactive hyperemia in fig. 10. The average relative error is evidently less by means of the FIR-filter-method with 0.087% than by means of the Kubelka-Munk-method with 0.66%.

At the light attenuation spectra in fig. 7 it is to be seen that after 60 s the amplitude of the spectra increases. This is exactly the time, at which the tourniquet was closed. Thus, briefly more blood was pressed into the fingertip and with it accompanies a higher hemoglobin concentration. Closing the tourniquet excludes the fingertip from further blood and oxygen supply. Due to the oxidative metabolism the tissue uses continual oxygen, which is now extracted from the limited blood volume. It comes therefore to a continuous desoxygenizing of hemoglobin, which becomes visible at a cyanotic discoloration of the skin. A complete desoxygenizing of hemoglobin is not reached, however, since the tissue reduces the metabolism with reduced oxygen partial pressure. Each cell reduces the oxygen consumption and thus ATP production, in order to sustain the aerobe Glycolyse as long as possible. This transition is to be seen also at the cam shape of the spectra. With 60 s one has the typical two peak cam shape, which changes increasingly into a one peak cam shape. After 360 s the tourniquet is opened and it comes to an exceeding influx from blood to the measuring point (reactive hyperemia), so that the oxygen saturation and the hemoglobin concentration increases again.

This already at the cam shape recognizable process of the oxygen saturation, represents the computed values with both methods in fig. 8.

The computed processes of the relative hemoglobin concentration are for both procedures similarly, however with the difference that the changes of the hemoglobin concentration are more pronounced with the Kubelka-Munk-method than with the FIR-filter-method.

The in-vivo-measurement shows that the micro measuring system presented here can be used for recording of tissue remission spectra and the algorithm on the basis of the FIR-filters, developed here, supplies meaningful results. Further investigations with increased experimental expenditure in the context of a field study will be necessary, in order to get secured statements, supported on a multiplicity of data records.

5. CONCLUSIONS

By means of the micro measuring system represented in fig. 1, tissue remissions spectra were recorded during a ischaemia with following reactive hyperemia.

From these tissue remission spectra the local oxygen saturation SO_2 and the relative hemoglobin concentration according to the standard method by Kubelka and Munk and according to the new represented method, which is called FIR-filter-method, were calculated. Both methods were compared and the advantages of the FIR-filter-method were pointed out.

The micro measuring system represented here, can be used thus for measuring tasks in medical diagnostics. Fields of application would be e. g. in the plastic surgery, the transplant and wound control, in the dermatology for evaluation of the tissue by carcinomas or inflammatory changes, in the heart surgery for intraoperative quality control, in the ophthalmology for the retina evaluation, or during the birth for the monitoring of the fetus, because the only accessible tissue is at the head of the fetus and thus only a tissue remission measurement is possible.

With this optical micro measuring system in fig. 1 and with a clearly improved method for the evaluation of tissue remission spectra (FIR-filter- method), which in section 2.2 were represented, the user has new possibilities for the non invasive determination of the local oxygen saturation and the relative hemoglobin concentration.

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