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An optical device to measure blood components by a photoplethysmographic method

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Abstract

The development of the photometric device described here is based on the realization of a photoplethysmography measurement device developed for the German Space Agency DLR. It is well known in biomedical engineering that pulsatile changes of blood volume in tissue can be observed by measuring the transmission or the reflection of light (Roberts 1982 *Trans. Inst. Meas. Control* **4** 101–6). The non-invasive multi-spectral method described here is based on the radiation of monochromatic light, emitted by laser diodes in the range 600–1400 nm, through an area of skin on the finger. After interaction with the tissue the transmitted light is detected non-invasively by photo-diodes. The method makes use of the intensity fluctuations caused by the pulse wave. The ratio between the peak to peak pulse amplitudes measured at different wavelengths and its dependence on the optical absorbability characteristics of human blood yields information on the blood composition. Deferrals between the proportions of haemoglobin and water in the intravasal volume should be detected photo-electrically by signal-analytic evaluation of the signals. The computed coefficients are used for the measurement and calculation of the arterial oxygenic saturation (SaO_2) and the relative haemoglobin concentration change. Results of clinical measurements are presented for a deoxygenation study with ICG-bolus injection (indocyanine green).

Keywords: non-invasive, photoplethysmography, photometric device

(Some figures in this article are in colour only in the electronic version)

1. Introduction

The absorption of whole blood in the visible and near-infrared range is dominated by the different haemoglobin derivates and the blood plasma that consists mainly of water [1, 2]. It is well known that pulsatile changes of blood volume in tissue can be observed by measuring the transmission or the reflection of light. This diagnostic method is called photoplethysmography, PPG [3].

The separation between arterial blood absorption and background absorption (mainly venous blood and tissue water) can be obtained by evaluating the relationship between the pulse signal component (AC part—alternating current) and the DC component (direct current) (figure 1). The DC component is calculated by subtraction of the AC component from the whole PPG signal. The pulsatile change of blood volume is caused by the heart beat [4].

Besides the measurement of oxidized (HbO_2) and reduced haemoglobin (Hb) for the calculation of oxygen saturation (SaO_2) in the arterial blood, the haematocrit value H (volume of red blood cells in whole blood) as well as the haemoglobin concentration are also important medical parameters.

The haematocrit absorption and scattering is influenced mainly by the total haemoglobin concentration. The absorption coefficient μ_a (in mm^{-1}), the scattering coefficient μ_s (in mm^{-1}) and the phase-function $p(s, s')$ are parameters necessary for the calculation of optical properties in turbid media like blood. The phase-function describes the probability

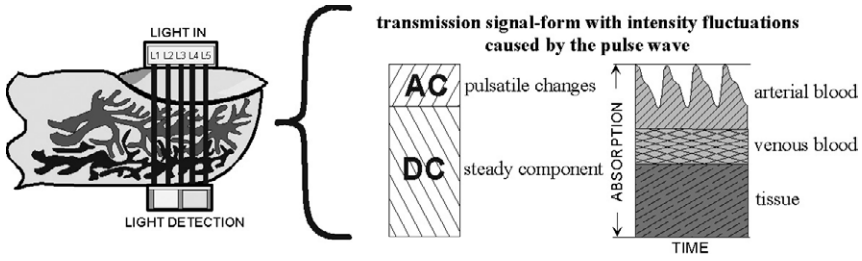

Figure 1. Principle of transmission measurement and signal-form.

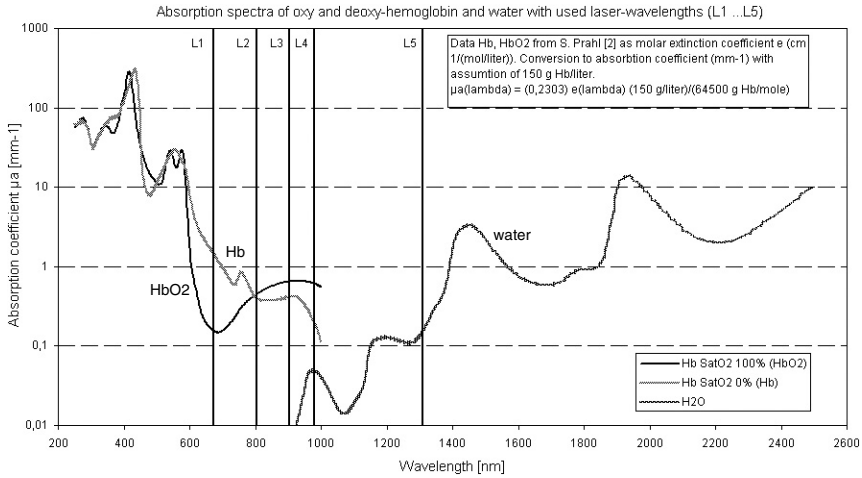

Figure 2. Absorption spectra for haemoglobin (Hb, HbO_2) and water [7].

of scattering for a photon travelling in direction s to be refracted in direction s'. Mathematical calculations can be simplified by using the anisotropy factor $g = E(\cos\ (s, s'))$ instead of the phase-function and the reduced scattering coefficient $\mu_s' = \mu_s(1-g)$ instead of the scattering coefficient. To take the influence of light scattering into account, we assume that the measuring volume is composed of tissue (v_{tis} tissue volume, μ_a^{tis}, $\mu_s'^{tis}$ absorption and scattering coefficient tissue), arterial blood (v^{art} arterial blood volume, μ_a^{art}, $\mu_s'^{art}$ absorption and scattering coefficient arterial blood), and venous blood (v^{ven} venous blood volume, μ_a^{ven}, $\mu_s'^{ven}$ absorption and scattering coefficient venous blood). The model assumes further that the measuring volume can be considered as a homogeneous distribution of scatterers and absorbers of the components mentioned [5]. Therefore, expressions for the coefficients are given in the following form:

$$\begin{aligned}\mu_a^{art} &= \mathrm{HSaO_2}\mu_a^{HbO_2} + \mathrm{H}(1-\mathrm{SaO_2})\mu_a\mathrm{Hb} + (1-\mathrm{H})\mu_a^{H_2O}\\ \mu_a^{ven} &= \mathrm{H}(\mathrm{SaO_2}-\delta\mathrm{SO_2})\mu_a^{HbO_2}\\ &\quad + \mathrm{H}(1-\mathrm{SaO_2}+\delta\mathrm{SaO_2})\mu_a^{Hb} + (1-\mathrm{H})\mu_a^{H_2O}\\ \mu_a &= v^{art}\mu_a^{art} + v^{ven}\mu_a^{ven} + v_{tis}\mu_a^{tis}\end{aligned} \tag{1}$$

and

$$\begin{aligned}\mu_s'^{art} &= \mu_s'^{ven} = \mu_s'^{blood} = H\mu_s'^{Hb}\\ \mu_s' &= v^{blood}\mu_s'^{blood} + v^{tis}\mu_s'^{tis}.\end{aligned} \tag{2}$$

2. Measurement method

The optical parameters of blood and its components depend on many factors, such as the haematocrit value, the oxygen saturation, the flow-velocity, the osmolarity and haemolysis [2].

The objective of the photometric device (PMD) described here is the non-invasive continuous measurement of light-absorbent blood components in the arterial blood of the human finger [6]. This non-invasive multi-spectral measurement method is based on the radiation of monochromatic light, emitted by laser diodes in the range 600–1400 nm, through an area of skin on the finger (figure 1).

The method takes advantage of the intensity fluctuations caused by the pulse wave. The ratio of the relative changes of the pulse sizes, when measured at different wavelengths after transmission through a finger, is directly related to the absorbance characteristics of blood components (figure 2).

After interaction with the tissue the transmitted light is detected non-invasively by photo-diodes. Figure 2 shows the absorption spectra for the main blood components and the wavelengths L1 to L5 of the five laser diodes of the PMD system. Suitable wavelengths were selected for the analyses of SaO_2 and relative haemoglobin/haematocrit concentration change.

Four of the five laser diodes emit light in the range of wavelengths between 600 and 1000 nm (670, 808, 905 and 980 nm).

This is the so-called therapeutic window region, in which the blood absorption is dominated by the haemoglobin derivatives [8]. At 980 nm, besides the haemoglobin absorption a weak absorption band also exists for water. An additional 1310 nm laser diode was integrated; at this wavelength the absorption of water is dominant (figure 2).

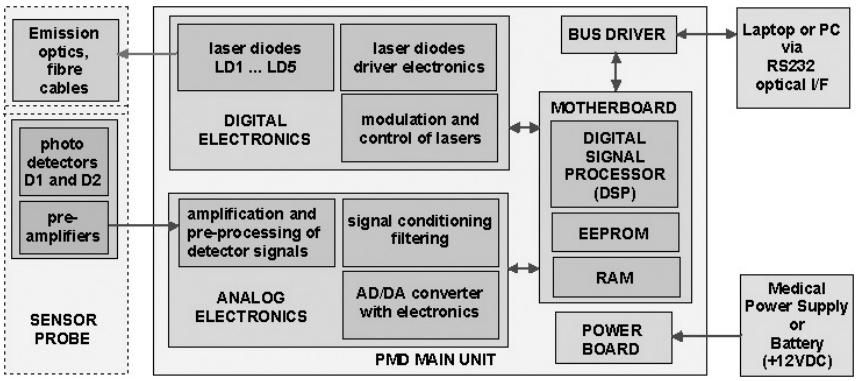

Figure 3. Block diagram of the photometric device (PMD).

The measurement method evaluates the waveforms of peaks, troughs, DC averages, and pulsatile averages (AC part). For a calculation of haemoglobin/haematocrit, the wavelengths are chosen to suit the absorbance peaks of water in blood [9] where the two components of blood have differing amounts of water (980 and 1310 nm).

To find a value corresponding to an isobestic point for absorbance of oxyhaemoglobin and deoxyhaemoglobin, a wavelength of 808 nm is chosen. A second relationship for the measurement and correction of oxygen saturation is calculated with the 670 nm (absorbance of deoxyhaemoglobin greatly exceeds the absorbance of oxyhaemoglobin) and 905 nm (absorbance of oxyhaemoglobin greatly exceeds the absorbance of deoxyhaemoglobin) transmission signals.

3. Photometric measurement device

Inside the measurement device the laser diodes are integrated together with the required control electronics (figure 3).

The device electronics consists of the components required for signal amplification, digitalization, and triggering of the laser diodes, which operate in a pulse mode. Each laser receives an activation signal every 8.5 ms and is in on-mode for 850 μs.

The sample frequency of the system is about 7 kHz. After software mean value calculations and subtraction of the dark-current inside the main unit, the transfer of the five photocurrents is achieved with a sample rate of about 100 Hz each.

A main component of the measurement device is a high performance DSP system with the floating-point processor TMS320C32, flash and memory. This enables DSP software-control and time-multiplexed operation of the five lasers and control of each of the two receive channels.

The evaluation of the data and the operation of mathematical algorithms for pre-processing, e.g. digital filtering and averaging, are achieved by using the DSP software. The data viewing and storing is achieved via a serial RS232 I/F connection on a laptop or personal computer. The application software is LabView® programmed.

The laser light is transmitted to a special optical transmission head by means of optical fibres inside the sensor probe (figure 4).

Two photo-detectors D1 and D2 are also contained in the sensor head together with the required pre-amplifiers;

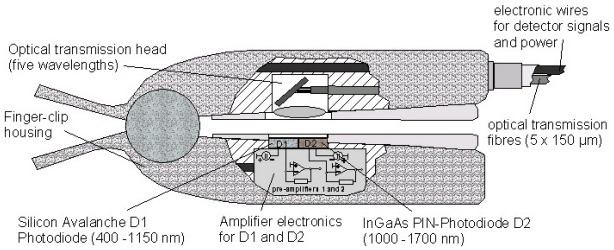

Figure 4. Schematic diagram of the PMD probe design.

the sensor signals detected here will be processed inside the measurement device.

To detect the transmission signals of lasers 1–4 (670–980 nm) a silicon avalanche photodiode is used with a spectral sensitivity of 400–1150 nm. For detection of the 1310 nm transmission signal an InGaAs-photodiode is required with a spectral sensitivity of 1000–1700 nm.

4. Applications and results

Previous measurements of the transmission signals of the five wavelengths had shown an apparent variation of the arterial pulse. The signal quality was sufficient to analyse the signal components and to calculate the relative attenuation coefficients of the arterial blood. With regard to the components at 1310 nm an evaluation of the relative portions of haemoglobin and water in the blood is feasible.

The measurement technique requires a pulse signal for the calculation of the relative attenuation coefficients. Vasoconstriction at the extremities can be a problem, as it decreases the signal amplitude, and therefore the signal to noise ratio. A small signal amplitude tends to give inaccurate results [10]. The PMD has, therefore, a minimum signal amplitude below which no value for the calculated coefficient is displayed. The lower limit for the pulse amplitude with the 1310 nm laser is of the order of 0.2% of the measured intensity. This may be a limitation when using the system on various patient groups (vascular disease, Raynaud's phenomenon, shock, etc).

Figure 6 shows the PMD time signals after the passage of the laser light through the finger. The measurements were performed with a person lying calmly in a horizontal position at room temperature.

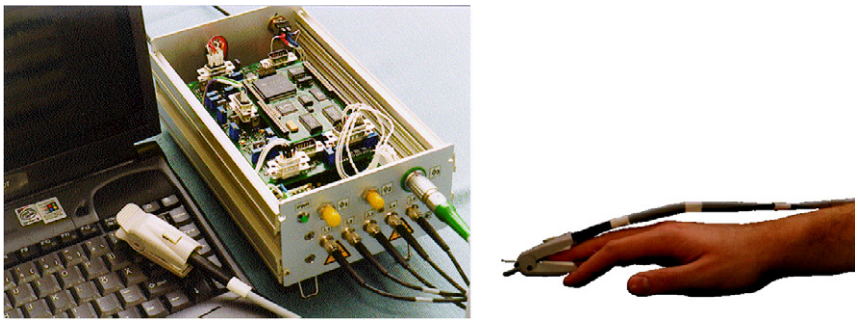

Figure 5. Photo PMD with sensor clip and application of the sensor on finger.

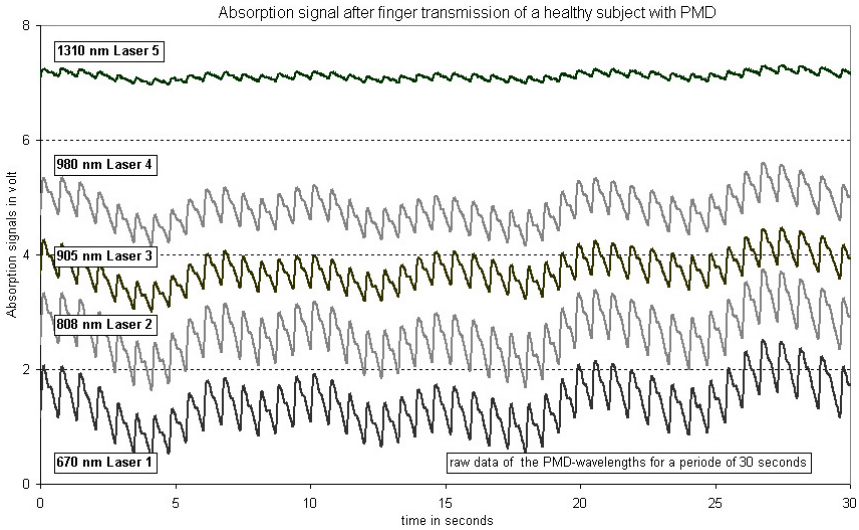

Figure 6. Pulse-wave signals of a healthy and dormant person for all five wavelengths.

The transmission values of the pulse waves processed are constant with breath-dependent periodical oscillations for all five laser wavelengths.

As an external reference measurement, a POX10 Oximeter device (Medlab GmbH, Germany), was used. The oximeter values (POX10) for the time window shown in figure 6 were 98% for SaO_2 and 89 bpm (beats per minute) for the heart rate.

The PMD is suitable for non-invasive continuous on-line monitoring of one or more biologic constituent values. The objective of this development is to reduce the dependence on measurement techniques which require that a sample of blood be withdrawn from the patient for *in vitro* analysis. Any invasive method used on the patient to obtain blood is accompanied by problems of inconvenience, stress, and discomfort. The patient is also exposed to the normal risks of infection associated with such invasive methods.

The non-invasive measurement method described in this paper might be applicable for clinical applications where an invasive method is undesirable or inconvenient. One particular application could be the monitoring of patients' vital signs in critical care medicine or anaesthesia. Another application might be in the monitoring of patients who are undergoing surgery, where presumably the loss of blood during surgery would produce a change of haemoglobin concentration. It may also be a useful tool during dialysis sessions for the monitoring of haemodialysis patients with end-stage renal failure [11]. By using a dialyser (haemofilter) the patient has dialysate (prevailing water) distracted. This deferral means a fluid reduction for the patient during the ultra-filtration. The change in blood volume involves a change of the haematocrit status.

5. Measurements and validation

It is necessary to compute a correction factor and the influence against deferrals in the arterial oxygen saturation for a photometric non-invasive measurement of haematocrit/haemoglobin. During a hypoxia study the device sensitivity was validated for SaO_2. The study took place in collaboration with the Institute of Biomedical Engineering at the University of Luebeck. Prior approval for the study was given by the medical ethics board.

The pulsatile changes in the intensity observed with the PMD were caused by changes in arterial blood volume. It has thereby been assumed that the arterial blood volume fluctuations do not introduce pulsatile changes of the venous (and capillary) blood volume fraction [12, 13]. The PPG signal intensity I of the 670 and 905 nm wavelength signals from PMD is used to compute the SaO_2 value. The following relations are used for the calculation of the coefficient C_{SaO2}, which correlates with the arterial blood oxygen saturation.

$$C_{SaO_2} = \frac{\Delta \ln[I(670\ \text{nm})]}{\Delta \ln[I(905\ \text{nm})]} \tag{3}$$

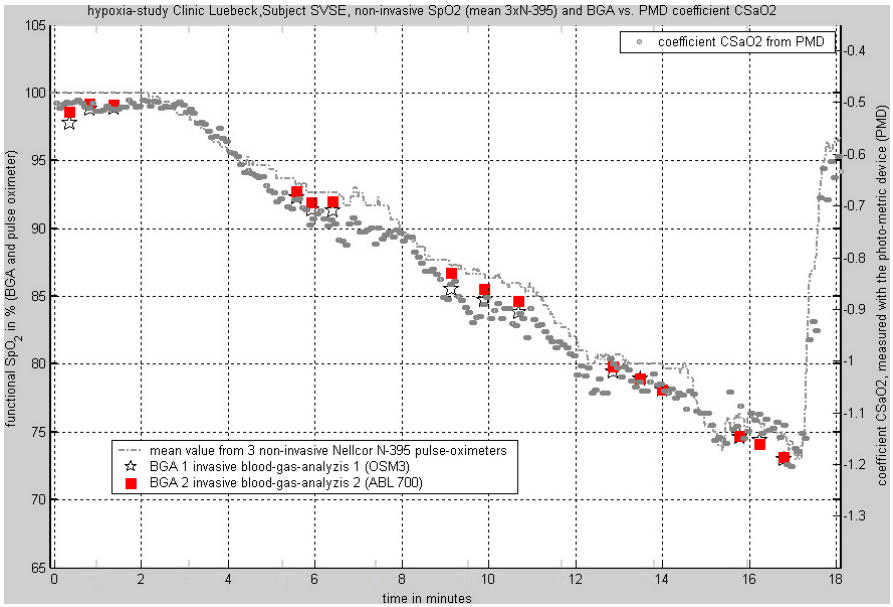

Figure 7. C_{SaO2} from PMD, compared with invasive BGA data and commercial pulse oximeter data during a hypoxia measurement.

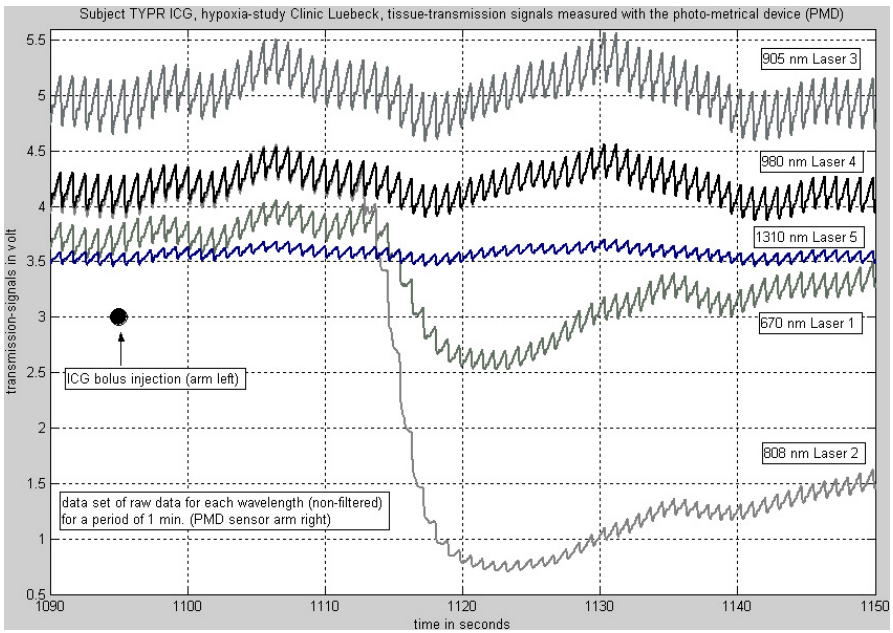

Figure 8. PMD signals after injection of an ICG bolus.

with

$$\Delta \ln(I(\lambda)) = \frac{\Delta I}{I} = \frac{\mathrm{AC}}{\mathrm{DC}}. \tag{4}$$

The intensity I corresponds with DC, the direct current or average part of the measured intensity, whereas the fluctuations ΔI correspond to AC, the alternating current or fluctuating part of the measured intensity. The coefficient C_{SaO2} can therefore be obtained from the measurement of the AC and DC components of the red (670 nm) and infrared (905 nm) intensity signals, as well as by the measurement of $\Delta \ln(I(\lambda))$ for these wavelengths [14].

Figure 7 shows a measurement during the hypoxia study for one subject. The arterial oxygen saturation was reduced to about 75%. Thereby the recorded data of the photometric device PMD was compared with the data of the blood-gas analysis BGA from the *A. radialis* (arterial oxygenic saturation—SaO_2 in per cent).

The results for four subjects showed a high sensitivity and high reproducibility for all measurements with the photometric device.

The PMD was also tested for a monitoring of indocyanine green (ICG) bolus injections. Indocyanine green is in general used for recording dye dilution curves, in particular for the determination of cardiac output. The principal advantages causing the rapid acceptance of the dye were the presence of an absorption maximum near the isobestic point of deoxyhaemoglobin and oxyhaemoglobin around 800 nm, the confinement to the vascular compartment by binding to plasma

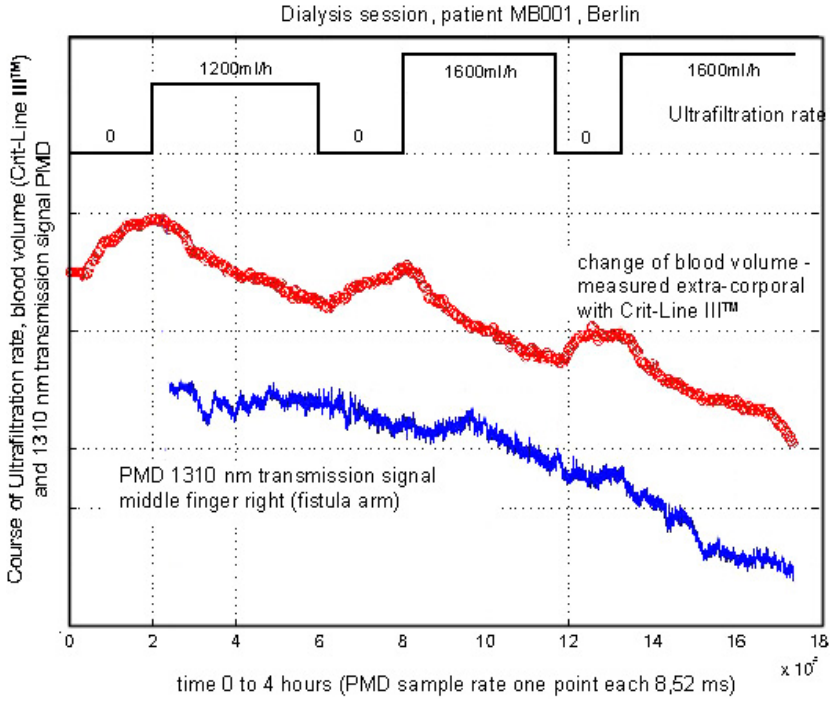

Figure 9. Measurement with the PMD during a dialysis session.

proteins, the very low toxicity, and the rapid excretion, almost exclusively into the bile [15].

Figure 8 shows the transmission signals of the five PMD laser wavelengths for a subject (ring finger, right arm), after the injection of an ICG bolus in the left arm.

The decrease in transmission signal intensities at 670 and 808 nm are caused by the high absorbability of ICG at these wavelengths. The other PMD laser signals at 905, 980 nm and 1310 nm remain unaffected by the ICG bolus.

The measurement method of the photometric device was also tested on haemodialysis patients with end-stage renal failure during dialysis sessions (figure 9).

In these studies the continuous recorded data of the photometric device was compared with the blood volume changes during the dialysis sessions.

The time signal of photo-current for the 1310 nm laser diode light transmission after its passage through the right middle finger (fistula arm) of a patient during a dialysis session is illustrated in the lower curve of figure 9. The ultra-filtration profile and rate of the dialysis is represented by the upper curve in figure 9. The change of the blood volume measured in the extra-corporal circuit of the dialysis device with the Crit-LineIII^{TM} device (In-Line Diagnostics Corp., Kaysville, UT, USA) is displayed in the middle curve of figure 9.

The time signal of the transmission value for 1310 nm shows an analogue course similar to the blood volume change during the dialysis of the patient.

6. Conclusions

In this paper a multi-wavelength photometric measurement method that provides non-invasive *in vivo* photoplethysmographic and spectral measurements in human blood and tissue has been described. A newly developed PMD device has been introduced that is able to measure PPG signals continuously at five different wavelengths from 670 nm up to 1310 nm.

The fact that the absorption coefficients μ_a and scattering coefficients μ_s for blood differ at different wavelengths has been exploited and is used for the calculation of the optical absorbability characteristics of human blood, yielding information on the blood composition.

In the first clinical measurements of the new measurement system, with injections of the green marker colour ICG, high sensitivity and spectral selectivity were demonstrated. A trial study to measure hypoxia showed that the sensitivity of the system for measurement of SpO_2 levels was very good.

The potential of this photometric method to measure changes in the tissue and blood water content was shown during a dialysis trial, and the possibility of non-invasive haematocrit and haemoglobin measurements with the system were proved. Future work will involve further clinical studies, optimization of the photometric measurement device, and evaluation of suitable statistical analysis algorithms.

References

[1] Roberts V C 1982 Photoplethysmography—fundamental aspects of the optical properties of blood in motion *Trans. Inst. Meas. Control* **4** 101–6

[2] Roggan A, Friebel M, Dörschel K, Hahn A and Müller G 1999 Optical properties of circulating human blood in the wavelength range 400–2500 nm *J. Biomed. Opt.* **4** 36–46

[3] Kamal A A R, Hatness J B, Irving G and Means A J 1989 Skin photoplethysmography—a review *Comput. Methods Programs Biomed.* **28** 257–69

[4] Woods A M, Queen J S and Lawson D 1991 Valsalva maneuver in obstetrics: the influence of peripheral circulatory changes on function of the pulse oximeter *Anesth. Analg.* **73** 765–71

[5] Niemz M H 1996 *Laser–Tissue Interaction* (Berlin: Springer)

[6] Kraitl J, Matz H, Ewald H and Gehring H 2003 Development of the diagnostic module—an optical measurement device for non-invasive determination of haemoglobin content in human blood *Sensors and their Applications XII, IOP Conf. (Limerick, Ireland)* (Bristol: Institute of Physics Publishing) pp 379–83 (ISBN 0-7503-0978-4)
[7] Prahl S *Optical Absorption of Hemoglobin* Oregon Medical Laser Center http://omlc.ogi.edu/spectra/
[8] Sadar D K and Levy L B 1998 Optical properties of whole blood *Lasers Med. Sci.* **13** 106–11
[9] Matcher S J, Cope M and Delpy D T 1993 Use of the water absorption spectrum to quantify tissue chromophore concentration changes in near-infrared spectroscopy *Phys. Med. Biol.* **38** 177–96
[10] Yoshida I, Shimada Y, Oka N and Hamaguri K 1984 Effects of multiple scattering and peripheral circulation on arterial oxygen saturation measured with a pulse-type oximeter *Med. Biol. Eng. Comput.* **22** 475–8
[11] Wabel P, Chamney P, Krämer M, Leonhardt S and Isermann R 1999 Patient-parameter identification during dialysis sessions *J. Int. Fed. Med. Biomed. Eng.* **37** (suppl. 2) 58–9
[12] Yoshida I, Shimada Y and Tanada K 1980 Spectrophotometric monitoring of arterial oxygen saturation in the fingertip *Med. Biol. Eng. Comput.* **18** 27–32
[13] Wukitsch M W, Petterson M T, Tobler D R and Pologe J A 1988 Pulse oximetry: analysis of theory, technology and practice *J. Clin. Monit.* **4** 290–301
[14] Payne J P and Severinghaus J W 1986 *Pulse Oximetry* (Berlin: Springer) pp 188–91
[15] Landsman M L J, Kwant G, Mook G A and Zijlstra W G 1976 Light-absorbing properties, stability, and spectral stabilization of indocyanine green *J. Appl. Physiol.* **40** 575–83