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Skin remittance kinetics in the spectral range of 550–790 nm

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Abstract. **An experimental method for remitted photon time-of-flight (PTOF) estimation in human skin *in vivo* is developed and tested. Remitted light signals are obtained using a picosecond broadband laser and a set of narrowband interference filters centred at 550, 600, 650, 700, 750 and 790 nm. Four different distances of 8, 12, 16 and 20 mm between the source and detector fibres are used. Measurements are performed at different wavelengths and distance combinations. Direct kinetic measurements make it possible to assess the mean photon path length in skin at various spectral regions. Results generally correspond to theoretical expectations; however, no clear dependence on wavelengths in the spectral range 650–790 nm is observed. This study shows a promising opportunity to use this kind of measurement system in the future research of PTOF in human skin.**

Keywords: *light scattering in skin, photon time-of-flight, optical path length, picosecond laser applications.*

1. Introduction

Light penetration depth in tissues and related photon scattering path length are important characteristics in optical diagnostics and therapy. Composition and density of chromophores influence the scattering path length of diffusely reflected (remitted) photons in various tissues, including human skin [1]. The remitted photon path length in skin at particular wavelengths is an essential parameter for clinical applications, e.g. for mapping of the chromophore concentration distribution in skin malformations [2–9].

The photon path length in tissues can be estimated theoretically using model calculations, e.g. Monte Carlo method [10]. However, a real tissue structure at specific body locations may not correspond to the model assumptions and therefore could lead to mistaken results.

Alternatively, direct photon time-of-flight distribution measurements allow estimation of scattering path length in the examined tissue if the refraction index of tissue is known. In particular, the remitted photon input–output time delay measured in classical double-fibre experiment [7] gives sense about the duration of photon travel in skin structures before it is back-reflected.

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PTOF spectroscopy has been widely used for *in vitro* studies of pharmaceuticals and turbid materials [7, 8], but there are only few *in vivo* PTOF studies in living tissues. In one of these studies, *in vivo* PTOF measurements of the skin excited at 1064 nm were performed [9]. The authors found the dependence of PTOF on the locations of the body, skin roughness, water diffusion, and no dependence on the age, sex, or measurement geometry (parallel or perpendicular to the Langer lines). *In vivo* PTOF measurements have been also performed at 405 and 510 nm in skin (phototype II by Fitzpatrick classification) and intradermal nevi [9]. A difference in PTOF distributions for healthy skin and nevi has been found. Such *in vivo* studies are clinically important and can contribute to improvements of algorithms for Monte Carlo simulations [10, 11] and other algorithms used for skin optical diagnostics [2, 3].

The aim of this work is to design new PTOF system to make *in vivo* skin measurements at several wavelengths in the range from 550 to 790 nm, with separated source and detector optical fibres and changeable distances between them. Similar existing time domain systems are used in diffuse optical tomography measurements [12–15].

In this work, an advanced picosecond laser system for PTOF *in vivo* measurements in human skin has been developed and experimentally tested on 8 volunteers. The system allows changing the wavelength of input pulses and the distance between emitting and receiving fibres at constant probe pressure to skin. The remitted pulse parameters at different combinations of wavelengths and the source–detector distances were analysed.

2. Method and materials

The proposed system is shown in Fig. 1. A broadband picosecond laser (Whitelaser micro supercontinuum lasers, Fianium, NKT Photonics, Denmark; pulse full width at half maximum (FWHM), 6 ps; repetition rate, 20 MHz) was used as a light source which emits in the spectral range from 400 nm to 2000 nm.

Spectrally narrowband input light was ensured by a set of bandpass optical filters (Andover Corporation, USA, mod. 500FS10-25, 550FS10-25, 600FS10-25, 650FS10-25, 700FS10-25, 750FS10-25, 790FS10-25, USA) with half-bandwidth of 10 nm and centre wavelengths of 500, 550, 600, 650, 700, 750, and 790 nm. For additional spectral filtering a grating monochromator with resolution 1 nm was used to exclude detection of skin fluorescence signals. To ensure optical signal between the input and output fibres (WF-400, Light Guide Optics International, Latvia; silica core diameter, 400 μm; and length, 1.05 m) at various distances between them, a spe-

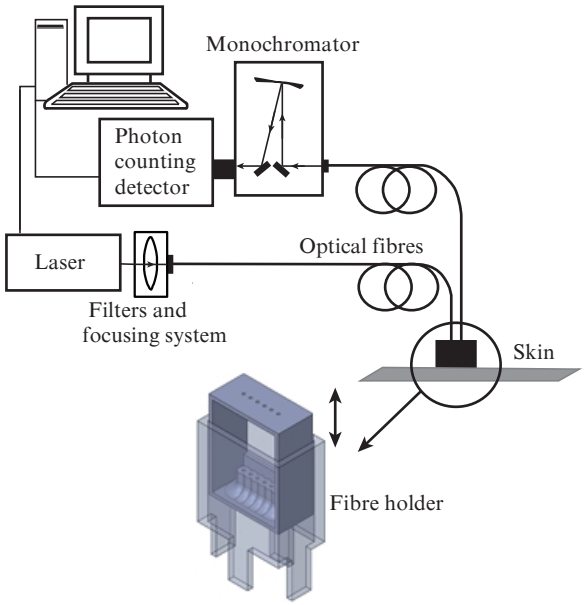

Figure 1. Setup for skin remission kinetics measurements.

cial fibre holding system was designed (Fig. 1) with inter-fibre distances of 8, 12, 16, and 20 mm.

A photon counting detector (PMC-100-4 photomultiplier) combined with a detector controller (DCC-100) and a data processing card (SPC-150, all Becker&Hickl GmbH, Germany) were used to process detected single-photon signals.

The goal of this pilot study was to measure shapes of the input and skin-remitted pulses in order to estimate the duration of passage of photons in forearm human skin and its wavelength dependence in the spectral range of 550–790 nm. The data were collected for each selected wavelength by changing the distance between fibres in the skin contact probe. To ensure equal pressure on the skin surface at all measurements, the probe was designed as a lift where the inside sliding part (with selected distances between the two fibres) lies on the skin providing a pressure determined by its weight, ~35 g cm^{-2} (the fibre's holding system is shown in Fig. 1). The outside part of the probe was fixed on the skin during the measurements.

Before *in vivo* measurements, the instrumental response function (IRF) was recorded (Fig. 2) by positioning the emitting and receiving fibres against each other. It determined the time scale for further measurements. Each measurement data was calculated as a mean value of three successively measurements, each 30 s long.

The dependences of the time delay of the pulse peak positions and the value of the full width at half maximum (FWHM) of signals on the wavelength and distance between optical fibres were analysed.

A special programme for signal processing was created in MatLab. The differences of both parameters were calculated for each measurement. The error of each measurement was calculated as a combination of the measurement setup error (approximately 6 ps) and quadric deviation of the average skin measurement value.

The measurements were performed on the left volar forearms of 8 volunteers aged from 25 to 68 with skin phototype II and III (Fitzpatrick classification) under permission of the

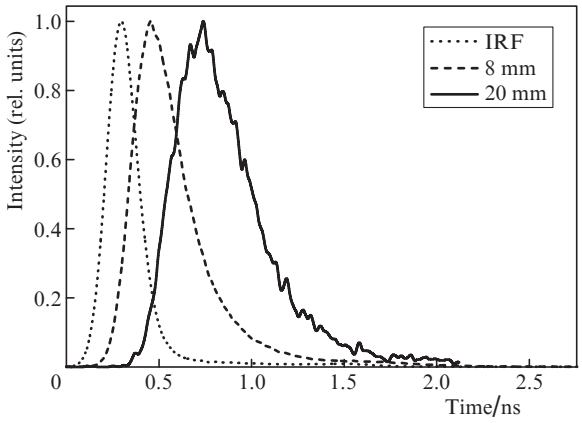

Figure 2. IRF signal and two remitted signals at λ = 650 nm from skin for 8 and 20 mm inter-fibre distances.

local Ethics Committee with written consent of the volunteers. The average spectral power density on skin was ~10 mW cm^{-2}, i.e. well below the skin laser safety limit of 200 mW cm^{-2} [16].

3. Results and discussion

Figures 3 and 4 illustrate the comparative results of input and output pulse parameters, i.e. the dependence of the time delays of pulse peaks (Δt) and differences of the FWHM values (Δ) on the distance between input and output fibres at different wavelengths, averaged by all volunteers. As expected, an increased distance between the fibres has led to broadening of the remitted pulse with a corresponding time delay of the pulse peak [5, 17].

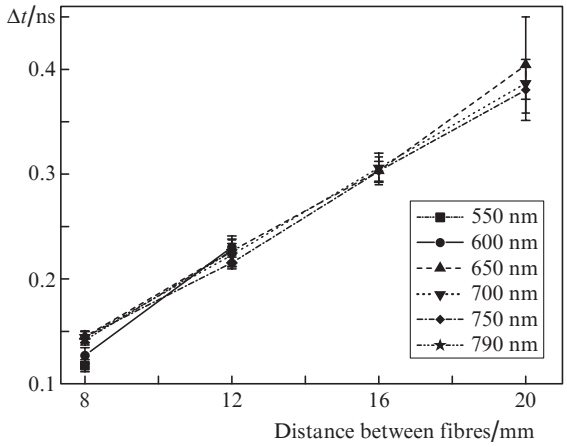

Figure 3. Dependence of the mean value of pulse peak time delays on inter-fibre distances for six particular wavelengths in the range of 550–790 nm.

We assumed that the values of Δt and Δ would notably increase at wavelengths longer than 600 nm when haemoglobin absorption significantly decreases. Data from a single volunteer shows the expected increase (Figs 5. and 6), but not for all distances between fibres because the remitted signals at 550 and 600 nm were extremely weak, comparable to noise.

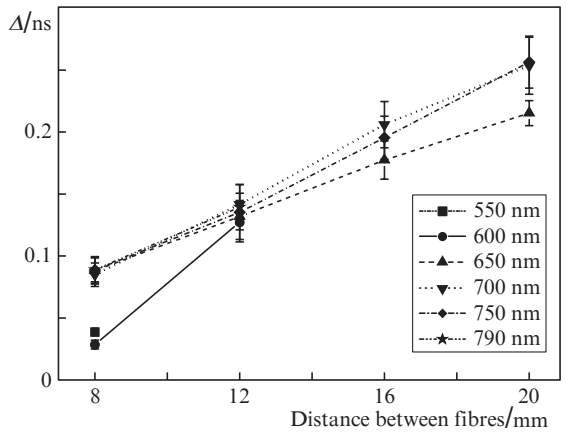

Figure 4. Dependence of the difference between the FWHM mean values of IRF and skin-remitted laser pulses on inter-fibre distances for six particular wavelengths in the range of 550–790 nm.

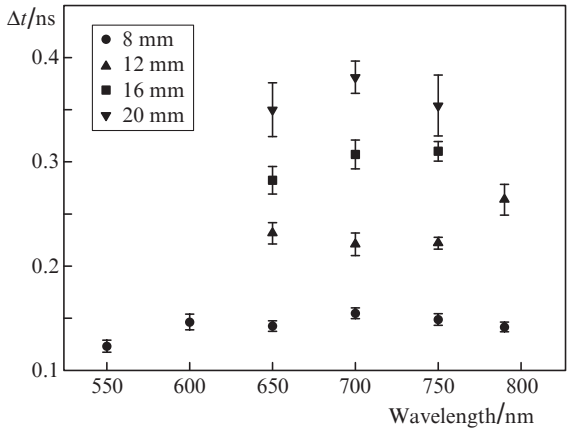

Figure 5. Dependence of pulse peak time delay (Δt) on wavelength at various inter-fibre distances for a single volunteer.

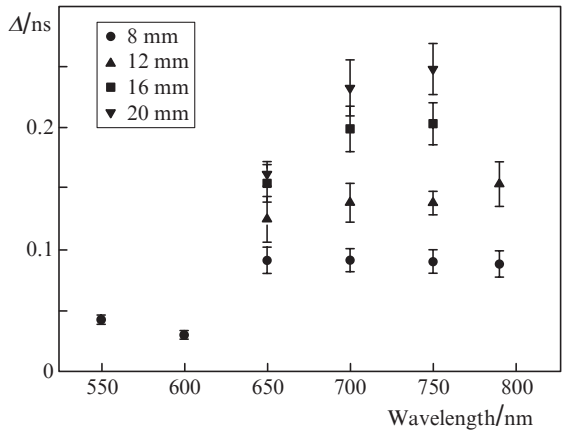

Figure 6. Dependence of differences between IRF and skin-remitted pulse FWHMs on wavelengths at various inter-fibre distances for a single volunteer.

The shortest photon path lengths in skin were calculated from the differences of arrival times of the first photons (i.e. very beginning of the pulse raising front) of the IRF and skin-remitted signals; refractive index $n = 1.36$ [10] was used in calculations. Calculation of path length of the 'first' backscattered photons is based on the time delay between the moments related to 10% level of the signal maximum at raising fronts of the IRF and skin-remitted pulses. Figure 7 shows the obtained 'first' photon path lengths at the inter-fibre distance of 8 mm for various wavelength bands.

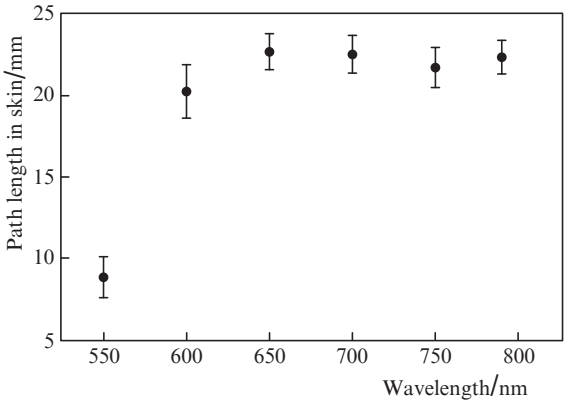

Figure 7. Path length of the 'first' backscattered photons for six particular wavelength bands at inter-fibre distance of 8 mm for a single volunteer.

The obtained shortest photon path lengths at longer inter-fibre distances have exceeded them for about 2.5 times. Reasons for this kind of difference could be explained with intensive absorption of oxy-haemoglobin (HbO_2) and deoxy-haemoglobin (Hb) [18]. The absorption of haemoglobin is much higher at 550 nm than in the 600–790 nm range; therefore, a signal for 550 nm is very weak and only few photons with 'ballistic' trajectories were registered. Meanwhile for 650 nm and longer wavelengths less photons are absorbed, and so as a result of scattering photons travel via deeper layers of skin with correspondingly longer path lengths.

4. Conclusions

An advanced picosecond pulse laser system for skin-remitted light kinetics measurements has been assembled and tested. It allowed using various wavelength bands at different distances between the input and output optical fibres. The pilot measurements from forehands of 8 volunteers provided the dependences of three parameters (input–output pulse peak delay, difference of pulse FWHM and arrival times of the 'first' photons).

The obtained results show general agreement with expectations based on skin blood haemoglobin absorption properties. However, no clear dependences on wavelengths of the above-mentioned parameters in the spectral range 650–790 nm have been observed. The signal-to-noise ratio of the measured signals was relatively low, especially at 550 and 600 nm spectral bands. To extract the skin-remitted photon path length distributions from the measured signals, sensitivity of optical pulse registration should be improved, first of all by reducing signal losses in the optical system. Further studies should be focused on improved signal processing, including calculations of the scattering-related transfer functions (responses to ultra-narrow delta-pulses) of the skin-remitted light to provide more detailed information how the input signal transforms when passing through the skin structures

[19, 20]. It would open the possibility to directly determine the distributions of the photon path length in human skin at particular wavelengths. Besides, a larger data set involving different skin types, ages and locations on the body should be collected for obtaining more extensive and precise experimental results.

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