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In vivo optical characterization of human prostate tissue using near-infrared time-resolved spectroscopy

Tomas Svensson
Stefan Andersson-Engels
Lund University
Department of Physics
SE-221 00 Lund
Sweden

Margrét Einarsdóttir
Katarina Svanberg
Lund University Hospital
Department of Oncology
SE-221 00 Lund
Sweden

1 Introduction

Interest in optical characteristics of the human prostate is mainly related to recent efforts in developing photodynamic therapy (PDT) into a modality for treatment of localized prostate cancer. Initial preclinical work on PDT of prostate carcinoma appeared during the 1980s (Refs. 1 and 2) and (first) clinical work was published as early as 1990. For the development and optimization of the technique, several studies have since been performed on canine and rat models. Clinical results from various research groups are now available. Altogether, these results have shown a great potential of PDT in the management of prostate cancer.

PDT, in general, relies on a process where light excites (i.e., activates) a photosensitizer, which induces cytotoxic oxygen species. Thus, PDT relies on the presence of light, a photosensitizer, and oxygen. The corresponding dosimetry is therefore a complex matter, involving measurement and prediction of sensitizer and oxygen concentrations as well as light fluence (light dose). Since full treatment of diseased regions is crucial, these issues must be carefully addressed. Accordingly, PDT dosimetry has been the main focus of numerous papers (see, e.g., Star), of which a few are closely related to the particular case of PDT of prostate cancer. One fundamental aspect in this context is the tissue optical properties, e.g., absorption and scattering coefficients. Not only do they together determine the light dose distribution, but in vivo access to absorption coefficients can be used to estimate, for example, sensitizer concentrations, hemoglobin concentration, and tissue oxygenation. In addition, due to dynamic changes during PDT treatment, it is sometimes argued that on-line monitoring is required to achieve optimal treatment.

PDT, in general, relies on a process where light excites (i.e., activates) a photosensitizer, which induces cytotoxic oxygen species. Thus, PDT relies on the presence of light, a photosensitizer, and oxygen. The corresponding dosimetry is therefore a complex matter, involving measurement and prediction of sensitizer and oxygen concentrations as well as light fluence (light dose). Since full treatment of diseased regions is crucial, these issues must be carefully addressed. Accordingly, PDT dosimetry has been the main focus of numerous papers (see, e.g., Star), of which a few are closely related to the particular case of PDT of prostate cancer. One fundamental aspect in this context is the tissue optical properties, e.g., absorption and scattering coefficients. Not only do they together determine the light dose distribution, but in vivo access to absorption coefficients can be used to estimate, for example, sensitizer concentrations, hemoglobin concentration, and tissue oxygenation. In addition, due to dynamic changes during PDT treatment, it is sometimes argued that on-line monitoring is required to achieve optimal treatment.

For the case of prostate tissue, several papers have addressed the issue of optical properties, e.g.,

Abstract. The development of photodynamic therapy into a modality for treatment of prostate cancer calls for reliable optical dosimetry. We employ, for the first time, interstitial time-resolved spectroscopy to determine in vivo optical properties of human prostate tissue. Nine patients are included in the study, and measurements are conducted prior to primary brachytherapy treatment of prostate cancer. Intra-subject variability is examined by measuring across three tissue volumes within each prostate. The time-resolved instrumentation proves its usefulness by producing good signal levels in all measurements. We are able to present consistent values on reduced scattering coefficients ($

Address all correspondence to Tomas Svensson, Department of Physics, Lund University, Box 118, SE-221 00 Lund, Sweden; Phone: +46-46-2223120; Fax: +46-46-2224250; E-mail: tomas.svensson@fysik.lth.se

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One option is to determine $\mu'_t = \mu_a + \mu'_s$ by measuring relative radiance at multiple, but short, source-detector separations. A second option, avoiding the need of short source-detector separations, is to measure absolute (instead of relative) fluence rate in the diffuse regime. Data evaluation, by means of curve fitting, then yields $\mu_{\text{eff}}$ and $\mu'_s$. Hence, both options enable estimation of absorption and scattering coefficients.

Previously published values on optical properties of the human prostate, together with the results obtained in this study, are given in Sec. 3. The first published work presents $\text{ex vivo}$ steady state measurements of $\mu_{\text{eff}}$ and $\mu'_s$ at 633 nm, in three whole, nonmalignant human prostates. Another $\text{post mortem}$ study estimates prostate optical properties ($\mu_a$, scattering coefficient $\mu_s$, scattering anisotropy $g$, and $\mu'_s$) at 1064 nm, by measuring through thin prostate slices. In $\text{in vivo}$ effective attenuation in prostates diagnosed with benign prostatic hyperplasia (BPH) or prostatic carcinoma (PC) has been estimated at 630, 633, and 665 nm, by employing steady state fluence rate measurements. Other studies rely on absolute fluence rate measurements to determine both absorption and scattering coefficients. By measuring fluence rates along a linear channel (5 mm away from the source fiber), one such study presents optical properties at 732 nm, before and after motexafin-lutetium-mediated PDT of locally recurrent prostate cancer. In a similar study, optical fibers are kept fixed (five source fibers and three detector fibers), and absolute fluence rate data is collected for 15 source-detector separations (at 762 nm). Measurements were in that case performed in connection with TOOKAD-mediated PDT of recurrent prostate cancer. Regarding hemoglobin monitoring in human subjects, only very limited data have been published.

This study employs interstitial time-resolved spectroscopy to characterize human prostate tissue in $\text{in vivo}$ context. Conceptually, this approach is very different from those chosen in previously published work related to prostate tissue. By analyzing the temporal broadening of picosecond laser pulses due to propagation through tissue, this technique provides reliable estimations of absorption and reduced scattering coefficients. The use of multiple wavelengths enables spectroscopic determination of total hemoglobin concentration (THC) and oxygen saturation ($S_O2$). The technique has previously been used in various areas of biomedical optics.

The aim of the paper is to provide information on parameters of dosimetric importance, as well as to introduce time-resolved spectroscopy as a tool in PDT research. In particular, the aim is to achieve separate information on $\mu_a$ and $\mu'_s$, as well as quantitative information on hemoglobin parameters. We also wish to give a proper indication on intra- and inter-subject variation, for the case of untreated prostate cancer. These parameters are important for PDT dosimetry calculations, and are not extensively explored in previous studies.

Nine patients diagnosed with prostate cancer were included in the study, and all measurements were performed before primary treatment. By inserting three optical fibers, three tissue volumes were probed, yielding information on prostate tissue heterogeneity.

### 2 Material and Methods

#### 2.1 Instrumentation

Time-resolved data were acquired using a compact ($50 \times 50 \times 30$ cm) and portable time-domain photon migration instrument primarily intended for spectroscopy of biological tissues in clinical environments. A schematic illustration is given in Fig. 1.

The system is based on diode laser technology and time-correlated single-photon-counting (TCSPC). A laser driver (SEPIA PDL 808, PicoQuant, Germany) controls four pulsed diode lasers (LDH, PicoQuant at 660, 786, 916, and 974 nm). Wavelengths are chosen to enable monitoring of important tissue constituents (hemoglobin, water, and lipids) and properties (tissue oxygenation). Lasers are typically operated at 1 to 2 mW, generating pulses about 70 ps wide (FWHM). Four wavelength pulse trains are generated at a repetition frequency of 40 MHz. This is accomplished by separating the individual pulses in time ($\sim$6 ns), using electric cables of different lengths.

The light emitted from each diode laser is individually coupled into a separate 200-µm graded-index (GRIN) fiber (G 200/280 N, ART Photonics, Germany). A four-to-one coupler is used to couple all light into a single 600-µm GRIN fiber (G 600/840 P, ART Photonics), which serves as the light source. A second fiber collects light and delivers it to the detector. Each of these two fibers is approximately 2 m long. To fit into 1-mm inner diameter brachtherapy needles, a thin polyimide layer acts as the only fiber coating at the fiber endings. Remaining parts are protected in polyolefin tubing.

Proper photon levels are achieved by sending collected light through an adjustable gradient neutral density (ND) filter. Remaining photons are sent to a cooled microchannel plate photomultiplier tube (MCP-PMT; R3809-59, Hamamatsu Photonics, Japan). A TCSPC computer card (SPC-300, Becker&Hickl, Germany) is used to obtain time-dispersion histograms with channel widths of approximately 25 ps.

Broadening in the fibers and the detector yields an instrument response function (IRF) that is about 100 ps wide. The IRF is measured by inserting source and detector fibers into a chamber so that the end facets are separated by 150 mm and face each other. The chamber is made of black delrin, and contains a pinhole that is inserted between the two fibers to block reflected stray light.

#### 2.2 Modeling

Experimental data are modeled using the diffusion approximation of transport theory. More specifically, data are fitted to the analytical solution of the time-dependent diffusion equation for the case of a homogenous and infinite medium.
Accordingly, the fluence rate $\Phi$ due to a infinitely short light pulse from an isotropic point source can be written as

$$\Phi(r,t) = c'E_0(4\pi c'Dt)^{-3/2} \exp\left(-\frac{r^2}{4c'Dt} - \mu_a c't\right).$$  \hspace{1cm} (1)

where $E_0$ is the pulse energy, $r$ is the distance from the point source, $c'$ is the speed of light within the material, and $D$ is the diffusion coefficient. The refractive index is assumed to be $n=1.4$. To comply with recent work concerning the diffusion coefficient, it is defined as in Eq. (2), rather than in the traditional (absorption-dependent) way\textsuperscript{42,43}.

$$D = \frac{1}{3\mu_s^*}.$$  \hspace{1cm} (2)

The form of Eq. (1) enables deduction of both $\mu_s^*$ and $\mu_a$ from experimental data without requiring absolute measurements of light fluence. This is achieved by considering temporal shapes only. That is, experimental data are fitted using an expression similar to that given in Eq. (1), but in which amplitude information is contained in a free parameter $k$. This expression is given as

$$y(\mu_a, \mu_s^*, k, t) = k r^{-3/2} \exp\left(-\frac{3\mu_s^* r^2}{4c' t} - \mu_a c't\right).$$ \hspace{1cm} (3)

The best fit is reached iteratively using a Levenberg-Marquardt algorithm, where $\mu_s^*$, $\mu_a$, and $k$ are varied to minimize a $\chi^2$ error norm.\textsuperscript{44} Final values of $\mu_s^*$ and $\mu_a$ are then estimations of the reduced scattering coefficient and absorption coefficient, respectively. Note that the IRF, being about 100 ps wide, cannot be regarded as infinitely short in comparison to the tissue response. Therefore, each iteration in the curve-fitting procedure involves a convolution of analytical data and IRF. It should also be noted that fitting is performed using the part of experimental data between times given by the 50% of maximum on the rising edge, and 20% on the falling edge (see Fig. 5 in Sec. 3).

The previous studies of human prostate optical properties mentioned in the introduction used the same theoretical framework. However, by employing steady state techniques, they rely on a stationary solution rather than the time-dependent solution given in Eq. (1). The stationary solution of the diffusion equation for an infinite homogenous medium is given as

$$\Phi(r) = \frac{3\mu_s^* P_0}{4\pi r^4} \exp(-\mu_{\text{eff}} r).$$ \hspace{1cm} (4)

Here, $P_0$ is the power of the point source, $r$ is the distance from the source, and $\mu_{\text{eff}}$ the effective attenuation. The effective attenuation is defined as

$$\mu_{\text{eff}} = (\mu_s^*/D)^{1/2} = (3\mu_a/2)^{1/2}$$ \hspace{1cm} (5)

From the form of Eq. (4), we see that $\mu_{\text{eff}}$ can be deduced from relative fluence measurements at multiple source detector separations. Only if absolute fluence rate is measured, it can be used to get information on reduced scattering and absorption separately. Absolute measurement of fluence rate in living tissue is of course a difficult matter (especially in interstitial settings). Note also that $\mu_{\text{eff}}$ is a primary parameter when using the steady state technique. Time-resolved experiments provide estimations of $\mu_a$ and $\mu_s^*$ which in a second step can be used to calculate $\mu_{\text{eff}}$.

### 2.3 Hemoglobin Spectroscopy

The spectroscopic evaluation employed in this study assumes that the absorption exhibited by prostate tissue originates from oxy- and deoxyhemoglobin ($\text{Hb}$ and $\text{HbO}_2$, respectively), water, and lipids. Since high absorption prevented the use of 974-nm data (see Sec. 3), water and lipid concentrations could not be estimated. Instead, absorption coefficients at 660 and 786 nm were used to extract oxy- and deoxyhemoglobin concentrations ([Hb$_2$] and [Hb], respectively). In this procedure, the prostate was assumed to contain 70% water and 10% lipids (note that due to relatively low absorption of water and lipids at 660 and 786 nm, the choice of these values is of very little importance). The extinction coefficients of these chromophores were taken from literature\textsuperscript{45–47} and are presented in Table 1.

In a second step, total hemoglobin concentration and oxygen saturation are calculated from [Hb$_2$] and [Hb].

### 2.4 Clinical Procedure

Clinical data were collected at the Lund University Hospital adhering to a protocol approved by the regional ethics committee. All nine patients involved in the study were undergoing primary treatment of prostate cancer. Measurements were performed in connection with brachytherapy (low-dose seed implantation). This fact limits our study to patients suitable for this treatment, that is, patients fulfilling the following requirements: (1) Gleason index <6, (2) prostate specific antigen (PSA) <10, (3) no tumor obstruction of urethra, and (4) prostate volume of the order of 20 to 40 cm$^3$. Such patients are often referred to as low-risk patients.

This type of brachytherapy of prostate cancer involves permanent implantation of radioactive seeds (internal radiotherapy). The procedure takes place in an operating theater, while patients are under general anesthesia. At the Lund University Hospital, the first step in this procedure is to image the prostate gland using transrectal ultrasound. When the physician has marked important boundaries in these images (prostate, urethra, and rectum), a radiotherapist performs dosimetric calculations. Dosimetric calculations open a time window of about 20 min in which the time-resolved measurements were performed without interfering with the routine proce-
time was approximately 3 min per patient. The total data acquisition was performed in 1 s increments. After completion of the tissue measurements, the IRF is measured using the same set of fibers. To ensure system stability (e.g., avoid drifts due to temperature changes), this step is conducted while the system is in the operating theater.

3 Results
Analysis of collected data reveals that time-resolved spectroscopy can provide consistent optical and physiological characteristics of human prostate tissue. Data were collected at four wavelengths: 660, 786, 916, and 974 nm. For each patient and wavelength, raw data consist of three dispersion curves (tissue response) corresponding to the three utilized fiber separations (in total, 27 time-resolved data sets from nine patients). Since the fiber separations within a patient are chosen to be different, e.g., 15, 20, and 25 mm, the obtained curves are significantly different. For the range of fiber separations used in this study, most detected light travels a time less than 1.5 ns through tissue (corresponding to maximum path lengths of about 30 cm). An example of raw data, for the case of 660 nm, is given in Fig. 4.

By means of curve fitting, optical properties were extracted for 660, 786, and 916 nm from all 27 time-resolved data sets. Unfortunately, high absorption at 974 nm prevented analysis of data acquired at that wavelength. A typical fitting example, for the case of 786 nm, is shown in Fig. 5.

A summary of the results of this study is given in Table 2, in which previously published data are presented for comparison. A detailed report on measured optical (at 660 and 786 nm) and physiological characteristics are given in Table 3. In the following, graphical representation of data from Table 3 is used to illustrate and support further analysis.

No pronounced correlation between estimated values of $\mu_a$ and $\mu'_a$ was seen for any wavelength, implying that results do not suffer from $\mu'_a$ to $\mu_a$ crosstalk. This is shown in Fig. 6, where results from 660 nm are presented. A general comment is, however, helpful in interpreting the pattern shown. Three measurements (21.5 and 26.4 mm from patient 6, and 25.0 mm from patient 8) resulted in very low values of both

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**Fig. 2** Two ultrasound images showing three needles inserted into the prostate. For clarity, the needles are marked by circles. Shadows behind the needles are due to their high reflectivity. The fairly homogenous image (a) originates from patient 5, while the heterogenous structure in (b) was exhibited by patient 6.

**Fig. 3** Schematic drawing of fiber positioning [compare with Fig. 2(a)]. The needle template has 5-mm grid spacing. Three optical fibers are inserted to the same depth. Fiber separations are 15, 20, and 25 mm. The three tissue volumes being probed are indicated using calculations of photon-hitting densities (PHDs) at the plane of fiber tips ($\mu'_a = 8.7$ and $\mu_a = 0.49 \text{ cm}^{-1}$). Iso-curves indicate where the PHD is 50% of the value exhibited halfway between source- and detector fibers.
absorption and reduced scattering (leading to a THC less than 100 \( \mu \text{M} \)). In these cases, note that an extremely low degree of bleeding was noted during the measurements (almost no blood stains on the used fibers, while the average fiber was heavily stained after a measurement). Thus, it is likely that these measurement outcomes correspond to a particular tissue composition, rather than a systematic error such as \( \mu_a' \) to \( \mu_a \) crosstalk.

The influence of fiber separation on derived optical properties is illustrated in Fig. 7. Although derived optical properties show no strict dependence on the fiber separation, there might be a tendency toward measuring lower \( \mu_a \) and \( \mu_a' \) for large fiber separations. Such a tendency may be related to the fact that the longest fiber separation (for each patient) corresponds to a central prostate volume, while the two shorter correspond to outer volumes. For comparison, note that studies on intralipid phantoms having prostate-like optical properties were performed. In these, the selection of fiber separation had no systematic influence on derived optical properties.

As we can see in Fig. 8, derived hemoglobin concentrations show a behavior similar to that of \( \mu_a \), while oxygen saturation showed no correlation to utilized fiber separation.

Turning to inter- and intrapatient variations, an important general observation is that patients 2 to 5 exhibited homogenous ultrasound images, while obvious inhomogeneities due to calcifications are found in images from patients 1 and 6 to 9. As we can see in Figs. 9–11, this fact correlates to measured intrasubject variations. A second general observation is that all derived parameters, except oxygen saturation, can be subject to a fairly large spread, even within individual patients. Figure 9 presents inter- and intrapatient variations of 660-nm optical properties.

Effective attenuation, being an important parameter in PDT dosimetry, was calculated from extracted \( \mu_a \) and \( \mu_a' \) and is presented in Fig. 10.

Hemoglobin concentrations and oxygen saturation were spectroscopically determined using absorption coefficients as measured at 660 and 786 nm. As we can see in Fig. 11, THC exhibit substantial inter- and intrasubject variations. Oxygen saturation, on the other hand, displays only minor variations.

One alternative to deriving hemoglobin concentrations from 660- and 786-nm data only, would be to include 916-nm data (still assuming that the prostate contains 70% water and 10% lipids). If this procedure is followed, resulting THC values deviate by +1.5±8.7\%, and \( S_\text{Ot} \) by +1.1±4.1\% from the values calculated from 660- and 786-nm data only. In terms of overall impact, derived THC changes from 215±65 to 215±56 \( \mu \text{M} \) (the rounded average is coincidentally conserved), and \( S_\text{Ot} \) from 76±4 to 77±4\%. These fairly small changes can also be understood by analyzing the difference between measured 916-nm \( \mu_a \) and the predicted 916-nm \( \mu_a \), as calculated from derived hemoglobin levels (estimated from 660- and 786-nm data) in combination with the assumption of 70% water and 10% lipids. Such analysis show that measured 916-nm \( \mu_a \) deviates by 2±11% from predicted values.

A synthetic absorption spectra can be constructed using derived hemoglobin concentrations and assumed concentrations of water and lipids. Figure 12 presents the average composite absorption spectra of the prostate, as derived in this study.

### 4 Discussion

Reliable optical dosimetry is an important issue in the development of PDT into a modality for treatment of prostate can-
Whether reliable dosimetry means on-line monitoring of certain parameters, individual pretreatment planning, or general knowledge on intra- and intersubject variations remains an open question. By employing time-resolved spectroscopy, this study was able to generate estimations of optical and physiological characteristics for all involved patients, and thus providing a reliable indication on intra- and intersubject variations for the case of untreated prostate cancer. In fact, as we can see in Table 3, all measurements resulted in quality data. In addition, by requiring only two fixed fibers for a single measurement, the time-resolved technique is accompanied with fairly simple clinical procedures.

Calculations of PHDs suggest that sampling volumes are kept within the prostate. Results are consistent and imply that the prostates of the patient group under consideration exhibit moderate intra- and intersubject variations. However, the following table summarizes the optical properties of human prostate tissue reported in various published studies:

<table>
<thead>
<tr>
<th>Study</th>
<th>Description</th>
<th>( \lambda ) (nm)</th>
<th>N</th>
<th>( \mu_a )</th>
<th>( \mu'_s )</th>
<th>( \mu_{\text{all}} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pantelides et al.(^{24}) (1990)</td>
<td>ex vivo steady state data, normal prostates 633 3 0.7±0.2 8.6±0.5 4.3±0.5</td>
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<tr>
<td>Whitehurst et al.(^{26}) (1994)</td>
<td>in vivo steady state data, untreated BPH and PC 633 11 3.6±0.2</td>
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<tr>
<td>Lee et al.(^{28}) (1995)</td>
<td>in vivo steady state data, untreated BPH and PC 633 11 3.9±0.5</td>
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<tr>
<td>Lee et al.(^{28}) (1995)</td>
<td>in vivo steady state data, untreated BPH and PC 665 11 3.2±0.5</td>
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<tr>
<td>Lee et al.(^{29}) (1999)</td>
<td>in vivo steady state data, untreated PC 630 7 3.5±0.7</td>
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<tr>
<td>This study</td>
<td>in vivo time-resolved data, untreated PC 660 9 0.5±0.1 8.7±1.9 3.6±0.8</td>
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<tr>
<td>Weersink et al.(^{30}) (2005)</td>
<td>in vivo steady state data, recurrent PC 762 22 0.4±0.2 3.4±1.6 2.0±0.6</td>
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<tr>
<td>Zhu et al.(^{34}) (2005)</td>
<td>in vivo steady state data, recurrent PC 732 13 0.4±0.2 11.8±8.2 3.3±0.5</td>
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<tr>
<td>This study</td>
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<tr>
<td>Essenpreis et al.(^{25}) (1992)</td>
<td>ex vivo integrating sphere data, normal prostates 1064 1.5±0.2 6.4</td>
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</table>

Table 2 Optical properties (given in inverse centimeters) of human prostate tissue as reported in various published studies; the number of involved patients is also given (N); note differences regarding wavelength.

![Fig. 7](image_url) Scatterplot showing the correlation between utilized fiber separation and derived optical properties at 660 nm. Correlation coefficients \( r \) are added for reference. The three points with \( \mu_s < 0.25 \) correspond to the three measurement cases exhibiting a low degree of bleeding and low THC.

![Fig. 8](image_url) Scatterplot showing the correlation between utilized fiber separation and THC and \( S_tO_2 \), respectively. Correlation coefficients \( r \) are added for reference.
Table 3 Detailed report on optical and physiological characteristics of human prostate tissue *in vivo*, as measured by our time-resolved instrumentation; prostate volumes (V) are determined from ultrasound images.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age (yr)</th>
<th>V (cm³)</th>
<th>Fiber sep. (mm)</th>
<th>Optical properties (cm⁻¹)</th>
<th>Physiological properties</th>
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<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>μₐ(660)</td>
<td>μₐ'(660)</td>
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<tr>
<td>1</td>
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<td>41</td>
<td>10.1</td>
<td>0.67</td>
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<td></td>
<td></td>
<td></td>
<td>14.9</td>
<td>0.50</td>
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<td></td>
<td></td>
<td>17.3</td>
<td>0.48</td>
<td>7.8</td>
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<td>15.1</td>
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<td>17.1</td>
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<td></td>
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<td>21.9</td>
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<td>26.6</td>
<td>0.49</td>
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<tr>
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<td>14.6</td>
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<td>0.50</td>
<td>8.4</td>
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<tr>
<td>5</td>
<td>70</td>
<td>33</td>
<td>14.8</td>
<td>0.56</td>
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<td>10.3</td>
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<td>16.0</td>
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Following observation calls for a separate discussion: derived $\mu'_s$ at 916 nm exceeds those at 786 nm in 23 out of 27 measurements. According to the general theory of tissue optics, a reduction of $\mu'_s$ is expected. Such behavior was only seen in 4 out of 27 measurements (three out of nine patients). The explanation is very likely to be related to the high absorption and low scattering at 916 nm. In fact, analysis of Monte Carlo simulations indicates a breakdown of diffusion approximation in this regime. Derived $\mu'_s$ is closely related to the rising flank of the dispersion curve (early photons), and may be particularly sensitive. Another aggravating circumstance is that such optical properties yield a low degree of broadening (and thus fewer data points), which in turn reduces the performance of the fitting procedure. Refined data evaluation may solve this problem, one option being implementation of Monte-Carlo-based evaluation.

When comparing our results to previously published data, two important facts must be noted. First, differences in patient groups must be taken into account (e.g., untreated or recurrent prostate cancer). The prostate physiology changes drastically upon radiation therapy. An experienced physician can feel such differences during the insertion of brachytherapy needles. Second, previous studies of prostate tissue have employed steady state techniques. Note that the ambiguous appearance of published steady state $\mu'_s$ data may indicate a difficulty in separating absorption and scattering using steady-state techniques (cf the published values of $\mu'_s$ stated in Table 2). Unfortunately, differences in patient groups prevent these results from being compared to the results of this study. It would thus be very interesting to employ time-resolved spectroscopy also in cases of locally recurrent prostate cancer. On the other hand, in the 600-nm range, $\mu_{\text{eff}}$ were measured using steady state techniques in patient groups similar to ours. In this respect, the two techniques seem to produce similar data (most derived $\mu_{\text{eff}}$ values are between 3 and 4 cm$^{-1}$).

To discuss the correctness of our approach, a few general remarks are given in the following. First, the model used in this study is valid for photon fluence rates. Detected light is collected by cleaved optical fibers, and is thus not a direct measure of fluence rate. Since data evaluation considers only temporal shapes, this fact is, however, not a source of error if the proportionality constant between actual fluence rate and collected light is the same for all photon times of flight. Moreover, the model applies for homogenous media. The prostate

Fig. 9 Inter- and intrapatient variations in 660-nm optical properties. Patients 2 to 5 exhibited homogenous ultrasound images. Solid lines mark the average values, and dashed lines correspond to ± one standard deviation.

Fig. 10 Inter- and intrapatient variations in 660-nm effective attenuation. Patients 2 to 5 exhibited homogenous ultrasound images. The solid line mark the average value, and dashed lines correspond to ± one standard deviation.

Fig. 12 Synthetic average absorption spectra (bold solid) together with derived absorption coefficients at 660, 786, and 916 nm (mean ± standard deviation). It is constructed using 215 μM THC at 76% oxygen saturation, 70% water and 10% lipids. Contribution spectra are also shown: 51.6-μM Hb (dashed), 163.4-μM HbO$_2$ (dotted), 70% water (solid), and 10% lipid (dash-dotted).
is, of course, heterogenous, and out of all detected photons, those that have traveled through regions of low attenuation will be overrepresented. One should therefore be careful in interpreting the derived values as true volume averages.

Second, the choice of data range involved in curve fitting occasionally has an influence on the outcome. Finding an optimal fitting range is, however, a complex issue, since the correctness of our model varies with the photon time of flight. For example, early photons are often disregarded since diffusion theory does not describe them very well. In this study, we focus on regions with fairly high photon count rates by choosing to cut the tail of late photons at 20% of maximum (see Sec. 2.2). The reason why later photons are excluded is to avoid a region with many data points at a low signal level, where even small absolute systematic measurement/model errors will have a significant impact on fitted parameters.

Third, as we can see in Eq. (3), the fiber separation is an important parameter in time-resolved spectroscopy. The inferring of fiber separations from ultrasound images is, of course, afflicted with some errors. However, it is unlikely that the error between true and measured separation, \( r_{\text{true}} \) and \( r_{\text{mean}} \), respectively, exceeds 1 mm. From Eq. (3), one finds that errors in fiber separation will produce errors only in derived reduced scattering coefficients. The optimal fit is expected when \( \mu_s' \) is selected so that

\[
\mu_s'_{\text{true}} \left( r_{\text{true}} + \delta r \right)^2 = \mu_s'_{\text{true}}^2, \tag{6}
\]

where \( \mu_s'_{\text{true}} \) is the true reduced scattering coefficient, and \( \delta r \) is the error in fiber separation. The impact of uncertainties regarding fiber separation can be described by the tuning coefficient

\[
\gamma_T = \left. \frac{\partial \mu_s'}{\partial \delta r} \right|_{\delta r=0} = \frac{-2 \mu_s'_{\text{true}}}{r_{\text{true}}}, \tag{7}
\]

which states the change in derived reduced scattering coefficients due to (small) errors in fiber separation. For the case of \( \mu_s'_{\text{true}} = 8.7 \text{ cm}^{-1} \) and \( r_{\text{true}} = 20 \text{ mm} \), one finds that \( \gamma_T \approx 0.87 \text{ cm}^{-1}/\text{mm} \). Hence, a nonnegligible part of the intra-subject variations in derived reduced scattering coefficients may be assigned to errors in fiber separation. However, correlation with ultrasound heterogeneity and the large variations measured in some patients imply that there are significant inherent intrasubject variations.

Finally, the time-resolved approach, in contrast to steady state techniques, should be rather insensitive to bleedings around the needle tips. These bleedings can be thought of as random filters, and may therefore disturb measurements of fluence rate. However, all detected photons must pass the region of bleeding, leaving the temporal shape unchanged.

5 Summary

This study clearly shows that time-resolved spectroscopy is a suitable tool for accessing information on human prostate tissue in vivo. By producing consistent estimations of absorption (\( \mu_a \)), reduced scattering (\( \mu_s' \)), effective attenuation (\( \mu_{\text{eff}} \)), hemoglobin concentrations, and tissue oxygenation, this technique was able to generate the most complete in vivo optical characterization of human prostate tissue published so far. By measuring across three tissue volumes in each prostate, both inter- and intra-subject variations were examined. All derived parameters, with the exception of tissue oxygen saturation, are subject to fairly large intra-subject variation. Interestingly, these variations correlate with the heterogeneity exhibited by acquired ultrasound images.

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References


