Simple instrument for the characterization of diffuse reflectance

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ABSTRACT

An apparatus for the measurement of the radial dependence of the continuous-wave diffuse reflectance of a tissue is described. It consists in a probe, which is applied on the tissue, and a detection unit. By employing an array of incoherent semiconductor light sources (LED's) and continuous-wave detection, probe structure is considerably simpler than other devices described in literature, allowing moreover an efficient coupling of the emitted light towards the tissue. The high responsivity so obtained permits fast and accurate measurements. Measurement speed, probe compactness and accuracy are potentially sufficient for the in-vivo application of the method to surgically exposed tissue. A preliminary set of data, measured on a scattering phantom and on non-exposed in-vivo tissue, is presented. Even though available models fitted to the measured data give the correct order of magnitude for light transport coefficients, in order to extract reliable absolute values they should be corrected for probe nonidealities. The availability of extensive high-quality in-vivo data is to this regard stimulating for further theoretical investigations.

Keywords: apparatus, diffuse reflectance, in-vivo, LED

1. BACKGROUND

Near infrared (NIR) spectroscopy is widely recognized as a potentially powerful technique for non-invasive quantitative determination of intrinsic and extrinsic tissue constituents. In its simplest form, it consists in the analysis of the extinction coefficient of the light backscattered by the tissue at (at least) two wavelengths in the NIR range. In order to extract meaningful information for the measured data, the technique however relies on an accurate characterization of the tissue optical behaviour for each wavelength employed in the measurement, at least as far as it regards optical transport coefficients. In order for the technique to be effective, such characterization must moreover be specialized to the specific tissues under examination, and should possibly be performed in-vivo.

In the relevant literature, we find only a few examples of instrumental setups for the explicit and extensive characterization of the optical transport coefficients of tissue. They are almost invariably based on a complex optical fiber probe, coupled to a consistently elaborate detection unit[1]. The responsivity, measurement time and reliability of the resulting apparatus is therefore typically quite unsuitable for measurements during a non-dedicated surgical exposure of tissues.

If we restrict the measurement to the determination of the radial dependence of the diffuse reflectance on a source-to-detector distance from a few millimetres to a few centimetres, it is possible to overcome these problems by an efficient design of the probe. However, we notice that, even though such measurement contains, in principle, all the relevant information[2], analysis of the measured data is still under discussion.

2. MATERIALS AND METHODS

Our apparatus is composed by a probe and a detection unit. The probe consists in a radial array of 10 sets of light emitting diodes (LED's), at different distances from a PIN photodiode, placed in the centre of the array. The probe wavelength is therefore fixed by the chosen LED's, and can be varied by changing the probe head. By using a photodiode area on the order of 1 mm^2, such as Centronic BPX65RT, the responsivity is sufficient to employ continuous-wave (CW) light emission. We can so use plastic case LED's, which cannot be modulated over a few tens of kHz, rather than metal-case diodes, thus reducing probe costs and improving manufacturability, or even LED arrays, such as those in common use for bargraph displays. All the measurements reported in this paper have been performed using Kingbright DC-10SRWA

Part of the EUROPTO Conference on Optical Biopsy • Stockholm, Sweden • September 1998
SPIE Vol. 3568 • 0277-786X/99/$10.00

Hyper-Red GaAlAs LED arrays, which emit at a peak wavelength of 660 nm. Within such arrays source intensity repeatability (uniformity) is sufficient to avoid individual source intensity calibrations. By encapsulating the probe in epoxidic resin (Flexane, Devcon, UK), it can be sterilized by Ethylene Oxide. A drawing of the probe is reported in Fig 1.

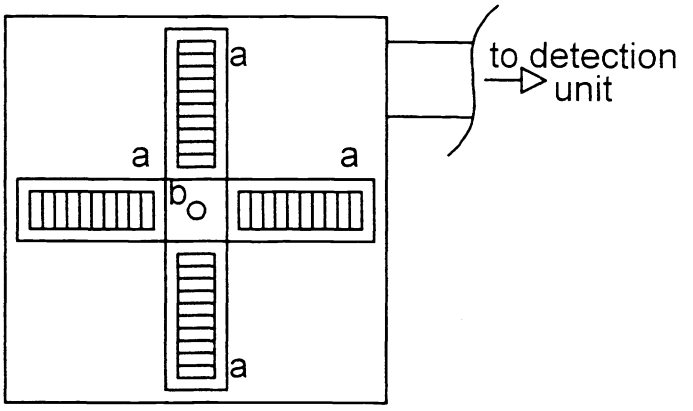

Fig.1.
Probe: a) LED bars, b) photodetector.

The sources are driven by a constant current CW driver, which can be selectively multiplexed to different sets of LED's. The detector current signal is preamplified in the detection unit by a transresistance stage and digitized by a standard benchtop multimeter. Since we do not need a high bandwidth, we can employ an elementary CW-coupled cold-resistor configuration[3], as long as we keep to low-input-bias (JFET input) opamps, such as National Semiconductor LF356. At the moment, the source multiplexer is electromechanical and manually driven; a fully automatic system is under development.

A first ("white") scattering phantom has been prepared according to ref.4, by dispersing aluminum oxide particles in a silicone matrix. Our particles have a nominal radius of 4.5 μm, and have been dispersed to a volume fraction of 1%. The silicone (Rhodorsil RTV141, Rhone-Poulenc, France) has a nominal refractive index of 1.41. Such index has been verified by spectroscopic ellipsometry (SOPRA ES-4G ellipsometer, Sopra, France) over the whole visible wavelength range, and at our wavelength agrees within 1% to the nominal value. With these parameters, the scattering coefficient for the phantom is calculated to be μ'_s =0.41 mm^{-1}. We do not yet have a complete characterization of the particles, which would permit a correction of the scattering coefficient for the particle polydispersity.

Measurements have been performed on the phantom and, through the skin, on a non-exposed large muscle (Vastus Medialis). The probe has been applied on the tissue under examination, and held in place with a rubber strip. The area has been obscured using a dark cloth. All the sources are switched off, and the offset due to background light, detector leakage current and preamplifier intrinsic offset is measured. The sources at different distances from the detector are then selectively switched on, and the measured signal corrected for the previously measured background. Signal level is sufficient for the measurement to be limited by the settling time of the readout unit.

A second ("gray") phantom has then been prepared, with the same structure as the first, with the addition of a small quantity of black aniline as a dye. The dye has been dispersed in ethanol, mixed to the phantom before curing, then the ethanol has been evaporated by gentle heating and the curing process started. The dye amount has been determined in order to obtain an average backscattering of the phantom on the same order of magnitude as the unexposed muscle. No nominal absorption coefficient has been calculated, since the employed dye is not completely stable in the silicone matrix. Measurements have then been carried out also on such phantom.

3. RESULTS

In Fig.2 are reported the measured backscattered intensities, as a function of source-to-detector distance, for both phantoms and Vastus Medialis.

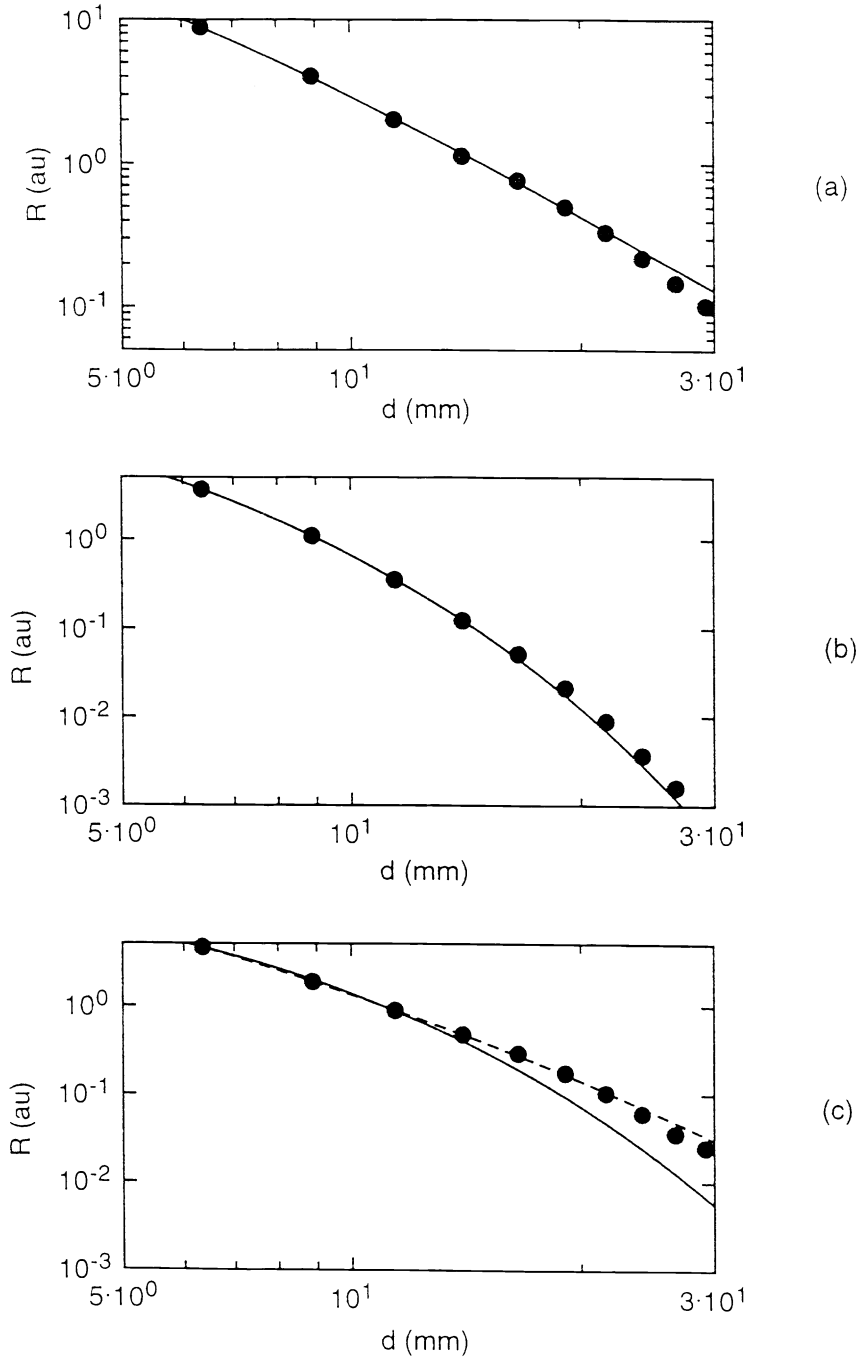

Fig.2
backscattered intensities, as a function of source-to-detector distance, for "white" (a), "gray" (b) phantoms and Vastus Medialis (c). The full and dashed lines indicate the regression curves, as explained in the text.

For a preliminary interpretation of the measured radial dependence of the diffuse reflectance a Levenberg-Marquardt regression with a diffusion theory model by Patterson et al.[2], which relates the absorption and scattering coefficients μ_a and μ'_s to the dependence of the absolute diffuse reflectance R to the source-to-detector distance d, was used:

$$R(d) = A \exp\left(\frac{-\mu_{eff}\sqrt{z_0^2 + d^2}}{z_0^2 + d^2}\right)\left(\mu_{eff} + \frac{1}{\sqrt{z_0^2 + d^2}}\right)$$

where A is the source amplitude, and

$$z_0 = \frac{1}{\mu'_s}$$

$$\mu_{eff} = \sqrt{3\mu_a(\mu_a + \mu'_s)}$$

In order to independently characterize all the transport coefficients with a single measurement, however, the model relies either on data taken at small d, which on our probe is inaccessible, or on an absolute calibration of the emitted light intensity. We therefore performed the first measurement on the "white" phantom and, in the regression, assumed μ_a=0, thus deriving μ_s and the light intensity. Keeping then the intensity fixed to the derived value, the regression was performed for μ_a and μ_s on the data from the "gray" phantom. The values so derived for μ_s on the "white" and "gray" phantoms agree, as they are expected to. The regression was finally performed, again assuming the intensity as derived from the first regression, on Vastus Medialis data (full line). If however we fit the last data also on the intensity, the fit quality is significantly improved (dashed line), but the derived parameters disgree with the frst fit. All the results are reported in Tab.1.

	White phantom	Grey phantom	Vastus Medialis (full line)	Vastus Medialis (dashed line)
Source intens. [au]	5930 ± 50	(5930)	(5930)	3400± 20
μ_a [mm^{-1}]	(0)	0.072 ± 0.004	0.049 ± 0.0015	0.004± 0.0007
μ'_s [mm^{-1}]	0.261 ± 0.006	0.254 ± 0.009	0.16± 0.006	0.30± 0.01

Tab.1

Results for the regressions represented in Fig.2. In parentheses are indicated the data assumed as fixed.

As a concluding remark, we notice that data quality is satisfactory, and indicates clearly the possibility of performing systematic in-vivo measurements, which are programmed for the next few months. Moreover the regression model, developed under the assumption of point sources and detector, gives the correct order of magnitude for the transport coefficients. Such model should however be corrected for the source and detector finite size and acceptance angle and for reflection losses at the tissue surface, and this could account for the two different results on Vastus Medialis. The availability of extensive high quality in-vivo data will certainly be stimulating for further theoretical investigations.

4. ACKNOWLEDGEMENTS

We whish to thank the Optical Spectroscopy Laboratory of the Department of Physics (Dip. di Fisica "A.Volta") of the University of Pavia for access to the spectroscopic ellipsometer, and Prof. Mario Corti (Dip. Chimica e Biochimica Medica, Università di Milano) for useful discussion.

5. REFERENCES

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