

An Optical Pathlength Meter for Near Infrared Spectroscopy

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ABSTRACT

Many commercial clinical Near Infrared Spectroscopy (NIRS) instruments rely upon estimates of the optical pathlength of tissue based upon the optode spacing multiplied by an average differential pathlength factor (DPF) for the tissue under investigation. Significant variation in the published DPF is the major source of error in the NIRS data quantification. To reduce this error, an inexpensive, stand-alone, single wavelength, single frequency optical pathlength meter based on the homodyne phase method was constructed for clinical use with an existing NIRO 300 instrument (Hamamatsu Photonics KK). The construction and testing of the optical pathlength meter is discussed.

Keywords: Spectroscopy, tissue diagnostics; Medical optics instrumentation Spectrometers and spectroscopic instrumentation; Light propagation in tissues; (170.3890) Medical optics instrumentation; (170.6510) Spectroscopy, tissue diagnostics; (300.6380) Spectroscopy, modulation

1. INTRODUCTION

Near infra red spectroscopy is an established research tool for the study of delivery of oxygenation and tissue haemodynamics. The NIRO-300 is a clinical NIRS instrument which provides quantitative estimates of tissue oxygenation and haematocrit based upon the technique of spatially resolved spectroscopy^{1,2}. In determining the concentration changes of chromophores in tissue it utilizes an estimate of the differential pathlength based upon the known optode spacing and the tissue DPF. There is an experimental uncertainty in DPF (and hence the optical pathlength) with tissue type, subject and age. For the head this variation can be up to 14% from the mean in adults and up to 10% from the mean in neonates^{3,4}.

In order to increase the accuracy of tissue chromophore concentration measured with such instruments, a simple optical pathlength meter has been developed.

2. THEORY OF OPERATION

The optical pathlength meter estimates the pathlength by measuring the phase shift experienced by an intensity modulated optical signal traversing the tissue. When a modulated optical signal is passed through a scattering medium like tissue, the phase of the received signal is different from that of the emitted beam because of the finite time taken for the light to transverse the tissue. The phase shift $\Delta\phi$ (in radians) is given by

$$\Delta\phi = \frac{2\pi f n_t \Delta L}{c} \quad (1)$$

where f is the modulating frequency, c is the speed of light, n_t is the refractive index of the tissue and ΔL is the mean path length change.

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2.1. Choice of Modulating Frequency

Accurate determination of the optical pathlength depends upon accurate measurement of the phase shift which increases for higher modulating frequencies. The maximum operating frequency is theoretically limited by the 2π ambiguity inherent in measuring phase shifts. The maximum frequency may be determined by considering the likely range in pathlength to be measured. For the NIRO-300 the minimum probe separation is 4cm and the minimum expected DPF of 3.6 (muscle) results in a minimum pathlength of 15cm. The maximum optode spacing is 5cm and with a DPF of 7 (brain) the expected pathlength is 35cm.

For a simple and rugged clinically usable instrument, we chose a photodiode rather than a photomultiplier tube detector. This also has the advantage of being essentially immune to damage from overexposure to light. Given the high levels of optical attenuation in tissue and the diffuse nature of the emerging light, a relatively large area detector is needed to provide adequate sensitivity.

A practical limitation to the operating frequency is then set by the large area photodiode (typically 100MHz). In the final system an avalanche photodiode was chosen to provide some additional signal gain. A Hamamatsu C5331-04 APD module was finally selected which has a -3dB roll off frequency of approximately 80MHz. A modulation frequency of 70MHz was eventually chosen because it is a common intermediate frequency used in communication equipment and many components operating at this frequency are commercially available.

2.2. Choice of Optical Wavelength

The laser wavelength was chosen to be 800nm as this lies in the operating range used in the NIRO instrument and is also an isobestic point for oxy and deoxy haemoglobin absorption. Our NIRO-300 instrument uses the following wavelengths 775, 813, 853, 910nm – each NIRO-300 uses slightly different wavelengths due to manufacturing spread in the laser diodes. The existing NIRO-300 software allows entry of a pathlength factor and adjusts this figure accordingly for the different wavelengths used^{2,5}. By providing a unique measurement of pathlength at 800nm before making measurements on the subject, uncertainty in the derived chromophore concentrations can be reduced. An extension in this system to measure the optical pathlength for all wavelengths used in the NIRO-300 would potentially further reduce error relating to the wavelength dependence of pathlength in each subject although this variation has been shown to be much smaller than the variation in absolute DPF.²

3. SYSTEM DESCRIPTION

The system is based on an 10mW, 800nm laser diode (Roithner Lasertechnik RLT80010MG) which is intensity modulated at 70MHz. Light output from the laser is delivered to the tissue by the 3mm diameter fibre bundle used in the existing NIRO-300 instrument. The emerging phase-shifted and attenuated optical signal is collected by an identical optode spaced 4cm from the first and is delivered to the avalanche photodiode (APD) detector (Hamamatsu C5331-04). The low NA of each fibre in the fibre bundle limits the effects of modal dispersion on pathlength measurement.

The output signal from the APD module is then further amplified and transferred to a digitally adjustable attenuator (Minicircuits ZSAT-31R5) and the resulting conditioned electrical signal is mixed with a sample of the modulating signal in an I&Q demodulator (Minicircuits MIQY-70D). The outputs of the I&Q demodulator are low pass filtered with a 40 Hz bandwidth and digitized by a microcontroller (Tern A-Core-86).

The microcontroller digitally filters the incoming I&Q signal to a bandwidth of 0.1 Hz. The phase and amplitude are computed by the microcontroller and the amplitude of the detected signal is used to maintain a constant RF signal power at the I&Q demodulator by means of the digitally adjustable attenuator. The phase delay in the adjustable attenuator is corrected in software from a lookup table.

The mean phase delay is converted to the corresponding optical pathlength in software using Equation (1), assuming the tissue refractive index is 1.4.⁷ The estimates of pathlength are then displayed on a liquid crystal display with an update rate of 1 Hz.

A block diagram overview of the complete optical pathlength meter is shown in Figure (1).

3.1. Measurement Calibration

Calibration of the pathlength meter before each tissue measurement is accomplished by placing both optodes on a pre-calibrated phantom with a known optical pathlength of $\approx 20\text{cm}$ and an optical attenuation of 7 OD. A pushbutton switch sets the calibration software in operation.

The calibration procedure allows for the subtraction of the phase delay due to the source and detecting fibres, associated delays in the electronics and due to phase imbalance in the I&Q demodulator. The phantom is designed to represent the expected measurement on the head with 4cm optode separation to minimize error in absolute pathlength.

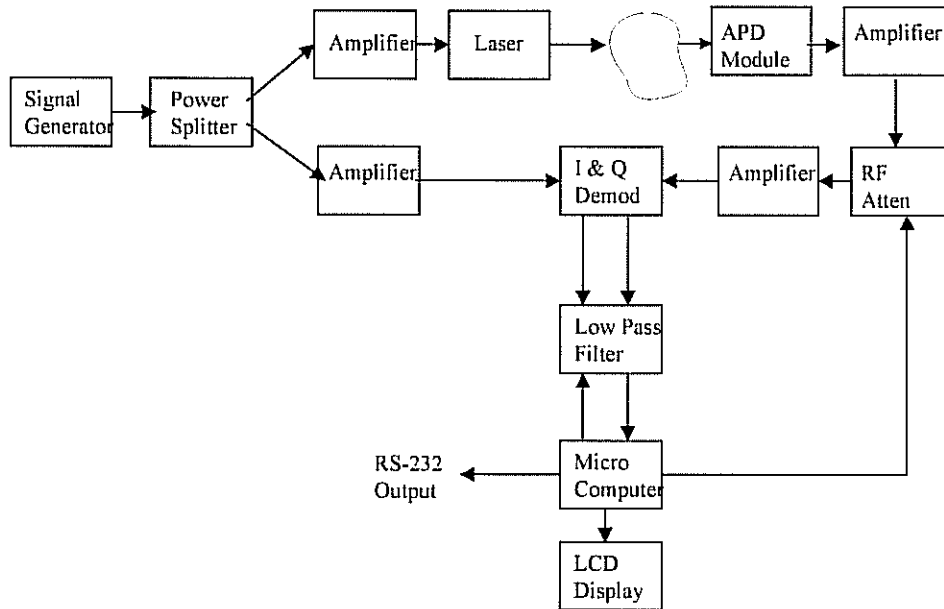


Figure 1. Schematic Diagram of Optical Pathlength Meter

4. PHANTOM DESIGN

The phantom design is shown in Figure (2). Light enters the symmetrical phantom through a thin diffusing slab of epoxy loaded with TiO_2 scatterers.⁶ The source and detecting optodes are made up of a fibre bundle and the scattering element is intended to reduce the sensitivity of the pathlength measurement on the phantom to the placement of the source and detecting optodes. Each of the diffusing slabs is $\approx 1\text{mm}$ thick, and a single mode fibre (Huber + Suhner 01-G62/K-G9) 13cm long is embedded $\approx 0.5\text{mm}$ into each. The other ends of the single mode

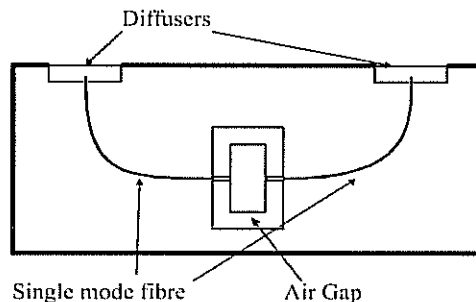


Figure 2. Schematic Diagram of Calibration Phantom

fibres are mounted onto a chamber of 1cm width which provides an optical loss because the detecting fibre captures only a small fraction of the emitted light. The inside of the cavity is painted black and the fibres are angled such that multiple reflections entering the detecting fibre are greatly attenuated. The overall optical loss of the phantom is adjusted to ≈ 7 OD.

Several optode holders of black silicon rubber were cast which hold the optodes at a separation of 4cm. Figure (3) shows a pair of calibration phantoms and optode holders.

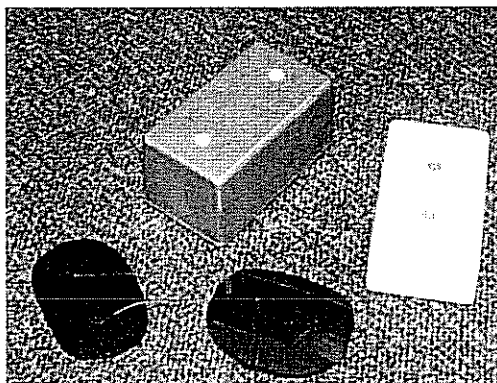


Figure 3. Photograph of Calibration Phantoms and Optode Holders

4.1. Calibrating the Phantom

The phantoms were calibrated by comparison to a pathlength in air. A calibration rig was constructed as shown in Figure (4). Thin layers of diffusing material were placed on the emitting and detecting optodes. The lens is adjusted to collimate the output beam and the attenuation of the air path is adjusted to the attenuation of the phantom. The optode separation is width adjusted to produce a phase shift identical to that in the phantom and the air path length measured. An estimate the path in lens and attenuators (assuming $n_{lens} = 1.5$) was made and added to the pathlength.

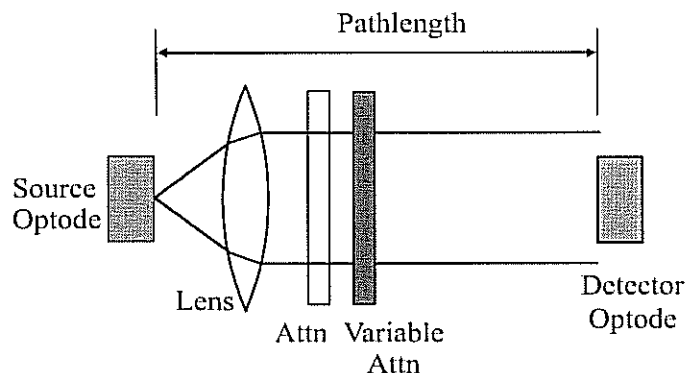


Figure 4. Schematic Diagram of Rig used to Calibrate Phantoms

This system was also used to measure the overall response by varying the pathlength in air (with constant optical attenuation) and the result is plotted in Figure (5).

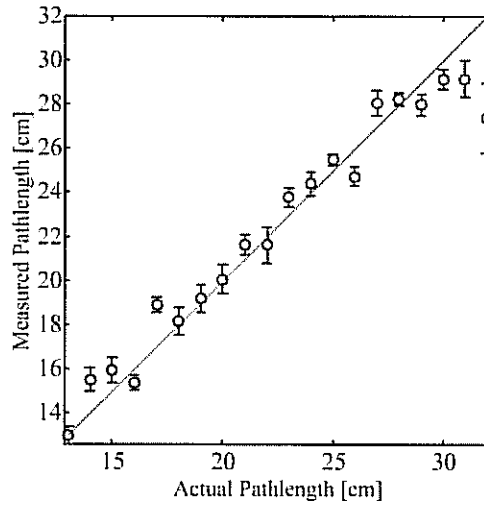


Figure 5. Pathlength response to air calibration rig.

4.2. Calibrating the Adjustable Attenuator

Amplitude to phase crosstalk is a common problem for phase measuring devices. The amplitude-phase crosstalk in the I&Q demodulator is reduced by the adjustable attenuator providing a constant (± 1 dB) amplitude over a maximum range of 30 dB in the RF input. All electronically adjustable attenuators however have an inherent phase shift and this phase shift is accounted for by a *one off* calibration of the attenuator. A portion of this calibration plot is shown in Figure (6).

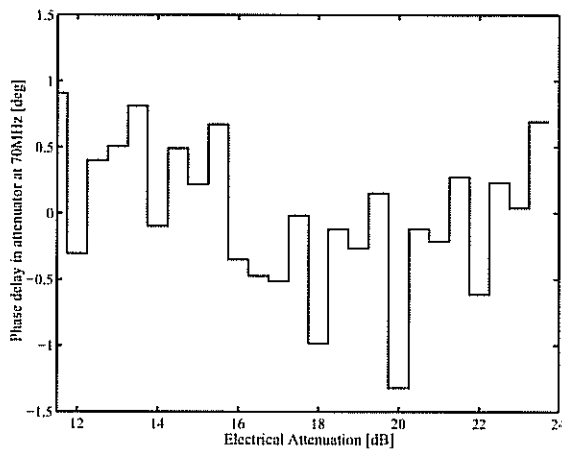


Figure 6. Dependence of phase delay on attenuator setting.

The calibration rig shown in Figure (4) allows adjustment of optical signal amplitude without any change in phase. The amplitude of the signal at the RF input of the demodulator is measured and kept constant by rotating the adjustable optical attenuator. The phase shift corresponding to each attenuator setting is recorded by this method and incorporated into a look up table for the microcontroller software.

5. PATHLENGTH MEASUREMENTS ON A LIQUID PHANTOM

To test the pathlength meter, a liquid phantom was constructed, consisting of intralipid solution together with a NIR absorbing dye (ICI S109564). The μ'_s of the initial emulsion was 2mm^{-1} and the μ_a was 0.01mm^{-1} . Liquid phantoms of this type are generally accepted to be a good approximation of tissue.⁸ The test solution was held in a opaque box ($180 \times 180 \times 110$ mm) into which was mounted windows made from microscope coverslips which allowed the mounting of the optodes of the pathlength meter with an optode center to center separation of 4cm. The pathlength meter monitored the optical pathlength in the emulsion for 500 seconds with a filter bandwidth of 0.1 Hz. Figure (7) shows measured optical pathlength plotted against time for four different values of μ'_s .

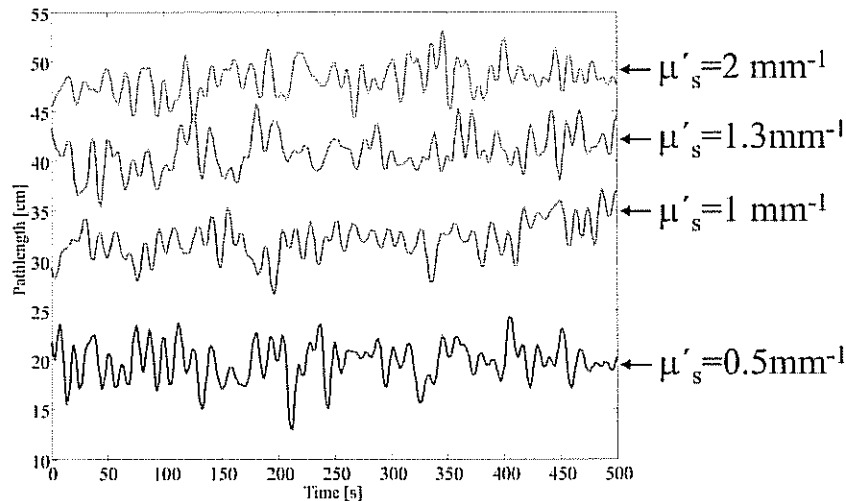


Figure 7. Pathlength output for a selection of scattering solutions with varying values of μ'_s .

The concentration of the emulsion was diluted by adding a solution of the dye dissolved in pure water. In this way, the value of μ'_s is reduced while the value of μ_a stays constant. Several dilutions of the solution were made with reduced scattering coefficients of 1.33, 1.0 and 0.5mm^{-1} .

6. CONCLUSIONS

A simple and robust instrument which measures the optical pathlength in tissues has been developed to improve the accuracy of a clinical oxygenation monitor (NIRO-300). A transfer method of absolute calibration of the pathlength provides a convenient method to calibrate the system against the pathlength in air. The short measurement duration and calibration before each measurement removes the effects of thermal drift and optode length from the pathlength measurement. The absolute error of the pathlength meter is $\approx 1\%$.

7. ACKNOWLEDGEMENTS

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