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A tissue-like optically turbid and electrically conducting phantom for simultaneous EEG and near-infrared imaging

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Abstract

We describe a phantom for simultaneous electroencephalography (EEG) and near-infrared imaging which consists of a solid, optically turbid and electrically conducting interface enclosing a tissue-mimicking aqueous scattering solution. The interface provides an electrical contact impedance comparable to that of the human scalp while the phantom as a whole has optical properties and electrical conductivity equivalent to that of head tissue. The construction is described and our design is evaluated experimentally using an optically absorbing target which also provides an EEG-equivalent electric field source. The results of this simultaneous EEG and near-infrared imaging experiment are presented.

1. Introduction

Systems which are capable of observing functional activity of the human brain can be divided into two fundamental categories: those which measure the electromagnetic dynamics of groups of active neurons such as electroencephalography (EEG) and magnetoencephalography (MEG) and those which measure changes in the state or concentration of certain molecules as a result of the increased metabolic demand of these active neurons such as functional magnetic resonance imaging (fMRI), positron emission tomography (PET), single photon emission computed tomography (SPECT) and near-infrared (NIR) imaging.

Recently, research has focussed on exploiting the complementary physiological information available via these two types of functional imaging. Simultaneous fMRI and EEG has, despite considerable technical challenges, been successfully performed and is increasingly used in studies of functional processing, epilepsy and neurovascular coupling (Kruggel et al 2000, Goldman et al 2002, Gotman et al 2006). There has been a similar surge in research combining EEG and NIR techniques, particularly for the study of epilepsy (Roche-Labarbe et al 2008, Gallagher et al 2008).

NIR imaging techniques exploit the differences in the optical absorption spectra of oxy and de-oxyhaemoglobin over NIR wavelengths to produce two- and three-dimensional images of cerebral haemodynamics (Gibson et al 2005). The concentration change of
de-oxyhaemoglobin during functional activation of the brain gives rise to the blood oxygen level-dependent (BOLD) signal observed in fMRI. NIR imaging systems are portable, and only require an array of optical fibres to be coupled to the head of the subject. As a result, NIR techniques can be applied at the bedside and are significantly less susceptible to movement artefact than fMRI.

The ability to combine EEG and NIR techniques in a versatile and robust manner is particularly desirable in neonatal medicine as the combined observation of electro-cortical and haemodynamic variations will provide new insight into the nature of neonatal seizure and other encephalopathies and potentially aid diagnosis (Toet and Lemmers 2009).

As the two observation systems are completely compatible, the most significant challenge in performing simultaneous EEG and NIR imaging is in optimizing the application of a sufficient number of EEG electrodes and optical fibre bundles. We have designed a combined electrode/optical fibre bundle (‘opto-electrode’) which allows optimal co-registration of the two modalities, is easy to apply and minimizes the occupied area of the scalp, thus maximizing our possible sampling density (Cooper et al 2009). To evaluate this and other probe designs, it was desirable to perform simultaneous EEG and NIR imaging experiments on an object which mimics the optical and electrical properties of the brain.

A broad variety of tissue-equivalent phantoms have been developed to evaluate optical imaging systems (Pogue and Patterson 2006). However, for the purposes of combined EEG and NIR imaging, we require a phantom which not only possesses tissue-equivalent optical properties but also provides a source of a varying electrical potential difference of an amplitude comparable to that of scalp-recorded electro-cortical activity. In the experiments previously described (Cooper et al 2009), we employed a liquid-based phantom and a submerged current dipole to provide the electric field and placed the opto-electrodes in direct contact with the solution. This successfully provided a test of simultaneous data acquisition but provided no measure of the quality of the electrical contact of the opto-electrode. This arrangement also risked damaging the opto-electrodes as they had to be completely submerged in the aqueous solution. We therefore explored whether a solid material could be developed which possesses scalp-like electrical conductivity and tissue-mimicking optical properties. Such a material could be used to provide an interface between the opto-electrode probe and a tissue-mimicking aqueous solution. This layered arrangement would allow different opto-electrode designs to be precisely assessed in terms of their optical and electrical coupling properties.

In this paper, we describe a phantom for simultaneous EEG and NIR imaging which employs such a solid, optically turbid, electrically conducting interface enclosing a tissue-mimicking aqueous solution.

2. Interface design

The optically scattering and electrically conducting interface is based on a polyester resin (Alec Tiranti Ltd, UK) to which an absorbing dye (pro-jet, Avecia Inc., USA) and a scattering compound (superwhite resin pigment, Alec Tiranti Ltd, UK) are added prior to curing so as to provide optical absorption and scattering properties equivalent to that of bulk tissue ($\mu_a = 0.01 \text{ mm}^{-1}$, $\mu_s = 1 \text{ mm}^{-1}$ at a wavelength of 800 nm). The resin mixture is poured into a shallow silicone rubber mould with dimensions of 145 × 100 × 4 mm and allowed to cure so as to produce a thin slab.

In order to make the interface electrically conducting, a dense array of 16 mm long, 0.25 mm diameter gold-plated copper wires (Griffin, Germany) is embedded in the rubber mould prior to the addition of the resin mixture. The lengths of wire are arranged vertically in a square grid with a separation of 2.5 mm. Once the resin has cured, each wire is left
protruding by approximately 6 mm from each surface of the resin slab. The wires on one face of the slab are then cut and abraded until each is flushed with the resin surface. In this arrangement, the gold-plated wires lie parallel to the optical path of the NIR light coupled into and out of the phantom via optical fibre bundles. Because the wires are fine, they do not significantly affect the absorption characteristics of the interface and therefore they have a negligible impact on NIR imaging measurements. A gold-plated wire was chosen because of its high conductivity, chemical stability and availability.

A typical silver/silver-chloride EEG cup electrode has a diameter of around 9 mm. When coupled to the surface of the interface, such an electrode will cover a minimum of nine wire cross-sections. The spatial density of wires embedded in the interface was chosen because previous experiments had indicated that it would provide an electrode contact impedance similar to that of the human scalp (see section 3.1). The final interface consisted of wire grid containing a total of 1189 lengths of wire distributed over an area of 100 × 70 mm. Figure 1 shows the interface design and the completed phantom.

3. Phantom design

To complete the dual-modality phantom, the interface was arranged as one of five sides of an open-top tank with dimensions 145 × 100 × 100 mm. The tank was filled with an intralipid

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**Figure 1.** The phantom interface design in plan (a) and cross-section (b). (c) The completed phantom.
and dye solution with the same optical properties as the interface (\( \mu_a = 0.01 \text{ mm}^{-1}, \mu' = 1 \text{ mm}^{-1} \) at a wavelength of 800 nm) and a 0.2% saline concentration so as to provide electrical conductivity similar to that of brain tissue (Tidswell et al 2001). To provide optical contrast, a resin cylinder 20 mm in diameter and 20 mm in height with an absorption coefficient of 0.04 \text{ mm}^{-1} was attached via an insulated arm to a three-axis micrometer stage and suspended in the solution. A current dipole (produced using two silver/silver-chloride pellet electrodes as described in Cooper et al (2009)) was embedded at the centre of the resin cylinder and used to mimic an EEG ‘source’. Our EEG ‘source’ and the optically absorbing target were thus co-located. The dipole was connected to the positive and common terminals of an 11 Hz sine wave generator with a peak-to-peak amplitude of 40 mV. This amplitude was found to produce an electrical potential difference oscillation at the phantom surface comparable in amplitude to scalp EEG measurements. An oscillation frequency of 11 Hz was chosen to mimic the dominant frequency component of normal, resting adult EEG.

3.1. Phantom contact impedance

In order to mimic the properties of tissue as closely as possible, it was desirable for the phantom to provide an electrode contact impedance of the order of 5 k\( \Omega \), a value which is regularly quoted as the desirable upper limit of electrode contact impedance for in vivo EEG recording. The electrode contact impedance was measured using two standard silver/silver-chloride EEG electrodes connected to an electrode impedance meter operating at 10 Hz (Checktrode\( ^{\circledR} \), UFI Inc., USA). The electrodes were positioned 30 mm apart and coupled to the interface surface using a standard EEG contact gel. The contact impedance was measured at 20 random locations on the interface. The mean contact impedance was found to be 6.1 k\( \Omega \) with a standard deviation of 1.1 k\( \Omega \).

3.2. Simultaneous NIR imaging and EEG recording

To evaluate the efficacy of the phantom, a simultaneous NIR imaging and EEG experiment was performed using an array containing 12 of the opto-electrode probes previously described by Cooper et al (2009). This array consisted of three rows of four opto-electrodes covering an area of 60 \( \times \) 40 mm. Six of these opto-electrodes were used as NIR sources and 6 as NIR detectors, with all 12 employed as active electrodes. The target cylinder/dipole source was placed in three positions: at the centre of the sampled area, 22 mm to the left and 22 mm to the right of the centre of the sampled area. The target was kept at a depth of 12 mm below the opto-electrode array and sets of NIR and EEG data were recorded at each of the three positions. A fourth NIR data set was recorded with the target absent from the tank so as to provide a reference for NIR image reconstruction.

NIR imaging was performed using the UCL optical topography system as described by Everdell et al (2005). This system employs up to 32 laser diode sources (16 at 690 nm and 16 at 850 nm) and 32 avalanche photodiode detectors. For this experiment, only six 690 nm sources were required. Optical images were reconstructed using a linear image reconstruction algorithm based on the Rytov approximation (Gibson et al 2005). An image of the change in optical absorption coefficient was computed using the difference in log-amplitude intensity recorded across each source–detector pair between each of the three sample data sets and the reference data set.

EEG data were recorded using a Grass-Telefactor H2O 32-channel EEG recording system (Grass Technologies, Astro-Med Inc.) with a sample rate of 400 Hz. The ground and reference electrodes were placed centrally at the rear of the phantom tank. Data were acquired using
3.3. Results

Figures 2(a)–(c) show the reconstructed images of the change in absorption coefficient measured at a depth between 15 and 20 mm using the UCL optical topography system for three positions of the cylinder target. Figures 2(d)–(f) are representations of the corresponding EEG measurements obtained by interpolation of the RMS amplitude of oscillation measured at each of the 12 active electrodes. The black cross in each figure indicates the actual position of the centre of the target cylinder/current dipole. As is apparent from these images, the spatial localization of the optical reconstructions is very good, with an average 2D error of approximately 2.1 mm from the peak change in absorption coefficient to centre of the target. The interpolated EEG images are a representation of the measured electric field rather than a position of the dipole and therefore are expected to be diffuse in nature. They do however show some spatial localization, with a peak position in each which corresponds approximately to the position of the dipole.

These experiments establish that the phantom successfully provides a target for combined NIR imaging and EEG recording whilst also possessing tissue-like optical properties and a solid contact surface with an electrical impedance comparable to that of the human scalp.
4. Discussion

The layered, electrically conducting, tissue-equivalent phantom described here is inexpensive, portable, re-usable and relatively easy to build. Having a robust and reproducible device on which to test the efficacy of combined NIR imaging and EEG contact systems will be of real benefit as the demand for and interest in such systems continues to grow. This phantom (and more complex versions of it, which may incorporate a series of layers of varying conductivity so as to further mimic the tissue of the head and brain) could also be employed to allow the acquisition and interpretation of dual-modality NIR imaging/EEG data to be optimized. For example, the phantom would allow software paradigms which enable NIR-informed EEG source localization to be tested in the laboratory in a simple and reproducible manner.

References


Toet M C and Lemmers P M A 2009 Brain monitoring in neonates *Early Hum. Dev.* **85** 77–84