Imaging of Motor Activity in Freely Moving Subjects Using a Wearable NIRS Imaging System

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Abstract: We present a miniaturized multi-channel NIRS imaging system for functional brain imaging in unrestrained settings suitable for any aspect of the head. Performance is demonstrated in a motor execution paradigm performed during bicycle riding.

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1. Introduction

Diffuse optical brain imaging’s potential for miniaturized, highly integrated, and unconstrained experimental settings has often been cited as one of the main advantages over other existing neuroimaging modalities. However, this claim has so far only partially be met; many (in fact, probably most) of the NIRS setups employed in neuroscientific research still offer a restrained setting because typically, the subject is tethered by (not very flexible or lightweight) fiber-optic cables to a more or less stationary instrument. Some developments towards miniaturized probe arrays or instruments have been described before; however, these are often restricted to specific portions of the head (i.e., forehead), and/or provide only a limited number of measurement channels and do not lend themselves readily to general-purpose large-area imaging of brain activity [1-3].

2. Instrumentation

The instrument utilizes a multi-purpose data acquisition (DAQ) board (NI-USB 6216, National Instruments, Austin, TX, USA (NI)) for all signal generation and measurement purposes. The device is connected to the host computer through a USB 2.0 connection for data transfer and power supply. The instrument control, data display and storage are accomplished through a graphical user interface developed under NI’s LabVIEW 2011 environment. All custom designed additional circuitry such as detector signal conditioning, source driving amplifiers, digital control, and power conditioning is contained on a single 160 mm x 100 mm printed circuit board which is powered by a second USB port. The DAQ board attaches to this main board so both form a single unit, which is housed in an aluminium enclosure (103 mm x 43 mm x 167 mm). The imager provides 8 illumination channels of 2 wavelengths each and 8 parallel detection channels.

Illumination is achieved by dual-wavelength light emitting diodes (LED, type: L760/850-36 by Epitex Inc., Japan), which deliver approximately 10 mW of optical power for each wavelength (760 nm and 850 nm). The LEDs are brought into direct contact with the tissue so as to minimize coupling losses. The source positions are time-multiplexed, yielding an overall sampling rate of 6.25 Hz. Both wavelengths are frequency-encoded in the low kHz range and distinguished by signal demodulation in the acquisition software. LEDs are housed in a custom-made cylindrical plastic housing, which can be attached to fabric caps such as worn for electroencephalography (EEG). The LEDs are powered through about 300 mm long segments of miniature multi-wire cables, which in turn are connected to a flat ribbon cable in groups of eight. The ribbon cable terminates at a 20-pin rectangular header in the instrument’s front.

To overcome the main drawbacks associated with fiber-optic based detection, we designed an active optical sensor (AOS), which places the photosensitive element outside of the main instrument, close to the skin (see, Fig. 1). In this way we aimed to avoid the losses, bulk, and weight usually imposed by fiber optic bundles. The optical detection is performed with Si photo diodes (PD, BPW34, Siemens, Germany) which are coupled to the tissue by means of 12-mm segments of plastic optical fiber (POF, 3-mm OD, NA = 0.50, type NT53-833 by Edmund Optics, Barrington, NJ, USA). PD and POF are housed in an enclosure similar to that for the LED. This enclosure is internally shielded and contains active amplification (10-MΩ transimpedance amplifier) and filtering circuitry to provide sufficient electronic noise-immunity for signal pickup and transmission. The active detectors are connected to the instrument by means of miniature wires and a ribbon cable, similar to the LEDs, which serve to power the AOS and to transmit the amplified electronic signal to the main instrument.
Both LED and AOS are shown in Fig. 2 below. The individually wired illumination sources and detectors can be inserted in fabric caps and provide great flexibility in creating application-specific source-detector-geometries. There is no fixed source-detector-channel relationship that needs to be maintained; as long as a detector is within measuring range (ca. 30 mm) of a source, this optode pair will create a valid measurement channel. This last fact is the consequence of employing source position multiplexing together with adaptable detector amplification. Each detector channel has programmable gain settings of 1x, 10x, and 100x which are engaged synchronously with the switching of source positions to maintain optimal signal-to-noise-ratio for each data channel. The principle of source-synchronous gain switching of the detectors follows the description found in [4].

3. Experimental Design

The NIRS imager and controlling notebook computer were contained in a backpack worn by the subject. The optodes were arranged in two groups, each consisting of four sources and four detectors, resulting in two clusters of 10 NIRS channels each. The optodes were placed in positions of the extended international EEG 10-20 system and were clustered around positions C3 and C4, thereby ensuring coverage of the primary motor areas of both hemispheres. The inter-optode distance between neighboring source-detector-pairs was 25±3 mm (see Fig. 3). A secondary fabric cap worn over the optical probes served to improve probe-tissue-contact and to provide stabilization against motion artifacts. In addition, a jacket hood worn over the imaging array provided additional shielding from ambient light.
The subject also was equipped with in-ear headphones connected to the control laptop over which auditory commands were issued.

After initiating the recording, the subject started biking along a pre-described, straight and level bike lane at a moderate speed. An initial baseline period of 3 min was allowed for warming up and settling of the cardiac rate. The LabVIEW program controlling the imager also served to provide the subject with auditory cues for the motion paradigm. Following the initial baseline phase, the auditory commands ‘start’ and ‘stop’ were issued to indicate onset and cessation of right-hand motion, respectively. Time-stamped markers were recorded synchronously with these events. The paradigm during biking consisted of 10 repetitions of 20 s of self-paced (approx. 1 Hz) left-hand clamping followed by 40 s of rest. To provide controlled resistance to the hand clenching, an additional dummy hand brake lever was mounted to the handle bar which the subject was squeezing without causing the bicycle to brake. For safety reasons and in case of needed assistance, the subject was accompanied by another bicycle rider at all times during the experiment.

All data analysis and display was performed in MATLAB (The MathWorks, Inc.). The raw signals were subjected to band-pass filtering ($f_{\text{cutoff, low}} = 0.016$ Hz, $f_{\text{cutoff, hi}} = 0.25$ Hz) to reject physiological artifacts such as heart rate oscillations and drifts. We then applied the modified Beer-Lambert law to estimate relative changes in oxy-(HbO) and deoxygenated hemoglobin (HbR) for each measurement position. To identify neural activation we obtained the temporal mean of the estimated hemoglobin concentration changes for each channel during the time of the supposed maximal neural activation, between 5 s and 25 s after motion onset for each of the 10 trials. Within each channel, these values were t-tested for significance, and the obtained $p$-values were corrected following Bonferroni, i.e. the significance level was set as $(p = 0.05) / (\text{no. of channels}) = 0.0025$.

4. Results and Discussion

Fig. 4 shows the results. The averaged concentration changes were topographically projected onto the cortical surface obtained from the structural magnetic resonance (MR) image of a healthy volunteer (male, age 41). The co-registration of the NIRS channels to the anatomy was performed using the extended international 10-20 system. A focal activation to left hand clenching is clearly revealed by the relative HbR decrease over the contralateral sensorymotor motor cortex (source 8 / detector 6: $T= -5.66$, $p < 0.001$). The time courses show the prototypical increase in HbO and decrease in HbR in response to neural activation. The resulting activation pattern to left hand clenching during bike riding demonstrate the feasibility of our compact NIRS device for realistic studies in the field.

Fig. 4. Left: Topographic projection of relative HbR changes onto a representative cortical surface map using the 10-20 EEG electrode system. Center: Estimated HbO (red) and HbR (blue) concentration changes averaged over 10 trials of hand clenching during biking ($t = 0$ s marks clenching; yellow: 20-s activation period). Right: Individual trial responses demonstrate excellent stability against motion artefacts.

This demonstration of a truly portable and miniaturized NIRS technique opens new perspectives to study sensory or cognitive paradigms in realistic environments and furthermore promise clinical uses as a monitoring tool in neuro-rehabilitation and intensive care units.

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6. References