Investigating hemodynamics in scalp and brain using high-resolution diffuse optical tomography in humans

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Abstract: We investigate prerequisites for developing a continuous wave NIRS bedside brain perfusion monitor. Based on the separation of superficial and cortical layers using 3D HR-DOT we investigate the power and distribution of low frequency oscillations in the brain and scalp.

OCIS: (170.6960) Tomography; (170.2655) Functional monitoring and imaging

1. Introduction

In neurointensive care, bedside monitoring of brain perfusion can be an attractive option to evaluate the patient's pathologic state and to guide treatment. Imaging facilities which are widely used in clinical cannot be used for a constant monitoring. High-resolution diffuse optical tomography (HR-DOT) as a 3D modality of near infrared spectroscopy (NIRS) has the potential to close this gap and serve as a quasi-continuous brain perfusion monitor.

One of the main challenges in NIRS is the separation of signals from superficial layers like scalp and skin from signals that originate from the brain. In contrast to topographical NIRS which uses next-nearest neighbor distances of optical fibers for a planar (2D) backprojection, we use a HR-DOT system with a dense grid of optical fibers with overlapping photon paths and an image reconstruction procedure which allows us to recover absorption changes in different tissue depths. Recently we presented results from a cw-NIRS experiment in humans using a HR-DOT setup and indocyanine green (ICG) as an exogenic contrast agent [1]. We showed the separation of intra- and extracerebral layers by visualizing the early arrival of ICG in deeper and the later arrival in superficial tissue.

This work presents a first step toward investigating if – instead of bolus kinetics – intrinsic spontaneous oscillations could serve as markers for brain perfusion monitoring. Based on the findings of the DOT-ICG experiment, we investigate low frequency oscillations (LFO) in the 0.1 Hz band from different compartments of the imaged volume i.e., skin and brain layers. LFO are a often described in human brain imaging studies (for a summary see [2]) and may represent autoregulatory processes of cerebral blood flow. Monitoring these hemodynamic features and detecting changes in the individual patterns could be an additional module to noninvasively detect changes in cerebral blood flow in neurointensive care patients.

2. Methods

We used data from the experiment that is described in Habermehl et al., 2011 [1]. Three healthy, voluntary subjects (2 male, mean age 38 years) were investigated. Relative absorption changes in a resting state were measured using the DYNOT tomography imager (NIRx Medizintechnik GmbH, Berlin, Germany), applying light of λ=760 nm & λ= 830 nm, 30 co-located optical fibers (inter-optode separation: 7.5 mm) in a 5 x 6 optical fiber grid (Fig. 1a) arrangement, obtaining 900 overlapping optical data channels with a sampling frequency of 1.81 Hz. We reconstructed a time series of relative absorption changes using the normalized difference method [3]. The weight matrix was determined using BrainModeler (NIRx Medical Technology LLC) which provides a library of subvolumes from a MRI-scan based finite element mesh (FE) with precalculated inverse parameters (based on diffusion equation, homogenous interior optical properties (μ_a= 0.06 cm^{-1}, μ_s= 10 cm^{-1})). The reconstruction procedure results in a 2D array containing reconstructed time courses for all nodes of the FE mesh. For visualization purpose, the results were later transformed into a volume using Matlab's standard routines.

To investigate LFO, we band pass filtered (0.083-0.125 Hz) the reconstructed time series and calculated, the power spectral density (PSD) for the reconstructed oxygenated (HbO) and deoxygenated hemoglobin (HbR) time courses in every node of the FE mesh (Welch method, hamming window size 90, 50% overlap).. Regions of interest (ROI) in the investigated FE mesh, based on the findings in [1] (Fig. 1c) were determined, belonging either to skin or brain (Fig. 1d). Finally the results of PSD for all nodes belonging to one ROI and throughout the three subjects were averaged.

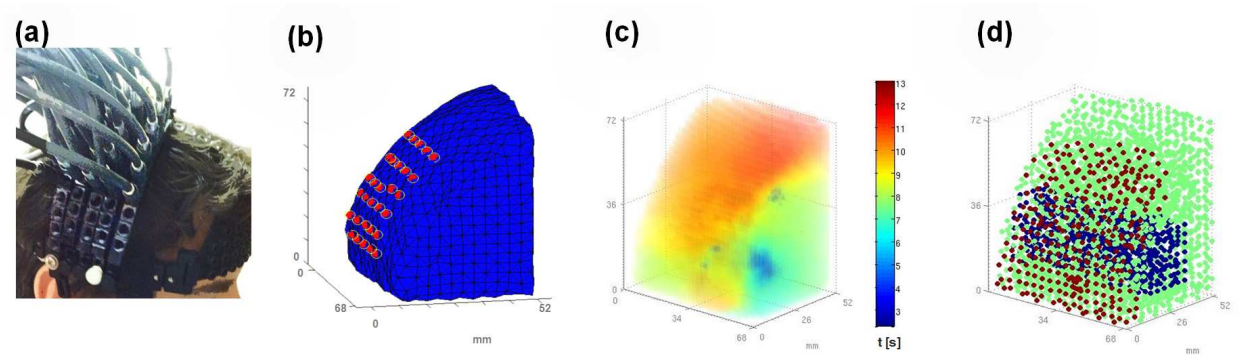

Fig. 1 (a) Imaging setup. Absorption changes were measured with a HR-DOT imaging system (DYNOT, NIRx Medizintechnik GmbH, Berlin, Germany). A 5x6 fiber grid with 30 co-located sources and detectors was placed pericentrally over the right hemisphere. (b) Finite element mesh that was used for image reconstruction of relative absorption changes. Red dots indicate the positions of the optical fibers in the forward geometry. (c) Result from an experiment, described in [1]. The arrival time of the absorber (ICG) is shown color-code in s after bolus injection. The bolus arrives first in the deeper (cortical) areas and few seconds later in the skin. (d) Based on the results shown in (c) different ROI (red = skin, blue = brain) were determined in the FE mesh.

3. Results

We determined the PSD for LFO for both skin and brain voxels in the defined ROI (Fig. 1d) for the non-normalized HbO and HbR time courses. Fig. 2a) depicts the averaged PSD. Regardless of the amplitude, we find LFO in both compartments of the volume, brain and skin. Not surprisingly, we see the highest power for HbO in skin voxels. Due to the higher absolute values for absorption changes in superficial layers and a bigger contribution of HbO to hemodynamic signals, we find the highest power for this chromophore$_{ROI}$-combination. This higher contamination of HbO with systemic signals is also seen in the 6 fold higher amplitude of HbO$_{SKIN}$ compared to HbR$_{SKIN}$ and additionally in the higher amplitude of HbO in the brain voxels.

Additionally to the analysis of LFO in averaged, pre-defined ROI we were also interested in the distribution of LFO in the reconstructed volume. Based on our previous results, showing distinct patterns of the arriving ICG in the skin [1], we assumed a non-homogenous distribution of these oscillations. Fig. 2b) shows the result from a calculation of the PSD for every voxel in the medium for one subject and HbO time courses. The PSD values were normalized for each voxel to reduce the influence of the much higher sensitivities on the surface. We find a highly non-homogenous result with focal high values in the skin but also in deeper layers indicting underlying structures with different perfusion.

Fig. 2c) shows the averaged time courses for brain and skin ROI for one subject. We see a slight phase delay of skin voxels compared to brain voxels in LFO which are confirmed by a cross-correlation of both time courses revealing a lag of 2 sampling points of the time course for the skin (Fig. 2d). This is an indication that oscillations first appearing deeper layers and arrive later in the skin. Fig. 2e) shows again the reconstruction volume with each voxel color-coded with the delay (in sampling points) for LFO compared to an averaged brain time course as a reference. Again we see inhomogeneous distribution of the oscillation phase.

4. Conclusions

In this work we present results from the analysis of LFO in different compartments of the head like skin/scalp and brain. We showed that LFO as hemodynamic features appear in the skin and also in the brain. The higher contamination with LFO in HbO time courses is in line with results from many groups using topographical NIRS, finding more systemic answers in HbO time courses. The appearance of LFO also in brain voxels shows the necessity of a careful preprocessing when functional NIRS data are analyzed. Analysis of the distribution revealed an irregular distribution of LFO in the investigated medium. Finding acute changes in this individual distribution pattern of LFO could be one possibility to monitor changes in perfusion in neurointensive care patients. Because the sampling rate of the imaging setup used was limited to 1.81 Hz,we could not reliably monitor the cardiac pulse. Currently, we are implementing an imaging setup with an increased sampling rate to allow us to extend the analysis to other hemodynamic frequency bands of interest, such as cardiac pulse.

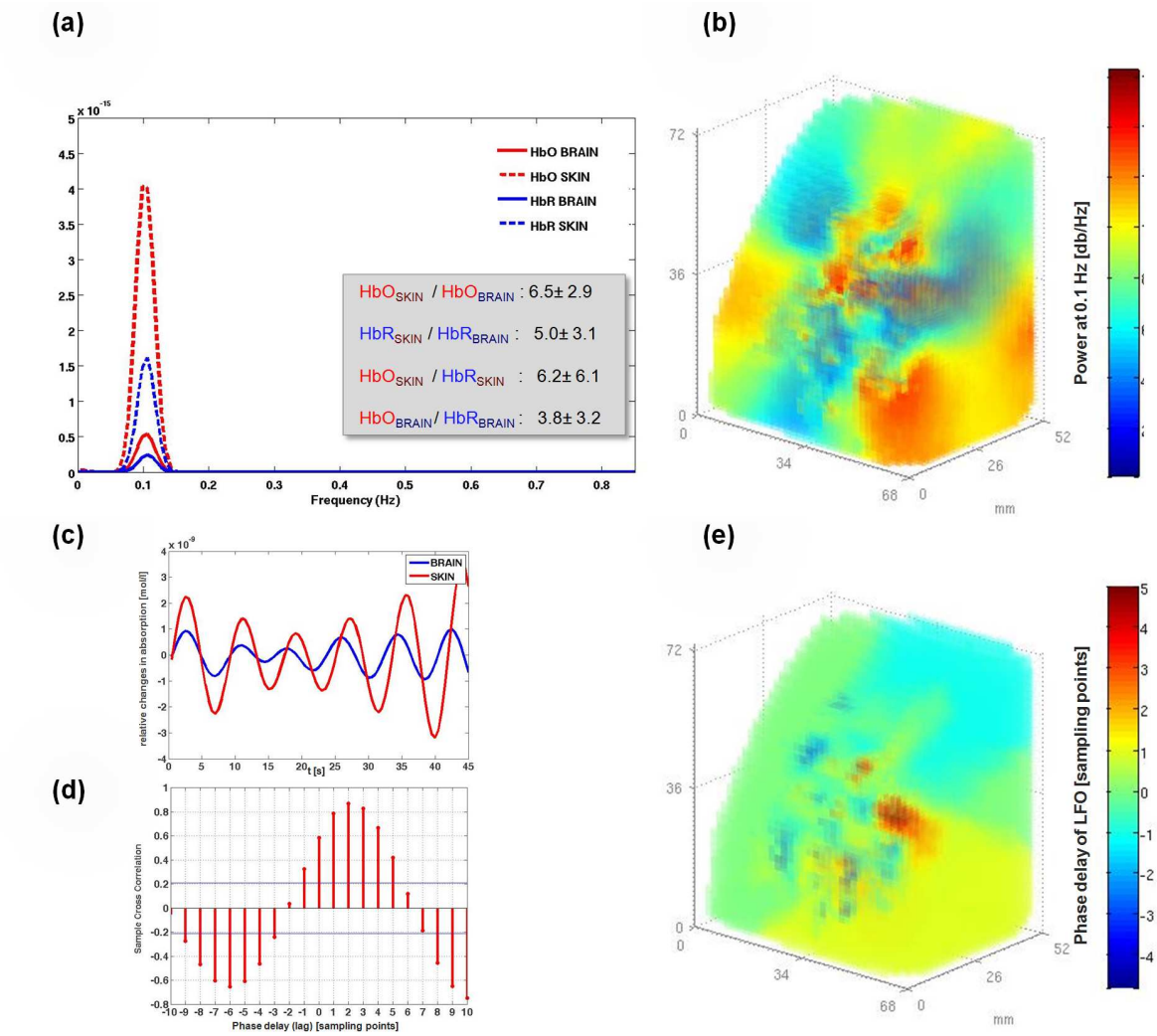

Fig. 2 Averaged PSD for time courses from skin and brain ROI and both wavelengths and three subjects. (b) Reconstruction volume. Voxels are color-coded by the normalized power at 0.1Hz. We find a non-homogeneous distribution of the power throughout the volume. (c) Averaged time courses for one subject from brain and skin ROI. A slight phase delay in skin can be observed. (d) Cross-correlation of the two time course reveal a lag of the skin compared to brain of 2 sampling points (~1s)(e) Reconstructed volume, each voxel is color-coded by the lag that was determined by cross-correlating each voxel's time course with an averaged brain time course serving as reference. Again, we find an inhomogeneous distribution of the arriving oscillation.

5. References

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