Noise reduction in combined EEG/fMRI using a vector beamformer

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Introduction: BOLD fMRI achieves excellent spatial localization of brain activity, however the latency and longevity of the haemodynamic response means that fMRI suffers from poor temporal resolution. Conversely, electroencephalography (EEG) directly measures the electrical potentials generated by neuronal activity, and therefore offers excellent temporal resolution, but the ill-posed EEG inverse problem and inhomogeneous conductivity profile in the head mean that its spatial resolution is limited. For these reasons, simultaneous application of fMRI and EEG represents an attractive way of imaging brain function with high spatio-temporal resolution. Simultaneous application is difficult due to the EEG artifacts caused by the MR scanner. Techniques are available to remove such artifacts [1,2], however, even after correction, noise remains in the data, particularly at high field. Here, we demonstrate a vector beamformer [2] approach that, if applied in addition to standard correction techniques, further reduces artifacts in EEG/fMRI. We show that, as well as significantly reducing noise in EEG data, this approach allows electrical source localization and extraction of virtual electrode timecourses.

Theory: Assume that $\boldsymbol{m}(t)$ represents an n dimensional column vector of measurements made at n EEG electrodes at time t in response to a current dipole of strength $\mathbf{Q}$ at location $\mathbf{r}$. An estimate of $\mathbf{Q}(\mathbf{r}, t)$ can be made by a weighted sum of sensor measurements such that $\hat{\mathbf{Q}}(\mathbf{r},t)=\mathbf{W}(\mathbf{r})\mathbf{m}(t)$ where $\mathbf{W}(\mathbf{r})$ is a matrix of weighting parameters. The weighting parameters are defined based on minimisation of the overall amount of power emanating from within the source space (i.e. the head) with the linear constraint that power originating from the location of interest ($\mathbf{r}$) must remain in the output signal. The weighting parameters can be shown to be given by $\mathbf{W}(\mathbf{r})^T = \left[\mathbf{L}(\mathbf{r})^T\{\mathbf{C}+\mu\boldsymbol{\Sigma}\}^{-1}\mathbf{L}(\mathbf{r})\right]^{-1}\mathbf{L}(\mathbf{r})^T\{\mathbf{C}+\mu\boldsymbol{\Sigma}\}^{-1}$. Here, $\mathbf{L}(\mathbf{r})$ contains a single lead field pattern for a unit dipole at location $\mathbf{r}$ for each of the three possible dipole orientations (x, y, and z). $\boldsymbol{\Sigma}$ is a diagonal matrix representing the white noise at each of the EEG channels and μ is a regularisation parameter. Spatial filtering in this manner can be thought of as a technique of noise rejection since, in principle, any signal that does not originate from the location of interest, in this case $\mathbf{r}$, will be rejected by the spatial filter and not appear in $\hat{\mathbf{Q}}(\mathbf{r})$. For this reason, in combined EEG/fMRI, imaging artifacts caused by the MR scanner can be largely eliminated. By constructing a statistical map showing stimulus induced changes in $\hat{\mathbf{Q}}(r)$ as a function of $\mathbf{r}$, the location of electrical activity can be compared to that of the BOLD response. In addition, for selected regions, timecourses of electrical activity can be determined with high (ms) temporal resolution. These are known as 'virtual electrode' measurements.

Methods: Two subjects took part in this proof or principle study. EEG data were recorded using a Brain Products EEG system (Brain Products, Munich) with an MR compatible electrode cap containing 32 electrodes (sampling frequency of 5kHz). fMRI data were acquired using a 3T Philips Achieva (TR = 2.2s; matrix size 64x64; TE = 35ms; in plane resolution = 3.25mm. 20slices with slice width 3mm.) The visual paradigm used a simple checkerboard flashing at 10Hz. A single trial comprised 10s of stimulation followed by 20s rest, during which time only a fixation point was shown to the subject. The experiment comprised 30 trials.

Initial correction of EEG data was performed using Brain Vision Analyzer (Brain Products, Munich) and involved averaging and subtracting the gradient and pulse artifacts [1,2]. fMRI analysis was performed using standard techniques in SPM. A custom-written EEG beamformer was applied to the EEG data in Matlab. Lead field calculations were based on a triple-sphere head model that has been described previously [4]. In order to compare source localization in EEG with the location of the maximum measured BOLD response, T-statistical images were derived in order to show the location of maximum change in oscillatory power. EEG statistical maps were compared to statistical maps showing areas of significant BOLD contrast (T = 0.05 corrected). EEG and BOLD timecourses were extracted from a region of interest derived from the BOLD statistical map. In order to show the noise reducing properties of the beamformer, power spectra from three occipital EEG channels were compared to those from the virtual electrode.

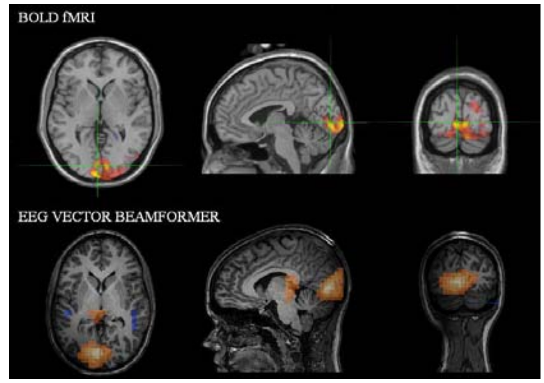

Figure 1: Localisation of brain activity using BOLD fMRI and the EEG beamformer.

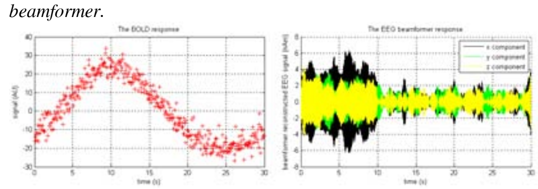

Figure 2: BOLD and EEG beamformer timecourses for a location of interest derived from the functional images shown in figure 1.

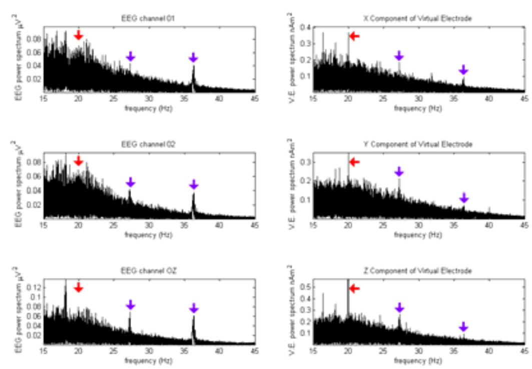

Figure 3: The noise reducing characteristics of beamformer analysis.

Results and discussion: Figure 1 shows localisation of brain activity using BOLD fMRI and the EEG beamformer. The two techniques show good spatial agreement, although the spatial resolution of the EEG beamformer remains low due to the simplistic head model used. Figure 2 shows both BOLD (left) and EEG (right) timecourses for a location of interest derived from the functional images. For EEG, the x-component is in black, the y-component in green and the z-component in yellow. An increase in activity in the virtual electrode during stimulus presentation ($0s<t<10s$) is shown. Figure 3 shows the noise reducing characteristics of the beamformer. The three traces on the left show power spectra of EEG responses from the three channels located close to the visual cortex (O1, O2 and Oz). The three traces to the right show the x, y and z components of the beamformer estimated timecourse. The 10Hz flashing checkerboard produces a maximum response at 20Hz, and whilst this 20Hz peak (marked by the red arrow) is clearly visible in the virtual electrode spectra, it is not visible in the raw electrode spectra. The peaks marked by the blue arrows were caused by residual gradient artifacts. These peaks are significantly reduced by beamformer analysis. Noise reduction by beamformers will increase with the number of available EEG channels [3]. This means that higher density MR compatible whole head EEG systems would further reduce the effects of scanner noise. The main limitation to the technique is accurate EEG forward modeling, however this is a general problem in EEG source localisation and is not specific to the EEG beamformer.

Conclusion: The EEG beamformer technique allows for source localization, timecourse extraction, and noise reduction in EEG/fMRI measurements. The technique has been proven here for the driven response to a visual stimulus, however it should also be possible to use this for comparison of non-phase-locked responses and the BOLD response. In separate but related work we have shown that this technique is especially valuable at higher fields (7T) where noise is more prominent.

References: 1) Allen *et al*. Neuroimage **8**:229-239,1998 2) Allen *et al*. Neuroimage **12**:230-239,2000. 3) Veen et al, IEEE Transactions on Biomedical Engineering 44(9), 1997. 4) Zhang, Phys. Med. Biol. 40:335-349. 1995.