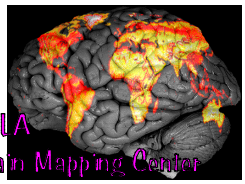
Abstract #6

Simultaneous EEG and *f*MRI Made Easy



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EEG and *f*MRI are independent windows to the brain that provide complementary information. EEG probes electrical potentials on the scalp as a marker for clinical, cognitive or neural states, but 3-dimensional localization of the EEG sources is challenged by the ambiguous relationship of the location of multiple electrical dipoles to the distribution of electrical potentials detected at the scalp. *f*MRI depends on variations in blood oxygenation content that are coupled indirectly to neural activity. Here we report a set of solutions to the technical problems in simultaneous recording of *f*MRI and EEG, which should aid our understanding and interpretation of both modalities.

Analog Processing

Differential Recording

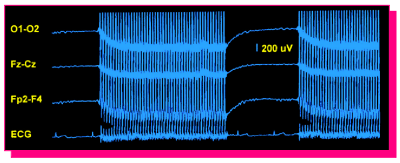
We have demonstrated that contamination of the EEG signal by the MR Gradients, Ballistocardiogram and Gradient Amplifier noise can be reduced by 6 to 10 dB with differential recording and the use of a chained and twisted set of lead connections [1]. Doing so minimizes the amplifier headroom requirements and reduces the possibility of saturation.

Low Pass Filter

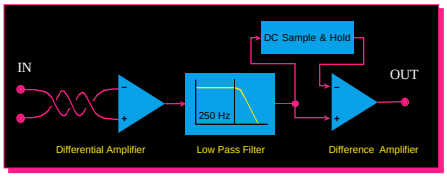
The gradient noise, however, is still nearly 100X larger than the EEG signal. In Echo Planar Imaging (EPI), most of the gradient energy is at more than 1 kHz. By inserting a 30 dB/octave Chebyshev low pass filter, centered at 250 Hz, we can reduce the gradient artifacts to be comparable to, or smaller than, the EEG.

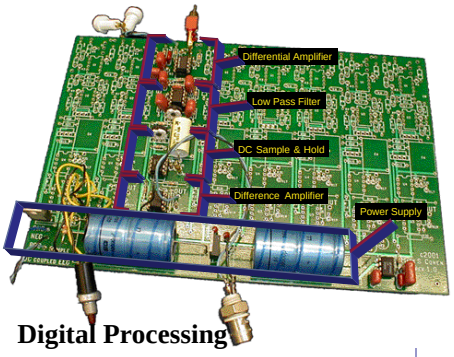
Differential Recording - Saturation Recovery

DC electrode potentials of a few milliVolts are typically present at the scalp. Because these can limit the useful dynamic range of the A/D converters, commercial systems AC couple the signals, usually with a high pass filter of 1 to 3 Hz. Such filters, however, introduce recovery transients with time constants of hundreds of milliseconds, during which the EEG signal is distorted heavily (as shown to the right, from published work of others).



To eliminate this problem we have developed a DC coupled circuit that samples, and then holds, the DC offset potential, and subtracts it from the input. In this manner, the EEG signal is centered about zero Volts in the analog to digital conversion. The circuit is shown in block form above with the components layout for one channel to the right. Each channel requires about 5 in^2.





Digital Processing

Ballistocardiogram

Like Lemieux, *et al.*, we have developed a cyclic averaging approach to eliminate the artifact from ballistocardiogram [1,2]. In our case, we record the EKG signal through a separate set of leads and use correlation measures to detect the QRS complex accurately with each heartbeat. We average the EEG signal, separately for each channel, following each heartbeat and then subtract this averaged signal from the raw EEG.

MR Field Gradients

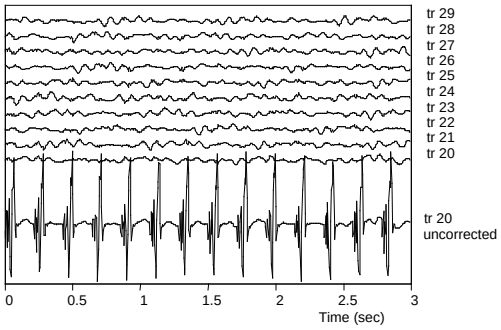
It is attractive to consider an approach similar to the ballistocadiogram suppression in removal of the gradient artifacts. However, small timing shifts of the digital sampling with respect to the gradient activity will result in incomplete removal of the artifact. With a sampling frequency f_s and an artifact frequency (e.g., due to gradients) of f_0, sampling errors of $\varphi = 2\pi f_0/f_s$ are to be expected. With ε as the residual artifact (the difference between the original and phase shifted signal) it is easy to show that:

$$\varepsilon = \cos(2\pi ft + \varphi) - \cos(2\pi ft)$$
$$= \cos(2\pi ft)\cos(\varphi - 1) - \sin(2\pi ft)\sin\varphi$$

which is essentially linear in φ for small errors, but can actually make the artifact up to twice as large as the original if the phase error is 180°. From this, we can calculate that the required sampling frequency to suppress the 1400 Hz (our EPI readout frequency) artifact by, for example, 100 fold is 880 kHz/channel, which is clearly prohibitive for a 32 channel system.

Synchronous Sampling

The above is not so much a problem of sampling rate, however, as one of sampling accuracy. If the ADC is linked precisely to the source of the artifact (the gradient), we can remove the artifact by subtraction even if we sample at low frequency. We have therefore implemented (in LabView 6i) a means of collecting a train of samples in each of the EEG channels that is resynchronized at each tr. Just as for the ballistocardiogram, we average the signal following each event and subtracted the cumulative average from the raw signal. The figure below demonstrates this for a single EEG channel.



On the bottom we show the "raw" signal differentially recorded from a pair of leads at the occiput during continuous echo planar imaging (tr=3s, 19 slices, 128x64 matrix), with the ADC triggered to synchronize with the start of each tr. The traces above, in blue, show 10 sequentially acquired periods of EEG after subtracting the average of 30 EEG traces and are essentially free of contamination by the scanning artifact.

Conclusions

Applications of this method to tomographic mapping of the spectrally analyzed EEG are the topic of a platform presentation by Goldman, *et al.*, [number 1291, Thursday 6/14] at this meeting. This approach is adaptable without modification to problems such as sleep staging and with minor modifications can be made equally effective in simultaneous ERP and fMRI.

1. R. I. Goldman, J. M. Stern, J. Engel, Jr., M. S. Cohen, Clin Neurophysiol **111**, 1974-80. (2000).
2. P. J. Allen, O. Josephs, R. Turner, Neuroimage **12**, 230-9 (2000).

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