Simultaneous Electromyography and fMRI

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Introduction

Measurement of complimentary behavioural data during fMRI has the potential to improve data analysis and interpretation. Electromyography (EMG), the measurement of bio-potentials produced during muscle contraction, is one such example. Simultaneous EMG during fMRI is critical to understanding behaviour in subjects where the motor output may not be intuitive, such as hemiparetic stroke patients. To date only interleaved EMG and fMRI has been demonstrated (Liu, Dai et al. 2000), principally because EMG signals are small (typically 0.1 – 2 mV) when compared to gradient noise that contaminates conventional EMG recordings. In addition, although of larger amplitude, EMG signals encompass wider frequency content (20-120 Hz) and are more stochastic than EEG signals. Consequently, investigation of optimal strategies to measure EMG signals during fMRI is required. Three different EMG pre-amplifiers are assessed in the present work. We hypothesize that MRI contamination of EMG will decrease as a function of the distance from the magnet iso-centre. This effect is explored for both wrist and ankle EMG during fMRI.

Methods

The EMG pre-amplifier was located 25 cm from the muscle of interest. To evaluate the effect of frequency range on EMG data quality, the following pre-amps were tested: #1 0.5 – 80 Hz (Gain: x1400), #2 15 – 80 Hz (Gain: x880), #3 33 – 80 Hz (Gain: x780).

Expt 1: EMG testing using a saline bath characterized the gradient-induced contamination as a function z-position from isocentre for the three pre-amps. **Expt 2:** Fifteen healthy control subjects (12 males, 3 females) participated in simultaneous EMG and fMRI experiments using an MR-compatible 8 channel optoelectronic system capable of recording EMG signals, and using a 3.0 T whole-body MRI system (Signa 3T/94 configuration, General Electric). Subjects performed flexion of the wrist (N=3) or ankle (N=12) in a 5 min event-related fMRI session (inter-task interval of 20 s) with spiral in/out k-space trajectories with T_2^*-weighted BOLD functional images (64 x 64 matrix, 14 slices, TR/TE/FA = 1000 ms/ 30 ms/50°). To test whether EMG data augment fMRI analysis, group activation maps for the ankle fMRI task were generated using 1) a conventional hemodynamic waveform based on visual cues and 2) a hemodynamic waveform that incorporated the EMG timing for each subject (i.e. the true task onset/offset).

Results

Expt 1: Gradient contamination was evident for all pre-amps, but to varying degrees. EMG power spectra revealed contamination at a principal harmonic (14 Hz, # slices/TR), as well as second harmonics. Fig 1 shows EMG contamination as a function of z and suggests all three pre-amps have low noise at z = 125 cm. **Expt 2:** Fig 2 shows representative EMG (ankle and wrist flexion) that demonstrates pre-amp#3 provides the best data. Fig. 3 shows increased brain activation by incorporating sub-second EMG timing.

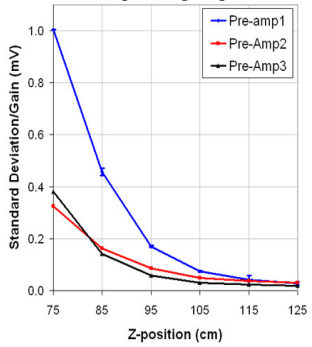

Fig 1. Expt 1: Gain normalized MRI-induced baseline noise in the EMG data as a function of proximity to the magnet iso-centre (z = 0 cm). The small standard deviation values depict minimal error during repeated measures.

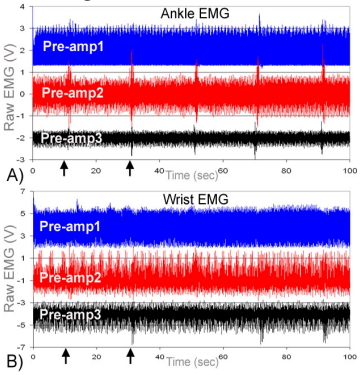

Fig 2. Expt 2: Raw EMG time series for A) 3 participants performing ankle movement, and B) 3 participants performing wrist flexion. Arrows indicate first two movement trials (i.e. inter-task interval of 20 s).

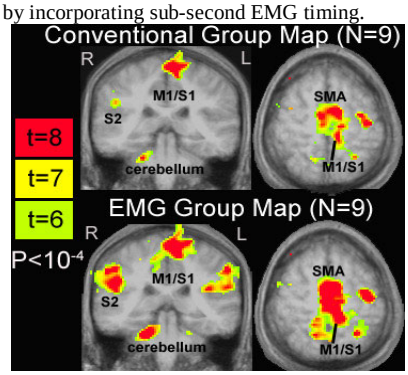

Fig 3. Group activation maps for 9 ankle fMRI participants. Incorporating precise EMG timing results in a significant increase in the activation volumes. Avg EMG burst onset occurred 1.2 ± 0.45 s after the visual cue and stopped at 2.6 ± 0.64 s later.

Discussion

These results show that after adopting an MRI-compatible EMG system with reduced lead wires, optoelectronics, and custom slew-rate limited pre-amplifiers, simultaneous EMG of the lower limb was possible in all three pre-amplifiers tested. Data from fig 2 shows only pre-amps #2 and #3 were capable of detecting wrist flexions and these signals were comparably small. Further development is required to improve our EMG system for measuring the upper limb.

Fig 3 demonstrates the significant improvement in fMRI activation patterns by using the precise EMG timing information compared to a conventional hemodynamic waveform. These results from healthy control subjects suggest that failure to account for individual, small differences in motor task performance can be detrimental, and that EMG recording is important for optimizing fMRI results.

Future EMG and fMRI experiments will be useful for biophysical characterization of BOLD signals, extending from work in which signals from MI neurons were used to predict EMG output (Morrow et al., 2003). Also, as part of our efforts in stroke motor recovery research, this system will play an important role in the interpretation of fMRI stroke data.

References

Liu, J. Z., T. H. Dai, et al. (2000). J Neurosci Methods **101**(1): 49-57. Morrow, M. M. and L. E. Miller (2003). J Neurophysiol **89**(4): 2279-88.