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Gradient distortions in EEG provide motion tracking during simultaneous EEG-fMRI

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Target audience: Users and developers of EEG-fMRI methods.

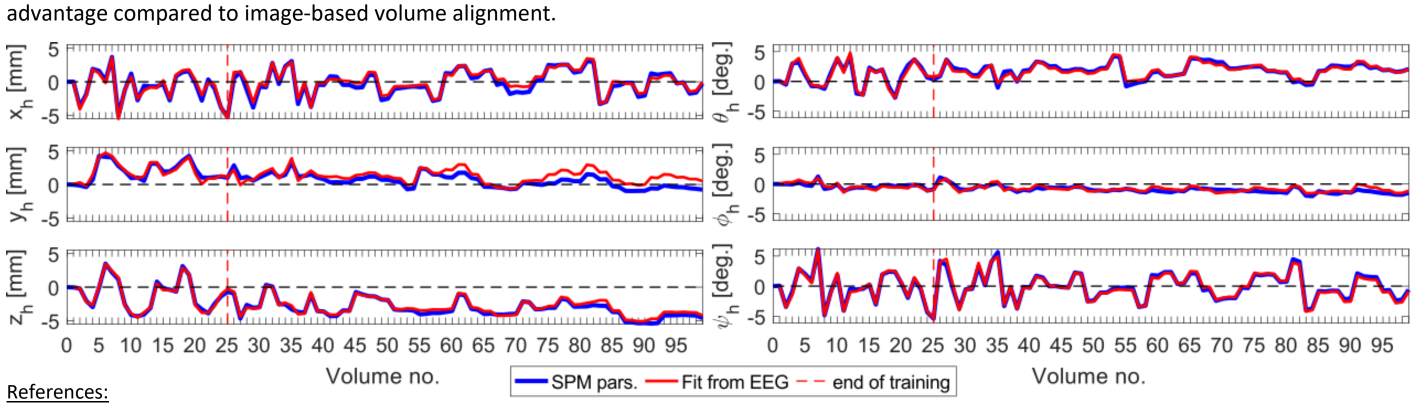
Purpose: Motion correction of fMRI using regressors derived from low-frequency components of EEG, was previously described [1]. We demonstrate that distortions in EEG caused by gradient-switching during normal EEG-fMRI can provide head motion tracking consistent with image-based retrospective analysis. Pilot studies [2] [3] [4] have shown that EEG electrode pairs interconnected to form high-impedance conductive loops fixed to the head can provide motion tracking. Here we use normal EEG-fMRI where the scanned subject is part of each conductive loop. Except for a calibration measurement, the method provides motion tracking “for free”. The method has no line-of-sight limitations and is applicable to setups with tightly fitting head coils. It requires no interleaved navigator modules or hardware beyond that needed to acquire simultaneous EEG-fMRI. It has potential for slice-wise prospective motion correction [3] with sub-millimeter spatial resolution [2] [4].

Methods: Changes of loop orientation following patient motion cause corresponding changes in gradient-induced electrode voltages. Faraday’s law describe the voltages from gradient switching: $V_i(t) = w_{ix}\frac{d\tilde{G}_x}{dt} + w_{iy}\frac{d\tilde{G}_y}{dt} + w_{iz}\frac{d\tilde{G}_z}{dt}$, where $\frac{d\tilde{G}_x}{dt}, \frac{d\tilde{G}_y}{dt}, \frac{d\tilde{G}_z}{dt}$ are altered versions of $\frac{dG_x}{dt}, \frac{dG_y}{dt}, \frac{dG_z}{dt}$ due to filters of the EEG-system and uncompensated eddy currents. The weights w_{ix}, w_{iy}, w_{iz}, depend on wire loop geometry, orientation, and position and therefore also on head position and orientation. Assuming rigid body movement, the change in weights, are by linear approximation (valid for small motion) described by $\Delta\boldsymbol{w} = \mathbf{A}\Delta\boldsymbol{r}_h$, where $\Delta\boldsymbol{w}$ is a $3\mathrm{N}$ element vector (N EEG-channels) with changes in weights relative to the starting position, $\Delta\boldsymbol{r}_h$ is a 6 element vector representing relative head position with six degrees of freedom $(x, y, z, \theta, \phi, \psi)$, and $\mathbf{A}$ is a $3\mathrm{N} \times 6$ matrix with partial derivatives of weights with respect to head position.

EEG-fMRI-like time series with long TR facilitating validation was acquired with a 61 channel MR-compatible EEG cap (Easycap, Germany) of which 11 channels were used (40kHz sampling rate, highpass-filtering at 32Hz to isolate gradient switching, NeurOne Tesla, Bittium Biosignals Ltd, Kuopio, Finland, & 3T Achieva MRI, Philips Healthcare, Best, Netherlands). A preparation scan of 3x30 volumes of Echo Planar Imaging (EPI) was acquired (TE/TR=30/4000ms, 42 slices of 3mm thickness, 80x80 matrix, 2.7mm in-plane resolution, tip angle 80°). During the preparation scan different pairs of gradients were deactivated so that induced voltages from gradients in x, y and z directions (physical scanner coordinate system) were measured individually. This was used to determine the time dependency of the EEG distortions. $\frac{d\tilde{G}_x}{dt}, \frac{d\tilde{G}_y}{dt}, \frac{d\tilde{G}_z}{dt}$. We then collected an EPI time series of 100 volumes using the above-mentioned parameters. To test the abilities of the setup, the subject was asked to perform random stepwise motion between volume acquisitions, this allowed the motion as estimated via the EEG data to be compared to movement estimated from the EPI data. The model was trained on the first 25 dynamics and tested on the remaining 75.

Results: Fig. 1 shows motion tracking with six degrees of freedom, using the EEG-model, and using SPM12 volume realignment (UCL, London, UK). A high correlation between motion parameters derived from images and from EEG, is seen.

Discussion & conclusion: The RMS error of the model is expected to increase over time, as continued patient movement may irreversibly change the loop geometries, or skin contact impedance. This is observed in tracking of y-translation in Fig. 1 (front-back, the degree of freedom with least movement in this experiment). Prospective motion correction [3] with continuous calibration of weights is expected to help. The RMS error of images aligned with the EEG-model is consistently ~10% higher than volume alignment with SPM, but this is not a concern considering the non-optimized setup. Further optimization, continuous calibration, and per-slice rather than per-volume tracking can potentially give the method an advantage compared to image-based volume alignment.



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