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Influence of Dense Array EEG Cap on fMRI Signal

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Abstract

Dense-array (>64 channel) EEG systems are increasingly being used in simultaneous EEG-fMRI studies. However, with increasing channel count, dense-array EEG caps can induce more severe signal dropout in the MRI images than conventional systems due to the radio frequency shielding effect of the denser wire bundle. This study investigates the influence of a 256 channel EEG cap on MRI image quality and detection sensitivity of BOLD fMRI signal. A theoretical model is first established to describe the impact of the EEG cap on anatomic signal, noise, signal-to-noise ratio (SNR) and contrast-to-noise ratio of BOLD signal. Seven subjects were scanned to measure and compare the T_2^* -weighted image quality and fMRI detection sensitivity with and without the EEG cap using an auditory/visual/sensorimotor task. The results show that the dense-array EEG cap can substantially reduce the anatomic signal in the brain areas (visual cortex) near the conducting wires (average percent decrease $\approx 38\%$). However, the image SNR with and without the EEG cap were comparable (percent decrease $< 8\%$, not statistically significant), and there was no statistically significant difference in the extent of BOLD activation. This suggests that the ability to detect fMRI signal is nearly unaffected by dense-array EEG caps in simultaneous EEG-fMRI experiments.

Keywords

EEG-fMRI; dense array; EEG cap; shielding; artifacts; SNR

Introduction

Simultaneous EEG-fMRI recording methods are beginning to achieve widespread use in functional neuroimaging studies (1, 2). In the detection of neural activity, EEG provides high temporal resolution ($\sim$ ms) but relatively low spatial localization due to regularization necessitated by the lack of a unique inverse solution. In contrast, fMRI is able to measure brain activation with superior spatial resolution, but the sluggish nature of the hemodynamic response limits the temporal resolution of the fMRI signal to the order of one second. Therefore, through simultaneous EEG and fMRI acquisition, the complimentary information provided by each may allow for improved characterization of neural activity in both the temporal and spatial domains under the assumption that the signals in the two modalities derive from the same neural source (1).

Although simultaneous EEG and fMRI signal recording offers many advantages, there are also some challenges in EEG-fMRI experiments. The greatest challenge has been the strong artifacts induced in the EEG recordings by the switching gradients as well as from ballistocardiogram effects (3, 4), but these concerns have been largely mitigated by synchronization of the scanner and EEG clocks (5-7) and algorithmic advances (3, 4, 8). An

important remaining concern is the influence of the EEG cap and its wiring on the image quality of MRI. The two components of EEG cap, the electrodes (even when made of non-metallic AgCl & plastic materials as used here) and the conducting wires, can produce susceptibility and electromagnetic shielding effects. These two effects can result in inhomogeneity of the main magnetic field (B_0) and radiofrequency (RF) field (B_1) respectively, and alter the MRI signal accordingly. Several groups (9-13) have evaluated the impact of the EEG cap on MRI image quality using EEG systems of up to 64 channels. Their results showed that the electrodes of the EEG cap can induce signal dropout and geometric distortion in MRI images, but the spatial extent of the artifacts is less than the minimum scalp-cortex distance in human subjects (~ 7.5 mm) (9). Negishi, et. al reported reduced radio frequency heating with the use of carbon fiber leads [13], but their system used only 32 channels and signal loss was not addressed. Thus, conventional EEG caps (less than 64 channels) have negligible influence on the MRI signal in brain areas, and therefore on BOLD signals.

Recently, dense array EEG systems (128 and 256 channels) have become available and are increasingly being used in EEG-fMRI studies (14) because they are able to provide higher spatial resolution for measuring the scalp EEG signal, which is thought to improve the accuracy of localizing cortical response (15). However, there are concomitantly more metallic conducting wires in the dense array EEG caps than the conventional systems of 64 channels or lower. In the typical experimental setup (Fig. 1a), the conducting wires are converged into a bundle at the back of subject's head and brought out of the coil for connection to the EEG amplifier. When the subject wears such an EEG cap, the dense wires act like a metallic conductive surface that can produce a strong electromagnetic shielding effect. This may result in a large MRI signal dropout in posterior brain regions close to the conducting wires, such as visual cortex. While previous studies have examined the MRI artifacts induced by conventional EEG caps, we know of no study that has investigated how dense array EEG caps will affect the image quality and detection sensitivity of the fMRI signal.

The purpose of the present study is two-fold: I. Establish a theory that quantitatively describes the effects of metallic conducting wires of the EEG cap on anatomic and BOLD fMRI signal detection; II. Measure and evaluate the effects of a 256 channel EEG cap through simultaneous EEG-fMRI experiments. The achievement of these two aims will not only resolve a fundamental question in simultaneous EEG-fMRI studies, but also provide a general framework for evaluating the influence of other metallic devices such as TMS coils on MRI/fMRI images.

In this paper, the theoretical model for quantifying the impact of EEG cap on the parameters of image quality and fMRI detection is described first. The parameters include anatomic signal, noise, signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) of BOLD signal. Then experiments to measure and compare the image quality in T_2^* -weighted images and the fMRI detection sensitivity without and with a 256-channel EEG cap are described and results presented.

Theory

CNR of BOLD signal

The detection sensitivity of BOLD fMRI signal is determined by its CNR, which is expressed as:

$$CNR = \frac{\Delta S}{\sigma} = (e^{-\Delta R_2^* TE} - 1) \cdot \frac{S}{\sigma} \approx \Delta R_2^* \cdot TE \cdot \frac{S}{\sigma}, \quad [1]$$

where ΔS and ΔR_2^* are the change of MRI signal and transverse relaxation rate induced by BOLD effects, respectively, σ is the noise, TE is the echo time, and S is the MRI signal at baseline. The approximation holds when $\Delta R_2^* \cdot TE \ll 1$, which is typically the case. The noise has thermal contributions from the MR system ($\sigma_{0,sys}$), thermal noise from the subject ($\sigma_{0,sub}$), and physiological noise (σ_p) (16):

$$\sigma = \sqrt{\sigma_{0,sys}^2 + \sigma_{0,sub}^2 + \sigma_p^2} \quad [2]$$

The physiological noise is proportional to the baseline MRI signal (16):

$$\sigma_p = \lambda S, \quad [3]$$

where λ is a constant and is equal to 0.01 approximately (16).

As shown in Eq. 1, the BOLD contrast is directly related to the SNR ($=S/\sigma$) and ΔR_2^* . Since the latter factor is a property of the neurovascular coupling and metabolism, the presence of an EEG cap affects the BOLD CNR only through its influence on the SNR. Therefore, we next examine shielding effects of the cap on the signal and then on the noise.

Influence of EEG cap on anatomic signal

When the subject wears the EEG cap in the MR scanner, the conducting wires of the cap will alter the strength of transmit (B_1^+) and receive (B_1^-) RF fields due to electromagnetic shielding effects (see Fig. 1b). The change of B_1^+ will result in a location-dependent change of flip angle, and consequently the transverse magnetization. The change of B_1^- will alter the electric signal induced in the receiver coil, i.e., the received MRI signal. Therefore, the shielding effect from the EEG cap influences the anatomic MRI signal in both the transmission and the reception processes. This influence can be described by the following equation:

$$S' = \kappa_r \cdot \Psi(\kappa_t, g_t, \alpha) \cdot S \quad [4]$$

Here S' and S are the anatomic MRI signals with and without EEG cap, respectively. κ_r is the reception shielding factor, which is the ratio of the B_1^- field strength with to without the EEG cap, and $\Psi(\kappa_t, g_t, \alpha)$ is the ratio of MRI signal change resulting from the impact of the EEG cap on B_1^+ . $\Psi(\kappa_t, g_t, \alpha)$ is a function of the flip angle when the cap is absent (α), the ratio of transmitter gain with to without the EEG cap (g_t), and the excitation shielding factor (κ_t). κ_t is the ratio of the B_1^+ field strength with to without the EEG cap when $g_t = 1$. For the gradient echo sequence (spoiled) typically used in fMRI experiments, $\Psi(\kappa_t, g_t, \alpha)$ is expressed as:

$$\Psi(\kappa_t, g_t, \alpha) = \frac{\sin(\alpha') / [1 - \cos(\alpha') e^{-TR/T_1}]}{\sin(\alpha) / [1 - \cos(\alpha) e^{-TR/T_1}]} \quad [5]$$

$$\alpha' = \kappa_t \cdot g_t \cdot \alpha, \quad [6]$$

where α' is the flip angle in the presence of EEG cap, T_1 is the longitudinal relaxation time and TR is the repetition time.

Influence of EEG cap on noise

Since the reception mechanism of the noise from the subject is same as the signal (see Fig. 1b), the receive RF field associated with the subject thermal ($\sigma_{0,sub}$) and physiological (σ_p) noise is also shielded by the conducting wires of the EEG cap, and the ratio of change in receive RF field strength (κ_r) is equal between noise and signal. The system noise is unrelated to the RF field, so it is not affected by the EEG cap. Thus, the total noise (σ') in the presence of EEG cap is expressed as:

$$\sigma' = \sqrt{\sigma_{0,sys}^2 + \kappa_r^2 \sigma_{0,sub}^2 + \lambda^2 S'^2} = \sqrt{\sigma_{0,sys}^2 + \kappa_r^2 [\sigma_{0,sub}^2 + \lambda^2 \cdot \Psi^2(\kappa_t, g_t, \alpha) \cdot S^2]} \quad [7]$$

Influence of EEG cap on SNR and CNR of BOLD

From Eqs. [1], [4] and [7], the SNR of image (SNR') and CNR of BOLD fMRI signal (CNR') with EEG cap can be expressed as:

$$CNR' \propto SNR' = \frac{\kappa_r \cdot \Psi(\kappa_t, g_t, \alpha) \cdot S}{\{\sigma_{0,sys}^2 + \kappa_r^2 [\sigma_{0,sub}^2 + \lambda^2 \cdot \Psi^2(\kappa_t, g_t, \alpha) \cdot S^2]\}^{1/2}} \quad [8]$$

If the system noise is negligible compared to the noise from the subject, i.e., $\sigma_{0,sys}^2 \ll \kappa_r^2 [\sigma_{0,sub}^2 + \Psi^2(\kappa_t, g_t, \alpha) \cdot \lambda^2 \cdot S^2]$, then Eq. [8] will be simplified as:

$$SNR' \approx \frac{\Psi(\kappa_t, g_t, \alpha) \cdot S}{[\sigma_{0,sub}^2 + \lambda^2 \cdot \Psi^2(\kappa_t, g_t, \alpha) \cdot S^2]^{1/2}} \quad [9]$$

This means the SNR will be independent of κ_r . In other words, the shielding effect on B_1^- will not influence the SNR of the MRI image and the CNR of BOLD signal. In a properly performing scanner at 1.5 T or higher, the system noise is in fact negligible compared to thermal and physiological noise from the subject (17).

Materials and Methods

MRI data acquisition

fMRI experiments were performed on human subjects to compare the image quality in T_2^* -weighted images and fMRI detection sensitivity with and without wearing a 256-channel EEG cap (HCGSN v.1.0; Electrical Geodesic, Inc., Eugene, OR). The cap uses non-metallic electrodes (plated AgCl and sponge soaked in KCl) and metallic wires. Two additional MRI experiments (measurements of the B_1 field and the thermal noise levels) were carried out to measure the values of the parameters that affect the SNR of MRI images with the presence of the EEG cap, such as, λ , κ_r , κ_t , etc (see Eq. [8]). These three experiments were performed on seven healthy volunteers, and were repeated twice on each subject, one set with the subject wearing the EEG cap and one set without the cap. The order of the experiments with and without the EEG cap was randomized in this study.

The MRI images were acquired using a 3.0 T GE MRI scanner (Discovery MR750; GE Healthcare Systems, Milwaukee, WI). The body coil and an eight channel head coil were utilized as transmit coil and receive coil, respectively. Prior to the MRI scanning, each subject provided informed consent in accordance with a protocol approved by the Stanford Institutional Review Board. The acquisition procedures are described as follows:

Experiment 1: Measurement of fMRI signal—In this experiment, fMRI images were acquired on the subjects using a block-design sensory task. Each block consisted of 15 s rest and 15 s stimulation. During the rest period, the subject viewed a fixation cross at the center of the screen. During the stimulation period, visual, auditory, and sensorimotor stimuli were delivered to the subjects simultaneously. The visual stimulus was a 4 Hz contrast reversing circular checkerboard. A randomized binaural tone sequence (synchronized with the visual stimuli) was used as the auditory stimulus. The sensorimotor stimulus was delivered by pneumatically driven plungers arranged in left- and right-hand gloves so as to push the fingers up and down in a randomized pattern at 4 Hz, in synchrony with the checkerboard reversals and auditory tones. This passive task controls for rate of finger motion and tends to be more reproducible than finger apposition. The whole task paradigm lasted a total of 255 s, which included 8 on/off blocks and an extra 15 s rest period at the end of the paradigm.

The functional MRI data were acquired using a gradient echo spiral in/out pulse sequence (18) with the following imaging parameters: TR = 2 s, TE = 30 ms, specified flip angle = 80° (Ernst angle), matrix size = 64 × 64, and FOV = 22 cm, 128 time frames preceded by 3 discarded initial time frames. Thirty one contiguous, axial slices with 4 mm thickness were selected to cover the whole brain. The scanner's autoprescan was utilized to adjust the transmitter gain to a value optimized according to its algorithm; however, the receiver gains were maintained constant for all functional acquisitions. It was hypothesized the transmit power could vary with and without the cap in place. For anatomical reference, corresponding high resolution T₂-weighted images were acquired using a fast spin echo sequence (TR=3000 ms, TE=68 ms, ETL=12, matrix size = 192 × 256, and FOV = 22 cm). To enhance statistical power, the fMRI scan was repeated twice on each subject.

Experiment 2: Measurement of thermal noise—In the second experiment, thermal noise was measured by utilizing the same pulse sequence and imaging parameters in the fMRI scan, except that the requested flip angle was 2°. As in the first experiment, a total of 128 time frames were acquired in this experiment. No stimulation task was delivered to the subjects during the scan.

Experiment 3: B₁ mapping—The transmit RF field (B₁⁺) was mapped using the double-angle method (19, 20). The sequence used in this experiment was the same as in the fMRI experiments. Two scans were performed using specified flip angles of 90° and 45°, respectively. The TR was 5 s, the number of time frames was 10, and all other parameters were identical to those used in the fMRI scans of experiment 1 and 2.

Data analysis

Evaluation of the influence of EEG cap on fMRI signal detection—The fMRI data with and without the EEG cap (experiment 1) were analyzed with in-house C and Matlab routines. In this experiment, there are two fMRI scans for both the case with EEG cap and that without EEG cap, and a total of 128 time frames were acquired in each scan. A second order detrending was used to remove the linear and quadratic temporal trends from the image time series. No spatial smoothing was applied to the images.

To detect the fMRI signal elicited by the sensory-motor task, a regressor was created first by convolving a rectangular waveform representing the stimulus periods with a canonical hemodynamic response function (21), and then a voxel-wise linear regression analysis based on the general linear model (22) was performed between the time courses of voxels and the regressor. The general linear model assumes that the measured fMRI signal is a linear combination of the mean baseline signal (i.e., the anatomic signal of T_2^* -weighted images averaged over time without any stimulation task), the BOLD signal induced by the stimulation task, and noise. The noise here is defined as the residual variance after model fitting. From this regression analysis, four parametric maps for each image slice were obtained: t-value, mean baseline signal (anatomic signal), noise, and SNR (= baseline signal/noise). A simple average was applied to corresponding maps from the two fMRI scans, resulting in the mean parametric maps. From the t-maps, BOLD activated foci were identified with the cluster analysis method as described previously (23). Firstly, a 3×3 sigma filter was used to cluster the pixels in the t-maps, and then the clustered t-maps were thresholded with $t > 3.5$ (corresponding to $p < 0.0003$). This cluster analysis method is an effective way to minimize the false positive error due to the multiple statistical comparisons (24). The BOLD activation maps were overlaid on the high resolution anatomical images to facilitate visual inspection and measurement.

With the parametric maps, a region-of-interest (ROI) analysis was performed to evaluate the influence of the EEG cap on fMRI experiments. From the high-resolution structural images, ROIs were drawn manually in multiple slices to enclose the whole visual, auditory, sensorimotor cortices, respectively. The binary images (mask images) corresponding to the individual ROIs were created by setting the pixel values as 1 inside and 0 outside the ROIs. These mask images were scaled to the same resolution as the parametric maps, and then the mask and parametric images were multiplied and averaged to obtain the mean anatomic signal, noise, and SNR values in the ROIs. The mean parametric values were averaged over the seven subjects to obtain the group averaged values. To evaluate the influence of the EEG cap on image quality, the ratios of ROI averaged anatomic signal, noise, and SNR with to without the EEG cap were calculated in the visual, auditory, and sensorimotor cortices, then these ratios were averaged over the seven subjects. Similarly, to examine the impact of the EEG cap on the detection sensitivity of fMRI signal, the ratios of averaged t-value and volume of the activated voxels ($t > 3.5$) in the ROIs with and without the EEG cap were measured in the BOLD activation maps. Then a group t-test was performed on the averaged results of all the subjects with and without the EEG cap.

Measurement of the parameters that affect SNR in EEG-fMRI experiments—

According to Eq. [8], the SNR in EEG-fMRI experiments is affected by the parameters λ , $\sigma_{0,sub}$, κ_b , κ_r , etc. These parameters were determined through the following procedure:

From experiment 1, the anatomic signal (baseline signal) and noise maps with and without the EEG cap were obtained correspondingly. Denote the signal and noise without EEG cap as S_1 and σ_1 , and that with EEG cap as S'_1 and σ'_1 . According to Eqs. [2], [3], [4], and [7], the following relationship exists between S_1 , σ_1 , S'_1 and σ'_1 :

$$S'_1 = \kappa_r \cdot \Psi(\kappa_t, g_t, \alpha) \cdot S_1 \quad [10]$$

$$\sigma_1^2 = \sigma_{0,sys}^2 + \sigma_{0,sub}^2 + \lambda^2 S_1^2 \quad [11]$$

$$\sigma_1'^2 = \sigma_{0,sys}^2 + \kappa_r^2 \sigma_{0,sub}^2 + \lambda^2 S_1'^2 \quad [12]$$

In experiment 2, the flip angle ($= 2^\circ$) used is close to zero, so its signal is very small and the physiological noise will be negligible compared to the thermal noise in the subject. Thus, the noise without and with the EEG cap in experiment 2 (σ_2 and σ_2') comprises only the thermal noise:

$$\sigma_2^2 = \sigma_{0,sys}^2 + \sigma_{0,sub}^2 \quad [13]$$

$$\sigma_2'^2 = \sigma_{0,sys}^2 + \kappa_r^2 \sigma_{0,sub}^2 \quad [14]$$

From the equations above, the ratio of physiological noise to baseline signal (λ) was obtained first by subtracting Eq. [13] from Eq. [11]:

$$\lambda = \frac{\sqrt{\sigma_1^2 - \sigma_2^2}}{S_1} \quad [15]$$

Since the system thermal noise is negligible compared to the subject thermal noise ($\sigma_{0,sys} \ll \sigma_{0,sub}$), then κ_r can be calculated from Eqs. [13] and [14]:

$$\kappa_r \approx \frac{\sigma_2'}{\sigma_2} \quad [16]$$

and then $\Psi(\kappa_b, g_b, \alpha)$ was determined directly from Eq. [10].

The time frames acquired in the same scan in experiment 3 were averaged first to produce an average image for each slice at the nominal flip angles of 90° and 45° , respectively. Then the B_1^+ maps without (α) and with (α') the EEG cap were calculated using the double-angle method, which has been detailed elsewhere (15, 16). Finally, the value of κ_t was computed using Eq. [6], and the transmitter gains with and without EEG cap, which were obtained from the header information in the image files.

Results

Figure 2a shows maps of the anatomic signal in T_2^* -weighted images, noise, and SNR with and without EEG cap from a typical subject. It can be seen that both the signal and the noise with EEG cap are smaller than that without the EEG cap in the same brain regions, which is especially noticeable in the posterior brain. Clearly, the degree of signal and noise reduction varies across the brain regions. Substantial decrease occurs in the posterior brain areas (e.g., occipital lobe) close to the conducting wires of the EEG cap such as the visual cortex, while the brain regions far from the wires (e.g., frontal lobe) display relatively small reduction in signal and noise. It is also observed that the percent decreases induced by the EEG cap are about the same at a given position in the signal and noise maps. Since the signal and noise are reduced by approximately the same rate, there is only small degradation of their ratio throughout the brain. In other words, there is little difference between the SNR without and with the EEG cap. This is confirmed by high similarity between their SNR maps (see Fig. 2a). All the subjects show consistent results, and results from ROI measurements in the three

brain regions averaged over subjects are shown in Fig. 2b and Table 1. The average percent decrease in signal varies from $8\pm 3\%$ to $38\pm 5\%$ ($N = 7$) in different brain areas (visual, auditory, and motor), while the percent decrease in SNR is only $8\pm 5\%$. This suggests that the 256-channel EEG cap has little influence on the SNR of T_2^* -weighted images, though it can result in large signal decrease in the vicinity of the conducting wires.

The fMRI activation maps from a single subject are illustrated in Fig. 3a. In the t-maps with and without the EEG cap, strong BOLD activation was detected in the visual, auditory, and sensorimotor cortices. In addition, the extent and pattern of activation with the EEG cap are comparable to that without EEG cap. As shown in Fig. 3b and Table 1, the ROI results for t-score and activation extent averaged over the seven subjects indicated that there is no statistically significant difference ($p > 0.05$) between the maps with and without the EEG cap. Hence, the dense array EEG cap does not significantly affect the detection sensitivity of the fMRI signal, even when there is substantial MRI signal loss in some regions.

The measured values of the parameters that affect the SNR with EEG cap are listed in Table 2. The data in the table are the average values from the seven subjects. As shown in the table, the measured ratios of physiological noise to signal (λ) and to thermal noise from the subject are 0.009 ± 0.002 and 1.98 ± 0.46 respectively. These values are comparable to those measured in the previous studies (16, 25). Also, it is found that κ_t value varies substantially in different brain areas, but κ_t value is approximately equal to 1 in the whole brain. This means that the shielding wires change the strength of transmit RF field (B^+), but do not influence the receive RF field (B^-). In addition, the g_t value in Table 2 demonstrates that the transmitter RF amplitude was increased by an average of 16% with the cap in place.

Figure 4 shows the B_1^+ maps in a subject with and without the EEG cap. Some heterogeneity is observed in the maps without the cap, as is normal at 3T (flip angle up to 100° in the center). However, there is substantially greater heterogeneity of the B_1^+ field in the maps with the EEG cap. In the region adjacent to the conducting wires, the B_1^+ field strength is found to be stronger with the EEG cap (up to 140°) than without EEG cap. In addition, it is found that significant loss of B_1^+ field occurred in other image regions (down to 60°), as indicated by the blue arrow in Fig. 4. Furthermore, the size and position of such regions varied in different subjects. These phenomena occurred in the B_1^+ maps of all the subjects.

Discussion and Conclusions

In this study, we established a theoretical model to describe the influence of a dense array EEG cap on anatomic image quality and fMRI detection sensitivity, and measured and evaluated this influence *in vivo* using a 256-channel EEG cap. In addition, the parameters relevant to the image SNR and the CNR of BOLD signal in EEG-fMRI experiments were also determined in this work.

The results show that the dense array EEG cap can considerably reduce the anatomic MRI signal in the areas near the conducting wires, and the signal reduction can be up to 38%. In contrast, the SNR is degraded by only a small amount ($\sim 8\%$) by the EEG cap. This phenomenon results because the image noise in the present experiments is mainly sourced from the physiological noise in the human subject. Under the imaging conditions (voxel volume = 46 mm^3 and $B_0 = 3 \text{ T}$) used in this study, the physiological noise greatly exceeds the thermal noise from the subject, the coil and other system components (16). Since the physiological noise is proportional to the baseline signal (anatomic signal) (16), the EEG cap results in the same ratio of decrease in the physiological noise and the signal. Therefore, the total noise is reduced by about the same ratio as the signal, and the SNR remains approximately unchanged. Also, according to the measurement results from the previous

study (25), the physiological noise dominates the image noise at 3 T if the voxel volume is larger than 30 mm^3 approximately. This implies that the SNR of MR images would not be significantly impacted by the EEG cap when using a voxel volume larger than 30 mm^3 . However, if the MR images are acquired with a higher spatial resolution (voxel volume $< 30 \text{ mm}^3$), the contribution from thermal noise to the image noise will be comparable or more than that from the physiological noise, i.e., $\sigma_{0,sub}^2 \geq \lambda^2 \cdot \Psi^2(\kappa_t, g_t, \alpha) \cdot S^2$ in Eq. [9]. In this case, the signal drop induced by the EEG cap will result in the decrease in the SNR, and then degrade the detection sensitivity of fMRI.

In addition, it was found that the value of the excitation shielding factor κ_t varied from 0.7 to 1.0 in the brain, while the reception shielding factor κ_r was always close to 1 in all brain regions. This means that the EEG cap altered B_1^+ field but not the B_1^- field. If the same one coil is used as both transmit and receive coil, then κ_t will be equal to κ_r according to the principle of reciprocity. In this study, the B_1 field is transmitted by the body coil and received by the head coil. The distance from body coil to the EEG cap/subject's head is much larger than the head's distance to the receiver head coil. Since the magnetic shielding effect produced by a metal plane decays rapidly with the decrease of the distance between the source and the plane (26), the conducting wires of the EEG cap shields the RF field of the body coil more effectively than that of the head coil. Consequently, the EEG cap resulted in a relatively large change in B_1^+ field but had little effect on the B_1^- field. The increase of 16% in transmitter RF amplitude (Table 2) required to achieve the nominal autoprescan flip angle is consistent with the influence of the cap on B_1^+ .

It is noted that when the EEG cap is present, the B_1^+ field is enhanced in the vicinity of the conducting wires (see Fig. 4). In the neighboring area of conducting wires, the B_1^+ field is amplified because the wires act as an “antenna”. This antenna receives and absorbs the energy of the RF field transmitted from the body coil, resulting in the increase of B_1^+ field near the antenna (conducting wires). In addition, it is also found that the B_1^+ field in some areas is much weaker than the average value of the whole brain (Fig. 4). This artifact arises from the dielectric standing wave effects and eddy currents, and it has been demonstrated that the size and position of this artifact depends on the size of the subject and the frequency of B_1^+ (27). In either case of too large a flip angle near the wires or reduced flip angle induced elsewhere, when the resulting flip angle differs from the Ernst angle the transverse magnetization will be diminished.

In addition to the 256-channel EEG system used in the present study, 128-channel systems are increasingly being used in EEG-fMRI studies. Since the number of conducting wires is halved in the 128-channel cap, the RF shielding effects will be weaker than those found here. Thus, the influence of the 128-channel EEG cap on anatomic image quality and fMRI detection sensitivity will be smaller than the results from the present study. Considering that the effect of the 256-channel cap on the image SNR and CNR of BOLD signal is small enough already, it can be concluded that the ability to detect fMRI signal will not be affected by the 128-channel cap in simultaneous EEG-fMRI experiments. Anecdotal experiments (not reported here) confirm this hypothesis.

It is worth mentioning that although the present study aimed to investigate the influence of the EEG cap on fMRI signal detection, the theoretical model and experimental procedures can be extended to evaluate the effect of other metallic devices placed in near the receive MRI coil, such as transcranial magnetic stimulation (TMS) coils. Similar to the EEG cap, the conducting wires inside TMS coils also produce electromagnetic shielding effect to the B_1 fields. So the impact of TMS coils on the anatomic image quality and detection sensitivity of fMRI can be investigated using the same theory and experimental methods

presented in this study. Again, we have observed anecdotally that, despite strong reductions in signal close to the coil, the BOLD sensitivity appears minimally reduced.

In summary, we investigated how the dense array EEG cap affects the anatomic image quality and fMRI signal detection through theoretical modeling and experimental measurements. Our results indicated that the EEG cap can induce a large decrease in the anatomic signal (nearly 40%), but the degradation in the image SNR is negligible ($< 8\%$), and the detection sensitivity of fMRI with EEG cap is comparable to that without EEG cap in terms of average t-value and activation volume. This fortuitous result occurs because the noise and signal are both affected approximately the same by the shielding influence of the wires.

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Figure 1.

(a) A photo of the setup of EEG cap in the EEG-fMRI experiments showing dress of the 256 wires. (b) Model of the influence from conducting wires of EEG cap on MRI image quality. Due to the RF shielding effect of the conducting wires, the transmit (B_1^+) and receive (B_1^-) RF fields will be altered when the EEG cap is presented between the subject's head and the RF coils. The ratios of the strength of B_1^+ and B_1^- with EEG cap to that without EEG cap are denoted κ_t and κ_r , respectively. This influence on RF fields will result in the change of anatomical MRI signal ($S \rightarrow S'$), thermal and physiological noise from the subject ($\sigma_{0,sub} \rightarrow \sigma'_{0,sub}$ and $\sigma_{0,p} \rightarrow \sigma'_{0,p}$). The system thermal noise ($\sigma_{0,sys}$) is not affected by the EEG cap.

Figure 2.

Influence of the EEG cap on image quality. (a) Maps of image parameters (anatomic signal, noise and SNR) without and with the EEG cap from a single subject. The parametric maps are obtained from the T_2^* -weighted images and overlaid on the corresponding high-resolution structural images to facilitate visual inspection. The data are normalized to the corresponding average values in the prefrontal brain region without the EEG cap. The shielding effect on the signal and noise is negligible in the frontal area but substantial in posterior brain regions (occipital lobe), while there is little difference between the SNR maps with and without the EEG cap. This suggests that the EEG cap has little effect on the

SNR. **(b)** Relative ratios of anatomic signal and SNR of T_2^* -weighted images without and with the EEG cap in different brain region ROIs. The results are averaged over the seven subjects and represented by mean $\pm$ standard deviation.

Figure 3.

Influence of the EEG cap on fMRI activation extent. **(a)** BOLD fMRI activation maps ($p < 0.0003$) without and with the EEG cap in a single subject. The BOLD responses were elicited by simultaneously delivering visual, auditory, and sensorimotor stimuli to the subject. **(b)** Relative ratios of average t-value and volume of the activated voxels (activation volume) with the EEG cap to that without the cap. The results were obtained separately in the three activated brain region ROIs (visual, auditory, and motor cortices). The data from individual subjects were normalized to the corresponding values when no EEG cap is present, and then averaged over the seven subjects. The error bars represent standard deviation.

Figure 4.

B_1^+ -maps with and without the EEG cap. The maps are overlaid on their corresponding high-resolution structural images. Each pixel value in the maps represents the flip angle at this pixel. The yellow and blue arrows indicate the areas in which the B_1^+ field strength are substantially enhanced and weakened, respectively.

Table 1
The normalized image values and fMRI activation extent with the EEG cap

Subject	Signal			SNR			Average t-value			Activation volume		
	vis	aud	mot	vis	aud	mot	vis	aud	mot	vis	aud	mot
1	0.56	0.90	0.70	0.96	0.98	0.89	1.00	0.96	1.02	1.01	0.97	1.00
2	0.65	0.92	0.77	0.98	0.93	0.99	1.01	0.92	1.04	0.94	1.14	1.62
3	0.57	0.87	0.87	0.91	0.88	0.97	1.02	0.93	0.98	1.16	0.84	1.16
4	0.65	0.91	0.79	0.87	0.87	0.88	1.02	1.00	1.00	0.86	1.21	1.44
5	0.63	0.92	0.84	0.92	0.95	0.94	0.95	0.98	1.02	0.90	0.84	1.03
6	0.68	0.97	0.87	0.88	0.98	0.94	0.92	0.94	0.93	0.82	0.88	0.84
7	0.61	0.90	0.83	0.90	0.91	0.90	0.95	0.99	0.99	0.98	0.87	0.89
Mean	0.62	0.92	0.81	0.92	0.93	0.92	0.98	0.96	1.00	0.95	0.96	1.14
S. D.	0.04	0.03	0.06	0.04	0.04	0.04	0.04	0.03	0.04	0.11	0.15	0.29

The data are the ROI averaged values in visual (vis), auditory (aud), and sensorimotor (mot) cortices. All the data are normalized to the corresponding values without the EEG cap. S. D.: standard deviation.

Table 2
The measured values of the parameters affecting the SNR with EEG cap

Brain Region	λ	σ_p/σ_0	κ_r	κ_t	g_t
Left Visual	0.012 ± 0.004	2.63 ± 0.96	1.07 ± 0.06	0.78 ± 0.27	
Right Visual	0.011 ± 0.003	2.52 ± 0.89	1.03 ± 0.08	0.98 ± 0.27	
Left Auditory	0.007 ± 0.001	1.59 ± 0.70	1.01 ± 0.05	0.68 ± 0.12	
Right Auditory	0.008 ± 0.002	1.68 ± 0.61	1.03 ± 0.04	0.83 ± 0.15	
Left Motor	0.006 ± 0.002	1.77 ± 0.51	0.98 ± 0.03	0.72 ± 0.10	
Right Motor	0.008 ± 0.002	1.71 ± 0.56	1.01 ± 0.04	0.77 ± 0.21	
Mean	0.009 ± 0.002	1.98 ± 0.46	1.02 ± 0.03	0.79 ± 0.10	1.16 ± 0.10