PERFORMANCE OF SQUID SENSOR ARRAYS FOR MRI OF THE BRAIN

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Introduction

While the state of the art in MRI has moved towards stronger magnetic fields, another approach has emerged, where the precession field is typically around 100 µT [1]. The signal is detected using a *superconducting quantum interference device* (SQUID), which outperforms an induction coil by orders of magnitude in sensitivity at kHz frequencies and below [2]. This ultra-low-field MRI (ULF MRI) has advantages, for instance, in better T_1 contrast and compatibility with other techniques such as *magnetoencephalography* (MEG) [1]. In MEG, magnetic signals due to neuronal activity are detected by an array of SQUID sensors. A present EU-funded project is developing a hybrid MEG-MRI scanner [3]. MEG-style 'whole-head' sensor arrays can improve the ULF-MRI signal-to-noise ratio (SNR), which is still insufficient for practical applications.

Fig. 1

In designing sensor arrays, ULF MRI has different optimization goals from both MEG and conventional parallel MRI; while a typical parallel MRI system is designed for accelerated imaging such as SENSE, in ULF MRI the SNR is more important. Moreover, MEG detectors are often gradiometers, which detect spatial variations of the magnetic field. SQUID systems behave differently from induction coils also in noise mechanisms. In this work, we addressed the lack of applicable studies, focusing on SNR in ULF MRI of the brain.

SQUID sensor configurations

A SQUID is typically small, ~ 100 µm, and the magnetic signal is coupled to it by a larger superconducting flux transformer, which consists of a pickup coil connected to an input coil that is on top of the SQUID. The magnetic flux seen by the SQUID is proportional to that through the pick-up coil. A magnetometer pickup coil can be just a wire loop, while gradiometers consist of two or more loops in different orientations. With an optimized input coil, the noise of the SQUID corresponds to a flux noise that increases with the pickup coil size R roughly as R^q, with $q = 0.5$. We consider signals and noise in terms of flux, which allows comparing magnetometers and gradiometers directly. The dominating noise source, however, is not always the sensor, and we show this to lead to different values for q.

One way to construct a sensor array for head imaging is to fill a helmet-shaped surface with N identical sensor units of size R. If the signal seen by a single sensor is proportional to R^p, we show that in an optimal linear reconstruction scheme, the reconstructed voxel SNR $\sim R^{p-q-1}$. However, with constant R, but a varying N, one has SNR $\sim N^{1/2}$ on average.

Numerical results

A number of assumptions was made in the above power-law approach. To investigate their validity and breakdown, numerical calculations were made. First, a filled 10-cm-radius spherical surface was used for studying the dependence of SNR on R and N. Figure 1 shows combined magnetometer sensitivities to sources at the center of the sphere and at a 6-cm distance from the center (representing the cerebral cortex) as a function of R.

In addition, SNR profiles of some existing SQUID sensor configurations and their variations were calculated. Figure 2 illustrates the SNR profile of (a) 102 magnetometer channels in the commercial Elekta MEG sensor array and (b) the Los Alamos 7-channel ULF-MRI setup [4] as $\log_{10}$ of an arbitrary unit—arbitrary in the sense that the absolute SNR is affected by, e.g., the imaging sequence. The numbers are, however, scaled using reported noise levels. Therefore, comparing different sensor configurations is possible.

Conclusions

With the SNR in an image voxel as the figure of merit, we investigated the influence of geometry, pickup coil configuration, dominant noise source, and noise correlation. In terms of combined array sensitivity, magnetometers are far superior to planar gradiometers of the same size. Using on the order of 100 sensors, as in typical MEG configurations, is reasonable, although increasing R and decreasing N may give slightly better results, depending on the dominant noise source.

Despite the emphasis on ULF MRI of the brain, our approach can be applied more generally to SQUID MRI and partly also to arrays of induction coils, when SNR optimization is in order.

References

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Fig. 2a

Fig. 2b