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Information content with low- vs. high-T_c SQUID arrays in MEG recordings: The case for high-T_c SQUID-based MEG

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HIGHLIGHTS

- We model low- and high-T_c SQUID arrays in MEG recordings of neural activity in the brain.
- We compare the total information available to these SQUID arrays.
- An MEG system based on high-T_c technology is capable of producing at least 40% more information than the state-of-the-art in low-T_c MEG systems.
- The gain in information provided by high-T_c MEG technology is a result of the closer source-to-sensor standoff distance.

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ABSTRACT

Background: Magnetoencephalography (MEG) is a method of studying brain activity via recordings of the magnetic field generated by neural activity. Modern MEG systems employ an array of low critical-temperature superconducting quantum interference devices (low-T_c SQUIDs) that surround the head. The geometric distribution of these arrays is optimized by maximizing the information content available to the system in brain activity recordings according to Shannon's theory of noisy channel capacity.
New method: Herein, we present a theoretical comparison of the performance of low- and high-T_c SQUID-based multichannel systems in recordings of brain activity.
Results: We find a high-T_c SQUID magnetometer-based multichannel system is capable of extracting at least 40% more information than an equivalent low-T_c SQUID system. The results suggest more information can be extracted from high-T_c SQUID MEG recordings (despite higher sensor noise levels than their low-T_c counterparts) because of the closer proximity to neural sources in the brain.
Comparison with existing methods: We have duplicated previous results in terms of total information of multichannel low-T_c SQUID arrays for MEG. High-T_c SQUID technology theoretically outperforms its conventional low-T_c counterpart in MEG recordings.
Conclusions: A full-head high-T_c SQUID-based MEG system's potential for extraction of more information about neural activity can be used to, e.g., develop better diagnostic and monitoring techniques for brain disease and enhance our understanding of the working human brain.

1. Introduction

Magnetoencephalography (MEG[1]) is one of many commercially available tools for studying brain activity in man. There are over 100 MEG systems used today worldwide (MEG Systems, 2013). Most of these systems are used purely for academic research, but clinical exploitation of the technique—such as for pre-surgical mapping of brain function and epilepsy source-localization—is growing (Hari and Salmelin, 2012).

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[1] MEG: Magnetoencephalography; the study of electrical activity in the brain via magnetic recordings around the head.
ms: milliseconds; 10^{-3} seconds.
pT, fT: picotesla = 10^{-12} tesla, femtotesla = 10^{-15} tesla; a unit of measure for magnetic field strength.
K: kelvin; the absolute temperature scale in which $0\,K = -273\,^{\circ}C$.
SNR: Signal-to-noise ratio.
T_c: Critical temperature; the temperature below which a material becomes superconducting.
SQUID: Superconducting quantum interference device; an extremely sensitive magnetic field detector based on superconducting technology.
RMS: Root mean square; a measure of the magnitude of a varying quantity.

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The magnetic signals recorded by MEG systems are generated by neural currents inside the brain (Cohen, 1968). An ideal MEG recording comprises sampling of the magnetic field with a spatial and temporal density that is well above its natural variation. Estimates place the spatial variation of the magnetic field generated by neural activity below the cm scale (Ahonen et al., 1993). Observed neuromagnetic frequencies reach at least 100 Hz (Hari and Salmelin, 2012). Furthermore, because the magnitude of magnetic fields decay as the inverse cube of the distance from the source (or faster), the ideal MEG recording would sample the magnetic field as closely to the sources as possible—i.e. at the scalp if one is to avoid opening the skull. Finally, because the magnetic fields generated by neural activity are extremely weak (<pT), the ideal MEG recording would be performed with sensors whose noise levels are far below that level. The ideal MEG recording thus combines magnetic:

1. sampling with spatial resolution below the cm scale,
2. recordings with sampling frequencies above 100 Hz,
3. sensors located close to the scalp, and
4. field noise levels far below ∼pT.

State-of-the-art MEG systems employ an array of low critical-temperature superconducting quantum interference devices (low-T_c SQUIDs) because they combine high temporal resolution (>kHz) with extremely low noise levels (magnetometers can achieve ∼1 fT (Clarke and Braginski, 2004)). Items (2) and (4) in the ideal MEG system are therefore satisfied by such MEG systems. Low-T_c SQUIDs are also highly reproducible from a fabrication point-of-view, making them practically and commercially attractive for MEG (Hari and Salmelin, 2012).

However, low-T_c technology requires liquid helium temperatures (∼4 K (kelvin)) for operation. While a low-T_c sensor might (with a lot of engineering) be able to operate within ∼1 cm of the scalp, achieving such a good stand-off distance for an array of these sensors is impractical (Clarke and Braginski, 2004). Extreme sensor operating temperatures thus prevent low-T_c SQUID-based MEG systems from satisfying condition (3). Condition (1) is also complicated by the high standoff distance as the spatial frequencies of the magnetic field decrease as the distance from the sources increases.

As compared to low-T_c, high-T_c SQUIDs have as good temporal resolution, poorer noise performance (typical magnetometer noise is ∼tens of fT), and much higher operating temperatures (liquid nitrogen, ∼77 K) (Clarke and Braginski, 2004). They therefore satisfy condition (2) of the ideal MEG system and, while they are not as quiet as low-T_c SQUIDs, they satisfy condition (4) as well. However, more moderate operating temperature means high-T_c SQUIDs can function within less than 1 mm of the scalp (Öisjöen et al., 2012). Furthermore, cooling to 77 K enables a flexibility in system design that could allow the entire array of sensors to satisfy condition (3).

The spatial variation of the magnetic field generated by neural activity is an open question. The spatial sampling of the magnetic field employed by state-of-the-art low-T_c SQUID-based MEG systems has therefore been optimized—according to Shannon's theory of communication (Shannon and Weaver, 1949)—to maximize the amount of information that can be extracted from brain activity. Early works compare the various competing multi-channel systems (Kemppainen and Ilmoniemi, 1990) whereas later works aim at optimizing the layout of an array of SQUIDs (Knuutila et al., 1993; Nenonen et al., 2004). Herein, we use this method of estimating information content in multi-channel MEG systems to compare a typical low-T_c SQUID MEG system to a hypothetical one based on high-T_c technology.

2. Theory

Following (Kemppainen and Ilmoniemi, 1990; Nenonen et al., 2004), Shannon's theory of communication can be used to estimate the information capacity, I, of a noisy channel as:

$$I = \frac{1}{2}\log_2(P+1) \tag{1}$$

where P is the power signal-to-noise ratio (SNR) of the channel and I is measured in bits (Shannon and Weaver, 1949). The noise is estimated from the literature in both the low- and high-T_c cases (Clarke and Braginski, 2004; Hari and Salmelin, 2012; Öisjöen et al., 2012). The signal level, however, is not as clearly defined because it depends on how the magnetic field generated by neural currents inside the brain couples to the sensors. This coupling is called the "lead field" of a sensor and can be defined as:

$$S = \int_V \vec{L}(r)\cdot\vec{J}(r)dr \tag{2}$$

where S is the magnetic field strength recorded by the sensor (the signal), $\vec{L}(r)$ is the spatial distribution of the sensitivity of the sensor (the lead field), $\vec{J}(r)$ is the neural currents, and the integration is over the volume of the sources (the brain) (Hansen et al., 2010). We include primary tangential currents only (as is common in the literature) because radial sources are magnetically silent in the spherical head model. Further, we assume the neural currents are generated by a random process (normally distributed in each non-radial direction), i.e.:

$$J_{\vec{\theta},\vec{\varphi}}(r) \sim \mathcal{N}(0, \sigma^2_{\text{signals}}) \tag{3}$$

where $J_{\vec{\theta},\vec{\varphi}}(r)$ are the θ- and φ-directed components of $\vec{J}(r)$, respectively, (in the spherical (r,θ,φ) coordinate system), and σ_{signals} is the magnitude of currents in the brain.

By combining Eqs. (2) and (3), the root mean square (RMS) of the magnetic field recorded by the sensor, S, can be re-expressed as:

$$S = \sigma_{\text{signals}} \int_V |\vec{L}(r)| dr \tag{4}$$

where $|\vec{L}(r)|$ is the magnitude of the lead field as a function of position in the source volume. The power SNR, P, is then:

$$P = \frac{S^2}{N^2} = \frac{\sigma^2_{\text{signals}}}{N^2}\left(\int_V |\vec{L}(r)| dr\right)^2 \tag{5}$$

where N is the noise in the channel, specified in the same units as S. We can then add up the information we receive from each of these channels. Like our signal sources, we assume the noise for each channel is normally distributed:

$$N_k \sim \mathcal{N}(0, \sigma^2_{\text{noise}}) \tag{6}$$

For a multi-channel MEG system, however, the situation is more complicated. Because the sensors can receive signals from the same sources (i.e. the lead fields overlap), we must generate a new set of independent/orthogonalized channels before assessing the information they receive. To this end, we calculate the lead-field inner-product matrix elements for each pair of sensors:

$$G_{jk} = \int_V \vec{L}_j(r)\cdot\vec{L}_k(r)dr \tag{7}$$

G_{jk} is a measure of the overlap of the j-th and k-th lead fields (see (Hämäläinen and Ilmoniemi, 1994) for an analytical approach to calculating the G_{jk}s). We orthogonalize the vectors in matrix G via singular-value decomposition such that $G = U\lambda U^T$ where U is a