

SQUID SYSTEM FOR MEG AND LOW FIELD MAGNETIC RESONANCE

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Abstract

A SQUID magnetometer system was developed for measuring sustained brain activity by magnetoencephalography (DC-MEG) and to record the free precession decay of protons (FPD) of the human brain at very low fields. The SQUID system has a white noise level of about $4 \text{ fT}/\sqrt{\text{Hz}}$. To generate the MR signal, two magnetic fields are used: a static polarisation field of a few mT and a static detection field of a few microtesla. To test the spectral resolution of the system, we measured the FPD of protons in distilled water having a spectral line width of about 156 mHz with an instrumental resolution of 2 mHz. The proton resonance line width of the human brain was found to be about 3.0 Hz. Using the same SQUID system we recorded a DC-MEG signal with an amplitude of about 1.5 pT upon motor stimulation. On the basis of these data, we discuss the possibility of detecting a shift of the resonance line due to the superposition of the neuromagnetic field generated by sustained brain activity.

Keywords: SQUID, MEG, low field MR, neuronal currents.

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1. Introduction

Superconducting Quantum Interference Devices (SQUIDs) are commonly used for biomagnetic research of brain activity. SQUID systems can also be used to investigate magnetic resonance (MR) phenomena at low magnetic fields around or below the earth field strength [1, 2]. Recently, even MR Images of the human brain anatomy were generated at 47 microtesla with a spatial resolution of $3 \times 3 \times 6 \text{ mm}^3$ [3].

In this paper we will describe a SQUID system used for both applications, for biomagnetic research and for magnetic resonance. A system like this enables the combination of the functional modality of the MEG with structural information of the brain's anatomy.

Another goal of the low field approach is the measurement of neuronal currents by their effect on the MR Image. At very low fields, the biomagnetic field generated by neuronal currents of a few picotesla should cause a non-negligible shift of the static field for the tissue around the nerve [4]. If this shift sustains for a few seconds, its effect on the resonance line might be measurable.

As a first step towards the realization of these two goals, we have employed our measurement system for the measurement of the resonance line of brain tissue and for the measurement of sustained brain fields.

2. Materials and Methods

We use a 3-channel SQUID magnetometer system consisting of three fully integrated helium-cooled multi-loop DC-SQUIDs [5]. The intrinsic white noise of the SQUID system is

about $4 \text{ fT}/\sqrt{\text{Hz}}$ with a $1/f$ corner frequency below 2 Hz. In Fig. 1 we show a picture and a schematic of our apparatus. The sensors with a sensitive area of 3.6 mm in diameter are sensitive in the y-direction (see schematic in Fig. 1) and are arranged vertically inside the helium cryostat. They are mounted at a distance of 0.5 mm, 60.5 mm and 120.5 mm with respect to the flat inner bottom of the helium cryostat. The three SQUID signals can be used to build different axial software gradiometers. The head of the volunteer is placed directly below the outer cryostat bottom with a minimum distance between the lowest sensor and the head of about 7 mm.

To operate our SQUID system at magnetic fields below the earth field we use it inside the Berlin Magnetically Shielded Room-2 (BMSR-2). This room has an extremely high magnetic shielding factor with a cubic inner volume of about $2.8 \times 2.8 \times 2.8 \text{ m}^3$ consisting of 7 layers of Mu-Metal and one layer aluminium (rf-shielding) resulting in a passive shielding factor of 7.5×10^4 at 0.01 Hz [6].

To record the Free Precession Decay (FPD) of protons we are using a coil setup consisting of a polarisation coil generating a field in the y-direction and a Helmholtz-like coil setup generating the homogeneous detection field in the z-direction.

The SQUID system was positioned with the Dewar tail closely above EEG-position C3, where we expected the maximum intensity of the DC-MEG signal. The frequency range of this recording was from DC to 250 Hz. For generating the sustained brain activity, the volunteer was asked to execute finger tapping movements of the right hand during a period of 30 seconds, followed by a period of 30 seconds rest. This procedure was repeated 30 times to average the signal.

At the same cryostat position, the MR signal was recorded after polarisation field intervals over 1 second (magnetic field strength of about 3 mT). The SQUIDs recorded the MR signal over 1 second with a static detection field of $20.7 \mu\text{T}$. MR recordings were repeated 999 times for averaging the signal before it was transformed into the frequency range. The line width (full width half maximum) of the frequency peak, positioned at 880.8 Hz, was determined by fitting it to a Lorentzian line shape.

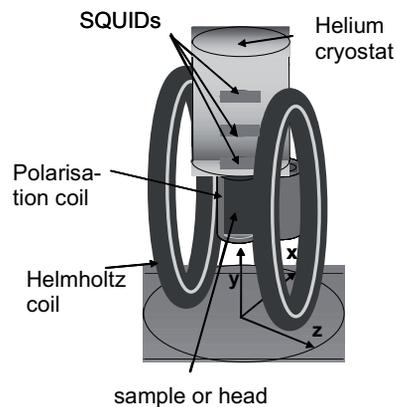
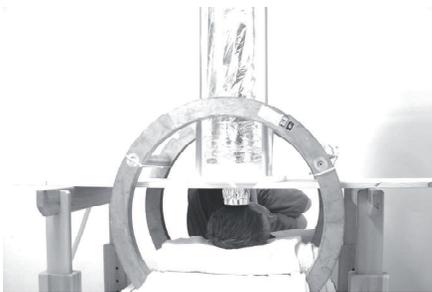


Fig. 1. SQUID system for MEG and Low field MR; left: picture for combined DC-MEG and MR recordings, right: schematic for sample recordings.

3. Results

To test the frequency resolution of our instrumentation we investigated the FPD of distilled water. After Fourier transformation the spectral line width was found to be 156 mHz with an instrumental resolution of 2 mHz [7].

In this study, we measured the proton resonance line of the human brain. Fig. 2 shows the absorptive part of the spectrum from which we determined the line width to be about 3.0 Hz. If it is assumed that the FPD is described by a single exponential decay, the corresponding T_2 – time is about 106 ms. With the given instrumental resolution of the setup, this is the natural lifetime (or line width, respectively) of the proton magnetization in brain tissue which is determined by the internal interactions of the nuclear magnetic moments with their molecular environment alone. It is more than one order of magnitude shorter than in pure water. This probably reflects that the water in the brain contains many ions and that brain tissue is by no means as homogeneous as a liquid.

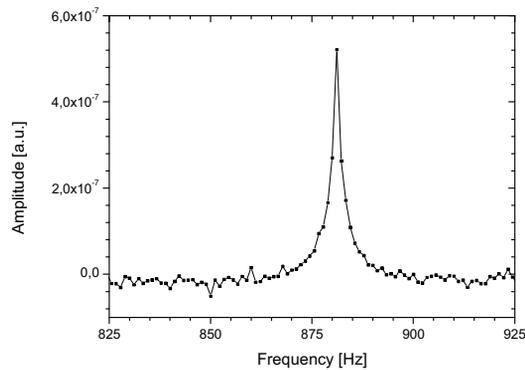


Fig. 2. Absorptive part of the spectrum of the FPD of a human brain at a field of 20.7 μ T.

Following the motor stimulation paradigm described above we recorded a DC-MEG signal having a temporal course that was found already in earlier studies on sustained fields [6] (see Fig. 3): It starts with a fast rise of a few seconds to a level elevated by about 1.5 pT. With a slight slope, the DC field remains constant for 30 s until it decays back to the baseline with some delay after the volunteer has stopped finger tapping.

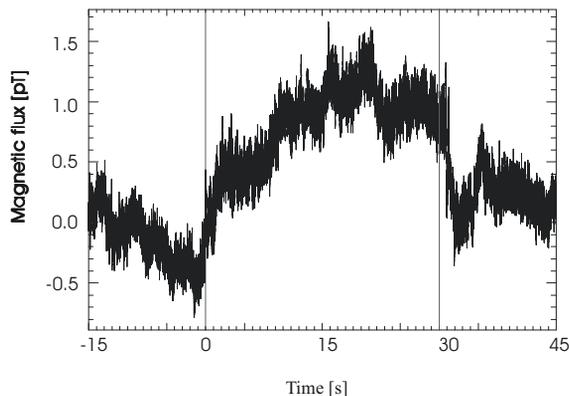


Fig. 3. DC-MEG after motor stimulation in the range from 0 to 30 seconds. The beginning and the end of the motor activity is marked. Number of averages = 30.

4. Discussion

The presented SQUID system enables the recording of MEG and low field MR signals of the human brain. This is the prerequisite for the combination of the recording of brain function by MEG with the recording of anatomical information by low field MRI in one session using the same recording setup.

A second aspect of the presented results is that they provide some data on the relation between the expected neuronal current field and the observed MR line width. It is this relation that decides whether the physical conditions in living brain tissue make neural current imaging feasible.

At a distance of about 40 mm above the motor cortex the detected brain field was about 1.5 pT. For the sake of simplicity let us assume for a moment that the neural current generator is a simple current dipole the magnetic field of which decreases with the inverse square of the distance. If we further assume 4 mm as a minimum distance for an MRI voxel from the neural current (*i.e.* \sim the voxel size in the LF-MRI study of Zotev *et al.* [3]), then we expect a neuronal magnetic field of up to 300 pT around the nerve. This corresponds to a frequency shift of 12 mHz.

Obviously, our SQUID technology with 2 mHz spectral resolution is sufficient resolution to detect frequency shifts of a few millihertz. This is the fundamental advantage of LF-MR over MRI at high fields where such instrumental resolutions are far beyond the technical limits. Now, our low field data show that brain signals have a natural line width of 3 Hz. To resolve a 12 mHz shift of a 3 Hz line is certainly a challenge, if not again beyond the limits. Improving the signal to noise ratio alone may not be sufficient to make this effect visible.

But note that our single current dipole model is certainly an oversimplification. In reality, neuronal currents in the brain may have a more complicated structure where close-by current vectors with opposed direction compensate their far magnetic field. Thus, our estimation of the intensity of the neuronal currents in the brain on the basis of the equivalent current dipole model is certainly too conservative. The data of our study provide a first lower limit of the effect of neuronal current on the MR image. But there is good reason to believe that the real situation is much more favourable due to locally larger magnetic fields, which are not seen in the MEG measured outside the head.

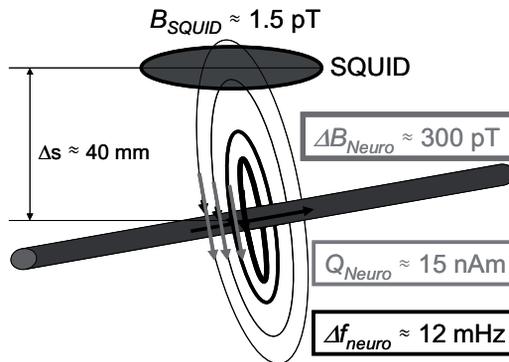


Fig. 4. Estimated expected effect of a shift of 12 mHz in the MR Larmor frequency generated by a neuronal current with a dipole moment of 15 nAm and a measured DC-MEG signal of about 1.5 pT.

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References

- [1] Ya.S. Greenberg: "Application of superconducting quantum interference devices to nuclear magnetic resonance". *Rev. Mod. Phys.*, no. 70, 1998, pp. 175-222.
- [2] R. McDermott, A.H. Trabesinger, M. Mück, E.L. Hahn, A. Pines, J. Clarke: "Liquid state NMR and scalar couplings in microtesla magnetic fields". *Science*, no. 295, 2002, pp. 2247-2249.
- [3] V.S. Zotev, A.N. Matlashov, P.L. Volegov, I.M. Savukov, M.E. Espy, J.C. Mosher, J.J. Gomez, R.H. Kraus Jr.: "Microtesla MRI of the human brain combined with MEG". *J. Magn. Reson.*, vol. 194, no. 1, 2008, pp. 115-120.
- [4] A.M. Cassarà, B. Maraviglia, S. Hartwig, L. Trahms, M. Burghoff: "Neuronal current detection with low-field magnetic resonance: simulations and methods". *J. Magn. Reson.*, 2009, doi:10.1016/j.mri.01.015.
- [5] D. Drung: "High- T_c and low- T_c dc SQUID electronics". *Supercond. Sci. Technol.*, vol. 16, no. 12, 2003, pp. 1320-1336.
- [6] M. Burghoff, T.H. Sander, A. Schnabel, D. Drung, L. Trahms, G. Curio, B.M. Mackert: "dc Megnetoencephalography: Direct measurements in a magnetically extremely-well shielded room". *Appl. Phys. Lett.*, vol. 85, no. 25, 2004, pp. 6278-6280.
- [7] M. Burghoff, S. Hartwig, L. Trahms, J. Bernarding: "Nuclear magnetic resonance in the nanoTesla range". *Appl. Phys. Lett.*, vol. 87, no. 5, 2005, pp. 054103.