Magnetocardiography and magnetoencephalography measurements at room temperature using tunnel magneto-resistance sensors

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Weak magnetic fields are generated in the heart and brain of humans by the electrical activity in various tissues. Measurements of such magnetic fields have great potential1-3 to improve the diagnosis of diseases and to clarify the biological functions of living organisms because of the non-invasive character and high spatial and temporal resolution of the imaging methods. However, such imaging requires the use of superconducting quantum interference devices (SQUIDs), which incur high equipment and operational costs, especially the purchase of liquid helium.

We have investigated the use of tunnel magneto-resistance (TMR) sensors that operate at room temperature to measure weak bio-magnetic fields.4-9 The TMR sensor is a sensor element consisting of a magnetic tunnel junction (MTJ)10-17 that operates at room temperature and has been developed for spintronic devices such as magnetic random access memory,18,19 and various magnetic sensors such as the read heads of hard disk drives and micro compasses.20-28

The present study succeeded in improving the signal-to-noise (S/N) ratio of TMR sensors by optimizing the film structure of the MTJs, integrating the MTJ devices, and developing a low-noise amplifier circuit. Our equipment successfully measured magnetocardiography (MCG) signals without averaging and magnetoencephalography (MEG) alpha waves with averaging.

MTJ multilayer films were deposited on thermally oxidized Si wafers using an ultra-high vacuum (P<sub>base</sub> < 3 × 10<sup>-6</sup> Pa) sputtering system. The stacking structure was Si/SiO<sub>2</sub>/Ta 5/5/Ru 10/Ni80Fe20 70/Ru 0.9/Co40Fe40B20 3/MgO 1.6/CoFe2Fe2B30 3/Ru 0.9/CoFe2B5 5/Ir22Mn78 10/ Ta 5/Ru 30 (in nm). The bottom NiFe and CoFeB free layers showed anti-ferromagnetic coupling, and the magnetization reversal process reflected that of the thick NiFe layer.34 MTJs were micro-fabricated by photo-lithography and argon ion milling; the size of each MTJ device was 210 × 105 µm<sup>2</sup>. To reduce the 1/f noise of the sensor,29,30 the MTJs were connected in 870 series and 2 parallel;31 the size of the integrated TMR sensor was 7.1 × 7.1 mm<sup>2</sup>. The annealing process was carried out after micro-fabrication to ensure orthogonal magnetic axis alignment of the free and pinned layers in the MTJs.41

Figure 1 shows the schematic diagram of a measurement system using the TMR sensors. The full-bridge circuit consisted of four TMR sensors, and the output voltage was input to an amplifier circuit. The gain of the amplifier circuit was variable to 120 dB, and the signal was passed through a band pass filter of 1 to 50 Hz. The sensitivity of the developed TMR sensors was 1.8 μV<sub>P-P</sub>/nT<sub>P-P</sub>, and the noise voltage was 420 nV<sub>P-P</sub>. The estimated detectable magnetic field was 233 pT with an S/N ratio of 1. In addition, the noise density was 14 pT/Hz<sup>1/2</sup> at the single frequency of 10 Hz. Finally, the output signal was input to an analog-to-digital (A/D) converter recorded and averaged using PC software. BPF: band pass filter.

![Fig. 1. Schematic diagram of the biomagnetic field measurement system using TMR sensors. Four integrated TMR sensors are connected in a bridge, the signal is input to the amplifier circuit, and the analog-to-digital (A/D) converted data recorded and averaged using PC software. BPF: band pass filter.](https://example.com/fig1.png)
ment using the TMR sensors. Both MCG and ECG measurements were taken at the same time. MCG measurements were taken at 40 to 80 mm to the left of the center of the chest, and 0 to ~20 mm downward, as shown in Fig. 2(a). The TMR sensor was placed several millimeters away from the body surface, and detected the transverse component of the magnetic field. Figure 2(b) shows the reference ECG signal, and Figs. 2(c) and 2(d) show typical MCG signals at x = 60 mm and y = 0 mm measured in a male subject (aged 33 years) in the supine position. Figure 2(c) clearly shows an R wave from the MCG signal without data averaging, which confirms that the positions of the R peaks for both MCG and ECG were almost the same. The amplitude of the magnetic field related to the R peak was approximately 70 pT, which is consistent with the reported value measured by SQUID.\(^2\) This is the first demonstration of real-time MCG measurement at room temperature using the TMR sensors. Additionally, Fig. 2(d) shows the waveform of the region where the ECG signal was observed but the MCG signal was not. In this study, the probability of obtaining a clear MCG R peak was about 50% because the S/N ratio of the present TMR sensor was about 0.3 with respect to the R peak. A strong R peak can be obtained when the noise and the signal resonate stochastically.

Averaging of the MCG signals was achieved by synchronizing the R peak of the ECG waveform to improve the S/N ratio. In this averaging, continuous MCG data were used whether a clear R peak was obtained or not. Figures 2(e), 2(f), and 2(g) show that noise apparently decreased with higher-number of averaging. The R peak, as well as the Q and S peaks were clearly observed in the MCG signal when the averaging was performed 16 to 64 times. Although the P and T peaks were not clearly observed, room temperature detection of the QRS complex within one minute can be very useful for the diagnosis of heart diseases. MCG mapping was also demonstrated as shown in Fig. 3. The MCG of each point was measured by moving the sensor module and averaging 100 times at each point. Different waveforms were observed at each point, which is an advantageous characteristic of MCG measurement. Such high spatial resolution is important for the diagnosis of diseases.

The successful measurement of the MCG signal without averaging encouraged us to try to measure the MEG signal using the same TMR sensor. MEG is a technique for measuring magnetic fields created by the electrical activity of the human brain, and is extremely challenging because the magnetic field intensity is several orders of magnitude lower than that of MCG. The present study tried to measure the alpha waves, which are thought to have relatively high intensity compared to other brain magnetic fields. Figure 4(a) shows the experimental setup for the MEG measurements. The TMR sensor was placed 60 mm to the left of the center of the occipital area on a male subject (aged 30 years) in the prone position, and the transverse component of the magnetic field was measured. Electroencephalography (EEG) was simultaneously performed using two Ag/AgCl electrodes placed on the midline occipital and forehead areas. Under voice instructions from outside the shielded room, the subject alternatively closed and opened his eyes for periods of about 10 s. Prominent peaks of EEG alpha waves at around 10 Hz during the eye-closure period were used as the trigger to average the simultaneously-measured MEG signals.

Figures 4(b) and 4(c) show the result of averaging the MEG and EEG signals at 3,750 times per second. Despite a phase shift, a MEG signal with almost the same frequency as the alpha wave from the EEG was detected. The amplitude of the magnetic field was approximately 2 pT, which is consistent with the reported value.\(^3\) Averaging 3,750 times reduced the noise density of the TMR sensor to 0.2 pT/Hz\(^{1/2}\) at 10 Hz, so that measurement was possible with an adequate S/N ratio for the 2 pT signal. Since the frequency of the alpha wave fluctuates, the signal attenuates as time elapses from the center of the trigger. The attenuation of the alpha wave due to this fluctuation can be seen in both the EEG and MEG measurements.
Figure 5 shows the EEG and MEG signals obtained by averaging 10,000 times. The phase change of the MEG signal was examined by changing the direction of the magnetic field for the TMR sensor to (b) 0° and (c) 180°, which also inverted the phase of the MEG signal by 180°. Figure 5 shows that 10-Hz MEG signals with the same frequency as those from the EEG were observed at both 0 and 180°, and that the amplitude of the magnetic field was approximately 0.5–2.0 pTP. On the other hand, the MEG signal and the EEG signal were out of phase by approximately 25 ms. We confirmed that the MEG signal was delayed by about 5 ms compared with the EEG signal due to differences in the filter circuits, but this could not explain a delay of 25 ms in the measured MEG signal. The phase shift between MEG and EEG may depend on the different sensor positions between the two modalities, because generator neurons of the alpha rhythm, which are distributed over a wide surface area of the brain, can produce signals that are out of phase. The phase change of the MEG signal was also observed by changing the direction of the magnetic field measurement direction of the sensor.
brain, are not all in phase. The Pearson’s product-moment correlation coefficient\(^{(1)}\) of the MEG signal with the EEG alpha wave was as high as 0.7 or more at both 0 and 180°.

In summary, we developed highly sensitive TMR sensors to measure both MCG and MEG signals at room temperature. The MCG R wave was detected without averaging, and the QRS complex was observed with a good S/N ratio by averaging signals for several tens of seconds. MCG mapping was demonstrated with high spatial resolution. Real-time MCG measurements with high spatial resolution at room temperature may lead to significant improvements in the diagnosis of heart disease. MEG of the human brain was achieved by averaging signals using the EEG signal as a trigger. The MEG alpha wave at approximately 10 Hz was observed with a correlation coefficient as high as 0.7 between MEG and EEG. This demonstration of MEG measurements at room temperature is a very significant advance in biomagnetometry. The present findings are expected to lead to various room temperature MEG applications using TMR sensors.

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