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SIMULTANEOUS MEASUREMENT OF THE MAGNETIC HEART VECTOR COMPONENTS WITH THE UNIPOSITIONAL LEAD SYSTEM

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Introduction

Interest in studying the magnetocardiogram is stimulated by the clinical applications and the possibility, that the MCG will provide new information about electrophysiological features of the heart. During the past decade many groups have reported results of their magnetocardiogram studies. In all research reports recordings have been made on one channel so that only one component of the magnetic vector can be measured at each instant of time (1,2,3). Full advantage of the use of magnetocardiography requires measurement of all three orthogonal components. This report describes a detector which simultaneously measures the three components of the magnetic heart vector.

Method

The time-varying magnetic field of the heart induces a voltage in a coil located over the chest of the subject. The voltage will be amplified and integrated. The output is then proportional to the magnetic field strength. Our device uses the unipositional lead system (4). In this system all three components, X, Y and Z of the magnetic field vector are measured simultaneously at the same point on the anterior side

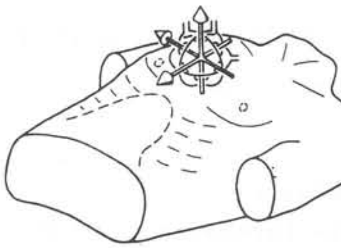

Figure 1. The principle of the measurement of the magnetic heart vector with the unipositional lead system.

of the chest, Figure 1. The patient and the amplifier are in a magnetically shielded room. The magnetometer is mounted on the inner wall of the room.

Magnetometer

The magnetometer coil system, which can be seen in Fig. 2., consists of three coils at the same point perpendicular to each other and it operates in the room temperature. The dimensions of the system are 298 x 276 x 138 mm and the weight is 15 kg. Y- and Z-coils are flattened so that the measurement point is close to the heart. The effective area N·A is 40 m^2 in X-coil and 80 m^2 in Y- and Z-coils. The different areas compensate the different sensitivities in the unipositional leads. The constructed mounting system allows the magnetometer to be raised and lowered and also to be moved aside over the chest. The magnetometer coils are shielded against electric fields with a grounded thin aluminium foil.

Amplifier

The amplifier has three similar channels. Each channel consists of an impedance transformer, an amplifier, filters and

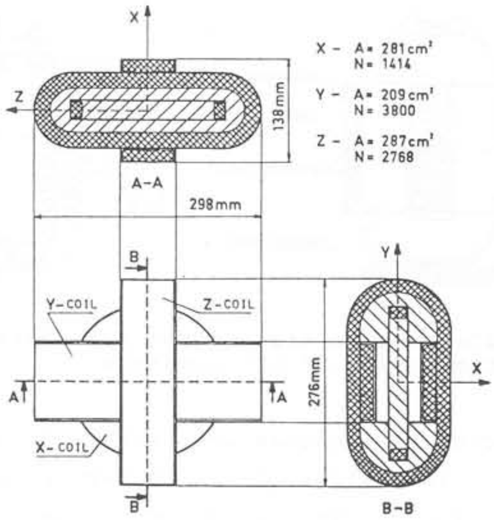

Figure 2. The construction of the coil detector.

an integrator. The impedance transformer adjusts the coil impedance to the amplifier input impedance so that the amplifier will operate with optimum source impedance. This leads to the minimum value in the noise figure NF (5,6). The transformer core is constructed from μ-metal with permeability of the order of 11 000. This makes the frequency response flat and decreases the low cut-off frequency. The core is assembled by pressing two pieces shaped like letter U together with a metal band. The construction of the transformer and the connection of the windings are presented in Fig. 3. The windings are divided into two equal parts to form a differential connection which reduces noise. The secondary voltages E_{e1} and E_{e2} induced by the external magnetic noise flux ϕ_e are equal and opposite and will be canceled at the output. The success in the cancellation depends on the balance between the windings. The voltages E_{p1} and E_{p2} induced by the primary flux ϕ_p are summed at the output. On the both branches of the

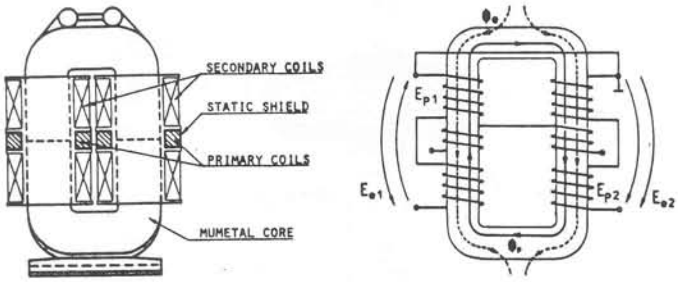

Figure 3. The construction of the transformer and the connection of the windings.

core the secondary windings are further divided into two sections and the two halves of the primary winding are placed between them, Fig. 3. This location of the windings decreases the leakage inductances. Because also the winding width is reduced the leakage capacitance decreases as well. This means that the higher cutoff frequency increases.

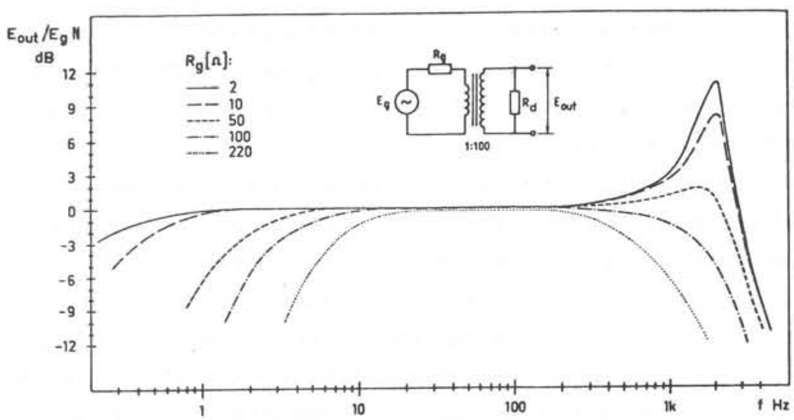

Figure 4. Frequency response of the matching transformer with different values of the source resistance R_g. The voltage of the source generator E_g was 6 mV_{pp} and the load resistance R_d was 10^9 ohms.

The ratio of the turns of the primary coil and the secondary coil of a transformer is 1:100. The frequency response of the transformer with different values of the source resistance is shown in Fig. 4.

As an amplifier in our system we have used an instrumentation amplifier LF352 which have a JFET input stage. The input impedance is $2 \cdot 10^{12}$ ohms and the input bias current and the input offset current are 3 pA and 0,5 pA respectively. The equivalent input noise current is 10 fA. The equivalent input noise voltage of the amplifier on the bandwidth 0,1-10 Hz is about 2 μVpp. The gain of the amplifier is 10 000. The secondary winding of the transformer is directly connected to the input of the amplifier, Fig. 5. There is not used any protective circuits for the over voltages at the input of the amplifier because they increase noise. Coaxial cables were used to connect the detector to the transformer and the transformer to the amplifier.

To limit the measurement bandwidth and cancel out the noise induced from electric power lines the amplifier consists of two low-pass filters and two bandstop filters (50 Hz and 100 Hz). The bandwidth can be adjusted between 50 and 100 Hz. The signal must be integrated because the voltage induced in the coil is proportional to the derivative of the magnetic field.

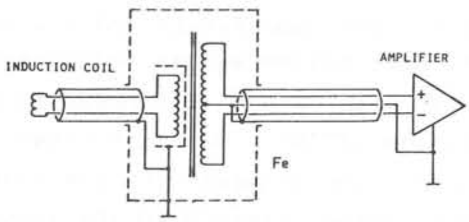

Figure 5. Connection between the transformer and the amplifier and the shielding of the cables.

Magnetically Shielded Room

The magnetic field strength of the heart when measured over the chest is about 50 pT peak to peak during the QRS-complex. The magnetic noise in our laboratory is of the order of 5 nT. To decrease the noise we use magnetically shielded room. It is made of 45 mm thick aluminium plate and has dimensions of 2 x 2 x 2 m. The shielding is based on eddy currents induced in the conducting wall by the magnetic field. The attenuation is 50 dB at 50 Hz and is proportional to the frequency (7).

Noise Characteristics

The noise characteristics of the amplifier system composed of the transformer and the amplifier were studied by replacing the detector coil with an equivalent resistance (low noise wire resistor with 120 Ω resistance) to eliminate the noise induced by magnetic field and to maintain the same operating point of the amplifier. The output noise of the system was measured with a Honeywell Saicor Model SA151B spectrum analyzer when the resistor was in room temperature (293 K) and when it was cooled with liquid nitrogen (77 K). The output noise was referred to the input of the system (the gain was 10^6) to get the equivalent input noise. The results are in Fig. 6.

We can see that the equivalent input noise is of the order of 1,8 $nV_{rms}/\sqrt{Hz}$ in room temperature and 1,2 $nV_{rms}/\sqrt{Hz}$ when cooled. From these values we can calculate the noise figures of the system in both cases. The noise figures are 2,2 dB (293 K) and 4,5 dB (77 K). We can further calculate the equivalent input noise generated by the transformer and the amplifier. This noise referred to the input is of the order of 1 $nV_{rms}/\sqrt{Hz}$. The system noise (in temperature 293 K) referred to magnetic field at the measurement location is as low

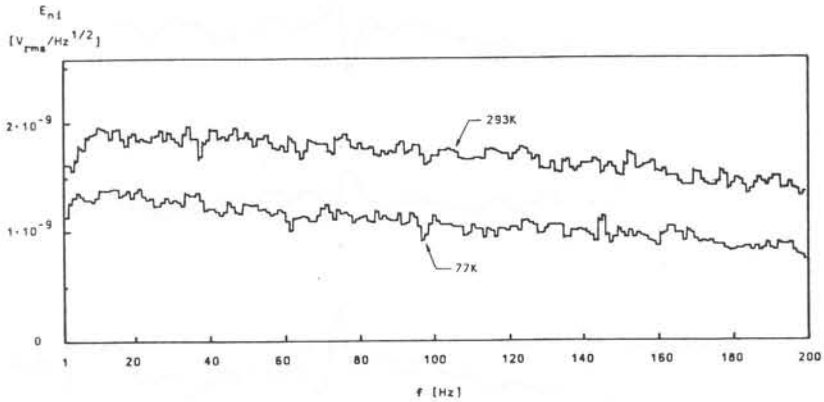

Figure 6. The equivalent input noise E_{ni} of the measurement system versus frequency with different temperatures of the source resistor.

as 180 $fT_{rms}/\sqrt{Hz}$ at 20 Hz.

Results

The first simultaneous recording of the three components of the magnetic heart vector is presented in Fig. 7. The signal is a direct recording without averaging. The noise originates from the thermal noise of the coils and from the low frequency magnetic noise in the shield. In this noise an unidentified 6 Hz noise is dominating. The 50 Hz noise from the line is effectively filtered from the recorded signal.

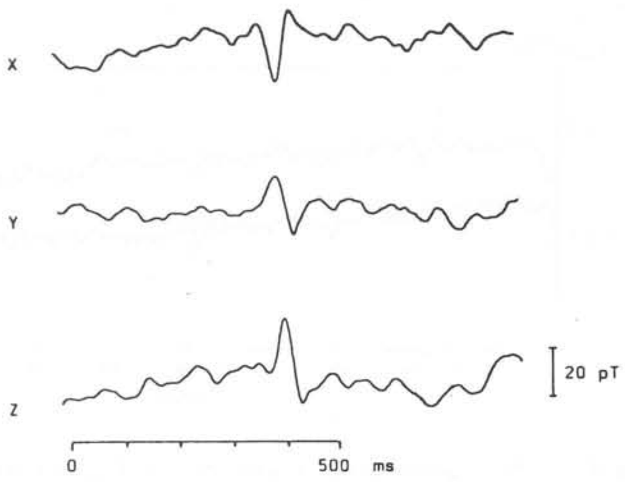

Figure 7. Components of the magnetic heart vector. The positive X, Y and Z axis point to anterior, left and superior directions.

References

1. Cohen, D., Edelsack, E.A. and Zimmerman, J.E.: Appl. Phys. Letters 16 7, 278-280 (1970).

2. Saarinen, M., Karp, P.J., Katila, T.E. and Siltanen, P.: Cardiov.Res. 8 6, 820-834 (1974).

3. Wikswo, J.P.Jr., Malmivuo, J.A., Barry, W.H., Crawford, G.E., Fairbank, W.M., Giffard, R.P., Harrison, D.C. and Roy, R.H.: Proc. San Diego Biomed. Symp. 14, 359-367 (1975).

4. Malmivuo, J.A.: Acta polytechnica scandinavica, Electrical engineering series 39, Helsinki 1976.

5. Motchenbacher, C.D. ja Fitchen, F.C.: Low-noise electronic design. John Wiley & Sons, New York 1973.

6. Nezer, Y.: Proceedings of the IEEE, Proceedings Letters, 3, 404-405 (1974).

7. Heinonen, P., Tuomola, M., Lekkala, J. and Malmivuo, J.: J. Phys. E: Sci. Instrum., 13 (1980).