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biomagnetic research

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## Multiplexed SQUID vectormagnetometer for biomagnetic research

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**Abstract.** A three-channel SQUID vectormagnetometer for biomagnetic studies is developed. The system is realised by using multiplexing where one of the three SQUIDS is monitored at a time. The system maintains continuously all the SQUIDS in the flux-locking condition. The construction of a gradiometer for recording the magnetic heart vector with the unipositional lead system is described. A low-noise, high-input impedance preamplifier and an RF-switch with nearly ideal properties at low signal levels are constructed and tested. The noise properties of the magnetometer system are estimated and measured. The usefulness of this vectormagnetometer is demonstrated by measuring the magnetic heart vector in real time from several healthy subjects. Two examples of these registrations are presented.

### 1. Introduction

Biomagnetic fields have been subject to intensive research over the past two decades. Sensitive magnetic detectors utilising superconducting quantum interference devices (SQUIDS) have been developed and used for biomagnetic measurements. At present growing interest in biomagnetism has increased the need for sensitive multichannel magnetometers. When using a multichannel system in magnetocardiography (MCG) the three magnetic heart vector components can be measured simultaneously. It means that the vector is recorded in real time and the changing of its parameters can be monitored during each heart beat.

The first vectormagnetometer suitable for real time registration of the magnetic heart vector (MHV) was constructed by the authors in 1980 (Lekkala and Malmivuo 1980). In this system the detector consisted of three mutually perpendicular induction coils working at room temperature. Each coil was coupled by means of an impedance transformer to a low-noise instrumentation amplifier in order to optimise the signal-to-noise ratio (Lekkala and Malmivuo 1981). A magnetic field sensitivity of  $180 \text{ fT}_{\text{RMS}}/\text{Hz}^{1/2}$  at 20 Hz was obtained. However, because the sensitivity of an induction coil magnetometer is proportional to the frequency, the noise level at 2 Hz was  $1.8 \text{ pT}_{\text{RMS}}/\text{Hz}^{1/2}$ . This is too high for real time recording of the P and T waves of the MCG.

The first multichannel SQUID magnetometer in this sensitivity range, which simultaneously measures three different magnetic field components was presented by Shirae *et al.* (1981). The system was based on the fact that a single RF circuit, normally coupled to one SQUID, can be used to monitor a number of RF SQUIDS simultaneously.

Ehnholm has used separate electronics units and specially constructed SQUIDS in his seven-channel, flux-guided, SQUID magnetometer designed for magnetoencephalographic studies (Ehnholm *et al.* 1981). Varpula has also reported a three-axis

differential SQUID magnetometer capable of detecting simultaneously the three components of the magnetic field vector with a sensitivity of  $28 \text{ fT}_{\text{RMS}}/\text{Hz}^{1/2}$  (Varpula *et al.* 1982). Both Ehnholm's and Varpula's systems consist of completely separate, commercial SQUID electronics units which work on different pump frequencies.

We have constructed a three-channel SQUID magnetometer by using conventional, commercially available RF SQUIDS. The multiplexing is applied so that only one of the three SQUIDS is monitored at a time and only one electronics unit is needed.

### 2. System description

To increase the sensitivity and to linearise the response of the SQUID a flux-locked loop is normally used. A typical circuit for the operation of the RF SQUID is shown in figure 1(a). The operation of this kind of circuit is presented in detail in a number of references (Lounasmaa 1974, Clarke 1977, Barone and Paterno 1982). The RF voltage across the tank circuit is amplified with a low-noise preamplifier. A square wave modulation flux at a frequency of a few tens of kilohertz and with a peak-to-peak amplitude of  $\sim \Phi_0/2$  is applied to the SQUID from the audiofrequency oscillator. This frequency is coupled also to the lock-in amplifier as a reference. When an external magnetic flux  $\Delta\Phi_e$  is coupled to the SQUID its operation point changes and the existing RF voltage will be amplitude modulated by the square wave. The phase of the modulation depends on the direction of the flux change. The modulated RF signal is detected and fed to the lock-in amplifier which acts as a phase comparator. Thus at the output of the lock-in amplifier there exists a DC voltage whose value depends on the modulation depth. This voltage is integrated and fed by a resistor back to the tank circuit. The magnetic flux which is generated by the feedback current in the tank circuit coupled to the SQUID, compensates the flux variation  $\Delta\Phi_e$  and drives the working point back to the starting value. The voltage change produced across the feedback resistance  $R_b$  is proportional to  $\Delta\Phi_e$ . This voltage is filtered by a low-pass filter and connected to the output.

The electronics unit which we have used as a base in our system is the commercial SUM-4 SQUID control unit (Instruments for Technology 1979†). Its working principle is the same as described above. It has two integrators, slow and fast. In slow mode the audiofrequency is set to 5 kHz and the slow integrator used. The time constant of the integrator depends on the gain of the system. In FAST mode the audiofrequency is 50 kHz. The slow mode is intended for situations where the maximum slew rate is not needed. The maximum voltage level of the integrator output is  $\pm 5\text{V}$ . If the voltage exceeds this value the level comparator resets the integrator in both modes. In the three gain set modes 1, 10 and 100 the full scale deflection of the output corresponds to  $\pm 100\Phi_0$ ,  $\pm 10\Phi_0$  and  $\pm \Phi_0$ , respectively.

The block diagram of the constructed multiplexed vectormagnetometer is presented in figure 1(b). The vector gradiometer detector and the SQUIDS are in liquid helium inside a Dewar. Each gradiometer is connected to one of the three SQUIDS. RF injection, audiofrequency drive and feedback are coupled separately to each SQUID. The common radiofrequency and audiofrequency oscillators are used for pumping all SQUIDS. The signals from the SQUID tank circuits are fed to three separate preamplifiers in the RF unit. After amplification RF signals are coupled alternately to the main RF amplifier by using fast RF switches which are driven by the multiplexing driver. The multiplexing frequency is one third of the audio oscillator's

† Instruments for Technology Oy, Ab. SUM-4 SQUID system. Instruction Manual and Data sheet 1979-E-1 Espoo 1979.

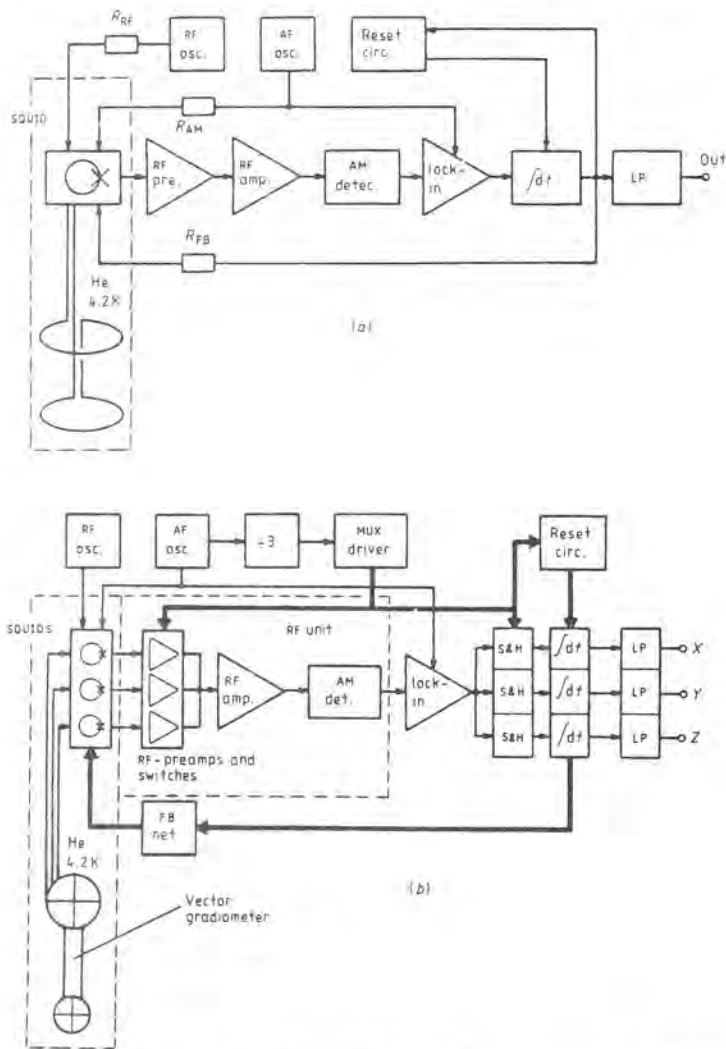


Figure 1. (a), Schematic of RF SQUID electronics in the flux-locked loop. (b), Block diagram of the constructed multiplexed vectormagnetometer. The units enclosed by the broken line (except the RF unit) are in liquid helium.

frequency. The multiplexed signal is amplified in the main RF amplifier, detected in an AM detector and coupled to the lock-in amplifier in the control unit. After passing this section the signals are separated by sample-and-hold circuits and integrated. The flux-locked loop is closed through the feedback network so that the working point is maintained. Thus there exist three separate parallel loops. The reset circuit is also multiplexed so that it resets the integrator in that channel which is turned on, if the maximum output value is exceeded. Each channel has a low-pass filter at the output. The main part of the electronics including RF amplifier, detector, lock-in amplifier, RF and AF oscillators and reset circuit, is common to the system.

The RF SQUIDS which we have used are commercial thin-film SQUIDS formed by four loops in parallel across a Nb-Nb oxide-Pb tunnel junction. These devices have been designed and

fabricated by Ehnholm *et al* (1975). The maximum usable RF frequency of these SQUIDS is about 50 MHz.

### 3. Detector

Our magnetometer is specially designed for vectormagneto-cardiographic studies with the unipositional lead system (Malmivuo 1976). In this system the magnetic heart vector (MHV) is recorded by measuring simultaneously the three orthogonal components of the magnetic field vector (MFV) in one point on the anterior side of the chest, figure 2. The detector is designed and optimised according to the dimensions of the volume source (heart) and the volume conductor (thorax). The detector coil system consists of three first-order asymmetric gradiometers wound on common fibreglass core with thin niobium wire. The X gradiometer, which measures the field

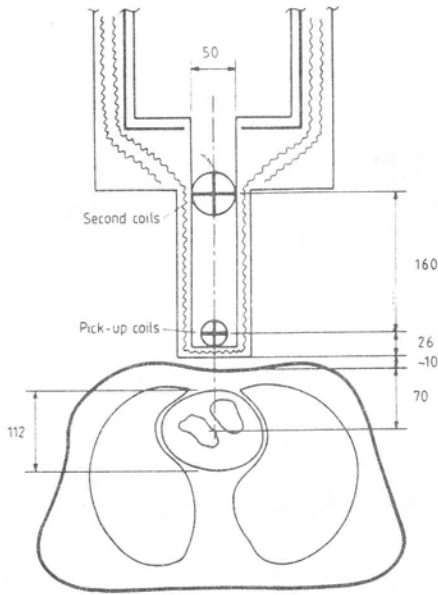


Figure 2. Measurement of the magnetic heart vector with the unipositional lead system. The vectormagnetometer is inside a Dewar which locates near the anterior thorax. Dimensions are given in millimetres.

gradient perpendicular to the patient's chest is constructed coaxially. In *Y* and *Z* directions there are coplanar gradiometers which measure the off-diagonal component of the magnetic field gradient, figure 3. All coils are circular. The number of turns are 15 and 5 and the coil lengths 4.5 mm and 2 mm in the pick-up and second coils, respectively. The baseline is 160 mm. No additional balancing methods are used for the gradiometers. The initial balance of the gradiometers was measured to be of the order of a few per cent. The vectormagnetometer is located inside the Dewar so that the distance from the centre of the pick-up coils to the outer surface of the Dewar is 26 mm. Thus the measurement distance to the centre of the heart is about 106 mm, figure 2. The calibration coefficient of the magnetometer with this vector gradiometer detector was measured to be  $57 \text{ pT V}^{-1}$ .

The detector core was constructed by using fibreglass tubes with thin walls. For the coplanar coils the tubes were cut and the pieces glued on the frame tube. The same glue (Super Epoxy; Plastic Padding) was used to fix the coils lightly on the core.

4. Multiplexing

According to the Nyquist sampling theory a continuous signal can be exactly reconstructed from samples which are taken from it at a rate which is at least two times the maximum existing signal frequency. Thus many signals can be transmitted in one channel by using a time division multiplexing (TDM) assuming that the bandwidth of the channel is so wide that the multiplexed signal is not distorted.

For magnetocardiographic measurements the maximum gain of the sum-4 system and the bandwidth up to 100 Hz are needed. In the control unit the slow mode with gain set 100 is the best alternative and the bandwidth is sufficient. However, it

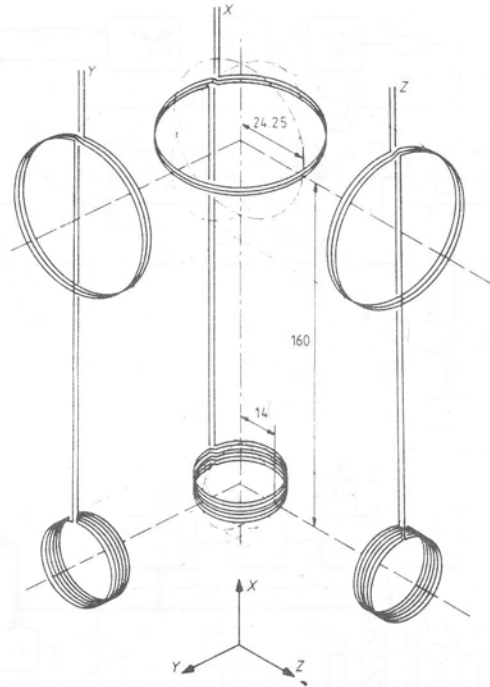


Figure 3. Coil construction of the asymmetric vectorgradiometer. For clarity different gradiometers are drawn apart. The *X* gradiometer is coaxial whereas the *Y* and *Z* gradiometers are coplanar constructions. Dimensions are given in millimetres.

is important to realise that in the situation where measurements are made in strong ambient noise (50 Hz for example) the slow rate of the system may be exceeded. To recover from such an interference it is necessary that the system is able to lock-in all signals existing at the input, including the noise. The given maximum slew rate of the system in FAST mode with gain set 100 is  $2.5 \times 10^4 \Phi_0 \text{ s}^{-1}$  (Instruments for Technology 1979). This corresponds to a maximum flux density of the order of 1.7 nT at 50 Hz with the detector used. The ability to track rapidly varying signals is greatest when the maximum slew rate is high (FAST mode). However, the SLOW mode can be used because the magnetic field noise level in the magnetically shielded room at Tampere, where the measurements were performed, is of the order of 40 pT in maximum at line frequency and is clearly below the maximum slew rate (Heinonen *et al* 1980).

When the multiplexing is applied to the flux-locked loop described above it is not sufficient that the samples transfer only the signal bandwidth (for example 100 Hz in MCG). The sampling rate must be so high that the whole bandwidth of the loop can be transferred without aliasing effect. Only in this case does the loop maintain its locking state and is the noise level of the system low. The loop bandwidth in our system is 4.5 kHz when the SLOW mode integrator is used. Thus the multiplexing frequency should be at least 9 kHz if we assume that the input signal to the multiplexer is filtered by an ideal rectangular filter with 4.5 kHz upper cut-off frequency. In real measurement situations anti-aliasing filters have not been used because the samples are taken from the RF signal. However, the detector coil



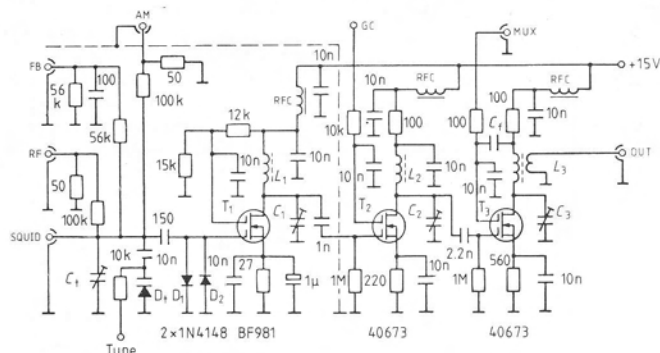


Figure 5. Circuit diagram of the low-noise preamplifier and switching stage of the prototype 2.

6 mA with the voltage of 8 V at gate 2. This voltage is used to gain control. The last stage is an RF switch.

### 6. The switching unit

The RF switch suitable for signal levels of few tens of microvolts and driving frequencies of tens of kilohertz requires a device with nearly ideal properties. There are two basic solutions to realise such an RF switch. Either an RF amplifier whose gain can be driven with high frequency or a device whose transfer conductance can be switched between high and low levels by a driving voltage, are usable. The properties of different RF switches are discussed in the following.

Diodes are often used as RF switches. With zero bias current a diode behaves like a high-impedance device for low-level RF signals. With sufficient bias the dynamic impedance decreases and thus the diode can be driven on and off with its bias current. However, the leakage capacitance of the pn-junction is relatively large and the RF signal leaks through the switch also when it is turned off. For PIN-diodes specially designed for RF switching the attenuation in the off-state is only about 30 dB. There is also another reason why diodes cannot be used in this application. Because the multiplexing must be performed at a frequency of the order of 10 kHz the switching current (about 50 mA in PIN-diodes) brings high transients to the output. These cannot be completely eliminated and the noise level of the system increases.

It is also possible to use another voltage-controlled resistor e.g. a FET. However, the analogue CMOS switches normally used for switching have too large input and transfer capacitances for high-frequency use. Also the crosstalk between channels is a serious problem. In the prototype 1 (figure 4) a high-frequency JFET was used as a switch. The switching unit consists of a buffer stage BF254 which lowers the impedance level and 2N3823 used as a switch. It is driven by the gate voltage limited between  $-3.5$  V and  $+0.4$  V. An attenuation of 28 dB in the off-state was obtained with this configuration. In figure 6 there can be seen a switched 20 MHz signal whose level at the switch input is  $200 \mu\text{V}_{\text{RMS}}$ . The switching rate is 10 kHz. The risetime of the output envelope is about  $1 \mu\text{s}$  but the fall time is as long as  $15 \mu\text{s}$ . The transient spikes, which are coupled through the gate leakage capacitance by the sharp edges of the driving voltage, are sufficiently attenuated.

The gain of a dual-gate MOSFET amplifier can be adjusted with the voltage on its second gate (figure 5). The switching on high frequency also causes the DC level of the output to change.

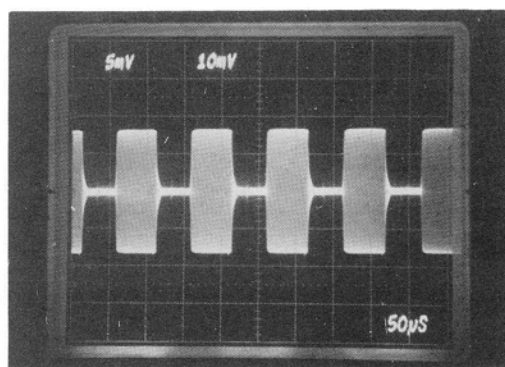
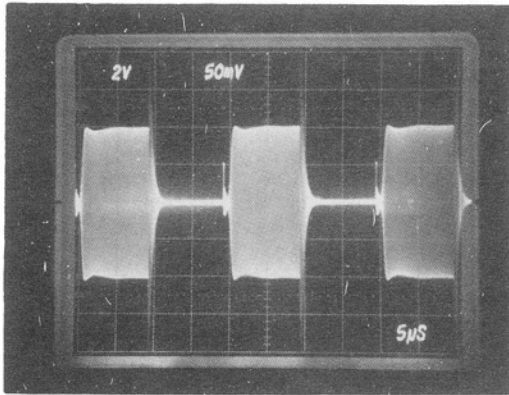


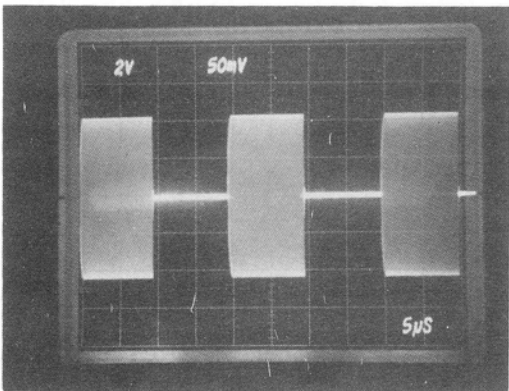
Figure 6. The output waveform of the prototype 1. The 20 MHz signal is switched at the frequency of 10 kHz. The signal amplitude at the input of the switching stage is  $200 \mu\text{V}_{\text{RMS}}$ . The horizontal scale is  $50 \mu\text{s div}^{-1}$

The voltage spikes caused by the DC level shifts were a difficult problem. Figure 7(a) shows a typical switching waveform obtained when the driving voltage was filtered for an optimum result. The risetime is delayed and oscillation occurs during the on-time. The fall time is long (about  $1 \mu\text{s}$ ) and an oscillation burst disturbs the switching. The multiplexing frequency is 50 kHz. Although the resonant circuit was left out the spikes occurred also in a resistively coupled circuit.

A very simple solution was found to this problem. By adding a capacitor  $C_f$  (figure 5) from the top of the resonant circuit to the second gate of the MOSFET a negative feedback path is formed. The feedback has no effect on radio frequencies because the gate 2 is grounded with a 10 nF capacitor. However, the spike in quiescent current couples through  $C_f$  to the gate 2 and eliminates bursts and oscillation. The value of  $C_f$  is 10–20 nF. Figure 7(b) shows the 20 MHz signal multiplexed with this switch at the relatively large frequency of 50 kHz. As can be seen the envelope is smooth and the rise and fall times are fast, 600 ns and 50 ns, respectively. The attenuation of the signal during off-state is 45 dB when compared to the on-state.



(a)



(b)

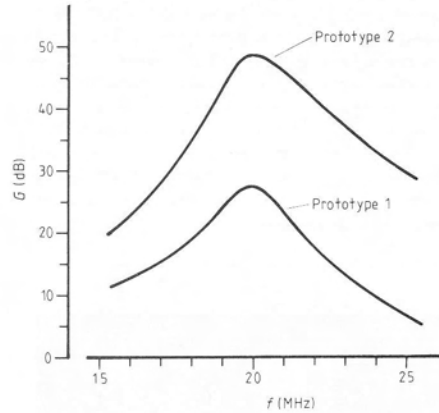
**Figure 7.** Output waveforms of two switched dual-gate MOSFET RF amplifiers. The signal frequency is 20 MHz and the switching frequency 50 kHz. (a). The driving voltage of the gate 2 is filtered for optimum result; (b). a feedback capacitor is added to smooth the output waveform. The RF-signal amplitude at the amplifier input is  $100 \mu\text{V}_{\text{RMS}}$ . The horizontal scale is  $5 \mu\text{s div}^{-1}$ .

### 7. Measured performances

The properties of the two constructed preamplifier units were tested by measuring the gain, the equivalent input noise and the sensitivity. All measurements were made by using  $50 \Omega$  load and source impedances. The plot of gain against frequency is presented in figure 8 for both systems. The maximum gains of the prototypes 1 and 2 at the centre frequency of 20 MHz are 27.5 dB and 48.5 dB, respectively. The bandwidths are 1.5 MHz and 1.8 MHz, respectively.

The noise and sensitivity measurements were carried out by using a spectrum analyser and an RF generator with a calibrated output. The equivalent input noise of the prototype 1 was measured to be  $2 \text{ nV}_{\text{RMS}} \text{ Hz}^{-1/2}$ . The sensitivity is  $8 \mu\text{V}_{\text{RMS}}$  with 10 dB S/N-ratio and the noise figure 7 dB with  $50 \Omega$  source resistance.

The second prototype was tested in the same manner. The equivalent input noise level was calculated to be  $1.4 \text{ nV}_{\text{RMS}} \text{ Hz}^{-1/2}$ . By taking into account the bandwidth of

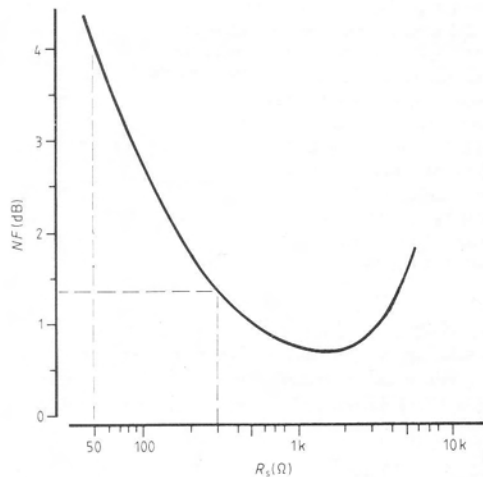


**Figure 8.** The gain of prototypes 1 and 2 as a function of frequency when tuned to 20 MHz.

1.8 MHz a sensitivity of  $6 \mu\text{V}_{\text{RMS}}$  with 10 dB S/N-ratio was reached. Because the source resistance was  $50 \Omega$  the noise figure of the amplifier is 4 dB, which is good. This value agrees well with the data given in figure 9 where the noise figure  $NF$  of BF981 is presented against source resistance  $R_s$ . The curve is calculated from the data given in BF981 Philips Data sheet 1982. Thus the equivalent input noise of the amplifier when measuring with the SQUID ( $R_s = 300 \Omega$ ) can be estimated to be about  $2.6 \text{ nV}_{\text{RMS}} \text{ Hz}^{-1/2}$ . As can be seen the noise properties of the two amplifiers are nearly equal. However, the prototype 2 has higher gain and it is also smaller.

### 8. Mechanical construction

The vectormagnetometer detector is attached to the lower end of a fibreglass tube which acts as a frame of the magnetometer.



**Figure 9.** Noise figure  $NF$  of BF981 as a function of source resistance  $R_s$ .

The SQUIDS are fixed symmetrically around the tube on an acrylic plate. The SQUIDS are coupled with coaxial cables to the BNC connectors on a bakelite plate attached on the upper end of the frame tube and screwed to the top of the Dewar. The case of the RF unit is made from 20 mm thick aluminium plate by milling out compartments for the printed circuit boards, figure 10. The case containing preamplifiers, RF switches, main RF amplifier, detector, RF oscillator and cable connector is fitted with the BNC connectors on the top plate. The constructed suspension system allows the Dewar to be moved horizontally and vertically. It can also be turned around the vertical axis and tilted 30° from the vertical direction.

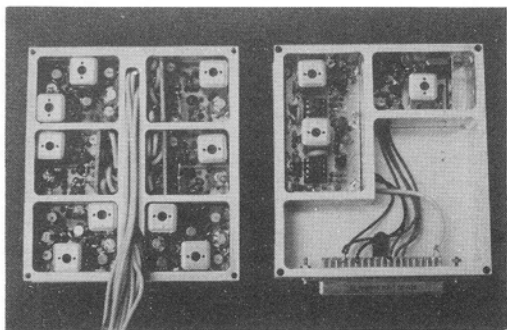


Figure 10. Mechanical construction of the RF unit's case. On the left there are seen the preamplifiers and switching units. The main RF amplifier and RF oscillator are shown on the right in the figure in a case which is placed on the top of the other one and assembled together.

### 9. Noise of the magnetometer

There exist several noise sources in measurements with a SQUID magnetometer. The internal noise originates from the measurement system. There is the intrinsic noise due to the SQUID, the noise associated with circuits directly coupled to the SQUID and noise generated by the preamplifier and the circuits in the control unit. External noise effects are the thermal noise field from the conducting materials surrounding the detector, low-frequency magnetic fields generated by different sources (the line frequency fields, moving ferromagnetic objects in Earth's DC magnetic field and geomagnetic fluctuations), vibration of the detector in DC magnetic field and radio frequency fields.

The intrinsic noise of the SQUID was determined from the 'staircase curve' (RF voltage against RF current) by measuring the slope of the 'riser' (Jackel and Burhmann 1975). The intrinsic noise flux has a value of

$$\langle \delta\Phi_1^2 \rangle^{1/2} \approx 2.2 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}. \quad (2)$$

The effective temperature of the tank circuit has a value which differs from 4.2 K because part of the tank circuit is at room temperature. A typical value is 200 K (Jackel and Burhman 1975). By using this value the equivalent noise flux caused by the tank circuit is calculated to be

$$\langle \delta\Phi_{tc}^2 \rangle^{1/2} \approx 4.3 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}. \quad (3)$$

When  $R_s$  is equal to 300  $\Omega$  the noise figure of the preamplifier is 1.4 dB. Thus its equivalent noise resistance is 114  $\Omega$ . The equivalent flux noise generated by the preamplifier can thus be

calculated to be

$$\langle \delta\Phi_p^2 \rangle^{1/2} = 4.8 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}. \quad (4)$$

The total equivalent flux noise of the system without multiplexing and assuming that the noise sources are uncorrelated is

$$\begin{aligned} \langle \delta\Phi_{tot}^2 \rangle^{1/2} &= \sqrt{\langle \delta\Phi_1^2 \rangle + \langle \delta\Phi_{tc}^2 \rangle + \langle \delta\Phi_p^2 \rangle} \\ &= 6.8 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}. \end{aligned} \quad (5)$$

This value agrees with the value of  $8 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}$  given by the manufacturer (Instruments for Technology 1979†). The system noise was measured without the detector coil (the signal coil open) and without multiplexing. Because the coupling of the external noise flux into the SQUID has a minimum in this situation, this equivalent flux noise represents the total internal noise of the system or the same as equation (5). The measured value is about  $7.5 \times 10^{-5} \Phi_0 \text{ Hz}^{-1/2}$ . The main part of the noise originates from the preamplifier and this noise was found to be equal for all SQUIDS. If the calibration coefficients for the vector gradiometer are used, the ultimate sensitivity of the magnetometer is 21 fT<sub>RMS</sub> Hz<sup>-1/2</sup>.

In figure 11 is shown the magnetic noise field spectrum measured in the vertical direction at the magnetically shielded room of Tampere University of Technology during a magnetically silent period. The ground noise when expressed as magnetic flux density is 40 fT<sub>RMS</sub> Hz<sup>-1/2</sup>. This is the practical limit of the system's sensitivity. The performance of the room at 50 Hz magnetic field has been studied by Heinonen *et al* (1980).

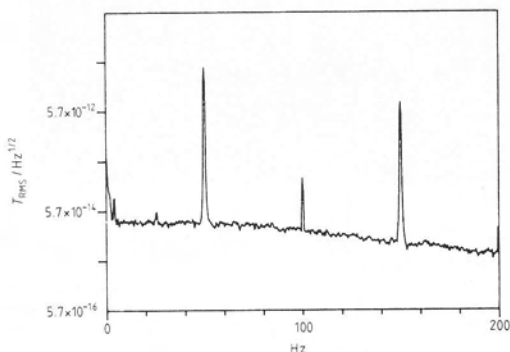


Figure 11. Magnetic noise spectrum in the magnetically shielded room at Tampere University of Technology. The measurement is made in the vertical direction with a first-order gradiometer.

They have found that in all three orthogonal directions at least 40 dB attenuation is achieved in about half of the enclosure's area. The maximum attenuation is about 50 Hz. The 50 Hz magnetic field inside the room varies from 5 to 60 pT depending on its direction and the time of day. We have used a digital notch filter locked into the line frequency (Heinonen 1982) to remove the interferences on the basic frequency and its harmonics. The Earth's magnetic field is compensated by a coil system (Suomaa *et al* 1980). The ripple of the driving current generator is so low that the magnetic low-frequency noise of

† Instruments for Technology Oy, Ab. SUM-4 SQUID system. Instruction Manual and Data sheet 1979-E-1, Espoo 1979.

$\pm 25$  nT caused by this current ripple is clearly lower than the magnetic fluctuations of Earth ( $\pm 100$  nT).

The measured noise levels of each channel of the fibreglass vectormagnetometer when the multiplexing is used are:

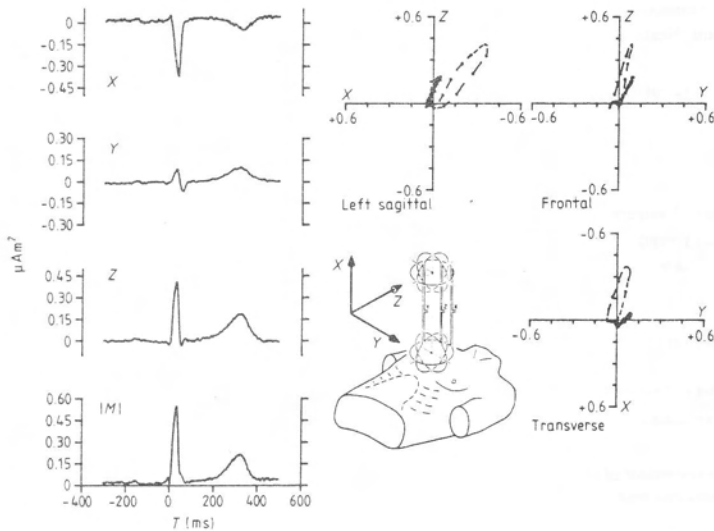
- X channel:  $105 \text{ fT}_{\text{RMS}} \text{ Hz}^{-1/2}$
- Y channel:  $85 \text{ fT}_{\text{RMS}} \text{ Hz}^{-1/2}$
- Z channel:  $100 \text{ fT}_{\text{RMS}} \text{ Hz}^{-1/2}$

Thus the noise level is 7–8 dB higher in the multiplexed system. This is caused by the fact that there always exists a reconstruction error which couples back to the SQUID through the feedback network. A field change of 50 fT corresponds to a 0.9 mV error in reconstructing.

Another reason is the aliasing effect. The anti-aliasing filter is normally used to limit the bandwidth of the sampled signal. However, in this vectormagnetometer the sampled wave is a radio-frequency signal which carries the information on the SQUID's state. So the anti-aliasing filter should be at the input of the SQUID. The RF foil around the gradiometer acts as a low-pass filter with the cut-off frequency of about 2 kHz. This limits the bandwidth of the externally coupled signals. However, the internally generated noise couples as such to the multiplexer.

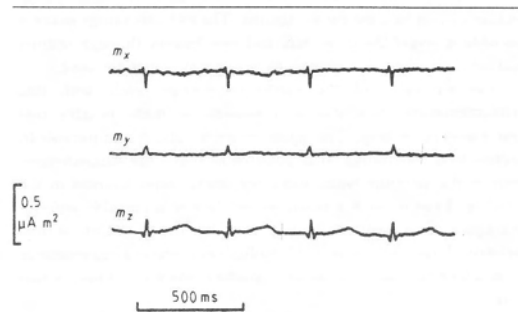
**10. Measured magnetocardiograms**

The constructed vectormagnetometer was tested by measuring the magnetic field of the heart from several normal subjects. All measurements were performed in the Tampere University's magnetically shielded room. Figure 12 shows the magnetic heart vector measured in real time from a normal 30-year-old male subject. We have used as the measurement point on the anterior chest wall the point  $V_2$  of the standard 12-lead ECG which is the optimal location for the unipositional lead system (Eskola 1983). On the left part of the figure, MHV components  $m_x, m_y, m_z$  and magnitude  $|m|$  are plotted as a function of time. On the right, MHV loops on the left sagittal, frontal and transverse planes are presented.



**Figure 12.** The magnetic heart vector (MHV) recorded in real time from a 30-year-old male subject by using the unipositional lead system. The scalar display and vector loops are shown. Detector's distance from the centre of the heart is 106 mm.

The fact that no electrodes are needed and that subjects can be measured with their clothes on, in addition to the use of the vectormagnetometer gives a useful tool to monitor the MHV from infants. In figure 13 there can be seen the MHV components recorded from a three-month-old child. The quality of the measurements is so good that a diagnosis from it is possible. The recording was made during a few seconds while the child was playing with her mother.



**Figure 13.** Scalar display of the MHV components recorded from a three-month-old child.

**11. Discussion**

Our three-channel SQUID vectormagnetometer is mainly designed for vectormagnetocardiographic studies. However, the method can be used in any multichannel magnetometer application. Shirae's system (Shirae *et al* 1981) works with frequency division multiplexing (FDM) whereas our system is time division multiplexed (TDM). The advantage of Shirae's method is that only one RF circuit and RF amplifier are needed.

However, to achieve acceptable performance the SQUIDS which are fabricated on the same substrate must have very closely matched characteristics. Shirae has reached a sensitivity of  $1 \text{ pT Hz}^{-1} \text{ cm}^{-1}$  when measuring three orthogonal magnetic field gradients. This noise level is a little too high for biomagnetic research. In our system one SQUID is monitored at a time but the feedback is continuously connected. The main advantages of this method are that no special SQUID construction is needed, only one SQUID is monitored at a time and the same pump frequency can be used for all SQUIDS. The two last things make it possible to avoid the cross talk and interference through SQUIDS and detector coils, as happens when separate units are used.

The examples of the vmCG recordings made with this instrumentation show that it is possible to make reliably real time vmCG recordings. The signal-to-noise ratio is comparable to routine ECG recordings. It is possible to make the measurement even in the daytime when there are many noise sources in the building. Even in such a situation the data is acceptable without averaging. The MHV recording can be made within a few seconds. Thus the constructed multiplexed vectormagnetometer gives a tool to make biomagnetic studies reliably and fast in real time.

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