2 Development of SQUID based MCG system

This chapter describes the Magnetocardiography (MCG) system that was developed as a part of the work carried out during the course of this thesis. This chapter also gives an account of the sensor calibration of the MCG system, evaluation of system noise and the standardized procedure adopted for carrying out MCG measurements on a human subject inside a magnetically shielded room.

2.1 MCG system set up

Fig. 2.1 shows the block diagram of the MCG measurement set up developed at IGCAR that comprises of different modules (or units) such as Magnetically Shielded Room (MSR), Superconducting QUantum Interference Device (SQUID) gradiometers, cryostat, SQUID electronics, data acquisition system, etc. The same SQUID based MCG system can also be used for probing the functional activities of a limited cortical area by measuring the magnetic field associated with neuronal currents, which is known as magnetoencephalography (MEG). The system for the measurement of MCG/MEG signals started initially as a single channel system and was upgraded progressively to house 4, 13 and 37 SQUID channels respectively [1, 2]. Although, MCG and MEG systems share the common infrastructure such as the Magnetically Shielded Room for measurements, the flat bottom cryostat used for MCG limits the cortical area probed during the MEG measurements using this system. For MCG measurements, a flat bottom cryostat (with sensors arranged in a plane at the same level) is used to match the approximately flat profile of the surface of the thorax. On the other hand, for whole head MEG measurements, a concave bottom cryostat wherein the sensors are arranged on the surface of a helmet shaped sensor holder that has the profile of an average adult human head is preferred. Such an MEG cryostat with 90 channels has been assembled and is being commissioned. Here, we particularly discuss the features of the MCG system used for the
work presented in this thesis.

Figure 2.1: Block diagram of the MCG system used for the present work. The liquid helium cryostat housing an array of axial gradiometers immersed in liquid helium is located inside the Magnetically Shielded Room (MSR). The MCG signals originating from a human subject are detected by an array of axial gradiometers. The SQUID output voltage of each sensor is amplified and processed by Flux Locked Loop (FLL) electronics inside the MSR. The output signals are then taken to the radio frequency shielded room (RFSR), via waveguides connecting the MSR to RFSR, for digitization using analog-to-digital converter (ADC) cards. All the electronic equipments that require electrical Mains Power are housed inside the RFSR. The digitized data is transmitted via an optical fibre link for storage and graphical display on a server class computer located in the unshielded environment.

In the following, we give a brief account of the basic principles governing the operation of the SQUID sensor and the associated Flux Locked Loop (FLL) electronics used for the linearization of the periodic output of the SQUID.

2.2 SQUID as a magnetic field sensor

Biomagnetic signals \((10^{-12} - 10^{-15} \text{ T})\) are extremely weak [3]; they are four to six orders of magnitude weaker than the fluctuations in the earth’s magnetic field \((\sim 2 \text{ nano-Tesla})\) and about six
Development of SQUID based MCG system

to seven orders of magnitude weaker than the earth’s static magnetic field (∼50 µT). The magnetic noise in an urban setting is generally caused by mains power line, vehicle movements, vibrations, etc., which may be in the range 1 nano-Tesla to 100 nano-Tesla depending on the site. The cardiac magnetic field (typically, ∼40 pT at the QRS peak) although relatively strong compared to the magnetic field produced by the human brain (100 fT - 2 pT), is completely masked by the ubiquitous ambient magnetic noise. In order to measure such a weak magnetic field associated with the biomagnetic sources, there are two essential prerequisites;

- The first is a magnetic field sensor, which is sufficiently sensitive to measure such extremely weak magnetic fields in the range of 100 femto-Tesla to 100 pico-Tesla.

- The second involves the reduction of the ambient magnetic noise to a level lower than the signal of interest.

The advent of the Superconducting QUantum Interference Device (SQUID) [4, 5] with a sensitivity of 1 to 10 femto-Tesla has therefore been crucial for the measurement and characterization of biomagnetic fields; indeed, these are the only sensors with adequate sensitivity to measure the biomagnetic fields.

Josephson effect and the flux quantization form the fundamental principles which govern the operation of the SQUID [4, 5]. The flux quantization implies that the magnetic flux threading through a closed superconducting loop is constrained to be an integral multiple of a flux quantum (Φ₀ = \frac{h}{2e}, 1Φ₀ = 2.07 \times 10^{-15} \text{ Wb, where } h \text{ is the Planck’s constant and } e \text{ is the elementary charge}). Josephson effect refers to the establishment of phase coherence between two weakly coupled superconductors, which allows super-current to flow from one superconductor to other without developing any voltage. The weak coupling may be realized in practice by separating the two superconductors by a thin tunnel barrier of an insulating material. Such a structure is known as the Josephson junction, and the quantum mechanical tunneling of cooper pairs across the junction assists in the establishment of phase coherence between the two superconductors.

Depending on the operating principle, SQUID sensors are broadly classified into two categories: DC SQUID (which is biased with direct current) and RF SQUID (which uses a radio frequency current bias). The DC SQUID consists of a superconducting loop intercepted by two weak links.
whereas in the RF SQUID, the superconducting loop is intercepted by a single weak link. Both types of SQUID sensors act as magnetic flux to voltage transducers [4, 5] and their output voltage is a periodic function of the magnetic flux coupled to them with the periodicity of a flux quantum $\Phi_0$. SQUID sensors based on Nb (having transition temperature, $T_c \sim 9$ K) are operated at liquid helium temperatures (4.2 K) and are known as low $T_c$ SQUIDs (LTSC SQUID), while SQUID sensors based on high temperature superconductors such as YBCO ($T_c \sim 90$ K) are operated at liquid nitrogen temperatures (77 K) and are known as high-$T_c$ SQUIDs (HTSC SQUID) [4]. LTSC SQUID sensors offer superior sensitivities in the detection of magnetic field changes compared to the HTSC SQUIDs. We use low temperature DC SQUID (based on Nb/$\text{AlO}_x$/Nb Josephson junctions) as a magnetic sensing device for MCG measurement and the working principle of the DC SQUID is briefly discussed in the following section.

### 2.2.1 Working Principle

In the absence of any external magnetic field, when a symmetric DC SQUID (which has two identical Josephson junctions) is biased with a DC current ($I_b$), the bias current $I_b$ divides equally ($I_1 = I_2 = I_b/2$) into the two branches as shown in fig.2.2(a) and flows through the two Josephson junctions. The critical current (which represents the maximum current the SQUID sensor can support at zero voltage) of the SQUID in this case would be equal to $2I_c$, where $I_c$ denotes the critical current of each Josephson junction. When a magnetic field ($B$) is applied perpendicular to the plane of the SQUID loop, a screening current ($I_s$) begins to circulate around the loop so as to produce the necessary magnetic flux required to satisfy the requirement of flux quantization [4, 5, 6].

$$\Phi_{\text{tot}} = \Phi_{\text{ext}} + LI_s = n\Phi_0 \quad (2.1)$$

Here $\Phi_{\text{tot}}$ is the total flux threading the SQUID loop, $\Phi_{\text{ext}}$ is the externally applied flux, $I_s$ is the screening current induced by the applied flux, $n$ is an integer which has to be chosen to minimize $I_s$, $L$ is the inductance of the SQUID loop, and $\Phi_0$ is the flux quantum. When the external applied flux is $n\Phi_0$, the screening current $I_s$ induced is zero and when the external applied flux is $(n + \frac{1}{2})\Phi_0$, the induced screening current is $\pm(\Phi_0/2L)$. Thus, as the magnetic flux coupled to the SQUID is monotonically varied, the induced screening current varies periodically with the periodicity of $\Phi_0$. 


Since the induced current circulates around the SQUID loop in a clockwise or counterclockwise direction depending on the magnitude and direction of the applied magnetic field, the direction of the induced current $I_s$ is the same as that of the bias current $I_b$ in one branch of the SQUID while its direction is opposite to that of the bias current in the other branch, and this gives rise to a total current of $I_1(= \frac{I_b}{2} + I_s)$ in one branch while in other branch, it is $I_2(= \frac{I_b}{2} - I_s)$ (see fig.2.2(a)). A voltage appears across the junction as soon as the total current in any one of the two branches exceeds the critical current of the Josephson junction in that branch. As a result of this, the critical current of the SQUID reduces from $2I_c$ to $(2I_c - 2I_s)$. The critical current of the SQUID is a periodic function of the externally applied magnetic flux and is given by [7].

$$I_{c_{\text{max}}}(\Phi_{\text{ext}}) = 2I_c|\cos\left(\pi \frac{\Phi_{\text{ext}}}{\Phi_0}\right)|$$  \hspace{1cm} (2.2)

A voltage is developed across the SQUID when the SQUID is biased with DC current slightly larger than $2I_c$; this output voltage is a periodic function of external flux with the periodicity of a flux quantum.
quantum $\Phi_0$ (see fig.2.2(b)), as the magnetic field applied perpendicular to the plane of the SQUID loop [4, 5, 6] is varied. Thus SQUID can be termed as a magnetic flux to voltage transducer, giving a measurable output voltage for extremely small variations in the applied magnetic field, which are not measurable by any other sensor technology. The linearization of the periodic output voltage of the SQUID sensor is essential for most of the SQUID based measurements and is generally accomplished using a Flux Locked Loop (FLL) or current locked loop electronics [4, 8]. The discussion on the FLL, that we use in the MCG system, is given in the following section.

### 2.2.2 Flux Locked Loop (FLL) electronics

Several SQUID electronics readout schemes [5, 7, 9], such as direct readout, flux locked loop mode, etc., have been used for the purpose of linearizing the otherwise periodic flux-voltage characteristics of the bare SQUID sensor. In many such schemes, the SQUID often serves as a null detector in the feedback control system by keeping either the flux threading through the SQUID loop (in the flux locked mode) or the screening current flowing through the gradiometer (in the current-locked mode) constant [8]. The working point $W$ on the $V$–$\Phi$ characteristic (see fig.2.2(b)) is maintained by cancelling the changes in the input flux or the screening current respectively using suitable feedback circuits in the SQUID electronics. In brief, the output voltage across the DC SQUID sensor resulting from the application of the signal flux is amplified, filtered and fed back as the compensating flux to ensure that the SQUID stays locked to the chosen working point. We now discuss in somewhat greater detail the operation of the SQUID in the Flux Locked Loop (FLL) mode [6], as shown in fig.2.3, which is extensively used for most of the DC SQUID readout systems (including the MCG system assembled during the course of this thesis).

Fig.2.3 shows the typical block diagram of FLL electronics [6] and this has been implemented in the present MCG system. An oscillator is used to pass a 100 kHz or higher frequency AC current with an adjustable amplitude through a modulation coil, which inductively couples a weak modulation flux of peak-to-peak amplitude up to $\sim \Phi_0/2$ into the SQUID loop. Depending on the flux-bias, the output voltage of the SQUID contains components at 100 kHz or 200 kHz or a mixture thereof with differing amplitudes of 100 kHz and 200 kHz components. A phase sensitive detector is arranged to detect the amplitude of 100 kHz component, which is indicative of the
2 Development of SQUID based MCG system

Figure 2.3: Typical schematic of the Flux Locked Loop (FLL) electronics circuit used for linearizing the periodic flux-to-voltage output response of the bare DC SQUID.

extent of deviation from the working point caused by the application of the input flux; the output of the phase sensitive detector is used to derive a feedback current to inject a compensating flux into the SQUID loop by passing the current through the feedback coil (and also the feedback resistor connected in series with the feedback coil). The output voltage across the feedback resistor ($R_f$) is a measure of the input flux signal, which is intended to be measured. The output voltage of the FLL is zero when no signal flux is coupled to the SQUID loop as SQUID remains at the working point $W$. When a signal flux is applied, however, there is a deviation from the working point and the FLL generates an output voltage which is proportional to the applied flux. The SQUID output voltage readout scheme based on flux locked loop electronics increases the dynamic range of measurement of magnetic flux, and also eliminates the low frequency noise arising from
low frequency drifts in the junction critical currents/other SQUID parameters and $1/f$ noise from the preamplifiers etc. as the signal of interest is moved to frequencies ($\sim 100$ kHz) above the $1/f$ threshold. Since the maximum feedback voltage is limited by the supply voltage (say, 12 V), the dynamic range is limited to the flux equivalent of the supply voltage at the selected gain of FLL and may be increased depending on the requirements by selecting a lower FLL gain, which may be easily tuned by changing the feedback resistor $R_f$, since it is given by the ratio $R_f/M_f$. The output voltage that appears across the feedback resistor $R_f$ is proportional to the external flux intended to be measured and thus the periodic response of the bare SQUID is converted into a linear one, with a flux-to-voltage conversion factor independent of the open loop gain of the amplifier [10].

Although, the signal bandwidth is limited by the modulation frequency and is often substantially lower than the modulation frequency, this is not a serious limitation in biomagnetic investigations since a bandwidth of 1kHz is adequate for most of the applications. In the FLL mode, the total magnetic flux noise of the dc SQUID is given by [7],

$$S_{\Phi}^{1/2} = \sqrt{S_{\Phi,SQUID} + \frac{S_{V,AMPL}}{V_{\Phi}^2}}$$  \hspace{1cm} (2.3)

where $S_{\Phi,SQUID}$ and $S_{V,AMPL}$ are the flux noise of the SQUID and voltage noise of the preamplifier respectively and $V_{\Phi}(=\partial V/\partial \Phi)$ is the flux-to-voltage transfer function of the bare SQUID. It may be noted that the decrease in the flux-to-voltage transfer function $V_{\Phi}$ deteriorates the attainable signal-to-noise ratio.

### 2.3 Reduction of ambient magnetic noise

Although SQUID sensors have the requisite sensitivity to detect the extremely weak biomagnetic fields, the ambient magnetic noise such as that associated with power line interference, vibrational noise, magnetic noise associated with vehicle movements etc., present at the site is orders of magnitude higher than the signal of interest and completely masks it. This poses a real difficulty in the measurement of biomagnetic signals, unless the ambient magnetic noise in the frequency bandwidth of interest (0-1000 Hz) is suppressed to a sufficiently low level.

Fig.2.4 indicates the strength of the typical biomagnetic signals of interest and compares it
Development of SQUID based MCG system with the strength of the magnetic fields associated with some of the parasitic sources of ambient magnetic noise in the neighbourhood [7, 12]. In order to detect such extremely weak biomagnetic signals, it is therefore essential to attenuate all the parasitic sources of the ambient magnetic noise by using some form of shielding [7, 13] such as that provided by a Magnetically Shielded Room (MSR) and by using pick-up loops in the form of gradiometers [14] which further reduce the contribution of the residual magnetic noise present inside the MSR by discriminating against distant sources of magnetic noise which are expected to produce a nearly uniform field over the relatively small volume of the gradiometer; nearby sources, on the other hand, are expected to produce a non-uniform magnetic field over the volume of the gradiometer and, hence, get preferentially detected. Some research groups have tried to avoid the use of the expensive MSR for biomagnetic measurements by using higher order hardware gradiometers [14] or synthetic gradiometers [15] for measuring biomagnetic signals in an unshielded environment; however, measured signals in this case are generally of much poorer quality. In the following sections, we discuss the basics of MSR and the gradiometers that we employed for the reduction of ambient magnetic noise.

Figure 2.4: Schematic illustration of the strength of the magnetic fields originating from different sources (adapted from [12]). Since the biomagnetic fields of interest are much weaker compared to the parasitic ambient magnetic noise, it is essential to attenuate the ambient magnetic noise using some form of shielding in order to measure the signal of interest.
2 Development of SQUID based MCG system

2.3.1 Magnetically Shielded Room (MSR)

As discussed earlier, Magnetically Shielded Room (MSR) is a prerequisite for biomagnetic measurements and is of crucial importance for all high quality measurements of biomagnetic signals [7, 11, 16]. MSR serves to attenuate the external ambient magnetic noise to a sufficiently low level so as to enable the SQUID sensors to measure the extremely weak biomagnetic signals, which are otherwise completely masked by the ambient magnetic noise. The magnetic shielding may be categorized into various types: ferromagnetic shielding using high permeability \( \mu \)-metal (low frequency magnetic shielding), eddy current shielding using high conductivity aluminum or copper [7] (high frequency magnetic shielding), active compensation [17] or superconducting magnetic shielding [18] etc. Here, we specifically describe the shielding configuration employed in our laboratory [19].

Our Magnetically Shielded Room (MSR), custom built by IMEDCO, Switzerland, inside which all the measurements reported in this thesis were carried out, has the internal dimensions 3m(width) \( \times \) 4m(length) \( \times \) 2.4m(height) and is constructed using two layers of ferromagnetic shielding (\( \mu \)-metal) and two layers of eddy current shielding (aluminum) on each of the six sides. The outer aluminum layer is 4 mm thick, while the outer \( \mu \)-metal layer is 2 mm thick. The inner aluminum layer is 8 mm thick while the inner \( \mu \)-metal layer is 3 mm thick. The top panel of fig.2.5 shows the photograph of the MSR erected in our laboratory, at IGCAR, for biomagnetic measurements such as MCG and MEG. The principles of magnetic shielding provided by \( \mu \)-metal and the aluminum layer may be explained [5, 11, 16] as follows:

High frequency magnetic noise is attenuated through the induction of eddy currents in the grounded high conductivity aluminum panels; by Lenz’s law the eddy currents are induced in a direction so as to oppose the original magnetic field fluctuations. While this eddy current shielding is quite effective at high frequencies, it ceases to be effective at low frequencies owing to an inevitable increase in skin depth at low frequencies. For low frequency magnetic shielding, ferromagnetic materials such as \( \mu \)-metal are used for the construction of the walls of MSR. Owing to the very high magnetic permeability of the \( \mu \)-metal compared to air, it offers a path of low magnetic reluctance to the magnetic lines of force, and consequently, the magnetic field lines, which would have otherwise
2 Development of SQUID based MCG system

Figure 2.5: (a). The photograph of the magnetically shielded room (MSR) used for shielding against the external ambient magnetic noise. The bottom panel (b) shows the measured shielding factor of the MSR as a function of frequency for each of the three mutually orthogonal components of the magnetic field along X, Y and Z directions. It may be noted that the MSR provides a shielding factor of 70 dB at 1 Hz and 110 dB at 100 Hz and beyond.

 passed through the shielded region, are bunched up and largely bypassed through the \( \mu \)-metal walls of the MSR. Thus, in the region enclosed on all sides by the \( \mu \)-metal walls, there is a considerable reduction (typically by a factor of 1000) in the density of magnetic lines of force. Magnetic shielding capability of a MSR may be quantified in terms of a Shielding Factor \( (S) \), which is taken to be the
ratio of the external field \( (H_A) \) outside the MSR to the corresponding residual field \( (H_R) \) in the interior of the MSR; since the attenuation is frequency dependent, the Shielding Factor varies with frequency \([20]\). It is usual to express the Shielding Factor of a MSR on a logarithmic scale (dB) as per eq. 2.4:

\[
S = 20\log\left(\frac{H_A}{H_R}\right)
\]  

The bottom panel of fig.2.5 shows the shielding performance of the MSR as a function of frequency for all the three mutually orthogonal components of magnetic field. MSR provides a shielding factor of 70 dB at 1 Hz, which steadily improves to 110 dB at 100 Hz and beyond as seen in fig.2.5(b). MSR is equipped with a pneumatically operated door and four waveguides to route the SQUID cables without noise pick up from the ambient into an adjoining Radio Frequency Shielded Room (RFSR), which houses the electronic instrumentation.

### 2.3.2 SQUID Gradiometers

For coupling the signal of interest into the sensing area of the SQUID, a superconducting flux transformer is conventionally used to enhance the attainable field sensitivity. The superconducting flux transformer typically consists of either a single superconducting pickup loop (magnetometer) or a set of superconducting pick-up loops connected in opposition (gradiometer); it serves to sense the instantaneous net magnetic flux at its location and couple it to the input coil of the SQUID so that the SQUID sensor generates a proportional voltage output. Gradiometer type of pick-up loops have emerged as the preferred flux transformer design inside most MSRs which provide only a moderate level of attenuation of ambient magnetic noise (40 dB to 80 dB) and, despite their somewhat lower sensitivity for the detection of deeper sources, offer the advantage of additional attenuation of contributions arising from distant sources of noise.

In the gradiometer configuration, the coil which is closer to the source of interest is known as the pickup coil and the other coils which are away from the source are known as the compensation coils. The first order gradiometer is obtained by separating a pickup coil and a compensation coil by a distance called the baseline of the gradiometer. The gradiometers may be classified as axial or planar depending on the spatial gradient of the magnetic field sensed by them. The axial gra-
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diometer measures the spatial gradient of the magnetic field along the axial direction perpendicular to the pick-up loop ($\partial B_z/\partial z$) and the off-diagonal or planar gradiometers measure the spatial gradient along the two orthogonal directions in the plane of the pick-up loop ($\partial B_z/\partial x, \partial B_z/\partial y$). Such gradiometers are commonly referred to as hardware gradiometers [7, 11], as distinct from synthetic gradiometers [14, 15, 16] which can be constructed electronically [13] or using software [15]. In the synthetic gradiometers, output signals of the lower order gradiometers are readout separately and subtracted. The different types of gradiometer configurations used conventionally in a SQUID based measuring set-up are illustrated in fig. 2.6.

![Figure 2.6: Schematic illustration of different types of pick-up coils used in a SQUID based measuring set-up (adapted from [4]).](image)

(a) to (e) show the different types of pick-up coils such as the magnetometer, first order axial gradiometer, second order axial gradiometer, planar gradiometer in the x-direction and planar gradiometer in the y-direction respectively. The magnetometer (a) measures the total magnetic field produced by both the signal of interest and any nearby as well as distant sources of magnetic noise. The gradiometers (b,c,d and e) measure the spatial gradient of the magnetic field ($\partial B_z/\partial z, \partial^2 B_z/\partial z^2, \partial B_z/\partial x, \partial B_z/\partial y$ respectively for (b)-(e)), and are relatively insensitive to distant sources of magnetic noise.

A first order axial gradiometer (used in our MCG system) consists of two equal radius loops of superconducting wire separated along the common axis by a distance $b$ (baseline of the gradiometer, which is typically 5 cm in our system) and wound in opposition (one loop is clockwise while the other loop is anti-clockwise). Since the gradiometer basically senses the difference in magnetic flux
threading through the two oppositely wound pick-up loops, the output of an ideally balanced gradiometer is zero if the applied magnetic field is uniform over the region occupied by the gradiometer. The variation of attenuation with the source distance $R$ is governed by the parameter $(b/R)^2$ for a first order axial gradiometer. The immunity of the gradiometer to the external sources of ambient magnetic noise can be further improved by using higher order gradiometers which comprise of two or more lower order gradiometers connected in opposition (which are designed to sense higher order spatial derivatives of the magnetic field); indeed using second or higher order gradiometers, some research groups have succeeded in measurement of biomagnetic fields in an unshielded environment, although the quality of the measured data is relatively poor.

2.4 MCG system modules

As some of the modules of MCG system such as SQUID gradiometers, MSR, and FLL electronics have been already discussed in the earlier sections, we now describe the remaining modules of the MCG system in the subsequent sections.

2.4.1 MCG cryostat and the SQUID insert

The SQUID sensors have to be operated at a temperature well below the superconducting transition temperature of the superconducting materials used for their fabrication. Niobium has a superconducting transition temperature of $\sim$9 K, and hence, LTSC SQUID sensors based on niobium (that we use) are usually operated by immersing them in liquid helium, which has a boiling point of 4.2 K at the normal pressure of 1 atm., and, hence, provides a stable thermal environment for the operation of the SQUID sensor. For storing cryogenic liquids such as liquid helium or liquid nitrogen, it is necessary to use a cryostat or a dewar, which is specially designed to reduce heat leaks associated with thermal conduction, convection and radiation and achieve a reasonably low boil-off rate and consequently a longer holding time [5, 16, 20].

The MCG measurements are carried out inside a MSR housing the liquid helium cryostat in which SQUIDs are immersed in liquid helium in order to operate them at the temperature of 4.2 K. The cryostat is supported on a non-magnetic gantry, which allows the cryostat to be moved...
vertically, along the z-axis, in order to adjust its position above the thorax of the subject, depending on the experimental needs. As mentioned earlier in the section 2.1, the MCG system at IGCAR was progressively upgraded from a single channel system developed initially to the 37 channels system presently operational. Each SQUID channel consists of an axial gradiometer connected to a DC SQUID, a flux-locked-loop module, an Analog-to-Digital Converter (ADC) module to digitize the data, which is eventually displayed on a computer as the channel output as a function of time. While the individual modules (or units) of the MCG system are identical for MCG systems with different number of channels, there are variations in the overall size of the fiber reinforced plastic (FRP) cryostat, the number of SQUID gradiometers, inter-sensor spacing, the number of channels in the data acquisition system hardware etc. The increase in size of the cryostat is necessary to accommodate a larger number of sensors with a view to increase the area covered by the sensors over the surface of the thorax. Table 2.1 shows the features of the different MCG systems developed during the course of this thesis; in each system, different cryostats and sensor configurations were used. The photographs of the 4 channel and 13 channel MCG systems have been respectively shown in fig.2.7 and fig.2.8(a), while a dimensioned sketch of the 37 channel cryostat is shown in fig.2.9(a).

### Table 2.1: The different MCG systems which were developed during the course of this thesis work.

<table>
<thead>
<tr>
<th>No. of channels</th>
<th>Liquid helium capacity (litres)</th>
<th>Warm-to-cold distance (mm)</th>
<th>Area covered on thorax ((cm^2))</th>
<th>LHe boil-off rate (litres/day)</th>
<th>Geometry of sensor array</th>
<th>Inter-Sensor spacing (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>11.5</td>
<td>10</td>
<td>40</td>
<td>2</td>
<td>square</td>
<td>42</td>
</tr>
<tr>
<td>13</td>
<td>13</td>
<td>10</td>
<td>106</td>
<td>3</td>
<td>hexagonal</td>
<td>28</td>
</tr>
<tr>
<td>37</td>
<td>18</td>
<td>16</td>
<td>300</td>
<td>5</td>
<td>hexagonal</td>
<td>30</td>
</tr>
</tbody>
</table>

The single channel and four channel MCG systems were assembled in a 11.5 litre capacity liquid helium cryostat, while the 13 channel system and the 37 channel system were assembled in liquid helium cryostats having a capacity to hold 13 and 18 litres of liquid helium respectively. The
Development of SQUID based MCG system

Figure 2.7: The photograph of the 4 channel MCG system developed during the course of this thesis. The inset in the photograph shows the SQUID holder in which the first order axial gradiometers are housed in a square lattice with an inter-sensor spacing of 42 mm.

Figure 2.8: (a) The photograph of the 13 channel MCG system developed during the course of this thesis. The inset in the photograph shows the SQUID holder in which the first order SQUID axial gradiometers are arranged in a hexagonal lattice with an inter-sensor spacing of 28 mm. (b) The photograph of the top view of the 13 channel cryostat showing the recesses (for locating the sensors) provided on the bottom plate of the liquid helium vessel of the cryostat, which serve to reduce the warm-to-cold distance.

The holding time of liquid helium depends on the evaporation rate and the latter is listed for MCG systems with different number of channels. The geometrical configuration of the sensors and the
inter-sensor spacing which eventually determine the coverage area are also different for the different MCG systems.

![Diagram of cryostat](image)

**Figure 2.9:** (a) A dimensioned schematic sketch of the 37 channel MCG cryostat. The portion enclosed by the oval at the bottom of (a) is enlarged at right to show the warm-to-cold distance which is about 16 mm. (b) The sketch of the top view of the 37 channel cryostat showing the recesses (for locating the sensors) provided on the bottom plate of the liquid helium vessel of the cryostat, which serve to reduce the warm-to-cold distance.

It is important for a MCG cryostat to have the warm-to-cold distance as low as possible so that the pick-up coil of the sensors at liquid helium temperatures could be brought as close as possible to the skin surface of the subject, which has to be at room temperature. To ensure this, each cryostat is provided with recesses on the bottom plate of the liquid helium vessel at the locations of the sensors so that each sensor would be actually fitted to sit inside the appropriate recess so that the respective pick-up coil is as close as possible to skin surface. Fig.2.8(b) and 2.9(b) respectively show the photograph and sketch of the top view of the 13 channel and 37 channel cryostats. The recesses can be seen at the bottom flange of the liquid helium vessel of the cryostat which serve to reduce the warm-to-cold distance. Since the magnetic field decreases rapidly as one moves away from the source, it is important that the sensors are as close to the source as possible so that the measurements are performed with a relatively high signal-to-noise ratio. Reduction of the standoff distance between the warm and cold surfaces of the cryostat to values \(\sim 10\) mm leads to an
improved signal-to-noise ratio [19]. The warm-to-cold distance is about 10 mm for the four and thirteen channel cryostats whereas, for the 37 channel cryostat, it is about 16 mm.

With cryostats having a lower number of channels, it was necessary to perform sequential measurements in multiple configurations of sensor positions over the thorax in order to cover the thoracic area of interest for MCG measurements with consequent uncertainties in sensor positions as well as the considerably long time required to complete a typical MCG scan; when the 37 channel system became operational, it was possible to record the MCG signals simultaneously over an array of 37 points arranged on a hexagonal lattice covering a much larger area (300 cm$^2$) on the anterior surface of the thorax, and the need to perform sequential measurements was obviated, making the MCG scans relatively faster. With the 37 channel cryostat, sensor positions were also accurately known, leading to a much smaller uncertainty in the coordinates of the sensors during the MCG measurement, thereby enhancing the reproducibility [21] of the MCG data recorded during different experimental runs; consequently, the error in the reconstruction of the source from the measured MCG data is also considerably lower [22].

A suitable insert was designed for each cryostat along with a SQUID holder on which the SQUID modules, comprising of the wire wound axial gradiometer coupled to a SQUID sensor, were mounted. The SQUID insert fabricated from the fiber-reinforced plastic (FRP) material, which goes inside the cryostat, is equipped with a set of mounting plates to support the SQUID gradiometers at the bottom and the insulated LEMO electrical connectors (which are used to connect the SQUID electrical leads) at the top [19]. Each axial gradiometer consisted of two superconducting loops of 15 mm diameter separated by a baseline of 50 mm and was connected via superconducting contacts to the input coil which was integrated on-chip. The insert is also equipped with suitable radiation baffles in the neck region, to reduce the boil-off of liquid helium as a consequence of radiative heat leak. Fig.2.10 shows the photographs of the insert. The SQUID insert and the SQUID holder are designed in such a way that as the insert is carefully lowered into the cryostat, each gradiometer sits snugly into the corresponding 10 mm deep recess specially provided on the bottom flange of the liquid helium vessel in order to bring the pick up loop of the gradiometer as close to the source as possible.
2.4.2 SQUID electrical leads

An Electro-Static Discharge (ESD) safe work bench and ESD safe work-practices were used during the assembly of the SQUID sensors onto the SQUID insert, to avoid the risk of any possible damage to the tunnel barriers of the SQUID sensor by electrostatic discharge during mounting of the sensors and soldering of the electrical leads. Each SQUID module was wired with four twisted pairs of electrical leads, one pair each for bias current, flux modulation, SQUID output voltage and heater. The use of twisted pairs of electrical leads offers substantial immunity against the possible contamination of the signal by inductively coupled noise. The 40 SWG low resistive (\(\sim 1 \, \Omega m^{-1}\)) copper wires were used for voltage and bias leads of the SQUID, while 40 SWG high resistive (\(\sim 35 \, \Omega m^{-1}\)) manganin wires were used for the modulation and heater leads. The use of low resistive copper wires for the SQUID output voltage reduces the possibility of any signal drop across the electrical leads [23], especially when the relatively low output impedance of the SQUID sensor (\(\sim 1 \, \Omega\)) is transformed using an impedance matching transformer to the values of source impedance at which the preamplifier yields the best noise performance. Use of low thermal conductivity manganin wires [23] for the other electrical leads (modulation and heater) serves to minimize the heat leak into the cryostat and consequently contributes to reducing the boil-off of liquid helium, thereby increasing the hold-time of the cryostat. The heater incorporated in the
sensor is occasionally used in case magnetic flux is accidentally trapped in the form of vortices in the vicinity of the Josephson junctions constituting the SQUID sensor and consequently, the critical current of the SQUID as well as the modulation depth of the SQUID are reduced. In such a case, the heater may be activated for a few seconds to pass a current to raise the temperature of the SQUID sensor above the superconducting transition temperature $T_c$, in order to detrap the trapped flux and restore the critical current as well as the modulation depth to their original values, in the cool down of the SQUID on switching off the heater. The SQUID output voltage signals from all the channels, which are first routed to the intermediate electrical connectors on the mounting plate of the insert, are eventually brought out of the cryostat to the electrical connectors at the top of the insert by means of twisted pairs of electrical leads [2]. The battery powered modules comprising of preamplifier and the FLL electronics are located close to the top of the cryostat.

2.4.3 Radio Frequency Shielded Room (RFSR)

The linearized FLL output voltages from all the SQUID channels are routed via shielded cables through the waveguides connecting the MSR to an adjoining Radio Frequency Shielded Room (RFSR) [19], where necessary electronic instrumentation for analog-to-digital conversion (ADC) and data acquisition is located. The RFSR houses all the electronic instrumentation requiring the Mains Power such as the control units for the FLL, ADC modules to digitize the SQUID output voltage signal in each channel, function generators, oscilloscope, spectrum analyzer and other test and measuring instrumentation required during an experiment. Fig.2.11(a) shows a photograph of the RFSR and shows how it is connected to the MSR via four 100 mm diameter waveguides, while fig.2.11(b) shows a photograph of the electronic instrumentation housed inside the RFSR. The RFSR is constructed using 2 mm thick aluminum sheets of high electrical conductivity and provides a shielding factor of 100 dB at frequencies of 1 MHz and beyond [19].

The digitized voltage output data from all the SQUID channels is transmitted outside the RFSR over a fiber optic cable to a data acquisition system PC (DAQ-PC) located in the unshielded area through a port provided for this purpose on the rear wall of the RFSR. The fiber optic cable supports data transfer rates upto 75 Mbps and provides immunity against noise during the transmission of data to PC. The RFSR is equipped with ten signal line feedthrough filters, which
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Figure 2.11: (a). The photograph of the Radio Frequency Shielded Room (RFSR) coupled to the Magnetically Shielded Room via four waveguides (left panel). The right panel shows a view of the electronic instruments kept inside the RFSR. The RFSR provides high frequency shielding against electromagnetic interference with an attenuation factor of 100 dB at 1 MHz and beyond.

allows the passage of low level, low frequency signals inside RFSR, if desired by the user. All the electronic instruments inside the RFSR derive the Mains Power from a special isolation transformer, located far away from the MSR and RFSR, and the 220V, AC power line enters the RFSR via a filter.

2.4.4 Data Acquisition System

Based on the experimental requirements, the SQUID output voltage signal in each channel, which is proportional to the instantaneous axial gradient $\partial B_z/\partial z$ at the location of the gradiometer, may be low pass filtered with user desired cut-off frequency which may be set at any of the four values: 30 Hz, 100 Hz, 300 Hz and 1 kHz. An individual Delta-Sigma ADC is used for digitizing the SQUID output voltage in each channel with a 24 bit resolution at any user desired sampling rate upto 200 kHz [19]. However it may be noted that for most measurements a sampling rate of 1 kHz is adequate. The digitized data received over the optical fibre link is stored in a server PC, which is equipped with a custom-built data acquisition and display software. The software, based on a LABVIEW platform enables real-time graphical display of the output voltage of each channel.
as a function of time, with modules for low-pass, high-pass or band-pass filtering depending on the needs of the user. Besides allowing the data to be visualized in the time domain, the software enables investigations of the spectral content in the frequency domain and is also equipped with modules required for trigger based epoching and averaging to suppress uncorrelated noise, which are important for studies of evoked response in the context of MEG and for visualizing weak effects such as those associated with the activation of His bundle in Signal Averaged MCG (SAMCG). The channel output data, together with information on the experimental details and the sensor coordinates, is stored in the computer for subsequent off-line analysis.

### 2.5 Field gradient-to-Voltage Calibration for SQUID gradiometer

The SQUID system needs to be properly calibrated, prior to the biomagnetic measurements so that the measured SQUID output voltage may be expressed in terms of magnetic field, or magnetic field gradient using the calibration factor. For calibration, it is customary to relate the output voltage of the magnetometer or gradiometer to the value of the external magnetic field or field gradient produced by a known reference system. In general, the calibration factor is expressed as voltage/field or as voltage/field gradient. It is important that the calibration factor is accurately known in a multichannel biomagnetic system, since even a few percent error in calibration may result in appreciable errors in the source localization [22]. Although several calibration procedures have been reported in the literature [20, 24], we have adopted relatively a simple procedure for calibrating the output voltage of the SQUID sensors, which is found to be effective.

We used a large circular current carrying coil, encircling the tail of the Dewar, whose diameter was 20 times larger than the diameter of the superconducting gradiometer coil, which was used for sensing the magnetic field (or field gradient) and coupling it to the SQUID sensor as a proportional flux. In this arrangement, the coil is aligned to be along the gradiometer axis and its position is adjusted until a maximum voltage from the SQUID is obtained. From Biot-Savart’s law, the vertical component of the magnetic field $B_z$ at a distance $z$ from the centre along the axis of the large circular coil is given by [24],

$$B_z(x = 0, y = 0, z) = \frac{\mu_0 NIa^2}{2(a^2 + z^2)^{3/2}}$$

(2.5)
where, $N$ is the number of turns in the large circular coil, $I$ is the current passed through the coil, $a$ is the radius of the large circular coil and $z$ is the distance between the sensor and the coil along the vertical axis. One can change either the amplitude of the current $I$ by adjusting the level of the sinusoidal voltage in the signal generator or the vertical distance $z$ by adjusting the height of the cryostat vertically to vary the input magnetic field values and the output voltage of the SQUID for the corresponding changes in the magnetic field are recorded. As the sensor used was a first order axial gradiometer, the first order derivative of $B_z$ at the sensor position is evaluated and the observed SQUID output voltage is plotted against the known value of the field gradient $\partial B_z/\partial z$ as shown in fig.2.12. The slope of the calibration plot yields the calibration factor. For instance, we estimated the calibration factor for the first order axial gradiometer, with a loop diameter of 15 mm and a baseline of 50 mm, as 22.2 pT/cm/V.

**Figure 2.12:** The flux-to-voltage calibration plot for a typical SQUID sensor. A known value of magnetic field (or field gradient) is presented to the sensor by passing a known amplitude of sinusoidally varying current through a large circular coil and the amplitude of the flux-locked-loop (FLL) output voltage for the corresponding magnetic field (or field gradient) is recorded. FLL output voltage is recorded for different values of applied magnetic field (or field gradient). The slope of the plot of FLL output as a function of applied magnetic field (or field gradient) is taken to be the calibration factor, which is used for converting the measured SQUID output voltage into the corresponding magnetic field (or field gradient).

### 2.6 Field gradient noise level in the SQUID system

It is imperative that, for a signal of interest to be measurable, its amplitude should be much larger than the sensitivity of the sensor used in the required bandwidth and the latter is limited
by the intrinsic noise generated within the SQUID sensors and the extrinsic noise associated with electronics units, residual ambient magnetic noise, vibrations etc [20]. Hence, the background noise (the environmental noise in the absence of the subject) measured in the shielded room with an axial gradiometer must be lower than the signal of interest and this condition implies that white noise level of the sensor should not be more than several $fT_{\text{rms}}/cm/\sqrt{Hz}$ which represents the typical spectral density of the noise content in unit frequency bandwidth. The minimum noise level to be achieved depends on the amplitude of the signal to be measured. Fig. 2.13(a) illustrates the typical variation of the spectral density of noise with frequency. The output noise of the SQUID comprises of a combination of white instrumental noise and the low frequency environmental noise. The distribution of system white noise for the 37 channel MCG system, above 10 Hz, is illustrated in fig. 2.13(b). The average field gradient noise level measured, above 10 Hz, for the first order axial gradiometer with 15 mm loop diameter and 50 mm baseline was measured to be $\sim 2.2 fT_{\text{rms}}/cm/\sqrt{Hz}$. The field gradient noise measured by the axial gradiometer could be converted

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure2_13.png}
\caption{(a) The spectral density of magnetic field noise for a typical SQUID sensor coupled to a first order axial gradiometer (with a baseline of 50 mm), measured with a low pass filter setting of 300 Hz. The low frequency noise peaks associated with vibration, and power line frequency of 50 Hz and its harmonics are seen. The right panel (b) shows the distribution of white noise level measured above 10 Hz, in the 37 channel SQUID based MCG system; the mean noise level ($\sim 2.2 fT_{\text{rms}}/cm/\sqrt{Hz}$ in field gradient noise or equivalently $\sim 11 fT_{\text{rms}}/\sqrt{Hz}$ in field noise) evaluated across all the channels is indicated by a horizontal line.}
\end{figure}
into an equivalent magnetic field noise by multiplying the field gradient noise by the baseline length of the gradiometer [20]. The average magnetic field noise is thus inferred to be $\sim 11 \, fT_{\text{rms}}/\sqrt{\text{Hz}}$ for our MCG system.

2.7 MCG Measurement

Before MCG recording, the SQUID sensors are tuned to the correct bias current at which the maximum modulation depth is observed, which usually corresponds to the minimum output noise in the Flux Locked Loop (FLL) mode. The subject is instructed to remove and deposit all magnetic and metallic objects (including currency notes, which are usually printed with a magnetic ink). The subject is then taken inside the Magnetically Shielded Room (MSR), and is positioned under the cryostat in a supine position by appropriately raising or lowering the cryostat along the vertical axis. It is important to adjust the vertical position of the cryostat in such a way as to minimize the distance between the sensor plane and the subject to realize a high signal-to-noise ratio during the MCG measurements; nevertheless, care should be taken to ensure that the body of the subject does not touch the cryostat and that there is a clear gap between them. After the position of the subject under the cryostat is adjusted to align the central axis of the cryostat with reference to selected anatomical reference points on the thorax of the subject, the pneumatic door of the MSR is closed. The MCG signal is then recorded using the SQUID sensor array and is displayed on the server PC of the data acquisition system located in the unshielded environment. For recording, we typically use a sampling rate of 1 kHz in each channel with a low pass filter setting of 300 Hz and the FLL is usually operated at a gain setting of $5 \, V/\Phi_0$. Fig.2.14(a) shows a photograph of the MCG recording on a human subject in progress inside the MSR using the 37 channel MCG system; fig.2.14(b) shows the recorded MCG on the Data Acquisition Computer.
Figure 2.14: (a). Photograph representing the MCG measurement, using 37 channel MCG system on a human subject, in progress inside the MSR. The position of the subject and the cryostat are adjusted in such a way that the sensors are just above the anterior thoracic surface. The bottom panel (b) shows a screenshot of the recorded MCG on the Data Acquisition Computer. A few channels have been shown on an expanded scale in the ordinate for clarity.
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2.8 MCG signal source analysis

The four channel SQUID system developed initially was used to record the magnetocardiograms on the subject’s chest sequentially at a total of 36 locations on a 6 \times 6 square lattice covering an area of 21 cm \times 21 cm as indicated in fig.2.15. The spatial distance between the adjacent SQUID sensors was 4.2 cm. Fig.2.15 also shows the grid pattern of the measurement locations superimposed on a representation of the chest surface to indicate the measurement positions relative to the heart, which is schematically shown as inlay. The signal at each location was preprocessed (removing breathing artifact and 50 Hz power line frequency) and averaged over a hundred cardiac cycles to improve the signal to noise ratio. The breathing artifact has been removed by using wavelet toolbox in MATLAB. The Daubechies (db10) wavelet was used with an appropriate user selectable level of decomposition to identify and eliminate the low frequency components such as breathing artifact (about 15 to 20 cycles/min) and baseline drift. The signal reconstructed was free of breathing artifact and baseline drift. Fig.2.16(a) shows the signal averaged MCG at the 36 locations and fig.2.16(b) shows the corresponding magnetic field map, which is the iso-field gradient contour map (MFM) constructed from the measured magnetic field gradient distribution at the instant of the

Figure 2.15: The two dimensional thoracic surface of a subject showing the location of the grid pattern (the jugular notch over the sternum provided the reference) used for the MCG measurement as well as other anatomical landmarks relative to the heart. The arrows represent the orientation of the respective dipoles obtained from a solution of the inverse problem at the instants of the R and T peaks.
R-peak of the ventricular depolarization, for a typical subject. From the MFM, our objective is to reconstruct the source parameters, which could account for the observed electrical activity of the heart, through a solution of the inverse problem. There are many algorithms (nonlinear least square optimization [10], MUSIC [25], simulated annealing [26], genetic algorithms [27], etc.) and cardiac source models (equivalent current dipole, multipole expansion model [28], etc.) which have been used by researchers to solve the inverse problem in the context of MCG. Here we have adopted the equivalent current dipole model (ECD) to represent the cardiac source and the inverse problem is solved by means of iterative nonlinear least square optimization described in chapter 4. The results and discussion of the data analysis are also presented in the fourth chapter.
As already noted, the use of a four channel system for experimental measurements imposes limitations on the possible accuracies in the positioning of the sensors exactly at the nodal points of the grid during the measurement of the MCG at each of the nine configurations. This problem could be circumvented if simultaneous measurements were made using a system with a larger number of channels and covering a larger area of the thorax. In view of this, subsequent studies have been carried out with 37 channel MCG system and also the advanced signal pre-processing methods such as ensemble empirical mode decomposition method (described in chapter 3) has been adopted for denoising since the conventional denoising methods like filtering, averaging, wavelet based denoising, etc. have some drawbacks which are discussed in the third chapter.
Bibliography


