BIOMAGNETISM

Biomagnetism describes the electromagnetic and magnetic phenomena that arise in biological tissues. These phenomena include

- Magnetic field at and beyond the body
- Response of excitable cells to magnetic field stimulation
- Intrinsic magnetic properties of the tissue

The magnetic field is generated either by the bioelectric currents or by magnetic material in the body. Similarly, by feeding magnetic energy to the body it induces electric currents that stimulate electrically excitable cells or it magnetizes magnetic material in the body. Both in measurement and in stimulation, the bioelectric currents in the body are the main subject of interest of the scientific work in biomagnetism.

The biomagnetic fields and the bioelectric currents are, of course, directly connected through Maxwell's equations. Therefore, to get new information from bioelectric sources by a magnetic method, the measurement sensitivity of the magnetic measurement must have a distribution different from that of an electric measurement. For more accurate information of the source distribution, the magnetic measurement must concentrate its measurement sensitivity on a smaller region than the electric measurement.

It is not self-evident that these two requirements are met by a biomagnetic measurement system. These questions are discussed in more detail here.

In addition to the theoretical requirements, there are technical reasons for using biomagnetic methods. They follow from the different technology used in magnetic detection. First, the magnetic field detector does not contact the body surface. Secondly, because superconducting technology is used, the magnetic detector is capable of measuring dc currents.

THEORY, BIOELECTROMAGNETIC BACKGROUND

Sources of Bioelectric Currents

Let us introduce the concept of the *impressed current density* $J^i(x, y, z, t)$. This is a nonconservative current that arises from the bioelectric activity of nerve and muscle cells due to the conversion of energy from chemical to electric form. The individual elements of this bioelectric source behave as electric current dipoles. Hence the impressed current density equals the *volume dipole moment density* of the source. Note that J^i is zero everywhere outside the region of active cells (1).

If the volume conductor is infinite and homogeneous and the conductivity is σ, the primary sources J^i establish an electric field E and a conduction current σE. As a result, the total current density J (2) is given by Eq. (1):

$$J = J^i + \sigma E \tag{1}$$

The quantity σE is called the *return current.* This current is necessary to avoid buildup of charges due to the source current.

Because the electric field E is *quasistatic,* it can be expressed at each instant of time as the negative gradient of a scalar potential Φ, and Eq. (1) may be rewritten as

$$J = J^i - \sigma \nabla \Phi \tag{2}$$

Because tissue capacitance is negligible (quasistatic conditions), charges redistribute themselves in a negligibly short time in response to any source change. Because the divergence of J evaluates the rate of change of the charge density with respect to time and because the charge density must be zero, the divergence of J is necessarily zero. (We refer to the total current J as being solenoidal, or forming closed lines of

current flow.) Therefore, Eq. (1) reduces to Poisson's equation:

$$\nabla \cdot \boldsymbol{J}^{\mathrm{i}} = \nabla \cdot \sigma \nabla \Phi + \nabla \cdot \boldsymbol{J} = \sigma \nabla^2 \Phi \tag{3}$$

The solution of Eq. (3) for the scalar function $\sigma\Phi$ for a region that is uniform and infinite in extent (3) is

$$4\pi\sigma\Phi = -\int_V \left(\frac{1}{r}\right) \nabla \cdot \boldsymbol{J}^{\mathrm{i}}\, dv \tag{4}$$

Because a source element $-\nabla \cdot \boldsymbol{J}^{\mathrm{i}}\, dv$ in Eq. (4) behaves like a *point source* in that it sets up a field that varies as $1/r$, the expression $-\nabla \cdot \boldsymbol{J}^{\mathrm{i}}$ is defined as a *flow source density* I_{F}. Because we seek the solution for field points outside the region occupied by the volume source, Eq. (4) may be transformed (3) to

$$4\pi\sigma\Phi = \int_v \boldsymbol{J}^{\mathrm{i}} \cdot \nabla \left(\frac{1}{r}\right) dv \tag{5}$$

This equation represents the distribution of potential Φ due to the bioelectric source $\boldsymbol{J}^{\mathrm{i}}$ within an infinite, homogeneous, volume conductor that has conductivity σ. Here $\boldsymbol{J}^{\mathrm{i}}\, dv$ behaves like a *dipole element* (with a field that varies as its dot product with $\nabla(1/r)$, and hence $\boldsymbol{J}^{\mathrm{i}}$ can be interpreted as a *volume dipole density*).

By using Green's theorem (4), Geselowitz (2) developed Eq. (6) which evaluates the electric potential anywhere within an inhomogeneous volume conductor containing internal volume sources:

$$4\pi\sigma\Phi(r) = \int_v \boldsymbol{J}^{\mathrm{i}} \cdot \nabla \left(\frac{1}{r}\right) dv + \sum_j \int_{s_j} (\sigma_j'' - \sigma_j')\Phi \nabla \left(\frac{1}{r}\right) \cdot dS_j \tag{6}$$

The current density $\boldsymbol{J}$ throughout a volume conductor gives rise to a magnetic field given by the following relationship (3,5);

$$4\pi H = \int_v \boldsymbol{J} \times \nabla \left(\frac{1}{r}\right) dv \tag{7}$$

where r is the distance from an external field point at which H is evaluated on an element of volume dv inside the body, $\boldsymbol{J}dv$ is a source element, and ∇ is an operator with respect to the source coordinates. Substituting Eq. (2) in Eq. (7) and dividing the inhomogeneous volume conductor into homogeneous regions with surfaces S_j,

$$4\pi H = \int_v \boldsymbol{J}^{\mathrm{i}} \times \nabla \left(\frac{1}{r}\right) dv - \sum_j \int_{v_j} \sigma_j \nabla\Phi \times \nabla \left(\frac{1}{r}\right) dv \tag{8}$$

Again using Green's theorem and making some vector manipulations, we obtain Eq. (9)

$$4\pi H(r) = \int_v \boldsymbol{J}^{\mathrm{i}} \times \nabla \left(\frac{1}{r}\right) dv + \sum_j \int_{s_j} (\sigma_j'' - \sigma_j')\Phi \nabla \left(\frac{1}{r}\right) \times dS_j \tag{9}$$

This equation describes the magnetic field outside a finite volume conductor containing internal (electric) volume sources $\boldsymbol{J}^{\mathrm{i}}$ and inhomogeneities $(\sigma_j'' - \sigma_j')$. It was first derived by Geselowitz (6).

It is important to note that the first term on the right-hand side of Eq. (9), involving $\boldsymbol{J}^{\mathrm{i}}$, represents the contribution of the volume source, and the second term is the effect of the boundaries and inhomogeneities. The impressed source $\boldsymbol{J}^{\mathrm{i}}$ arises from cellular activity and hence has diagnostic value, whereas the second term is a distortion due to the inhomogeneities of the volume conductor. These same sources were identified earlier when the electric field generated by them was being evaluated by Eq. (6). (Just, as in the electric case, these terms are also called primary and secondary sources.)

Similarly, as discussed in connection with Eq. (6), it is easy to recognize that if the volume conductor is homogeneous, the difference $(\sigma_j'' - \sigma_j')$ in the second expression is zero, and it drops out. Then the equation reduces to the equation of the magnetic field due to the distribution of a volume source in a *homogeneous* volume conductor.

Nature of Biomagnetic Sources

Equation (9) shows that the physiological phenomenon that is the source of a biomagnetic signal is the *electric* activity of the tissue $\boldsymbol{J}^{\mathrm{i}}$. Thus, for instance, the source for the magnetocardiogram (MCG) or magnetoencephalogram (MEG) is the *electric* activity of the cardiac muscle or nerve cells, respectively, as it is the source of the electrocardiogram (ECG) and electroencephalogram (EEG).

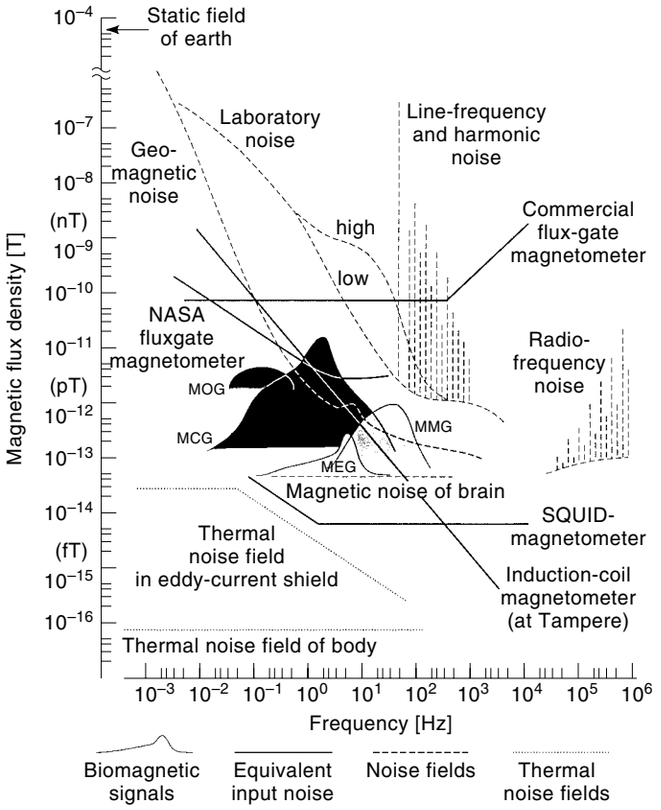
The difference between biomagnetic and bioelectric signals is seen from the form of their mathematical equations. When comparing Eqs. (6) and (9), one can note that the magnetic field arises from the curl and the electric field from the divergence of the source. This distinction holds for the first component on the right-hand side of these equations that arises from the distribution of impressed current, and for the second component that arises from the boundaries of the inhomogeneities of the volume source.

In the design of magnetic leads, one must keep in mind the electric origin of the magnetic signal and the characteristic form of the sensitivity distribution of the magnetic measurement. If the lead of a magnetic measurement is not carefully designed, it is possible that the sensitivity distribution of a magnetic lead will be similar to that of another electric lead. In such a case the magnetic measurement does not provide any new information about the source.

Note that the biomagnetic signal previously discussed is not assumed to arise from magnetic material because such material does not exist in these tissues. There are special circumstances, however, where biomagnetic fields are produced by magnetic materials, for example, in the case of the signal due to the magnetic material that contaminates the lungs of welders or the iron accumulated in the liver in certain diseases. Such fields are not discussed in this article.

Instrumentation

Biomagnetic fields have very low amplitude compared with the ambient noise fields and to the sensitivity of the detectors. A summary of these fields is presented in Fig. 1 (7). The figure indicates that it is possible to detect the MCG with induction coil magnetometers, albeit with a reasonably poor signal-to-noise ratio. However, even the most sensitive induction coil magnetometer built for biomagnetic purposes (8) is not sensitive enough to detect the MEG for clinical use. The Superconducting QUantum Interference Device (SQUID) is the only



Biomagnetic signals
MCG = magnetocardiogram
MMG = magnetomyogram
MEG = magnetoencephalogram
MOG = magneto-oculogram

Noise fields
Static field of the earth
Geomagnetic fluctuations
Laboratory noise
Line frequency noise
Radio frequency noise

Equivalent input noise
Commercial flux-gate magnetometer
Ring-core flux-gate (NASA)
Induction coil magnetometer
SQUID-magnetometer.

Thermal noise fields
Eddy current shield
The human body

Figure 1. Magnetic signals produced by various sources.

instrument sensitive enough for high-quality biomagnetic measurements. The instrumentation for measuring biomagnetic fields is not discussed further in this article, but a good overview of instrumentation appears in Williamson et al. (9).

LEAD-FIELD THEORY

Concept of Lead Field

The bioelectromagnetic differences between bioelectric and biomagnetic signals may be explained by the different sensitivity distributions of electric and magnetic measurement methods (10). The lead field is an electric current field in the volume conductor generated by feeding a unit current to the lead. (Because the volume conductor is not superconducting, in magnetic leads an alternating current that has a unit time derivative is used.) According to Helmholtz's reciprocity theorem (11), *the current field produced in this manner in the volume conductor is identical to the distribution of the sensitivity of the lead.*

Differences in the Information Content of the Bioelectric and Biomagnetic Recordings

The initially optimistic view of the new information content of magnetic recordings was based on consideration of Helmholtz's theorem which states that: *"A general vector field, which vanishes at infinity, can be represented as the sum of two independent vector fields; one that is irrotational (zero curl) and another which is solenoidal (zero divergence)."* These vector fields are often called the *flow source* and the *vortex source,* respectively.

In an idealized case where the head or the thorax are modeled by concentric conducting spheres, it can be shown that the electric field generated by the bioelectric current sources arises from a flow source and the associated magnetic field from a vortex source. At the beginning of biomagnetic research, it was believed that because of the Helmholtz theorem these two fields are independent and that as much new information could be obtained from magnetic recordings as is already present in electric recordings. However, experimental studies demonstrated that these signals look very much the same and are not fully independent. Malmivuo resolved this apparent contradiction in the following way (10).

The sensitivity of a lead system that detects the dipolar term of the flow source consists of three orthogonal components, each of which is linear and homogeneous (Fig. 2) (10). Orthogonality means that none of them can be obtained as a linear combination of the other two. Thus, the three sensitivity distributions are fully independent. However, the electric signals each lead records cannot be completely independent because each represents a different aspect of the same volume source.

Similarly, the sensitivity distribution of a lead system for detecting the dipolar term of the vortex source also has three orthogonal components. Each component can be represented by a set of concentric circles, so that the lead sensitivity is always tangential to the axis of symmetry. The magnitude of

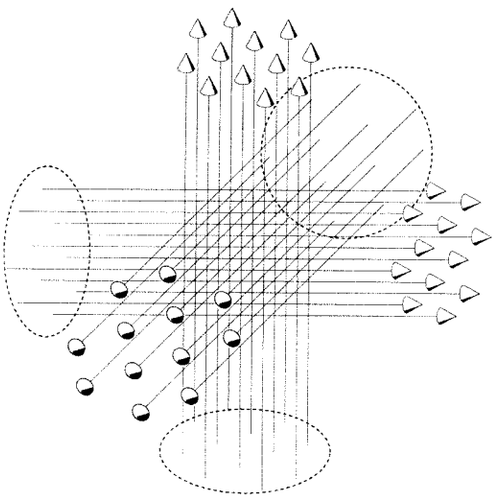


Figure 2. Sensitivity distribution of a detector that detects the electric dipole moment of a bioelectric volume source.

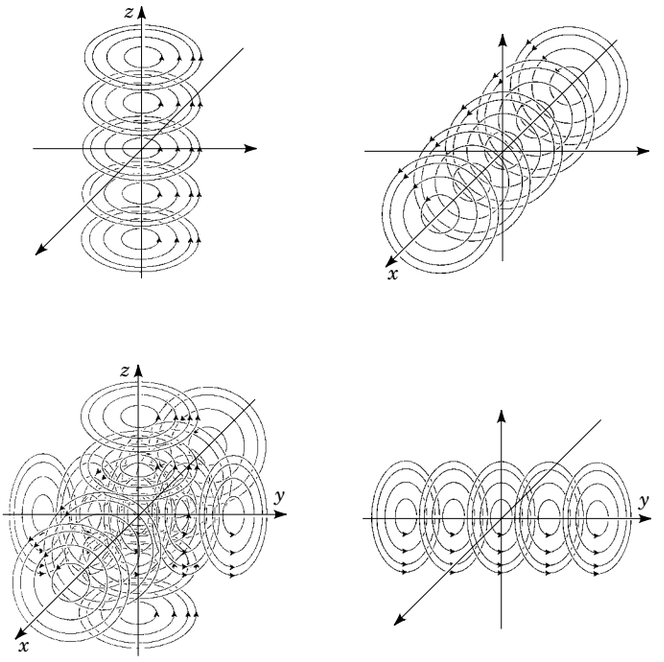


Figure 3. Sensitivity distribution of a detector that detects the magnetic dipole moment of a bioelectric volume source.

the sensitivity is proportional to the radial distance from the symmetry axis (Fig. 3) (10). Again, because the sensitivity distributions of these three components are orthogonal, none of them can be constructed as linear combinations of the other two. Thus, all three magnetic sensitivity distributions are also fully independent. However, as before, the three signals detected by the magnetic leads are not fully independent because each represents a different aspect of the same volume source.

Now it is possible to resolve the paradox involving Helmholtz's theorem. What the Helmholtz theorem expresses is not the independence of electric and magnetic signals, but the independence of the *sensitivity distributions* of the recordings of the flow and vortex sources, that is, the electric and magnetic lead fields. It indicates that the three electric lead fields are orthogonal to the three magnetic lead fields. However, the six signals, measured by the dipolar electric and magnetic leads, cannot be completely independent because they all arise from different aspects of the underlying current source where the activation of the cells is strongly interconnected.

On the other hand, if the sensitivity distributions of two detection methods, regardless of whether they are electric or magnetic, are identical in the source region, the signals and their information contents are also identical.

We may illustrate this principle with a mechanical analog. Though we live in a three-dimensional world, the *movement* of a body is not restricted to only three directions. In addition to the *linear movement* in the directions of the three coordinate axes, a body may also *rotate* around these three coordinate axes. These six directions are mutually independent and are analogous to the sensitivity distributions of the measurements of dipolar electric and magnetic sources.

Ability of a Lead to Concentrate Its Measurement Sensitivity

Although the geometric form of certain electric and magnetic leads might be similar, if one of these had its measurement sensitivity concentrated in a smaller region, that is, if it were capable of measuring a source region with smaller dimensions or of localizing an equivalent dipole with better accuracy, it would be considered superior for brain research.

Localization of a source is not possible with a lead whose sensitivity is homogeneously distributed. Such a lead can be used to determine only the magnitude and orientation of the source. Therefore, the electric and also magnetic leads used for localizing the source have forms different from those described in the previous section.

Application of the Results to Electric and Magnetic Stimulation

Because of reciprocity, the sensitivity distributions of electric and magnetic leads can be directly applied to electric and magnetic stimulation. In that case the sensitivity distributions can be understood as stimulation energy distributions. This is easy to understand because what is done when calculating the lead fields is actually feeding a unit current to the lead which can be thought of as a stimulating current.

In practice the physical dimensions of the coils in magnetic stimulation are much larger than those used in measuring biomagnetic fields. Therefore the results of this article concerning the calculation of magnetic lead fields are not as directly applicable to stimulation problems as are those of the electric lead fields.

EQUIPMENT AND EXPERIMENTS

Magnetocardiography

Selection of the Source Model for MCG. In ECG and MCG the clinical problem is to solve the inverse problem, that is, to find the source of the detected signal so as to obtain information about the anatomy and physiology of the source. Although the actual clinical diagnostic procedure is based on measuring certain parameters, such as time intervals and amplitudes from the detected signal, and actually not to display the components of the source, the selection of the source model is very important from the viewpoint of available information.

In the clinical ECG, the source model is a dipole. This is the model for both the 12-lead ECG and vectorcardiography (VCG). In 12-lead ECG, the volume conductor (thorax) model is not considered, which causes considerable distortion of the leads. In VCG, only the form of the volume conductor is modeled. This decreases the distortion in the lead fields but does not eliminate it completely. Note that today the display systems used in these ECG and VCG systems do not play any role in the diagnostic procedure because the computerized diagnosis is always based on the signals, not on the display.

In selecting the source model for MCG, it is logical, at least initially, to select the magnetic source model on the same theoretical level with the ECG. Only in this way is it possible to compare the diagnostic performance of these methods. It is clear, of course, that if the source model is more accurate, that is, has more independent variables, the diagnostic performance is better. But when comparing ECG and MCG, the comparison is relevant only if their complexity is similar (10).

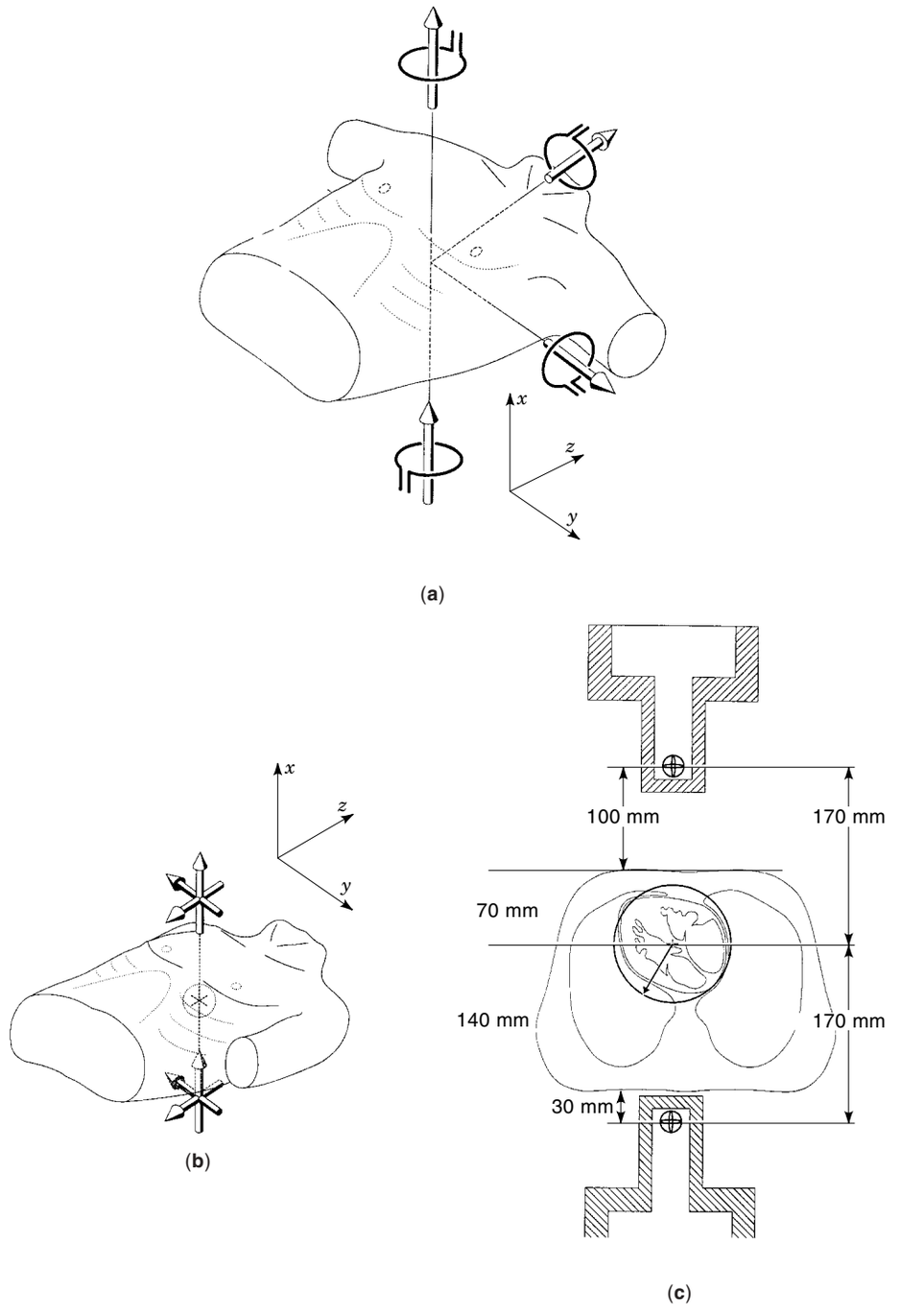


Figure 4. Various methods for detecting the magnetic dipole moment of the heart. (a) The basic principle, the *XYZ*-lead system. (b) Symmetrical *XYZ*-lead system. (c) Symmetrical unipositional lead system.

Detection of the Equivalent Magnetic Dipole of the Heart. The basic method for detecting the equivalent magnetic dipole moment of a volume source is to measure the magnetic field on each coordinate axis in the direction of that axis [Fig. 4(a)]. To idealize the sensitivity distribution throughout the volume source, the measurements must be made at a distance that is large compared with the source dimensions. Of course, this decreases the signal amplitude. The quality of the measurement increases considerably if bipolar measurements are used, that is, measurements are made on both sides of the source [Fig. 4(b)]. Measurement of the magnetic field on each coordinate axis, however, is difficult to perform in MCG because the geometry of the human body. It would require either six sequential measurements with one magnetometer (dewar) or six simultaneous measurements using six dewars.

It has been shown (12) that all three components of the magnetic dipole also can be measured from a single location. By applying this unipositional method symmetrically so that measurements are made on both the anterior and posterior sides of the thorax at the same distance from the heart, only two dewars are needed, and a very high quality of lead fields is obtained [Fig. 4(c)] (10).

Diagnostic Performance of ECG and MCG. The diagnostic performance of ECG as well as MCG was compared in an extensive study made at the Ragnar Granit Institute (13). The

study was made using the asymmetrical unipositional lead system, that is, making measurements only on the anterior side of the thorax. The patients were selected, however, so that myocardial changes were located dominantly on the anterior side.

This study consisted of 290 normal subjects and 259 patients with different myocardial disorders. It was found that the diagnostic performance of ECG as well as MCG is about the same (83%). Then diagnostic parameters were selected from both ECG and MCG. With this combined method, called *electromagnetocardiogram* (EMCG), diagnostic performance of 90% was obtained. This improvement in diagnostic performance was obtained without increasing the number of parameters used in the diagnostic procedure. Moreover, this improvement is significant because it means that the number of incorrectly diagnosed patients was reduced by approximately 50%.

This important result may be explained as follows. The lead system recording the electric dipole moment of the volume source has three independent leads. (This is also the case in the 12-lead ECG system.) Similarly, the lead system detecting the magnetic dipole moment of the volume source has three independent leads. Therefore, the diagnostic performance of these methods is about the same. However, because the sensitivity distributions of electric and magnetic leads are different, the patient groups diagnosed correctly with both methods are not identical.

As stated before, the electric leads are independent of the magnetic leads. If the diagnostic procedure simultaneously uses both the ECG and the MCG leads, we obtain $3 + 3 = 6$ independent leads, and the correctly diagnosed patient groups may be combined. Thus the diagnostic performance of the combined method is better than that of either method alone. This is the first large-scale statistically relevant study of the clinical diagnostic performance of biomagnetism.

Technical Reasons to Use MCG. The technical differences between ECG and MCG include the MCG's far better ability to record static sources, sources on the posterior side of the heart, to monitor the fetal heart, and to perform electrodeless recording. As a technical drawback, it should be mentioned that the MCG instrument costs two to three times more than the ECG. An important feature of MCG is that, unlike the MEG instrument, it does not need a magnetically shielded room. This is very important because a shielded room is very expensive and also limits application of the technique to a specific laboratory space.

Theoretical Reasons to Use MCG. It has been shown that MCG has clinical value and that it can be used alone or in combination with ECG as a new technique called the *electromagnetocardiogram* (EMCG). The diagnostic performance of the combined method is better than that of either ECG or MCG alone. With the combined method, the number of incorrectly diagnosed patients may be reduced by approximately 50%.

Magnetoencephalography

Similarly as in the cardiac application, in magnetic measurement of the electric activity of the brain, the benefits and drawbacks of the MEG can be divided into theoretical and technical categories. First, the theoretical aspects are discussed.

The two main theoretical aspects in favor of MEG are that, because the skull is transparent to magnetic fields, the MEG should be able to concentrate its measurement sensitivity in a smaller region than the EEG, and the sensitivity distributions of these methods are fundamentally different. These issues are discussed in the following: The analysis is made by using the classic spherical head model introduced by Rush and Driscoll (14). In this model, the head is represented by three concentric spheres, where the outer radii of the scalp, skull, and brain are 92, 85, and 80 mm, respectively. The resistivities of the scalp and the brain are 2.22 $\Omega \cdot m$ and that of the skull is 80 times higher, 177 $\Omega \cdot m$.

The two basic magnetometer constructions in use in MEG are axial and planar gradiometers. In the former, both coils are coaxial, and in the latter, they are coplanar. The minimum distance of the coil from the scalp in a superconducting magnetometer is about 20 mm. The coil radius is usually about 10 mm. It has been shown (10) that with this measurement distance, decreasing the coil radius does not change the distribution of the sensitivity in the brain region. In the following the sensitivity distribution of these gradiometer constructions is discussed.

To indicate the magnetometer's ability to concentrate its sensitivity in a small region, the concept of *half-sensitivity volume* has been defined (15). This concept means the region in the source area (brain) where the detector sensitivity is one-half or more of the maximum sensitivity in the source region. The smaller the half-sensitivity volume, the better is the detector's ability to focus its sensitivity in a small region.

In magnetocardiography, it is relevant to detect the magnetic dipole moment of the volume source of the heart and to make the sensitivity distribution within the heart region as independent of the position in the axial direction as possible. In magnetoencephalography, however, the primary purpose is to detect the electric activity of the cortex and to localize the regions of certain activity.

Sensitivity Distribution of the Axial Magnetometer. In a cylindrically symmetrical volume conductor model, the lead field flow lines are concentric circles and do not cut the discontinuity boundaries. Therefore, the sensitivity distribution in the brain area of the spherical model equals that in an infinite, homogeneous volume conductor.

Figure 5 illustrates the sensitivity distribution of an axial magnetometer. The thin solid lines illustrate the lead-field flow lines. The dashed lines join the points where the sensitivity has the same value, the so-called isosensitivity lines. The half-sensitivity volume is represented by the shaded region.

Sensitivity Distribution of the Planar Gradiometer. Figure 6 illustrates the sensitivity distribution of a planar gradiometer. Again, the thin solid lines illustrate the lead-field flow lines, and the dashed lines represent the isosensitivity lines. The half-sensitivity volume is represented by the shaded region. The sensitivity of the planar gradiometer is concentrated under the center of the two coils and is mainly linearly oriented. Further, two zero-sensitivity lines exist.

Half-Sensitivity Volumes of Electro- and Magnetoencephalography. The half-sensitivity volumes for different EEG and MEG leads as a function of electrode distance and gradiometer baselines are shown in Fig. 7(a). The minimum half-sensitiv-

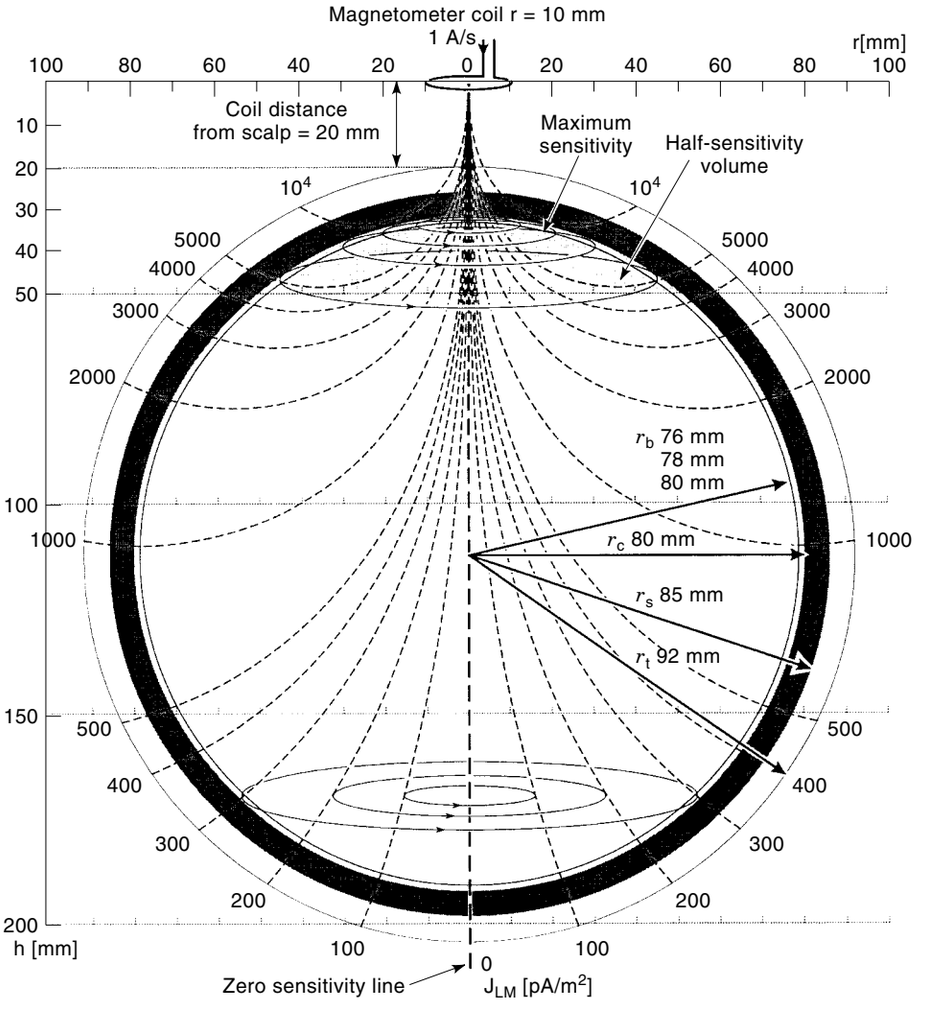


Figure 5. Sensitivity distribution of an axial gradiometer in the inhomogeneous spherical head model.

ity volume, of course, is achieved with the shortest distance/baseline. For three- and two-electrode EEG leads, the half-sensitivity volumes at 1° of electrode distance are 0.2 and 1.2 cm^3, respectively. For 10 mm radius planar and axial gradiometer MEG leads, these volumes at 1° of coil separation (i.e., 1.6 mm baseline for axial gradiometer) are 3.4 and 21.8 cm^3, respectively.

The 20 mm coil distance from scalp and 10 mm coil radii are realistic for the helmet-like whole-head MEG detector. However, MEG devices exist for recording in a limited region where the coil distance and the coil radii are on the order of 1 mm. Therefore the half-sensitivity volumes for planar gradiometers with a 1 mm coil radius at 0 mm to 20 mm recording distances are also illustrated in Fig. 7(b). These curves show that when the recording distance is about 12 mm and the distance/baseline is 1 mm, such a planar gradiometer has about the same half-sensitivity volume as the two-electrode EEG.

Short separation, of course, also decreases the signal amplitude. An optimal value is about 10° of separation. Increasing the separation to 10° increases the EEG and MEG signal amplitudes to approximately 70% to 80% of their maximum values, but the half-sensitivity volumes do not increase considerably from their values at 1° of separation.

Thus, contrary to general belief, the EEG can focus its sensitivity better on a small region in the brain than the whole-head MEG. At about 20° to 30° of separation, the two-electrode EEG lead needs slightly smaller separation to achieve the same half-sensitivity volume as the planar gradiometer. The sensitivity distributions of these leads, however, are similar. Note that if the sensitivity distributions of two electric or magnetic different lead systems, are the same, they detect exactly the same source and produce exactly the same signal. Therefore, the planar gradiometer and two-electrode EEG lead detect similar source distributions.

Sensitivity of EEG and MEG to Radial and Tangential Sources. The three-electrode EEG has its maximum sensitivity under the electrode that forms the terminal alone. This sensitivity is mainly directed radially to the spherical head model. With short electrode distances, the sensitivity of the two-electrode EEG is directed mainly tangentially to the spherical head model. Thus with the EEG it is possible to detect sources in all three orthogonal directions, that is, in the radial and in the two tangential directions relative to the spherical head model.

In the axial gradiometer MEG lead, the sensitivity is directed tangentially to the gradiometer axis of symmetry and

thus also tangentially to the spherical head model. In the planar gradiometer, the sensitivity has its maximum under the center of the coils and is directed mainly linearly and tangentially to the spherical head model. The MEG lead fields are oriented tangentially everywhere to the spherical head model. This may be easily understood by recognizing that the lead field current does not flow through the surface of the head because no electrodes are used. Therefore, the MEG detects only sources oriented in the two tangential directions relative to the spherical head model.

EVALUATION

The biomagnetic measurement technology is and will always be more expensive than the bioelectric technology. Especially this holds on MEG due to its very low signal amplitude. Therefore the biomagnetic measurements must have verified benefits over the bioelectric measurements to be worth to apply.

Magnetocardiography

There exists one study comparing the clinical diagnostic performance of ECG and MCG (13). This study did demonstrate that the diagnostic performance of these methods is about the same. But because the sensitivity distributions of the methods, when applied correctly, are fundamentally different and independent, the patient groups diagnosed correctly with both methods are not identical. Therefore, by combining the methods, we may combine the patient groups and increase the diagnostic performance even so much that the number of incorrectly diagnosed patients may be decreased to half.

Magnetoencephalography

As discussed before:

- The MEG is spatially not more accurate than the EEG.
- The planar gradiometer MEG does not measure a source complementary to the EEG and therefore does not provide information essentially different from that of the EEG.
- From the electric sources the MEG does not measure the component radial to the head. The EEG measures separately all three orthogonal components of the electric sources.
- Though there is no need to fix electrodes with the MEG, the dewar restricts the movement of the patient. At pres-

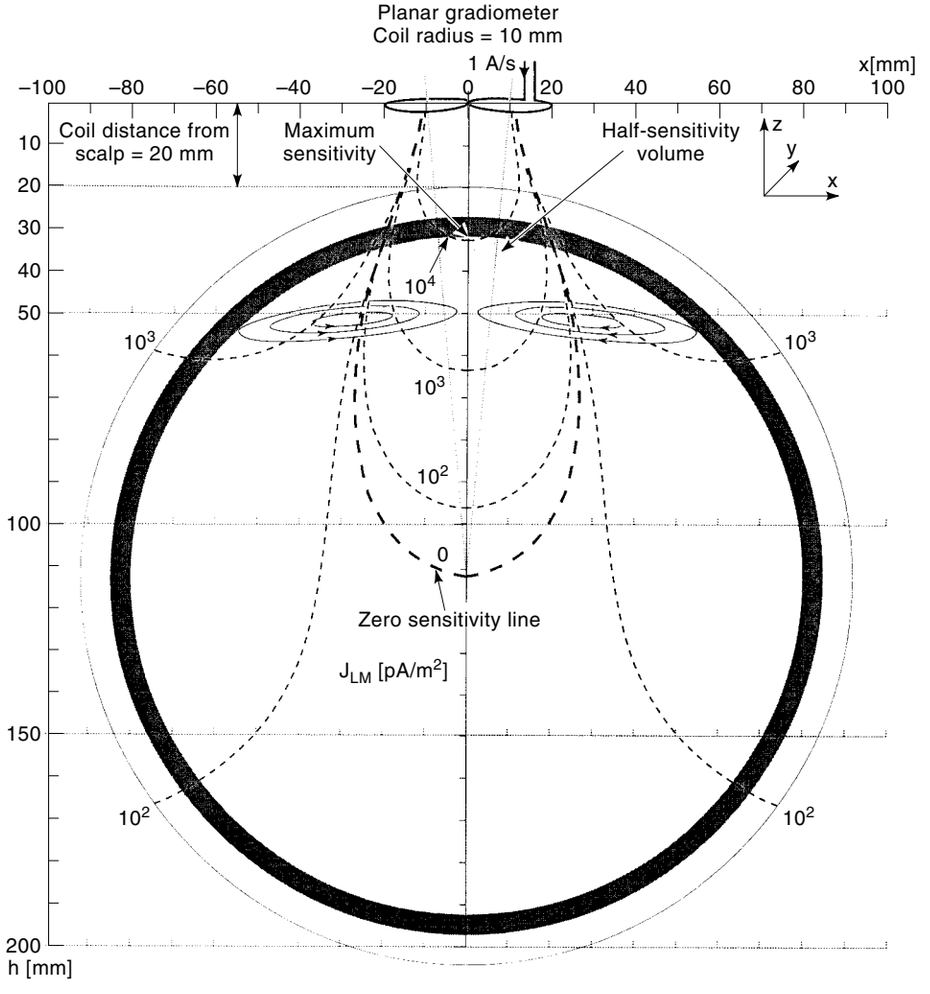


Figure 6. Sensitivity distribution of a planar gradiometer in the inhomogeneous spherical head model.

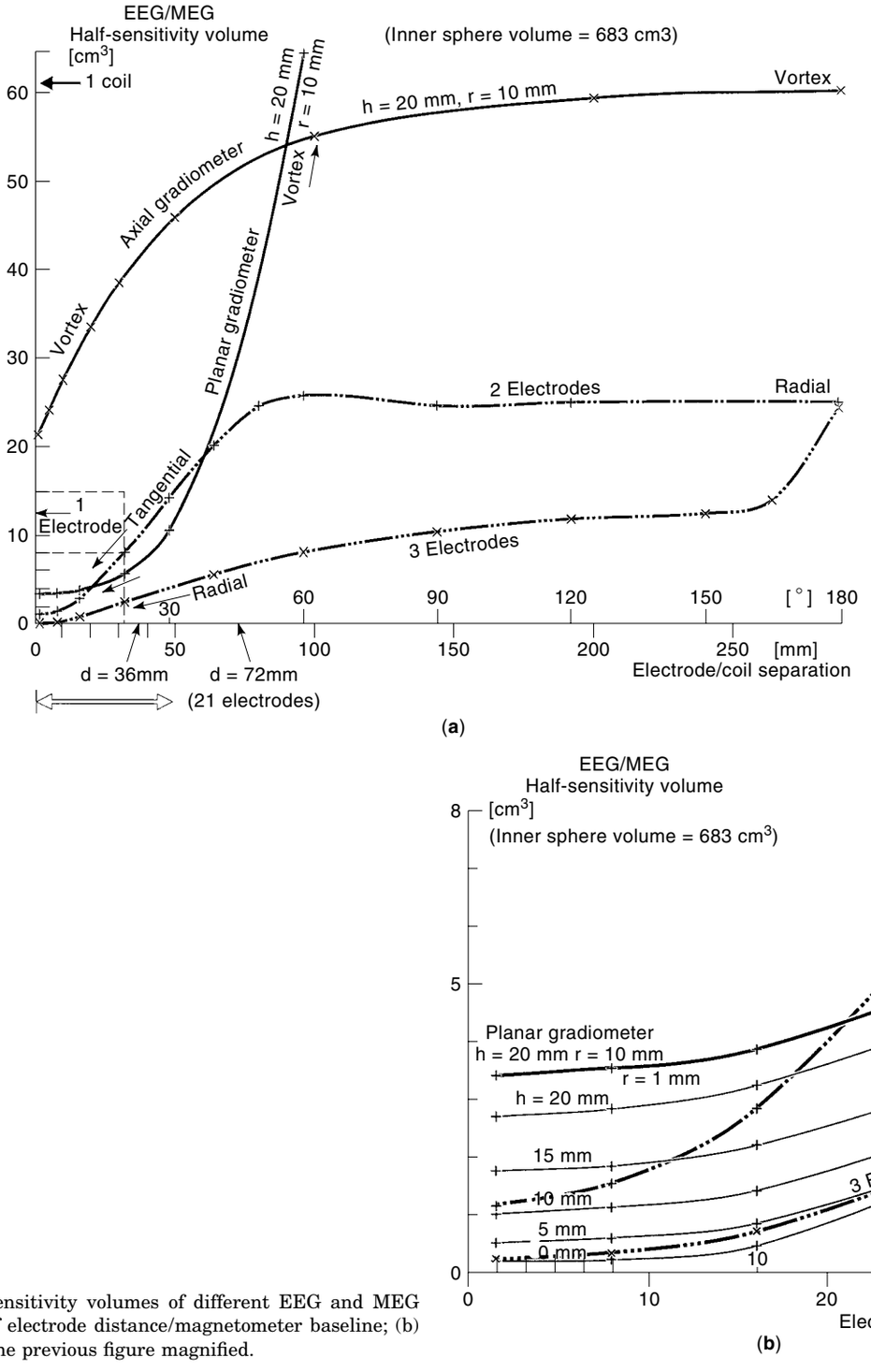


Figure 7. (a) Half-sensitivity volumes of different EEG and MEG leads as a function of electrode distance/magnetometer baseline; (b) Lower left corner of the previous figure magnified.

ent there are available electrode caps which allow fixing of over 100 electrodes to the head within some 10 min.

- The MEG needs, at least at present, a magnetically shielded room whose size due to the size of the MEG dewar exceeds normal laboratory height. This restricts the application of the MEG to certain locations. Instead, the EEG can be recorded at any location making it more easily accessible for patients.

Even though there exist more than 40 MEG installations in the world, there does not exist any clinical study with a relevant number of patients where the diagnostic perfor-

mances of the MEG and EEG were compared. Therefore, at least at the moment, there does not exist any theoretical or clinical evidence on the superiority of the MEG over the EEG which would justify its use.

CONCLUSION

Biomagnetic measurements have been performed for 35 years. SQUID-technology, which is essential for measurement of ultralow intensity biomagnetic fields, has existed for almost 30 years. At present there exist some 40 large-scale installations for MEG research in the world. In addition to those, several smaller groups make biomagnetic studies. The number of active scientists in biomagnetism is several hundreds.

The groups working with MEG have been able to demonstrate that the technology can be used for detecting the electric activity of the brain and for obtaining new important research results in neurology. However, none of these groups has demonstrated that the results have been obtained with the MEG more accurately than with the EEG or are only available with the MEG.

In MCG research most of the research activities are concentrated on localization of the arrhythmogenic foci of the heart or finding indicators for the risk of sudden cardiac death. Neither of these approaches have been more successful than the ECG. The number of researches on utilizing the unique sensitivity distribution of the MCG in relation to the ECG is very limited. However, it is only just this application which has been successful in biomagnetism.

Before biomagnetic methods can be accepted for clinical use there must exist several clinical studies demonstrating the superiority of these methods.

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