9.8. THEORETICAL ASPECTS OF THE ECG–MCG RELATIONSHIP

John P. Wikswo, Jr.

Department of Physics and Astronomy
Vanderbilt University
Nashville, Tennessee, U.S.A.

In this section we examine a topic that has become the subject of much debate. The question at issue is whether the MCG could <u>in principle</u> indicate different aspects of myocardial electrophysiological activity than the ECG. If the MCG contains information about cardiac sources that is not present in the ECG, the MCG might help us gain new insight about normal or abnormal cardiac function.

During the past decade, much of the experimental work in magnetocardiography has been directed towards determining the capabilities and limitations of magnetocardiography relative to electrocardiography. Three types of differences have been explored: frequency response, spatial sensitivity, and information content. Because magnetometers can measure steady or low–frequency bioelectric currents in intact animals more readily than can conventional voltage amplifiers, the MCG has been used to clarify the relation between ST segment elevation and diastolic injury currents during infarction (Cohen and Kaufman 1975), as discussed in Section 9.4. This difference between the MCG and the ECG is one of instrumentation and will not be discussed further in this section.

The difference in spatial sensitivities of the ECG and the MCG arises from fundamental differences in the source–field relationships for the electric and magnetic fields associated with current sources in conducting media. These differences were first described by Baule and McFee (1970) using "lead field" theory (see Section 6.3.2). The sensitivity pattern of an electric or magnetic measurement can be described by the pattern of the corresponding "lead current density," which takes account of non–uniformities in electrical conductivity of the body. Figure 9.8.1 illustrates simple ECG and MCG leads and lead currents. For the ECG, I_{RE} is the current used to reciprocally energize the electric lead and produce the electric "lead current density" J_{LE}. For the MCG, the detection coil (or "magnetic lead") is reciprocally energized by a time–varying current I_{RM} that produces a time–varying lead magnetic field B_R. By Faraday induction, this field produces the "magnetic lead current density" J_{LM}. In measurements of the ECG and MCG, V_E is the output of the electric lead, and V_M is the induced output of the magnetic lead. They can be determined by summing the projection of each element of the primary current density J^P along the corresponding lead current densities. If a SQUID detection coil is used
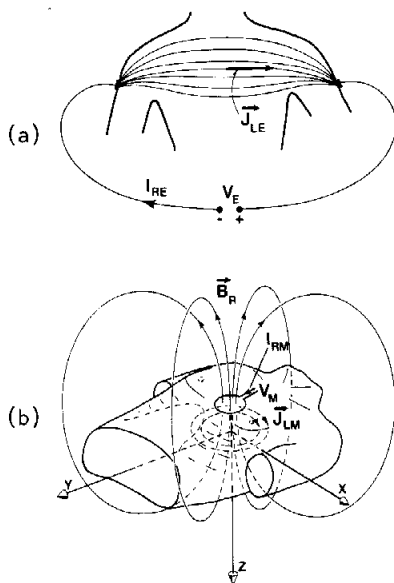
Fig. 9.8.1.   Schematic of (a) ECG lead current density J_{LE} for
              electrodes attached to both arms and (b) MCG lead cur-
              rent density J_{LM} when a single-turn detection coil is
              placed above the chest.

as the magnetic lead, instead of an induction coil, the sensitivity
pattern is unchanged, the only difference being that the magnetic
flux within the coil and not induced voltage is sensed.

     As pointed out by Plonsey (1972) and illustrated in Fig. 9.8.1
(Wikswo et al. 1979), the ECG lead currents must always satisfy La-
place's equation, with the electrodes at the body surface providing
the boundary conditions. For this reason, ECG lead current lines
J_{LE} must begin and end on the body surface (Fig. 9.8.1a). In con-
trast, the MCG lead current lines represent volume currents induced
throughout the conductor by the time-varying lead magnetic field
produced by the detection coil. Thus MCG lead current lines J_{LM}
must satisfy Poisson's equation, and will form closed loops (Fig.
9.8.1b). The transverse nature of the ECG lead current lines, cou-
pled with their distortion by the intracardiac blood mass, causes
the ECG to be more sensitive to radial current flow within the
heart whereas the MCG will be more sensitive to tangential cur-
rents.

     This difference in sensitivity is evidenced by the vector car-
diograms in Fig. 9.8.2. Figure 9.8.2a shows the ECG and MCG for a
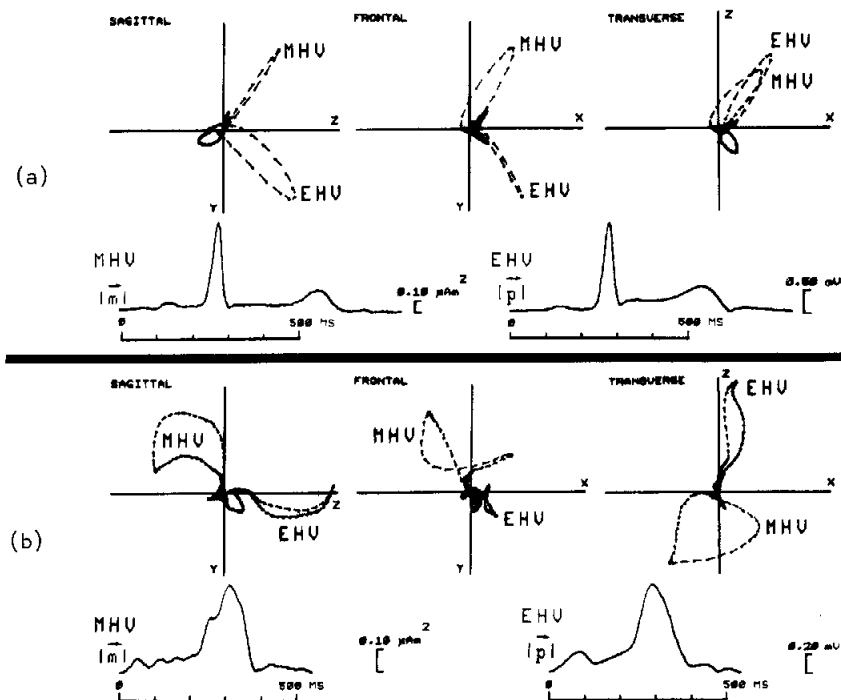
Fig. 9.8.2.   A computer-generated display of the magnetic heart vector (MHV) and electric heart vector (EHV), obtained from (a) a normal subject and (b) a patient with left bundle branch block. The vector magnitudes are also shown (Barry et al. 1977).

normal subject, where activation of both the right and left ventricles proceeds in a predominantly radial direction. Figure 9.8.2b is from a subject with left bundle branch block (LBBB), in which the right ventricle depolarizes in a normal, radial manner while the left ventricle depolarizes in a slow, tangential fashion. The fact that the later part of the QRS complex in the LBBB MCG is larger relative to that in the ECG indicates that the MCG is more sensitive to tangential currents. The differences between the spatial sensitivities of the ECG and MCG may also provide for higher spatial discrimination by the MCG, as suggested by recent measurements of the magnetic fields produced by electrical activity in the bundle of His (Farrell et al. 1980a), discussed in Section 9.5.

The equations that describe these spatial sensitivity differences also suggest the possibility that the ECG and MCG may contain

independent information, but the presence of such information in either the ECG or the MCG has yet to be proven or disproven theoretically or experimentally. The question of the relative information content of the ECG and the MCG has proven difficult to answer completely for three reasons:

1) cardiac activation is complex at both the whole-heart and cellular levels;

2) sophisticated mathematical models would be needed to answer the question;

3) the presence of radial-tangential sensitivity differences described above makes it difficult to quantify the information content of each signal in a manner that allows analytic comparison.

Since the question of relative information content of bioelectric and biomagnetic measurements is an important and unanswered one, the theoretical basis for this question will be reviewed in the following sections.

9.8.1. Comparison of Source-Field Relationships

To explain what we have learned about the MCG-ECG relationship, we must first summarize the theory that governs the electric and magnetic fields associated with cardiac electrical activity. In a very simple model of the heart the electrically active region can be approximated by a cup-shaped wavefront that propagates outward at approximately 1 m/s, as was shown in Fig. 6.1.5. This wavefront serves as the battery that provides a "primary current density" $J^P(r)$ that drives the current in the passive conductor. We shall first assume that the electrical contributions of the wavefront are determined by J^P. Later, we shall use a more formal definition of J^P and specifically identify the contribution of the cellular inhomogeneities to the field produced by the wavefront. In our simple model the total current density at any point is the sum of the primary current density (if the point is within an active region) and the passive current density, which obeys Ohm's law

$$J(r) = - \sigma(r) \nabla V(r) + J^P(r) \tag{9.8.1}$$

The primary current density can be viewed as a current dipole density. As summarized in Section 6.2.4 a pair of equations describe the electric potential and the magnetic field produced by primary current sources in a finite, piecewise homogeneous conductor (Geselowitz 1970):

$$V(\mathbf{r}) = - [4\pi\sigma(\mathbf{r})]^{-1}\{ \int |\mathbf{r} - \mathbf{r}'|^{-1} \nabla'\cdot\mathbf{J}^{p}(\mathbf{r}') \; dv'$$

$$- \sum_{j} \int (\sigma-\sigma') \; V(\mathbf{r}') \; (\nabla'|\mathbf{r} - \mathbf{r}'|)\cdot d\mathbf{S}'\}$$

(9.8.2)

$$B(\mathbf{r}) = (\mu_{o}/4\pi)\{ \int |\mathbf{r} - \mathbf{r}'|^{-1} \nabla'\times\mathbf{J}^{p}(\mathbf{r}') \; dv'$$

$$+ \sum_{j} \int (\sigma-\sigma') \; V(\mathbf{r}') \; (\nabla'|\mathbf{r} - \mathbf{r}'|)\times d\mathbf{S}_{j}'\}$$

(9.8.3)

The summation is over all surfaces separating regions of differing conductivity. In each equation, the first integral is taken over the entire volume containing the primary current – the bioelectric sources – while the second integral is over each distinct surface to account for the effects of inhomogeneities in the conducting medium.

In an infinite homogeneous conductor, the second integrals vanish, so that the electric and magnetic fields are determined only by integrals of the primary current; the integrals need extend only over the region where $\mathbf{J}^{p}$ is non-zero. As discussed in Chapter 6, these equations then provide a simple explanation of the ECG-MCG sensitivity differences described above: the electric potential, recorded as the ECG, is determined by the mathematical divergence of $\mathbf{J}^{p}$; the magnetic field, recorded as the MCG, is determined by the mathematical curl of $\mathbf{J}^{p}$. As demonstrated in model calculations by Cuffin (1978), cardiac current distributions with more "curl," or tangential activation components, may be specified more accurately by magnetic measurements than by electric ones. The possible implications of the divergence-curl differences toward the information content question will be addressed in later sections.

In the case of finite, inhomogeneous conductors, the second integral in each equation may become significant, and in fact may dominate the first term. An ideal current dipole in a conducting sphere is the classic example of how both the conductor's geometry and the sources affect the ECG and the MCG. In an infinite conducting medium, a dipole produces an electric field that decreases as r^{-3} and a magnetic field that falls off as r^{-2}. If instead this dipole is placed inside a conducting sphere, the dipole will produce electric potentials within and on the surface of the sphere. Because of the effects of the spherical boundary, the r^{-3} dependence for the electric field applies only close to the dipole. A magnetic field is also produced inside the sphere, and a magnetic field will exist outside the sphere if the dipole has a component tangential to the surface of the sphere (Grynzspan and Geselowitz 1973; Tripp

1977). Since a radial dipole will produce an electric potential on the surface but no magnetic field outside the sphere, one can state that the magnetic field does not contain all of the information present in the surface potential distribution. Since few biological conductors are perfect spheres, this does not appear to represent a fundamental limitation to magnetocardiography, but it may to magnetoencephalography.

In actual practice, the determination of the electrical potential $V(\mathbf{r})$ using Eq. 9.8.2 for piecewise homogeneous conductors becomes complicated because V appears not only on the left side of the equation but also within the second integral on the right. For this reason, this equation does not have a general analytic solution except in certain simple geometries, and it must be solved using iterative numerical techniques. The appearance of V on both sides of the equality sign leads to conceptual problems which have been addressed by ascribing the effects of the inhomogeneities and boundaries to "secondary sources" that would have to be added to an infinite homogeneous conductor so that the current distribution in the infinite homogeneous conductor would be the same as in the finite inhomogeneous one. It is then possible to compare the field contributions due to the primary sources (the first integral) with those of the secondary sources (the second integral). This type of analysis has proved useful in assessing the relative sensitivities of the ECG and MCG to intracardiac and thoracic inhomogeneities (Horacek 1973; Hosaka et al. 1976; Cohen and MacArthur 1975; Wikswo et al. 1978). It is important to realize that the secondary sources depend upon the primary ones since the secondary source strength is determined by V. Interpretation of the effects of inhomogeneities, either as written in Eqs. 9.8.2 and 9.8.3 or in terms of secondary sources, is complicated by the fact that the boundary geometry affects the electric and magnetic equations differently: the second integral in Eq. 9.8.2 involves a scalar (dot) product of the normal to the surface with the gradient of $|\mathbf{r}-\mathbf{r}'|^{-1}$, while the corresponding magnetic term involves a vector (cross) product of the same two terms.

9.8.2. The Helmholtz Decomposition

We have just seen that the electric potential in an infinite homogeneous conductor is determined solely by the divergence of the primary current, while the magnetic field is determined by its curl. This leads to an interesting, and somewhat controversial, observation about the primary current distribution. If $\mathbf{J}^P(\mathbf{r})$ is an arbitrary vector field, it can be represented by a Helmholtz decomposition (Plonsey 1972)

$$\mathbf{J}^P(\mathbf{r}) = \mathbf{J}_F^{\ P}(\mathbf{r}) + \mathbf{J}_V^{\ P}(\mathbf{r}) \qquad\qquad (9.8.4)$$

where

$$\nabla \times J_F^P(r) = 0 \quad \text{and} \quad \nabla \cdot J_V^P(r) = 0 \qquad (9.8.5)$$

Thus J_F^P has no curl and is called a "flow field," while J_V^P has no divergence and is called a "vortex field." In an infinite homogeneous conductor, represented by only the first terms in Eqs. 9.8.2 and 9.8.3, the electric potential is thereby determined solely by the pattern of J_F^P and the magnetic field only by the pattern of J_V^P; if J_F^P and J_V^{PF} are independent, then V and **B** will likewise be independent. This observation has been the basis for the controversy regarding possible independent information in the MCG. As we will see, the answer to this controversy may lie in the clarification of either the physiological constraints on J^P, since any constraint that prevents J^P from being the most general form of vector field will also cause J_F^P and J_V^P to be related, or in the effects of inhomogeneities at the cellular level.

The question of constraints on J^P can be addressed using a simple variable-counting argument (Wikswo et al. 1979). If the primary current distribution is a general vector field, it will have three degrees of freedom. The Helmholtz decomposition is simply a statement that such a vector field can be described by a scalar potential with one degree of freedom (consistent with J_F^P having no curl), and a vector potential with two degrees of freedom (consistent with J_V^P having no divergence). As first pointed out by Rush (1975), there are physiological constraints on J^P that leave it with less than three degrees of freedom. As a result, the electric and magnetic fields cannot be completely independent.

The question of independence becomes more complicated in a conductor with boundaries or internal inhomogeneities. Then V is still determined solely by J_F^P, but **B** now receives contributions from both J_V^P and J_F^P through the appearance of V in the second integral of Eq. 9.8.3. The magnetic field contribution from J_V^P may still contain independent information consistent with our preceding discussion, but the contribution associated with the inhomogeneities is determined by J_F^P. This would provide a mixing of the information in the MCG and ECG, which could be particularly important in situations where the second term of Eq. 9.8.3 dominates the first. At first glance the second term is governed by V, which is described by J_F^P. However, as we discussed before, the second integral in both equations incorporates the boundary geometry differently and the effect of the geometry on the relative information content of these two integrals is not yet understood. Using the variable counting approach, the secondary source contribution should contain at least two degrees of freedom, since V contains one and the boundary that specifies the orientation of the secondary sources will contain another. How these two or more degrees of freedom couple to the ECG and the MCG is in need of clarification.

The ideal way to demonstrate the independence (or dependence)

of the two integrals on the boundary geometry would be to demonstrate (or to prove the non-existence of) examples in which a particular boundary or internal inhomogeneity affects either the electric field or the magnetic field from a particular source but not both, i.e. an inhomogeneity that would be silent to only one of the two fields. A simple example is a horizontal current dipole in a conducting halfspace beneath a horizontal boundary: the normal component of the magnetic field in the air above the boundary is not affected by the inhomogeneity but the electric field above it is. Many theoretical and experimental studies have compared the effects of the geometry of macroscopic inhomogeneities on the MCG and ECG with the hope of determining which measurement is least sensitive to the boundary and most sensitive to the primary sources. As we will see later, the situation is reversed at the cellular level, since the inhomogeneity of the membrane dominates both the electric and magnetic fields and the active currents do not directly affect either field to an appreciable degree.

9.8.3. Uniform Double-Layer Model

We can extend our discussion of the ECG-MCG relationship by examining a simple physiological model for the cardiac current sources. The cardiac activation wavefront can be modeled as a cup-shaped primary current source as was shown in Fig. 6.1.5. The relationship between the ECG and MCG signals that would be produced by such a generalized source can be determined from Eqs. 9.8.2 and 9.8.3. A great simplification is possible if one assumes that the direction of this primary current is everywhere perpendicular to the surface of the cup and that the primary current density and the wavefront thickness are the same over the entire cup, as shown in Fig. 9.8.3a. The source reduces to a uniform electric double layer, so that the electric potential at any point in a homogeneous medium is proportional to the solid angle subtended at that point by the rim of the double layer (Section 6.2.6). We have shown (Wikswo et al. 1979) that the magnetic field is given by a particular line integral over the rim of the double layer. Since the rim determines both signals, one can argue that the two signals contain the same information, but that they have differing spatial sensitivities. Thus this source is an example of one in which physiological constraints cause the flow and vortex sources to be directly related. Using our variable counting approach, this source is seen to have only a single degree of freedom and it follows that the ECG and the MCG would have to be related.

If we relax the assumption that the primary currents are parallel to the cardiac activation wavefront, current sources such as the one in Fig. 9.8.3b can be considered (Wikswo and Barach 1982). This type of source might result from the spiral geometry of cardiac muscle fibers within the heart, but the existence of such a source has not yet been demonstrated. Since this source has the
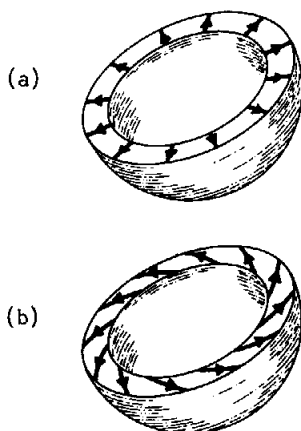
Fig. 9.8.3.   Two hypothetical models of the primary current density
              in the cardiac activation wavefront with (a) the pri-
              mary currents perpendicular to the wavefront and (b)
              having a tangential component as well (Wikswo and Bar-
              ach 1982).

same divergence as that in Fig. 9.8.3a, the two sources will pro-
duce identical electrical fields. The second source differs from
the first only by a divergence-free vortex component of $\mathbf{J}^p$ parallel
to the wavefront surface. This component will result in the two
sources having differing curls and thus differing magnetic fields,
so that only magnetic measurements could distinguish between such
sources if they indeed exist in the heart.

    9.8.3a. Multipole Expansion. One way to describe mathemati-
cally these concepts is by the current multipole expansions for the
electric potential and magnetic field introduced by Katila and Karp
in Section 9.3.4. These authors emphasized that the leading term of
each expansion has the current dipole as its source, and therefore
to first order the two measures of cardiac activity yield identical
information. Independently we have noted that the next term of the
expansion has the current quadrupole coefficients a_{ij} appearing as
the source, but different combinations of a_{ij} enter the expression
for $V(\mathbf{r})$ than for $B(\mathbf{r})$. In particular, the field depends on asymme-
tric terms of the quadrupole tensor that do not contribute to the
potential. We shall briefly elaborate on this point of view.

    The a_{ij} represent particular integrals of each Cartesian com-
ponent of the current density, as given by Eq. 9.3.8. The set of
quadrupole terms can be arrayed in a matrix denoted by the symbol
[a]:

$$[a] = \begin{bmatrix} a_{xx} & a_{xy} & a_{xz} \\ a_{yx} & a_{yy} & a_{yz} \\ a_{zx} & a_{zy} & a_{zz} \end{bmatrix} \qquad (9.8.6)$$

Each of the terms can be represented by a pair of opposing dipoles placed side-by-side or in a line, with an appropriate distance of separation. The terms of the quadrupole tensor of Eq. 9.8.6 can thus be displayed pictorially as in Fig. 9.8.4a. This tensor at first glance appears to have nine independent terms. However, the sum of a_{xx}, a_{yy}, and a_{zz}, known as the "trace" of the tensor, must be set equal to zero, because this sum must be electrically silent, i.e. the electric potential outside this combined source would be zero everywhere, and the sum of a_{xx}, a_{yy}, and a_{zz} would otherwise be indeterminant. Because of this constraint, there are only eight independent terms in the tensor. While the act of setting the trace to zero will also modify the pictures in Fig. 9.8.4a (Wikswo 1978c), we shall not do this so as to simplify the discussion.

As was also noted in Section 9.3.4a, we can write [a] as the sum of symmetric and antisymmetric tensors

$$[a] = [a]^s + [a]^a \qquad (9.8.7)$$

where $[a]^s$ and $[a]^a$ are given by combining [a] with its mathematical transpose, indicated by $[a]^T$:

$$[a]^s = (1/2)\{[a] + [a]^T\}$$
$$[a]^a = (1/2)\{[a] - [a]^T\} \qquad (9.8.8)$$

The pictorial representation of $[a]^s$ and $[a]^a$ are shown in Figs. 9.8.4b and c, respectively. The dipoles shown for $[a]^s$ and $[a]^a$ must have one-half the strength of those in [a] to be consistant with Eq. 9.8.7. The symmetric tensor $[a]^s$ contains combinations of a_{ij} that appear in the current multipole expansion for the electrical potential distribution from a primary current source (Eq. 9.3.7). It contains five independent terms, when we take account of the constraint that the trace must be zero. The antisymmetric tensor $[a]^a$ contains three independent terms, since the three terms in the lower left corner of Fig. 9.8.4c are simply the negative of the respective three terms in the upper right corner across the diagonal. We also see that these three terms in the <u>current quadrupole</u> tensor represent current loops (<u>magnetic dipoles</u>) oriented normal to the three axes of the Cartesian coordinate system. Each is electrically "silent," as illustrated in Fig. 9.8.5, so it is apparent that these three terms can be determined only by magnetic

measurements and not by electric ones. One advantage of analyzing a
field pattern with the current multipole expansion, as opposed to
the magnetic multipole expansion, is that it allows extraction of
the maximal information from the magnetic field while also provid-
ing equivalent primary current generators that can be readily in-
terpreted in terms of various physiological current source distri-
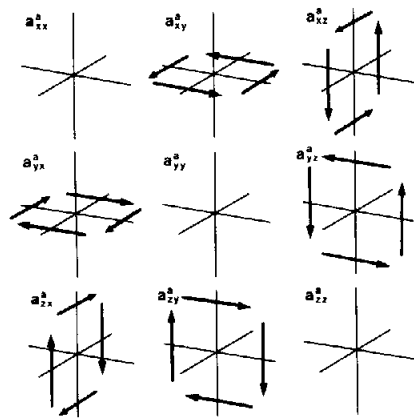butions.
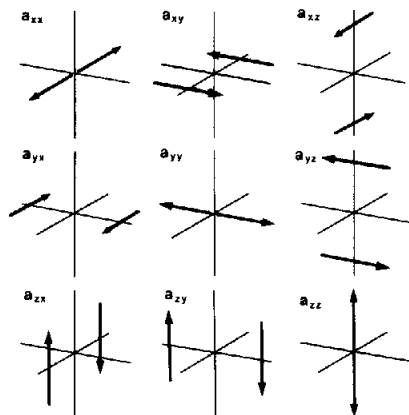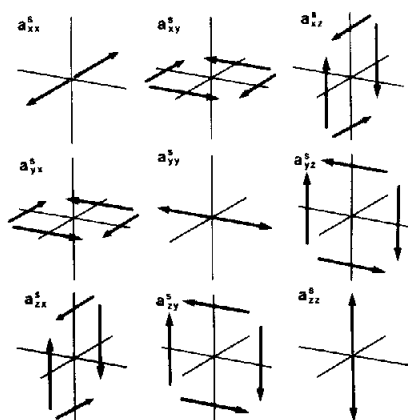
Fig. 9.8.4. Pictorial representation of the current quadrupole
tensor: (a) the complete tensor; (b) symmetric part of
the tensor; and (c) antisymmetric part of the tensor.

9.8.4. Possible Tests for New Information

A simple test for independent information in the MCG has been conducted at Stanford (Barry et al. 1977; Wikswo et al. 1979; Leifer et al. 1981). Sufficient simultaneous ECG and MCG data were recorded to allow determination of the components of the electric heart vector P_H and the magnetic heart vector M_H that describe the ECG and MCG. If the ECG and MCG were produced by the source in Fig. 9.8.3a in a spherical conductor, then one would expect that

$$M_H = (1/2)\ r \times P_H \qquad\qquad (9.8.9)$$

where r is the location of the center of the rim of the wavefront. This equation requires P_H and M_H to be perpendicular, as is seen to be approximately the case in the vector displays in Fig. 9.8.2a. In a group of 20 normal subjects, the mean angle between the peak electric and magnetic heart vectors was found to be $77° \pm 21°$. This relationship has been confirmed by studies in France and Finland (Denis et al. 1978; Karp et al. 1981). In subjects with abnormal cardiac activation, as shown in Fig. 9.8.2b, the relationship between the signals becomes more complicated. While the observed departures from the expected relationship could be due to sources such as Fig. 9.8.3b, it is more likely that they result from the non-dipolarity of complex sources based on Fig. 9.8.3a and from the irregular shape and internal inhomogeneities of the human body.

The ideal test of the relationship between the ECG and the MCG would be to record both signals and to attempt to predict one from the other. As an example, suppose that an isolated heart is maintained in a spherical tank filled with electrolyte. Measurement of the potentials on the surface of the tank could be used to determine the electric multipole expansion that describes these potentials. Similarly, magnetic field measurements outside the tank could be used to determine the multipole expansion that describes the scalar magnetic potential of the magnetic field. One test of independence of information would be to use the electric multipole expansion to predict the magnetic multipole expansion. The difficulty with this step lies in the fact that there is no unique scalar magnetic multipole expansion for an electric multipole expansion (Geselowitz and Miller, pers. comm.). As an example, the two current sources in Fig. 9.8.5a and b have identical electrical multipole expansions but differing magnetic ones (K.R. Swinney, pers. comm.). It appears that the only way to relate the scalar electric and magnetic multipole expansions is through an intermediate physiological model. It should in principle be possible to create two physiological models, one as in Fig. 9.8.3a and one as in Fig. 9.8.3b, and determine which would be consistent with both the ECG and MCG data. In practice, the complexity of the cardiac activation wavefront, uncertainties in the model, and noise in the data may make this approach impractical for measurements on entire hearts.

(a)   (b)   (c)

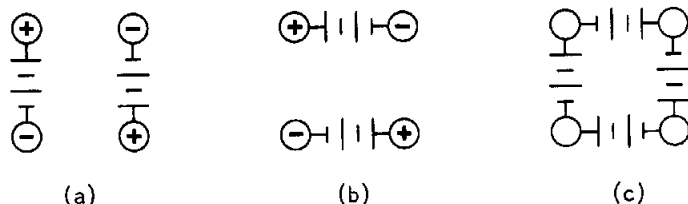Fig. 9.8.5.   Two simple electric current sources (a) and (b),   each
              composed of a pair of current dipoles, that have
              identical electric fields but differing   magnetic
              fields. (c) An electric current source constructed by
              superimposing source (a) with the negative of source
              (b). Because (a) and (b) produce the same electric
              fields, (c) will be electrically silent. Since (a) and
              (b) produce different magnetic fields, (c) will have a
              non-zero magnetic field, equivalent to that of a mag-
              netic dipole.

It may also be possible to use the current multipole expansion to
simplify comparison of the electric and magnetic fields, but the
electric measurements will be unable to specify the asymmetric
quadrupole moments, and in a spherical tank the magnetic measure-
ments could not be used to determine the axial quadrupole moments.
It may also be possible to make quantitative measurements based on
this idea using only small, isolated samples of tissue, but it will
be important to ensure that samples with appropriate cellular geom-
etries were included.

     One such experiment would be to place a bipolar stimulating
electrode within the myocardium of the exposed heart of an experi-
mental animal, as shown in Fig. 9.8.6.   If the electrode is placed
midway between the endocardium and the epicardium, stimulation of
the myocardium will create a closed activation wavefront that will
expand radially from the stimulation site. The wavefront remains
closed while it expands until it reaches the endocardial or epicar-
dial surfaces. In the uniform double-layer model (Fig. 9.8.6a)
there should be no external electric or magnetic fields from this
closed wavefront. Recent electrical measurements on closed wave-
fronts (Corbin and Scher 1977, Colli-Frazone et al. 1982) indicate
that cellular anisotropies result in significant departures from
the uniform double layer model, as shown in Fig. 9.8.6b. The use of
a SQUID magnetometer may help clarify differences between the two
models concerning such intramyocardial current flow.

9.8.5. Analysis at the Cellular Level

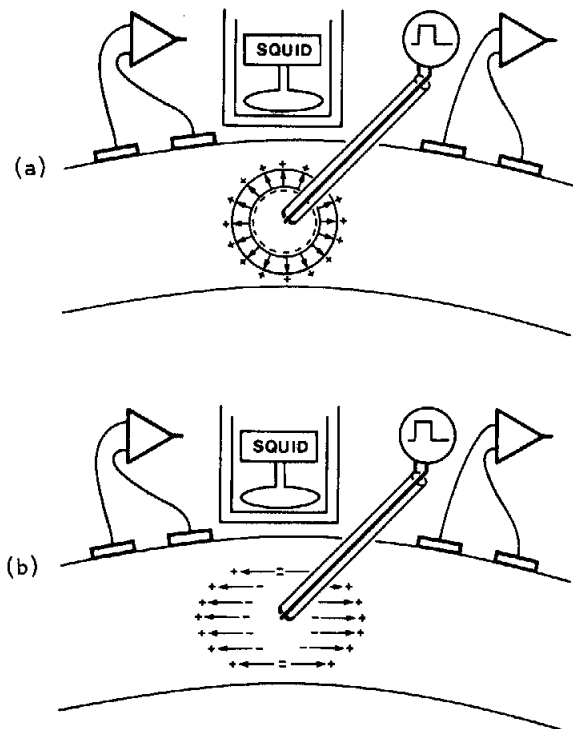   The discussion of the information content of biomagnetic meas-

Fig. 9.8.6. A schematic representation of electric and magnetic measurements of a closed, expanding actuation wavefront based on (a) the uniform double-layer model, and (b) an anisotropic model.

urements has recently been extended to the level of a single cell. In a single cell, the impressed currents exist only in the cell membrane which is several nanometers thick and has a very high resistivity. All other currents both inside and outside the cell are ohmic. Plonsey (1982) has pointed out that his calculations (Plonsey 1981) and those by Swinney and Wikswo (1980) demonstrate that the contribution of these impressed currents to both the electric and magnetic fields outside a single cylindrical nerve cell are many orders of magnitude smaller than the contributions associated with the secondary sources at the inner and outer surfaces of the membranes. Plonsey also examined the ratio of contributions from impressed and secondary sources for a collection of cardiac cells in which the activation wavefront is assumed to propagate primarily across fiber directions. He concluded that the impressed current sources still did not contribute in multi-cellular systems. Because

the secondary sources for both fields are determined by V, the only
possibility for truly independent information in the cellular mag-
netic field lies in the way that the membrane geometry enters into
the equations. The situation is very much more complex than that of
the macroscopic inhomogeneities examined in previous studies, since
in systems of cardiac cells the surfaces bounding the inhomogenei-
ties will be complex and multiply connected. Development of an
understanding of the geometric contribution to the secondary elec-
tric and magnetic sources may be the key to understanding the rela-
tionship of the electric and magnetic fields produced by multicel-
lular systems.

        One difficulty with relating the analysis by Plonsey to that
presented by Wikswo and Barach (1982) occurs because in their des-
cription of the cardiac activation wavefront as a region with a
"primary" current, this equivalent current source represents both
impressed and secondary sources at the cellular level, while Plon-
sey limits consideration to the impressed currents in the cell mem-
brane and relegates all other effects to secondary sources. If in
fact we realize that the secondary sources constrain the current to
flow along the cell axis, the arrangement of the cells will deter-
mine whether this current and the corresponding primary current is
parallel or perpendicular to the activation wavefront. The analysis
by Wikswo and Barach indicates that if new information were to
exist, it may be in the form of primary current source distribu-
tions that have a component tangential to the surface bounding the
source. In this perspective, the key to the question of indepen-
dence lies in determining whether the primary current being forced
through the wavefront goes straight across or at an angle, not
whether this effect is due to impressed or secondary cellular
sources. As long as you are outside the activation wavefront, it
seems acceptable to describe the wavefront as a primary current
source, even if it is really an "equivalent" impressed current
source describing both impressed and secondary sources. Cellular
models will be invaluable to develop a description of how current
flows within the wavefront; we can then use this description to de-
termine whether the pattern of the current being forced across the
wavefront is such that it might affect the magnetic field differ-
ently than the electric one. It is important to remember that at
the level of the law of Biot and Savart, the magnetic field is de-
termined by the total current density, not by the individual prima-
ry, secondary and ohmic contributions.

9.8.6. Summary

        It has been clearly demonstrated over the past decade that the
ECG and MCG have several significant differences. Instrumental ones
have allowed the MCG to measure currents of injury during myocardi-
al infarction; spatial sensitivity differences have been studied
and may lead to differences in the diagnostic specificity and sen-

sitivity of the two techniques. The possibility of information in
the MCG which is independent of that in the ECG has been discussed
at length, and while much progress has been made in understanding
the theoretical possibilities and restrictions, the existence or
absence of such information has yet to be proven either experimen-
tally or theoretically. There appears to be universal agreement
that such sources are not likely to exist in single cells, and if
they did their contribution would be orders of magnitude too small
to see. The possibility that new information may exist in connec-
tions of cells is just now being explored. We must remember that if
such information dominated the MCG signal its presence should have
been detected by the simple pattern recognition and modeling stud-
ies that have already been conducted. Therefore if new information
exists in the MCG, it may contribute only a small fraction of the
total energy in the signal. While it is impossible to predict the
scientific value of such information if it were found, it is clear
that the search for it will broaden our understanding of both car-
diac electrophysiology and the electromagnetic fields produced by
biological systems.

ACKNOWLEDGEMENTS

Bibliography

Barry, W.H., Harrison, D.C., Fairbank, W.M., Lehrman, K., Malmivuo, J.A.V., and Wikswo, J.P., Jr., 1977, "Measurement of the human magnetic heart vector," Science, 198: 1159-1162.

Baule G., and McFee, R., 1970, "The magnetic heart vector," Am. Heart J., 79: 223-236 (1970).

Cohen, D., and Kaufman, A.L., 1975, "A magnetic determination of the relationship between the S-T segment and the injury produced by coronary occlusion," Circ. Res., 36: 414-424.

Cohen, D., and MacArthur, J., 1975, "Magnetocardiograms from an insulated in vivo canine heart," in Report of the Low-Field Group: The Magnetocardiogram, (Francis Bitter National Magnet Lab, MIT, 1975), pp 117-122.

Colli-Frazone, P., Guerri, L., Ticcardi, B., and Viganotti, C., 1982a, "The inverse potential problem applied to the human case," in Models Measurements of the Cardiac Electric Field, E. Schubert, ed., Plenum Press, New York, pp. 19-33.

Colli-Franzone, P., Guerri, L., Viganotti, C., Macchi, E., Baruffi, S., Spaggiari, S., and Taccardi, B., 1982b, "Potential fields generated by oblique dipole layers modeling excitation wavefronts in the anisotropic myocardium," Circ. Res., 51: 330-346.

Corbin, L.V.,III, and Scher, A.M., 1977, "The canine heart as an electrocardiographic generator: Dependence on cardiac cell orientation," Circ. Res., 41: 58-67.

Cuffin, B.N., 1978, "On the use of electric and magnetic data to determine electrical sources in a volume conductor," Annals of Biomed. Eng., 6: 173-193 (1978).

Denis, B., Machecourt, J., Favier, C., and Martin-Noel, P., 1978, "Interpretation of the normal and pathological magnetocardiogram, based on 140 thoracic recordings," Ann. de Cardiologie et D'Angeiologie, 27: 81-86.

Farrell, D.E., Tripp, J.H., and Norgren, R., 1980a, "Magnetic study of the His-Purkinje conduction system in man," IEEE Trans. Biomed. Eng., BME-27: 345-350.

Geselowitz, D.B. 1970, "On the magnetic field generated outside an inhomogenous volume conductor by internal current sources," IEEE Trans. Mag., Mag-6:346-347.

Grynszpan, F., and Geselowitz, D.B., 1973, "Model studies of the magnetocardiogram," Biophys. J., 13: 911-925.

Horacek, B.M., 1973, "Digital model studies in magnetocardiography," IEEE Trans. Mag., MAG-9: 440:444.

Hosaka, H., Cohen, D., Cuffin, B.N., and Horacek, B.M., 1976a, "The effect of torso boundaries on the magnetocardiogram," J. Electrocardiography, 9: 418-425.

Karp, P.K., Katila, T.E., Saarinen, M., Siltanen, P., and Varpula, T.T., 1981, "The normal magnetocardiogram II. A multipole analysis," Circ. Res., 47: 117-130 (1981).

Leifer, M.C., Griffin, J.C., Iufer, E.J., Wikswo, J.P., Jr., Fairbank, W.M., and Harrison, D.C., 1981, "An integrated system for magnetic assessment of cardiac function," in Biomagnetism, S.N. Erne, H.-D. Hahlbohm and H. Lubbig, Eds., (de Gruyter, Berlin, 1981) pp 123-137.

Plonsey, R., 1972, "Capabilities and limitations of electrocardiography and magnetocardiography," IEEE Trans. Biomed. Eng., BME-19: 239-244 (1972).

Plonsey, R., 1981 "The magnetic field resulting from action currents on cylindrical fibers," Med. Biol. Eng. and Comp., 19: 311-315 (1981).

Plonsey, R., 1982, "On the nature of sources of bioelectric and biomagnetic fields," Biophys. J., 39: 309-312 (1982).

Rush, S., 1975, "On the independence of magnetic and electric body surface recordings," IEEE Trans. Biomed. Eng., BME-22: 157-167 (1975).

Swinney, K.R., and Wikswo, J.P., Jr., 1980, "A calculation of the magnetic field of a nerve action potential," Biophys. J., 32: 719-732 (1980).

Tripp, J.H., 1977, "Effects of torso geometry on the magnetocardiogram," Biophys. J., 18: 269-273.

Wikswo, J.P., Jr., Griffin, J., Leifer, M., Barry, W.M., and Harrison, D.C., 1978, "Vector magnetocardiography: The effect of the heart-lung boundary," Clinical Research, 26: 486A (1978).

Wikswo, J.P., Jr., Malmivuo, J.A.V., Barry, W.H., Leifer, M.C., and Fairbank, W.M., 1979, "The Theory and Application of Magnetocardiography," in Cardiovascular Physics, D.N. Ghista, Ed., (S. Karger: Basel, 1979) pp1-67.

Wikswo, J.P., and Barach, J.P., 1982a, "Possible sources of new information in the magnetocardiogram," J. Theoretical Biol., 95: 721-729 .