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COMPARISON OF THE MAGNETOENCEPHALOGRAM AND ELECTROENCEPHALOGRAM [1]

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The brain-generated currents that produce the potentials measured by the electroencephalogram (EEG) also produce magnetic fields which can be measured by the magnetoencephalogram (MEG). However, since the MEG is a different type of measurement than the EEG it may provide new or better information about electrical activity in the brain. Because of this, a number of MEG measurements have rencently been made (Cohen 1968, 1972; Brenner et al. 1975, 1978; Teyler et al. 1975; Hughes et al. 1976, 1977; Reite et al. 1976; Reite and Zimmerman 1978). Although the main features of the MEG resemble those of the EEG, there are differences in frequency spectra, phase, etc. For example, the 5 c/sec theta waves present in the EEG are absent in the MEG of one particular subject (Cohen 1972). Also, MEGs appear to detect a more localized source of somatically evoked brain activity than the corresponding EEGs (Brenner et al. 1978). These results suggest that the MEG may detect a more restricted group of sources than the EEG, i.e., sources which occupy a smaller region or which have only particular angular orientations.

This suggestion is supported by some previous studies. Both theoretical and experimental studies of the EEG have shown that the different tissues and bone surrounding the brain attenuate and/or 'smear' the potentials that reach the scalp (Geisler and Gerstein 1961; De Lucchi et al. 1962; Cooper et al. 1965; Vaughan 1969). As a result, the EEG is sensitive to sources in a large region of the brain and cannot easily detect a source which occupies a small region (Regan 1972). On the other hand, a theoretical study has shown that concentric inhomogeneous layers in a sphere do not affect the magnetic fields produced by sources (dipoles) in the sphere (Grynszpan and Geselowitz 1973). Since a sphere is a reasonable approximation to the head, it follows that the MEG may not be subject to 'smearing' and hence should be able to detect sources which occupy small regions. In addition, it has been shown that dipoles oriented perpendicularly to the surface of a sphere will produce zero magnetic field (Baule and McFee 1965). Therefore, the MEG will tend to detect only those sources parallel to the surface of the head; in contrast, the EEGs can detect dipoles with any orientation.

To date, no detailed comparison of the angular orientations and locations of sources in the brain which are detected by the MEG and EEG has been made. It is, therefore, the purpose of this paper to make this comparison. This is done by computer calculations of MEG and EEG measurements on a spherical model of the head. This model consists of 4

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concentric regions with different conductivities that represent the brain, cerebrospinal fluid, skull, and scalp. Various types of MEGs and EEGs are considered; these are unipolar, bipolar, and quadrupolar which use 1, 2, and 4 measurement points respectively.

These MEGs and EEGs are compared by a comparison of their lead fields. A lead field is a vector field in a volume conductor; this field is the pattern of sensitivity of a MEG or EEG to sources at various locations and orientations in the volume conductor. Lead fields can be used to calculate the MEG or EEG signal magnitude, defined as M and E respectively, produced by a source in the following manner (Plonsey 1972):

$$M \text{ or } E = \iiint \overline{L}_{M \text{ or } E} \cdot \overline{J}^s \, dv \qquad (1)$$

where $\overline{L}_M$ and $\overline{L}_E$ are the MEG and EEG lead fields and $\overline{J}^s$ is the source current density. This source current density is a vector field of current dipoles; these are the current sources themselves and not the currents that they produce in the volume conductor. The dot (or scalar) product in Eqn. (1) shows that source dipoles which are in the region of the largest lead field and parallel to it will contribute most to a MEG or EEG measurement. Therefore, lead fields give a visual pattern of the sensitivity of a MEG or EEG. Since each different type of MEG or EEG has a different lead field pattern, these fields provide a good method for comparison among these types. Note that Eqn. (1) refers to a static situation, but can be used to calculate a time-varying MEG or EEG by allowing $\overline{J}^s$ to change with time. Since it has been stated by Adrian and Yamagiwa (1935) and subsequently proven by many others that the main portion of scalp EEGs is produced by sources in a thin layer of the cortex, lead fields are determined in this model only on a spherical cap near the surface of the brain region.

As Eqn. (1) shows, the magnitude of a measurement depends not only on the characteristics of the lead field but also on the characteristics of the source, $\overline{J}^s$. For example, source dipoles which are all aligned will produce a different magnitude of M or E than dipoles which are randomly oriented. We here asumme that various types of brain activity can be represented by either aligned or randomly oriented sources or some combination of both. Therefore, in cases where we calculate M or E, these two basic types of sources are used.

Theory and methods

The model of the head used in this study is illustrated in Fig. 1. Lead fields are determined at points on a spherical cap with a radius of 7.5 cm which represents the cortex. Eqn. (1) is used to determine the lead field as follows. The source is considered to be a single discrete current dipole so that $\overline{J}^s dv$ becomes $\overline{P}$, the current dipole, and Eqn. (1) for M then becomes

$$M = \overline{L}_M \cdot \overline{P} \, . \qquad (2)$$

When $\overline{P}$ has unit amplitude, then M has a value equal in magnitude to the component of the lead field in the direction of $\overline{P}$. Hence, the 3 components of the lead field at any point can be determined by calculating the M's produced by 3 orthogonal unit amplitude current dipoles at that point. The expression for M (magnetic field) produced by current dipoles in the spherical model have been derived previously (Cuffin and Cohen 1977a); the expressions for E (electric potential) are given in Appendix A. Unipolar MEG and EEG lead fields are calculated by using a magnetic or potential measurement at one point. Bipolar and quadrupolar lead fields are calculated by vector summation of appropriate unipolar lead fields. For example, bipolar MEG lead fields are determined by summation of the lead fields produced by a magnetic field measurement at one point minus a measurement at another point.

In general, if the lead fields are known, then Eqn. (1) can be used to calculate M or E produced by either an aligned or random source. However, in this study lead fields are

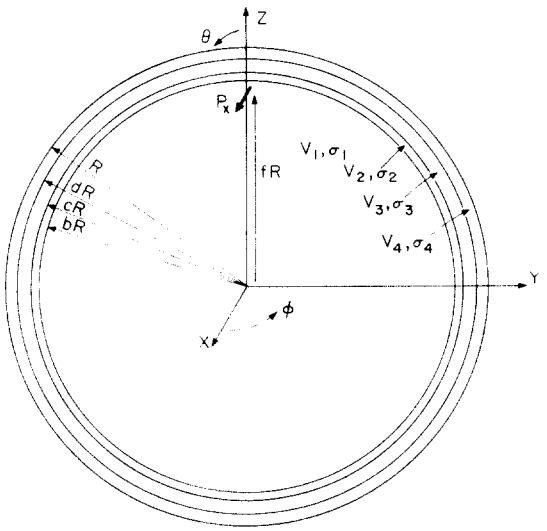

Fig. 1. Four-layer spherical model of the head. The inner sphere (region 1) represents the brain; the successive layers represent the cerebrospinal fluid (2), skull (3) and scalp (4). The V's and σ's are the potentials and conductivities in the 4 regions; P_x is a current dipole. In the calculations, R = 8.8 cm and f, b, c, and d are coefficients such that fR = 7.5, bR = 7.9, cR = 8.1, and dR = 8.5 cm. The conductivities used are: $\sigma_1 = \sigma_4 = 3.3 \times 10^{-3}$, $\sigma_2 = 10.0 \times 10^{-3}$, and $\sigma_3 = 4.2 \times 10^{-5}$ ℧/cm.

determined only at discrete points on a spherical cap and hence only approximations to Eqn. (1) can be calculated here. For these approximations, it is assumed that the sources are located only at the same points on the cap where the lead fields have been determined. For the aligned source, the approximation to Eqn. (1) is made by dividing the cap into a number of area elements, A_i, and considering all sources on each element to be gathered into an equivalent source dipole, S_i, at the centroid of the element. The magnitude of each $\overline{S}_i$ is equal to PN_AA_i where N_A is the areal density of aligned dipoles with amplitude P. Each $\overline{S}_i$ is parallel to the surface of the model and oriented so that in terms of the coordinate system in Fig. 1 it has a component in the positive y direction and no x component; the z component is positive or negative depending on the location of the element. The approximation is then given by

$$M \text{ or } E \approx \sum \overline{L}_{M_i} \text{ or } {}_{E_i} \cdot \overline{S}_i \qquad (3)$$

where $\overline{L}_{M_i}$ and $\overline{L}_{E_i}$ are the lead fields at the centroid of A_i and the summation is over the total number of area elements. For the random source, RMS values must be calculated. It has been shown that if N randomly oriented dipoles, all with the same amplitude P, are located at a point, they will combine to form a vector component in any direction which has an RMS value over time of $\pi P\, N^{1/2}$ (Cuffin and Cohen 1977b). For dipoles distributed over an area, this RMS value at large distances from the area is approximately given by $\pi P(N_A A)^{1/2}$. Therefore, for the random source the approximation to Eqn. (1) is given by

$$M \text{ or } E \approx \pi P(N_A \sum |\overline{L}_{M_i} \text{ or } \overline{L}_{E_i}|^2 A_i)^{1/2} \qquad (4).$$

Results

Fig. 2 presents the unipolar MEG and EEG lead fields on the spherical cap which represents the cortex. The term unipolar is strictly accurate for the MEG since only one measurement point is used. However, for the EEG the difference between two measurement points (electrodes) must be used to obtain actual measurements. By locating one of the electrodes on the bottom of the model, the lead field on the cap is almost completely associated with the electrode on the top. The lead field of a different type of unipolar MEG which measures the component of magnetic field tangential to the surface of the model is presented in Fig. 3. As an example of how these lead fields represent sensitivity to sources, the MEG with the lead field in Fig. 2 is seen to be most sensitive to a dipole at $\theta = 10°$ which is parallel to the lead field arrow. The EEG is seen to be most sensitive to a dipole directly under the electrode at $\theta = 0°$ and perpendicular to the surface.

There are large differences in these 3 uni-

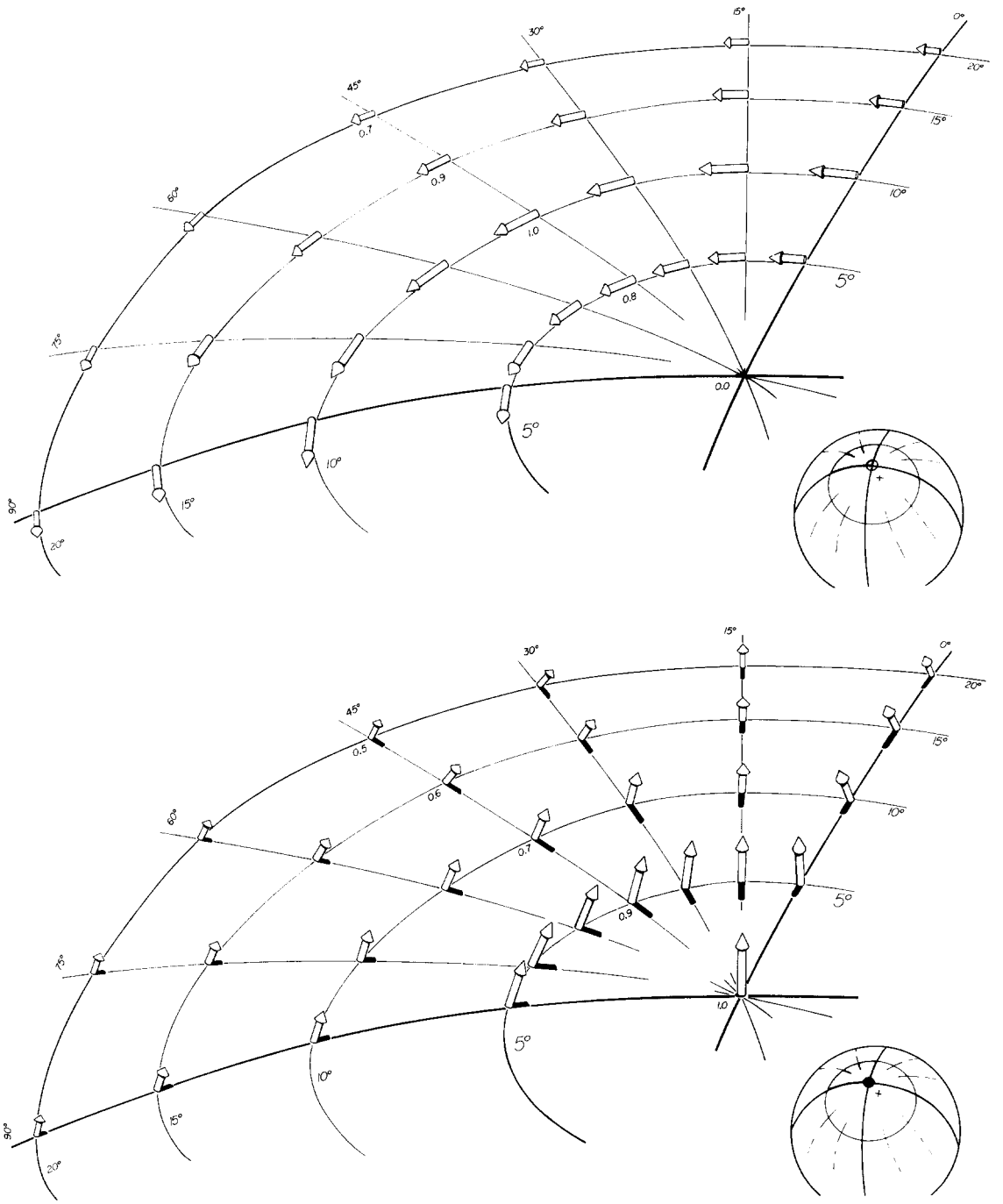

Fig. 2. Unipolar MEG (top) and EEG (bottom) lead fields. Each hemispherical insert shows the location and polarity of the measurement point at the surface of the model. The open circle represents a small coil parallel to the surface of the model at z = 9.5 cm (0.7 cm from the scalp) which measures the radial component of magnetic field. The outward direction of the magnetic field is defined as positive. The closed circle represents an EEG electrode at z = 8.8 cm on the scalp; the second or 'indifferent' electrode is located on the bottom of the model. The shaded area is the expanded area on which the lead fields are plotted. The lead fields, represented as arrows, are plotted on the spherical cap which represents the cortex. The lead fields are always normalized so that the largest arrow has unit value. The angles shown correspond to the spherical coordinates θ and ϕ in Fig. 1. For reference, the distance from the point on the top of the cortical cap to the $\theta = 5^\circ$ circle is 0.65 cm; to the 10° circle, 1.30 cm; etc. The numbers along the $\phi = 45^\circ$ line are the lead field magnitudes at those points.

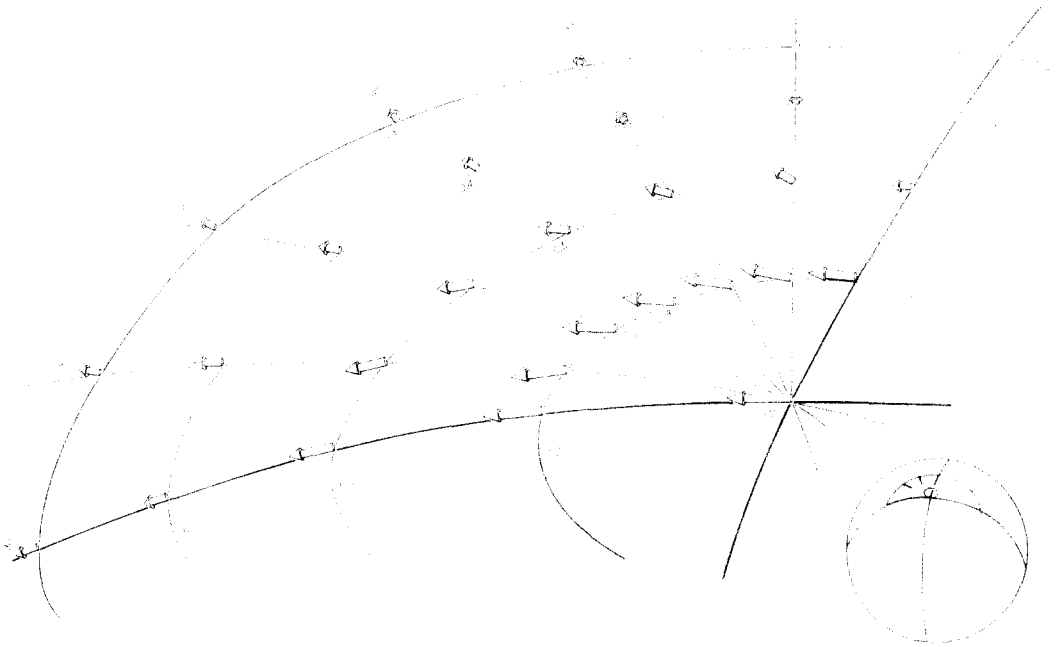

Fig. 3. Unipolar MEG lead field associated with the measurement of a tangential magnetic field component. This component is measured by the small coil at z = 9.5 cm shown in the hemispherical insert. In contrast to the other figures, this coil is upright rather than parallel to the surface of the model.

polar lead fields; the radial field MEG lead field (of Fig. 2) has a null directly under the measurement point while the EEG and tangential field MEG (of Fig. 3) lead fields have maxima at that point. In addition, these and all other MEG lead fields are parallel to the surface of the model while the EEG lead field is primarily perpendicular. Note that the magnitude of the tangential field MEG lead field decreases more rapidly with θ than that of the EEG; stated otherwise, the MEG lead field pattern is narrower. Calculations were also made of the MEG lead field associated with taking the difference between radial magnetic field measurements at z = 9.5 cm and z = 11.5 cm; such a measurement is made by a common form of in-line magnetic gradiometer. This lead field, not presented here, shows the same circular pattern but is somewhat narrower than the unipolar z = 9.5 cm lead field. For example, at $\theta = 30°$ the unipolar and in-line gradiometer lead fields are 40% and 27% of their respective maxima. For all results presented here (with the exception of the gradiometer discussed above) the MEG measurements are made at z = 9.5 cm which is 0.7 cm from the surface of the model since this is as close to the head as it is possible to make such measurements with present magnetic detectors.

Bipolar MEG and EEG lead fields are presented in Fig. 4. This EEG lead field has somewhat the same configuration as the MEG lead field; the maximum for each is located at the center, midway between the two measurement points, and is parallel to the surface. The primary differences are: the MEG lead field has no perpendicular component; the MEG lead field has a null at $\theta = 10°$ and $\phi = 0°$ with a direction reversal beyond that point which the EEG lead field doesn't have; and the MEG lead field pattern is narrower than that of the EEG with approximately one-half its magnitude for $\theta > 5°$.

The lead field of a bipolar MEG that measures tangential magnetic field components is

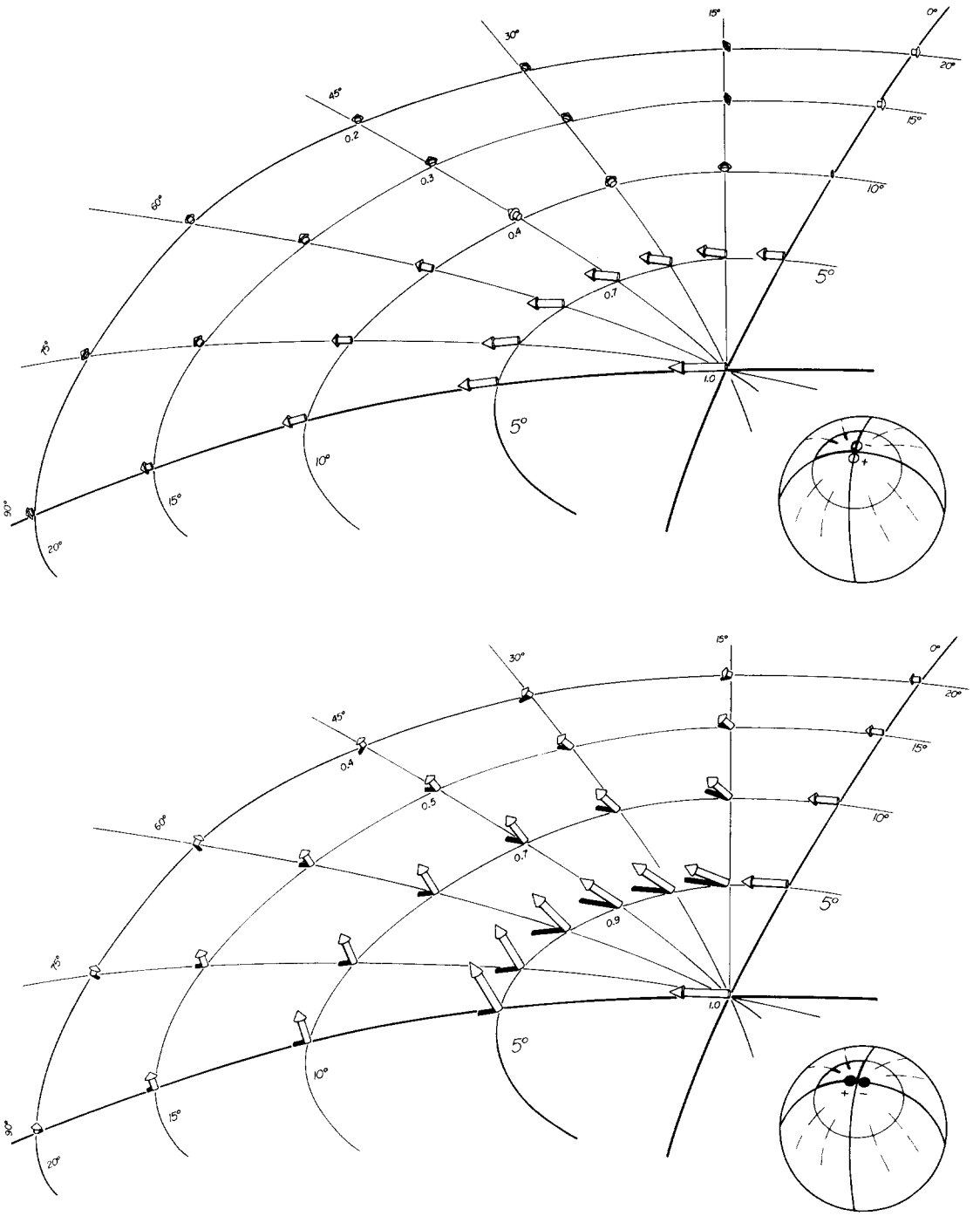

Fig. 4. Bipolar MEG (top) and EEG (bottom) lead fields. The MEG lead field is associated with the difference between two radial magnetic measurements which are 1 cm apart at z = 9.5 cm. The EEG lead field is associated with the difference between two measurements 1 cm apart on the surface of the model.

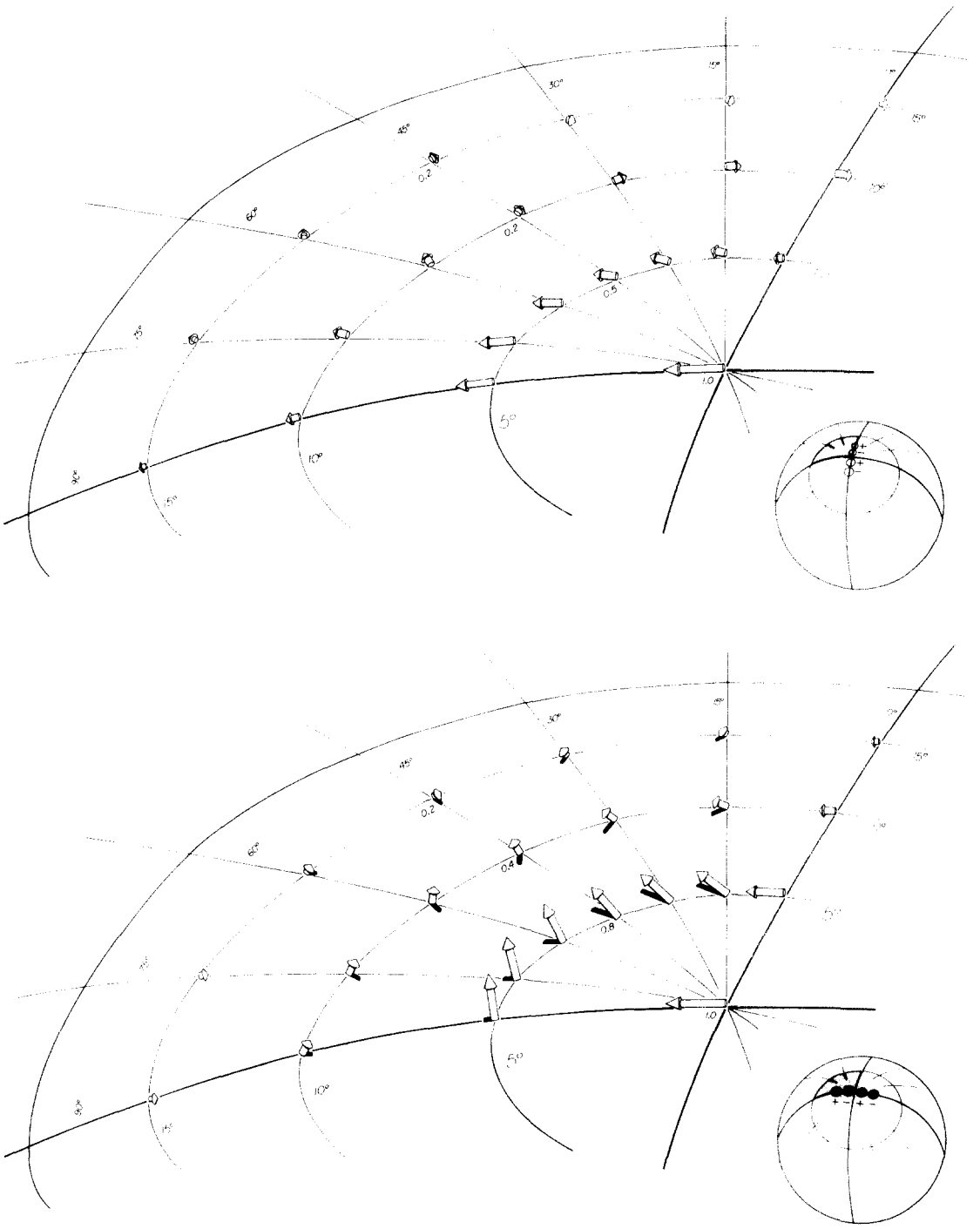

Fig. 5. Quadrupolar MEG (top) and EEG (bottom) lead fields. The MEG lead field is associated with the weighted sum of two bipolar measurements. The first bipolar measurement consists of the inner two radial field component measurements which are 1 cm apart at z = 9.5 cm; the second, the outer two radial field component measurements that are 2 cm apart and also at z = 9.5 cm. The quadrupolar field is the vector summation of the 1 cm lead field plus one-half of the 2 cm lead field. Similarly, the EEG lead field is the sum of a 1 cm lead field plus one-half of a 2 cm lead field.

not as narrow as one that measures radial components. When a combination of radial and tangential measurements is used, as is the case when measurements of the magnetic field component at an angle to the surface of the sphere are made, a slightly narrower lead field was obtained for small (<10°) angular changes away from the radial. However, for larger angles the tangential measurement lead field predominates and this produces a broader lead field.

Quadrupolar MEG and EEG lead fields are presented in Fig. 5. While many other quadrupolar EEG measurement site arrangements are possible, this particular arrangement was used here because it has a lead field which is similar to that of the bipolar EEG and because theoretical arguments show that it is narrower than those of other arrangements. This quadrupolar MEG arrangement was chosen because it directly corresponds to the EEG arrangement and because it has the narrowest lead field of the various arrangements investigated in this study. In particular, an arrangement in which two of the measurements were at z = 9.5 cm and two at z = 11.5 cm, i.e., located at the corners of a rectangle, was investigated and found to have a broader lead field than this four-in-line arrangement. Comparison of the two lead fields in Fig. 5 shows that for $\theta > 5°$ the quadrupolar EEG and MEG lead field magnitudes are approximately equal.

The sensitivity of the lead field patterns to small changes in the model parameters was also investigated. The conductivity of the cerebrospinal fluid was reduced from 10.0×10^{-3} to 6.7×10^{-3} ℧/cm and this produced only minor changes in the normalized bipolar EEG lead field. Even the total removal of the cerebrospinal fluid and its replacement by equal regions of brain and skull produced only an average change of approximately 13% in the normalized bipolar lead field magnitude and little or no change in orientation. In addition, when the radius of the spherical cap representing the cortex was varied, approximately the same changes were produced in the MEG and EEG bipolar lead fields. For example, when the radius was increased from 7.50 to 7.75 cm the average MEG and EEG lead field magnitudes on the $\theta = 5°$ circle dropped by 9% and 2% respectively; on the $\theta = 30°$ circle, by 27% and 24%. When the radius was reduced to 7.25 cm, the magnitudes at 5° increased by 7% and 2%; at 30°, by 28% and 26%. The corresponding changes are, therefore, small or nearly equal. Hence, bipolar MEG and EEG lead fields are either insensitive to these small changes or change together so that their relationship does not significantly alter.

The MEG lead fields presented so far have been for idealized measurements, i.e., the measurement of the magnetic field with infinitely small coils. In practice, however, MEG measurements are made by detecting the magnetic flux passing through a coil with some finite area. Therefore, the lead field of a realistic bipolar MEG detector was calculated in order to determine if the results presented so far have practical validity. We selected a particular bipolar detector, called a '2-D' gradiometer for this calculation. This consists of two 'D'-shaped coils placed back to back to form a circle and lying in a plane at z = 9.5 cm. The bipolar measurement is produced by taking the difference of the magnetic flux passing through the two coils. The lead field was calculated by dividing the area of each 'D' coil, which has a radius of 1.5 cm, into 84 area elements and determining the lead fields associated with a perpendicular magnetic field component at the centroid of each element. The total lead field at a point in the model is then the vector summation of the lead fields for each element weighted by the area of the element. This lead field is somewhat broader than the idealized one in Fig. 4, with the following increase in the average magnitudes: at $\theta = 5°$, 12%; 10°, 30%; 15°, 15%; 20°, 27%; 25°, 36%; and 30°, 44%. There are no significant differences in the lead field orientations. While these '2-D' gradiometer lead field magnitudes are larger than those of the idealized lead fields, they are still approxi-

mately 60% of the bipolar EEG lead field magnitudes. The lead field of the '2-D' gradiometer can be made narrower by reducing the radius of the coils. For example, when coils with a radius of 1.0 cm are used, the increase in the average lead field magnitudes over the ideal ones is: $\theta = 5°$, 6%; 10°, 12%; 15°, 6%; 20°, 11%; 25°, 16%; and 30°, 20%.

The responses of idealized bipolar MEG and EEG measurements to distributed sources which are aligned and random are presented in Fig. 6. For all curves, the number of area elements into which the spherical cap was divided for use in the equations was increased until no significant changes in the curves were produced. This ensured that the approximations used in the equations were not a significant source of error. For all cases, the contributions of the annuli decrease more rapidly with θ for the MEG than for the EEG. In considering the results for the aligned source, it is important to bear in mind that a small area of source located at $\theta = 0°$ directly under the measurement points produces a larger signal than the same area located further out. However, since the area of an annulus near $\theta = 0°$ is much smaller than that of one at a larger angle, it is this latter annulus which gives the largest contribution. It is useful to note that the curves in Fig. 6 can be used to calculate the total value of M or E produced by a particular source by simply adding the contributions from appropriate annuli.

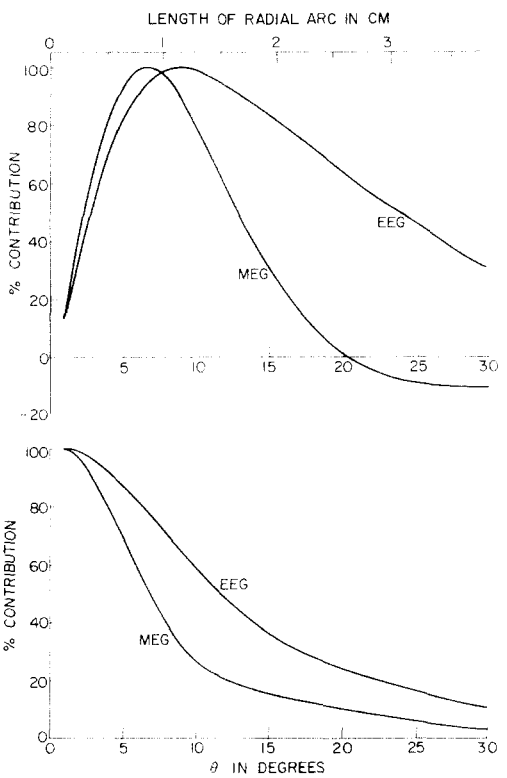

Fig. 6. Bipolar MEG and EEG responses to distributed aligned (top) and random (bottom) sources on the spherical cap which represents the cortex. The curves represent the contributions from annuli formed by 1° increments of the spherical coordinate θ on the cap; hence the curves start at 1°. Each curve is normalized so that the maximum contribution is 100%. The scale at the top of the figure gives the distance from the point on the top of the cortical cap for increasing θ.

Another way of comparing the MEG and EEG responses to distributed sources is to compare the 'effective area' on the cortex that each 'sees'. We define this area to be the section of the cap inside the 50% boundary as determined from the curves in Fig. 6. For the random source, this is the section of the cap bounded by the $\theta = 6.5°$ circle for the MEG and by the $\theta = 11.5°$ circle for the EEG. These sections have areas of 8.7 and 28.8 cm^2 respectively; this is a ratio of 0.30 for the areas. For the aligned source, the section is in the form of a broad annulus bounded by two circles. For the MEG these are the $\theta = 2.5°$ and 13° circles; for the EEG, the $\theta = 3°$ and 23.5° circles. The areas of these sections are 2.3 and 7.1 cm^2 for the MEG and EEG respectively; this is a ratio of 0.32 for these areas. Since the lead field associated with the realistic '2-D' gradiometer is not as narrow as that of the idealized bipolar MEG, the areas seen by this detector increase to 3.4 cm^2 for the random source and 9.3 cm^2 for the aligned source. For these areas the ratios of areas seen by the MEG and EEG are 0.48 for the random source and 0.32 for the aligned source.

It is of interest to consider the case where both aligned and random sources are present. In this case, we should recognize that for both the MEG and EEG, the random source will make the largest contribution from a different section of the cap than the aligned source. The random source will contribute most from a circular section at the center of the cap; the aligned source will contribute most from a broad annular section of the cap. The term 'effective area', as we have defined it, has no meaning for this combination of sources; one should treat each source separately.

Discussion

These results allow comparisons of MEG and EEG sensitivities to sources at different orientations and locations. In comparing the orientations of the sources, the results show that all MEG measurements are sensitive to only those sources which are parallel to the surface of the model. On the other hand, the orientation of the sources to which an EEG measurement is sensitive depends on the type of measurement. The unipolar EEG is primarily sensitive to perpendicular sources; the bipolar and quadrupolar EEG, primarily to parallel sources.

The location of sources to which MEG and EEG measurements are most sensitive is dependent on the type of measurement. A comparison of these locations for the radial field measurement unipolar (and in-line gradiometer) MEG and unipolar EEG cannot be made because of the large differences in their lead fields. However, the locations for the tangential field unipolar MEG and unipolar EEG can be compared; this comparison shows that this MEG has high sensitivity to sources over a smaller area on the cortex than the unipolar EEG. The bipolar MEG and EEG have maximum sensitivities to sources at the same location and orientation. However, the sensitivity of the MEG decreases approximately twice as rapidly with θ as does that of the EEG. Hence, the bipolar MEG has high sensitivity over a smaller area on the cortex and therefore should be better able to search for and locate a source than the bipolar EEG. The advantage of the MEG is due to the fact that the magnetic fields which it measures are not 'smeared' by the layers of bone and tissue surrounding the brain while the potentials measured by the EEG are. The quadrupolar MEG and EEG have sensitivities with approximately similar spatial variation. The advantage of the quadrupolar MEG over the EEG has been lost because the MEG measurement must be made at a greater distance from the head; the magnetic detector is a cryogenic device and hence must be in a vessel containing liquid helium.

The lead fields that have been presented here allow comparisons of the sensitivity of the various MEGs and EEGs to a single discrete source at various locations and orientations on the cortex. In order to compare their sensitivities to distributed sources, the curves in Fig. 6 have been calculated. The shape of these curves is determined by the choice of comparing the contributions of the sources from annular regions. If differently shaped regions were considered, then the curves would have a different shape and this in turn would change the 'effective areas' associated with the measurements. However, the choice of annuli is the logical one since it corresponds to circular 'fields of view' as seen from the measurement points. Likewise, the effective areas would change if some criteria other than the 50% boundary were used, e.g., the 'half-power' or 70.7% boundary. However, for reasonable choices of region shape, the ratio of the effective areas of the MEGs and EEGs would not differ significantly from the value of 0.3 calculated here.

It is important to bear in mind that the results presented here are for a particular spherical model. An evaluation of the effects of changes of some parameters in this model, e.g., conductivity and location of cortex, has been made. It was found that for bipolar MEGs and EEGs, the effects of these changes were either not significant or similar for both

measurements. Therefore, we have compared only similar MEGs and EEGs, i.e., unipolar with unipolar, bipolar with bipolar, etc., so that errors in the assumptions contained in the model will tend to affect the results in the same way. One such error is that the head is not truly spherical. However, a previous study of the effects caused by differences between spherical and oblate and prolate spheroidal volume conductors indicates that this error should cause only a small effect (Cuffin and Cohen 1977a). A more significant error is that associated with the convolutions on the cortex; the lead fields near the convolutions will be distorted by the high conductivity cerebrospinal fluid contained in them. Elaborate computations would be required to evaluate the magnitude of this distortion.

The differences in frequency spectra, phase, etc. that have been reported in experimental MEG and EEG measurements can be explained by the differences in their lead fields. Since unipolar or in-line gradiometer MEGs and unipolar EEGs were used in the reported measurements, these measurements were sensitive to different sources. If the electrical activity in these sources was not related, this would explain some observed differences in the experimental results such as the absence of the 5 c/sec theta waves in the MEG of one particular subject (Cohen 1972), the low correlation (except at alpha frequencies) between MEG and EEG recordings (Hughes et al. 1976), etc.

Such difference in sources detected may also explain the localization of the somatically evoked MEG as compared to the EEG that has been reported by Brenner et al. (1978). Somatically evoked brain activity may consist of a small area of sources parallel to the surface of the head which are detected by the MEG and a larger area of perpendicular sources which are detected by the EEG. It may also be that both the MEG and EEG are detecting parallel and perpendicular sources with the same area and the localization of the MEG is a manifestation of its better spatial resolution. On the other hand, this apparently better spacial resolution may be due to the difference in the pattern of MEG and EEG recordings produced on the surface of the head by such sources. As a unipolar or in-line gradiometer MEG detector is moved over a parallel source it will produce a signal that has maxima of opposite signs on either side and a null directly over it. However, a moving unipolar EEG electrode will produce a signal that has a peak directly over a perpendicular source. Because of this difference, it is difficult to compare the spatial resolution of the MEG and EEG. The comparison becomes even more difficult when a source which contains both parallel and perpendicular components is considered. More careful measurements, particularly of simultaneously recorded EEGs, and analysis must be done to evaluate the spatial resolution capabilities of such MEG and EEG measurements. The lead fields presented in this paper as well as the methods presented for the calculation of the lead fields of various types of practical MEG detectors provide the means for performing this evaluation for proposed models of electrical activity in the brain.

Some practical aspects of MEG measurements should now be considered. In this study we have used distances of 0.7 cm between the magnetic detector and the scalp and 1 cm between the magnetic measurement points. We consider these distances to be at or near the practical limits. The 0.7 cm distance may be somewhat conservative; it can be achieved with present detector equipment and it is possible that it can be reduced to 0.5 or 0.4 cm in a few years. This would result in significant improvements in the MEG lead fields. For example, the bipolar MEG lead field magnitude for a measurement made at the surface of the model decreases approximately twice as fast with increasing θ as that of a measurement made at 0.7 cm from the surface. The 1 cm distance between measuring points is possible with present equipment. However, coils with large areas are required and these broaden the lead field. In a few years improvements in technology will alow

the use of smaller coils and this broadening will be reduced (Reite and Zimmerman 1978). Another practical aspect that should not be overlooked is that present MEG detectors employ superconductor detecting devices. Such devices require a supply of liquid helium and the attendant cryogenic technology. However, in the forseeable future closed-cycle systems will become available so that the user need not be concerned with this technology and difficulty.

Finally, it is useful to compare presently available MEG detectors with EEG measurements. The MEG with the narrowest lead field is the quadrupolar. However, because of the low sensitivity to magnetic fields that such detectors would have, this is not a practical device at present. The '2-D' MEG detector is the best practical device at the present time. It should be compared with the bipolar or quadrupolar EEG. In comparison with the bipolar EEG, this MEG is advantageous in the sense that it is sensitive to a smaller area on the cortex. Compared with the quadrupolar EEG, it is worse. However, we must bear in mind the quadrupolar EEG would require the application of 4 electrodes to the surface of the head; this is an awkward and difficult arrangement to move around the head in attempting to search for a source. Therefore the bipolar MEG has the advantage that it is able to search for and detect a source on the cortex much more rapidly than the bipolar and certainly the quadrupolar EEG.

From a practical point of view, it appears that the MEG will be useful for detecting a more restricted group of sources than the EEG. (An addendum, Appendix B, shows that the MEG is also more restrictive in terms of the depth of the sources in the brain which it detects.)

Summary

The spatial response of the magnetoencephalogram (MEG) to sources in the brain's cortex is compared with that of the electroencephalogram (EEG). This is done using computer modeling of the head which is approximated by 4 concentric spherical regions that represent the brain and surrounding bone and tissue. Lead fields are calculated at points on the cortex for unipolar, bipolar and quadrupolar MEG and EEG measurements. Since lead fields are patterns of the sensitivity of these measurements to a source at various locations and orientations, they provide a convenient means for comparison. It is found that a unipolar MEG has a very different lead field than a unipolar EEG. Hence, this type of MEG detects sources at different locations and orientations than this EEG. Although bipolar MEG and EEG lead fields are found to have similar patterns, the MEG lead field is narrower than that of the EEG and hence 'sees' a smaller area on the cortex than the EEG. This is because the potentials measured by the EEG are 'smeared' by the low-conductivity skull; the magnetic fields measured by the MEG are not smeared. Quadrupolar MEG and EEG lead fields are found to be about the same. The responses of bipolar MEGs and EEGs to distributed sources, which are composed of aligned and randomly oriented dipoles, are compared. It is found that for both types of sources, the MEG 'sees' an area on the cortex which is approximately 0.3 times that for the EEG. Hence, the MEG appears to be useful for detecting a more restricted group of sources than the EEG.

Résumé

Comparaison du magnétoencéphalogramme et de l'électroencéphalogramme

La réponse spatiale du magnétoencéphalogramme (MEG) à des sources situées dans le cortex cérébral est comparée à celle de l'électroencéphalogramme (EEG). A cette fin, la tête est assimilée à un modèle constitué par 4 sphères concentriques représentant le cerveau, l'os et les tissus qui l'entourent. Les

champs de dérivation sont calculés à différents points du cortex pour des mesures MEG et EEG unipolaires, bipolaires et quadripolaires. Puisque les champs de dérivation sont un indice de la sensibilité d'une dérivation à des sources de topographie et d'orientation différentes, ils fournissent un moyen approprié de comparaison. On observe qu'un MEG unipolaire a un champ de dérivation très différent de celui d'un EEG unipolaire. Donc, ce type de MEG détecte des sources à des topographies et des orientations différentes de l'EEG. Bien que les champs de dérivation du MEG et de l'EEG bipolaires s'avèrent avoir une forme identique, le champ du MEG est plus étroit que celui de l'EEG et donc 'voit' une aire plus petite sur le cortex que ne le fait l'EEG. Ceci est dû au fait que les potentiels mesurés par l'EEG sont estompés par la faible conductivité de la boîte crânienne; les champs magnétiques mesurés par le MEG ne sont pas estompés. Les champs de dérivation MEG et EEG quadripolaires s'avèrent être à peu près similaires. Les réponses du MEG et de l'EEG bipolaires à des distributions de sources composées de dipôles orientées en ligne ou de façon aléatoire sont comparés. Il est observé que pour les deux types de sources, le MEG 'voit' une aire du cortex qui est à peu près 0.3 fois celle de l'EEG. Ainsi, le MEG semble détecter des groupes plus restreints de sources que l'EEG.

Appendix A

Potentials produced by current dipoles in an inhomogeneous sphere

The general expression in spherical coordinates for the potential produced by a P_x current dipole in a region of the inhomogeneous sphere shown in Fig. 1 is given by

$$V_i = \frac{P_x \cos\phi}{4\pi\sigma_i} \sum_{n=1}^{\infty} \left(\frac{A_{in}}{r^{n+1}} + B_{in} r^n\right) P_n^1(\cos\theta) \quad \text{(A1)}$$

where i varies with the number of the region, i.e., 1, 2, 3, or 4. Since there are 4 regions and the expression for the potential in each region has two unknown coefficients (A_{in} and B_{in}) this means that for each value of n there is a total of 8 unknown coefficients. However, it can be shown that $A_{in} = (fR)^{n-1}$ (Arthur and Geselowitz 1970). The remaining 7 coefficients can be determined by applying the boundary conditions that potential and current flow must be continuous across the boundaries between regions of different conductivity. This results in a system of 7 simultaneous equations that can be solved to yield the following expression for the potential on the surface of the sphere

$$V = \frac{P_x \cos\phi}{4\pi\sigma_4 R^2} \sum_{n=1}^{\infty} \times \frac{(2n+1)^4 f^{n-1} (cd)^{2n+1} P_n^1(\cos\theta)}{n\Gamma} \quad \text{(A2)}$$

where

$$\begin{aligned}\Gamma = {} & d^{2n+1} \{b^{2n+1} n(k_1 - 1)(k_2 - 1)(n+1) \\ & + c^{2n+1}(k_1 n + n + 1)(k_2 n + n + 1)\} \\ & \times \{(k_3 n + n + 1) + (n+1)(k_3 - 1) d^{2n+1}\} \\ & + (n+1) c^{2n+1} \{b^{2n+1}(k_1 - 1)(k_2 n + k_2 + n) \\ & + c^{2n+1}(k_1 n + n + 1)(k_2 - 1)\} \\ & \times \{n(k_3 - 1) + (k_3 n + k_3 + n) d^{2n+1}\}\end{aligned}$$

and $k_1 = \sigma_1/\sigma_2$, $k_2 = \sigma_2/\sigma_3$, and $k_3 = \sigma_3/\sigma_4$.

For a P_y dipole it is only necessary to change $\cos\phi\ P_x$ to $\sin\phi\ P_y$ in Eq. (A2). For a P_z dipole, $\cos\phi\ P_n^1(\cos\theta)$ is replaced by $P_n(\cos\theta)$ in Eq. (A1) and $A_{in} = n(fR)^{n-1}$ and this yields

$$V = \frac{P_z}{4\pi\sigma_4 R^2} \sum_{n=1}^{\infty} \times \frac{(2n+1)^4 f^{n-1} (cd)^{2n+1} P_n(\cos\theta)}{\Gamma} \quad \text{(A3).}$$

For the calculations, n up to a value of 60 was used.

Appendix B

Lead field magnitude vs. depth in brain

It is known (Grynszpan and Geselowitz

1973) that a source at the center of a spherical volume conductor produces zero magnetic field but non-zero surface potential. Therefore, the ratio of maximum MEG to EEG lead field must decrease with increasing depth into the conductor. Maximum lead fields were calculated in the 4-layer head model and the results are shown in Fig. 7. For the unipolar EEG, a curve for the tangential lead field component in addition to the one for the radial component is presented for comparison with the MEG lead field which is only tangential. The EEG curves are seen to fall off relatively slowly; this is partially due to the 'smearing' of potentials by the skull. The EEG electrodes were on the surface of the model while the MEG detector was 0.7 cm from the surface; future MEG detector improvements reducing this spacing will produce more rapid fall-off of the MEG curves. Quadrupolar MEG and EEG curves (not shown) are about equal because of the 0.7 cm spacing. In general, as compared to the EEG, the MEG is seen to be sensitive to sources primarily on the surface of the brain. This is, therefore, another way in which the MEG is more restrictive than the EEG.

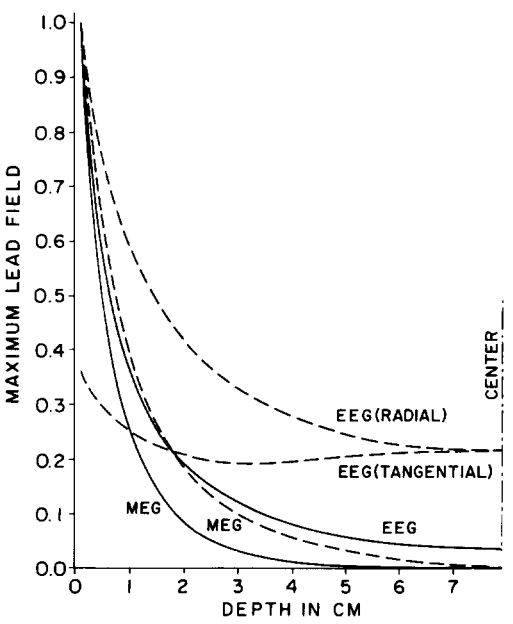

Fig. 7. Maximum unipolar (dashed) and bipolar (solid) lead fields vs. depth below brain surface. All curves are normalized with respect to the value at 0.1 cm except the tangential which has the same scale as the radial.

References

Adrian, E.D. and Yamagiwa, K. The origin of the Berger rhythm. Brain, 1935, 58: 323—351.

Arthur, R.M. and Geselowitz, D.B. Effect of inhomogeneities on the apparent location and magnitude of a cardiac dipole source. IEEE Trans. biomed. Engng, 1970, BME-17: 141—146.

Baule, G. and McFee, R. Theory of magnetic detection of the heart's electrical activity. J. appl. Phys., 1965, 36: 2066—2073.

Brenner, D., Williamson, S.J. and Kaufman, L. Visually evoked magnetic fields of the human brain. Science, 1975, 190: 480—482.

Brenner, D., Lipton, J., Kaufman, L. and Williamson, S.J. Somatically evoked magnetic fields of the human brain. Science, 1978, 199: 81—83.

Cohen, D. Magnetoencephalography: evidence of magnetic fields produced by alpha-rhythm currents. Science, 1968, 161: 784—786.

Cohen, D. Magnetoencephalography: detection of the brain's electrical activity with a superconductivity magnetometer. Science, 1972, 175: 664—666.

Cooper, R., Winter, A.L., Crow, H.J. and Walter, W.G. Comparison of subcortical, cortical, and scalp activity using chronically indwelling electrodes in man. Electroenceph. clin. Neurophysiol., 1965, 18: 217—228.

Cuffin, B.N. and Cohen, D. Magnetic fields of a dipole in special volume conductor shapes. IEEE Trans. biomed. Engng, 1977a, BME-24: 372—381.

Cuffin, B.N. and Cohen, D. Magnetic fields produced by models of biological current sources. J. appl. Phys., 1977b, 48: 3971—3980.

De Lucchi, M.R., Garoutte, B. and Aird, R.B. The scalp as an electroencephalographic averager. Electroenceph. clin. Neurophysiol., 1962, 14: 191—196.

Geisler, C.D. and Gerstein, G.L. The surface EEG in relation to its sources. Electroenceph. clin. Neurophysiol., 1961, 13: 927—934.

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

Hughes, J.R., Hendrix, D.E., Cohen, J., Duffy, F.H., Mayman, C.I., Scholl, M.L. and Cuffin, B.N. Relationship of the magnetoencephalogram to the electroencephalogram. Normal wake and sleep

activity. Electroenceph. clin. Neurophysiol., 1976, 40: 261—278.

Hughes, J.R., Cohen, J., Mayman, C.I., Scholl, M.L. and Hendrix, D.E. Relationship of the magnetoencephalogram to abnormal activity in the electroencephalogram. J. Neurol., 1977, 217: 79—93.

Plonsey, R. Capability and limitations of electrocardiography and magnetocardiography. IEEE Trans. biomed. Engng, 1972, BME-19: 239—244.

Regan, D. Evoked Potentials in Psychology, Sensory Physiology, and Clinical Medicine. Chapman and Hall, London, 1972, 17 p.

Reite, M. and Zimmerman, J. Magnetic phenomena of the central nervous system. Ann. Rev. Biophys. Bioengng, 1978, 7: 167—188.

Reite, M., Zimmerman, J.E., Edrich, J. and Zimmerman, J. The human magnetoencephalogram: some EEG and related correlations. Electroenceph. clin. Neurophysiol., 1976, 40: 59—66.

Teyler, T.J., Cuffin, B.N. and Cohen, D. The visual evoked magnetoencephalogram. Life Sci., 1975, 17: 683—692.

Vaughan, H.G. The relationship of brain activity to scalp recording of event-related potentials. In: E. Donchin and D.B. Lindsley (Eds.), Average Evoked Potentials. NASA, Washington, D.C., 1969: 45—94.