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The relative merits of EEG and MEG

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Abstract. After the first detection of the magnetoencephalogram, MEG, it was believed that it has important benefits over the electroencephalogram, EEG. We have, however, earlier shown that the planar gradiometer MEG has measurement sensitivity distribution very similar to bipolar EEG. Here we show that, unlike initially believed, MEG is not complementary to EEG but gives only partially independent information. We have also earlier shown that MEG does not have better spatial resolution than EEG. Based on the newest information on the resistivity of the skull, we have recalculated the spatial resolution of EEG. These calculations show that EEG has even better spatial resolution than MEG.

Key words: EEG, MEG, lead field, spatial resolution.

1. INTRODUCTION

The first detection of the magnetic field, induced by the electric activity of the brain, the magnetoencephalogram (MEG), was made by David Cohen in 1968 [1]. It was then believed that MEG will have superior properties over the electroencephalogram (EEG) in the following aspects: 1) magnetic measurement should be complementary to EEG and it gives new independent information on the electric activity of the brain; the amount of new information was assumed to be of the same order as the information already given by EEG; 2) the spatial resolution of MEG should be better because the skull is transparent to the magnetic field; it has been well known that the high electric resistivity of the skull strongly decreases the spatial resolution of EEG.

2. THE NATURE OF ELECTRIC AND MAGNETIC LEAD FIELDS

The issue on the complementary nature of EEG and MEG leads is based on their visual appearance. In a bipolar electric lead, the lead field flows mainly

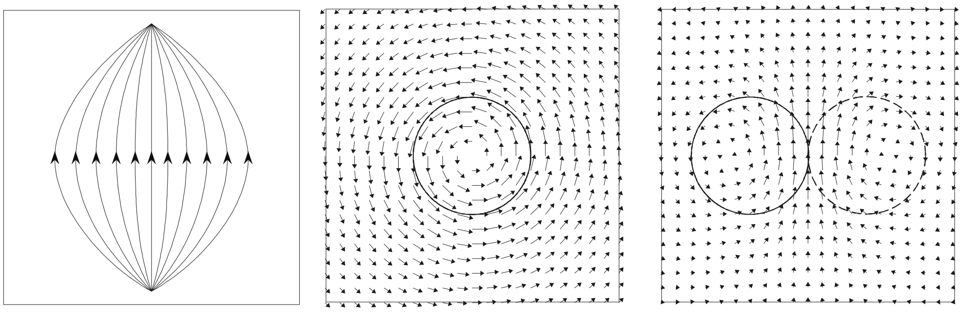


Fig. 1. Sensitivity distributions of bipolar EEG (a), axial MEG (b) and planar gradiometer MEG (c) leads; because bipolar EEG and planar gradiometer MEG methods have similar lead fields in the source region, the source configurations and thus the signals, which they detect, are very similar.

linearly in the region between the electrodes (Fig. 1a). The lead field of a single magnetometer coil is tangential, flowing around the symmetry axis of the coil (Fig. 1b). A single magnetometer coil is, however, not used in MEG due to its very poor spatial resolution, as will be explained later. The basic magnetic lead is the planar gradiometer, whose spatial resolution is much better. The lead field of such lead (Fig. 1c) is, however, in its highest sensitivity region between the coils, very similar to a bipolar EEG lead and therefore it does not measure very different aspect from the underlying volume source. Therefore the complementary nature of EEG and MEG leads remains questionable.

3. INDEPENDENCE OF ELECTRIC AND MAGNETIC MEASUREMENTS

The electric activity of the brain induces a magnetic field. Thus the measurements of the electric and magnetic fields detect the same physiological phenomenon: the electric activity of the brain.

The independence or interdependence of the electric and magnetic fields is a fundamental issue in the discussion whether MEG brings any new information on the electric activity of the brain which is not available in EEG. This issue is discussed here on the basis of the Helmholtz theorem.

The Helmholtz theorem states: "*A general vector field which vanishes at infinity, can be represented as a sum of two independent vector fields, one that is irrotational and another which is solenoidal*". The electric sources represented by the impressed current density $\overline{J}^{\mathrm{i}}$ form such a general vector field and due to the Helmholtz theorem $\overline{J}^{\mathrm{i}}$ may be divided into two components as follows:

$$\overline{J}^{\mathrm{i}} = \overline{J}^{\mathrm{i}}_{\mathrm{F}} + \overline{J}^{\mathrm{i}}_{\mathrm{V}}. \tag{1}$$

These vector fields are referred to as *flux source* and *vortex source*. Bioelectric signals originate from the flux source (Eq. (11.50) in [2]):

$$V_{\mathrm{LE}} = \int_v \Phi_{\mathrm{LE}} \nabla \cdot \overline{J}^{\mathrm{i}}_{\mathrm{d}v}. \tag{2}$$

Here $\nabla \cdot \overline{J}^{\mathrm{i}}$ is the strength of the impressed current source. It is called the *flow* (*flux*) *source*, $\overline{I}_{\mathrm{F}}$. In Eq. (2) Φ_{LE} is the reciprocal electric scalar potential field of the electric lead.

Biomagnetic signals originate from the *vortex source* (Eq. (12.16) in [2]):

$$V_{\mathrm{LM}} = \frac{\mu}{2} \int_v \Phi_{\mathrm{LM}} \overline{r} \cdot \nabla \times \overline{J}^{\mathrm{i}}_{\mathrm{d}v}, \tag{3}$$

where the expression $\nabla \times \overline{J}^{\mathrm{i}}$ is defined as the *vortex source* $\overline{I}_{\mathrm{V}}$ and Φ_{LM} is the reciprocal magnetic scalar potential field of the magnetic lead.

Flux and vortex sources are general concepts, characteristic not only to bioelectromagnetism.

Robert Plonsey published in 1972 a paper [3], which strongly stimulated the biomagnetic research. He claimed, on the basis of the Helmholtz theorem: "*Since the flux and vortex sources are independent, ECG and MCG are similarly independent*". If this had been the case, MCG and MEG would include as much new independent information as ECG and EEG, respectively.

Three years later, in 1975, Stanley Rush published a paper, where he expressed a completely opposite view [4]. He stated: "*The independence of the flow and vortex sources is only a mathematical possibility; the flow and vortex sources are one-to-one with each other*". This fundamental controversy is explained in the following way [2].

4. INDEPENDENCE OF ELECTRIC SIGNALS MEASURED WITH INDEPENDENT LEADS

If a new lead is a linear combination of the existing leads, the information it gives on the source is a linear combination of the existing information and is not new or independent information. It is self-evident that to obtain maximum amount of additional information on the source, a lead must be used that detects a component of the volume source, which is not detected by the existing lead(s).

Let us discuss first the independence of orthogonal leads and the signals which they detect, with an example in two dimensions. Assume that an elementary dipole is first detected with only one (dipolar) lead in the horizontal direction. This lead detects one component of the elementary electric source, namely the horizontal component (Fig. 2a). This lead does not detect the other

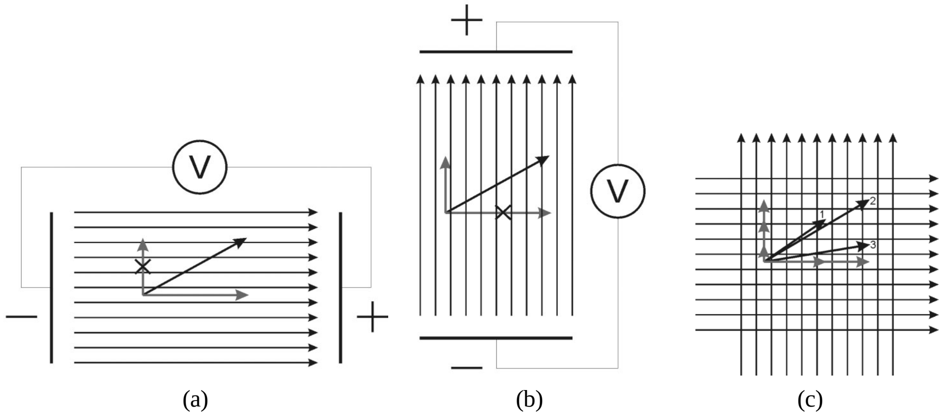


Fig. 2. (a) the horizontal lead detects the horizontal component of the elementary electric source; this lead does not detect the vertical component; (b) the vertical component of the source is detected with a lead in the vertical direction; (c) changes in the amplitude or angle of the source affect both signals (cases 1 and 2); in the rare occasion that only one component of the source changes (from case 2 to case 3), the signal changes only in one lead.

component which is normal to this, i.e., the vertical component. The vertical component of the source is detected with a lead in the vertical direction. That lead, of course, does not detect the horizontal component of the source (Fig. 2b). Even though these orthogonal leads detect orthogonal components of the source, the signals are not fully independent, because changes in the amplitude or angle of the source affect both signals (cases 1 and 2, Fig. 2c).

In the rare occasion that only one component of the source changes (from case 2 to case 3), the signal changes only in one (vertical) lead and the signals may be considered independent (case 3, Fig. 2c). This example demonstrates that even though the two leads were independent, the signals which they detect are only partially independent. All this holds also within magnetic leads and between the electric and magnetic leads on the volume source level.

5. INDEPENDENCE OF EEG AND MEG

EEG and MEG are both generated by the bioelectric activity of the brain tissue. EEG is a result of the flux source and MEG is a result of the vortex source. What Helmholtz theorem states is, that the *lead fields* of EEG and MEG are independent. In other words, EEG and MEG *leads* are not linear combinations of each other. But, as demonstrated above, EEG and MEG *signals* are only partially independent. Thus, by recording MEG, the diagnostic information obtained is only partially new [2].

6. IMPORTANCE OF THE SPATIAL RESOLUTION OF EEG AND MEG

The brain consists of 10^{10}–10^{11} neurons that are very closely interconnected via axons and dendrites. The neurons themselves are vastly outnumbered by glial cells. One neuron may receive stimuli through synapses from as many as 10^3 to 10^5 other neurons [5]. To be able to get an accurate image of the electric activity of the brain, the spatial resolution of the recording system must be good.

There are two important issues related to the spatial resolution: 1) which method has better spatial resolution, EEG or MEG and 2) what is the maximum number of channels which give new information on the source?

7. THE CONCEPT OF HALF-SENSITIVITY VOLUME

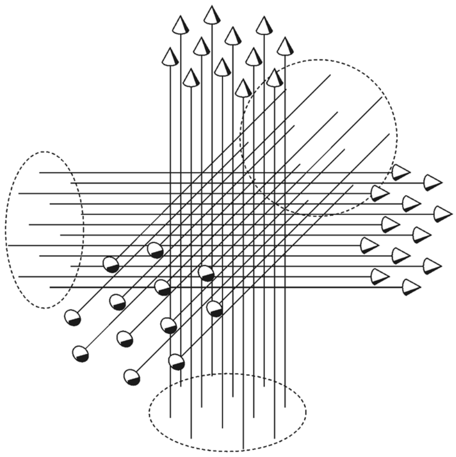
7.1. Half-sensitivity volume as the source model

Let us start the discussion on the ability of EEG and MEG detectors to concentrate their measurement sensitivity by discussing the sensitivity of a single surface electrode. In the brain region its sensitivity is the highest just under the electrode. Let us assume that the brain is a homogeneously distributed source, i.e., throughout the brain the neuronal sources have the same probability to be activated at any time and in any direction. In such situation most of the signal comes from the region where the sensitivity is the highest, i.e., from under the electrode. The faster the sensitivity decreases as a function of the distance from the electrode, the smaller is the region from where the signal comes, i.e., the better is the spatial resolution. To find a relationship between the fall-off of the sensitivity as a function of distance and the spatial resolution, we define *the half-sensitivity volume,* HSV: *HSV is the volume of the source region in which the magnitude of the detector sensitivity is more than one half of its maximum value in the source region* [6]. The smaller the HSV is, the smaller is the region from which the signal of the detector originates. The HSV concept concerns primarily the spatial resolution on the surface of the brain. In this paper we do not discuss the detection of deep sources.

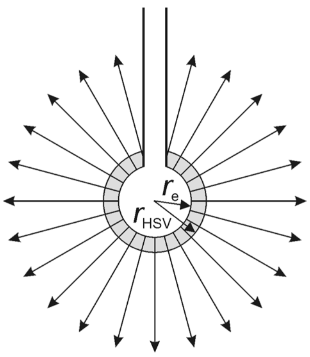
To clarify the concept of HSV we give some examples.

7.2. Dipolar leads

A dipolar lead, like the *x, y* and *z* leads of vector cardiography, has homogeneous sensitivity in the direction of the coordinate axes throughout the source region (Fig. 3a). The HSV of a dipolar lead is the whole source region and recording such dipolar leads gives no information on the source location.



(a)



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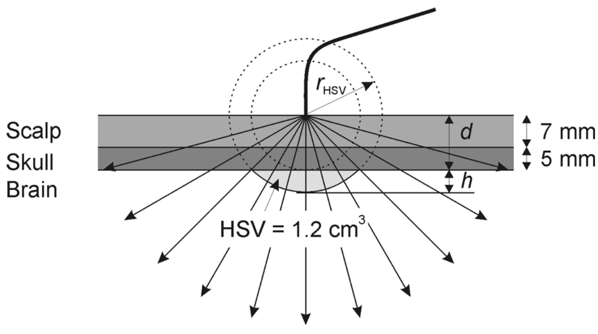


(c)

Fig. 3. Examples of the lead fields and HSVs for different lead configurations: (a) a dipolar lead has homogeneous sensitivity and HSV equals the whole source region; (b) for a deep electrode HSV $\cong 12 r_e^3$; (c) if a point electrode is on the surface of the scalp and the head is homogeneous, HSV $\cong 0.688\, d^3$; with the scalp and skull thickness of 1.2 cm, HSV is approximately 1.2 cm^3.

7.3. Deep electrodes

When recording the electric activity of the brain with an electrode, which is in the brain region, the lead field current density decreases and therefore the measurement sensitivity decreases proportionally to the square of the distance from the electrode centre (Fig. 3b). If the spherical electrode tip radius is r_{e}, its surface is $4\pi r_{\mathrm{e}}^2$ at the distance r_{HSV}, where the surface of the sphere is doubled and the sensitivity is one half of that on the electrode surface, we have

$$4\pi r_{\mathrm{HSV}}^2 = 2\cdot 4\pi r_{\mathrm{e}}^2, \quad r_{\mathrm{HSV}} = \sqrt{2r_{\mathrm{e}}}, \tag{4}$$

where r_{HSV} is radius of the HSV sphere segment.

HSV is then

$$\frac{4}{3}\pi r_{\mathrm{HSV}}^3 = \frac{4}{3}\pi(\sqrt{2r_{\mathrm{e}}})^3 = 11.84 r_{\mathrm{e}}^3 \cong 12 r_{\mathrm{e}}^3. \tag{5}$$

If the electrode tip radius is 1 μm or 1 mm, HSV is of the order of 12 μm^3 or 12 mm^3, respectively. If the electrode is located on the surface of the cortex, HSV is half of that.

Note that for the deep electrode Eq. (5) gives the total HSV including also the electrode. To indicate accurately the half-sensitivity source volume, the electrode volume in that region must be subtracted. The electrode occupies a volume of

$$\frac{4}{3}\pi r_{\mathrm{e}}^3 = 4.2 r_{\mathrm{e}}^3.$$

Thus HSV in the brain region is $7.8 r_{\mathrm{e}}^3$.

7.4. Model of the homogeneous head

Assume that the scalp, skull and brain have the same resistivity. We approximate the head with a half-space model (Fig. 3c). Assume that a point electrode is on the surface of the scalp. The maximum sensitivity of this electrode in the brain region is on the cortex just under the electrode. On the surface of a sphere with radius $\sqrt{2}d$, where d is the thickness of the scalp and skull, the sensitivity is one half of the maximum sensitivity. HSV is the sphere segment whose volume is

$$\mathrm{HSV} = \frac{1}{3}\pi h^2(3r_{\mathrm{HSV}} - h), \tag{6}$$

where h is the height of the sphere segment.

Inserting into Eq. (6) $r_{\mathrm{HSV}} = \sqrt{2}d$ and noting that $h = r_{\mathrm{HSV}} d$, we obtain

$$\mathrm{HSV} = \frac{1}{3}\pi(4\sqrt{2} - 5)d^3 \cong 0.688 d^3. \tag{7}$$

In the Rush–Driscoll model $d = 1.2$ cm. Then $\mathrm{HSV} = 0.688 \cdot 1.2^3\ \mathrm{cm}^3 \cong 1.2\ \mathrm{cm}^3$.

7.5. Inhomogeneous model of the head

In the inhomogeneous model of the head the high resistivity of the skull spreads out the EEG lead field and therefore HSV will increase. Because of the tangential direction of the MEG lead fields, the high resistivity of the skull does not have any effect on them. In the next chapter it will be shown with theoretical calculations that even though the relative resistivity of the skull were 80/1, as predicted by Rush and Driscoll, HSV of EEG is of the same order as that of MEG. According to the present information, the skull resistivity is much lower and therefore EEG has better spatial resolution than MEG.

8. METHOD

8.1. Model of the head

For the head we used the Rush–Driscoll model with concentric spheres of 80, 85 and 92 mm radii for the outer surfaces of brain, skull and scalp, respectively [7]. Like Rush and Driscoll, we assumed that the brain and the scalp have the same resistivity. We will first repeat our earlier results [8] from the calculations where we used the relative resistivity of the skull equal to 80 times that of scalp and brain tissues, as predicted by Rush and Driscoll.

Because the relative value of 80/1 of the skull resistivity of Rush and Driscoll has recently been seriously questioned [9,10], we recalculated the results with the resistivity values of 5, 10, 15, 20, 40 and 80 times that of the brain and scalp. This gives the reader the possibility to evaluate the HSVs of EEG and MEG with a resistivity value which is considered to be the correct one.

8.2. EEG and MEG leads used

We used in our calculations two- and three-electrode leads for EEG and axial and planar gradiometer leads for MEG (Fig. 4, Table 1). Due to the different nature of the electric and magnetic detection, various electrode and coil configurations are not exactly comparable. The bipolar EEG lead is the basic EEG lead. With short baselines it has tangential sensitivity. With long baselines it may be considered unipolar and it has radial sensitivity. With long baseline the bipolar (unipolar) EEG corresponds to the single coil magnetometer or axial gradiometer. That has vortex-form sensitivity. The bipolar EEG with short baseline corresponds to the planar gradiometer which has linear tangential sensitivity.

The three-electrode EEG is discussed because it generates a radial lead field also with short baselines. The magnetic equivalent for the three-electrode EEG system is the three-coil or second-order planar gradiometer. The discussion is not extended to that level. It is important to note that radial lead fields are not at all possible to generate with magnetic measurements.

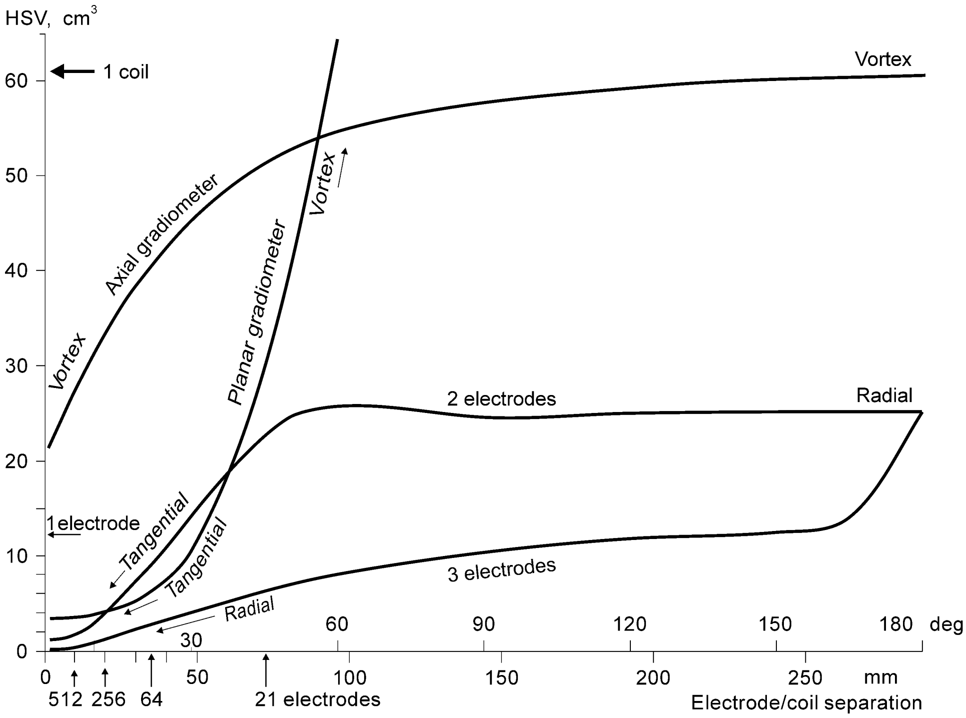


Fig. 4. The HSVs of two- and three-electrode EEG leads and axial and planar gradiometers as a function of electrode distance and gradiometer baseline; the gradiometers have a measurement distance of 20 mm from the scalp and coil radii of 10 mm; arrows indicate the electrode distances for a EEG systems with 21, 64, 256 and 512 electrodes having electrode distances of 72, 35, 20 and 10 mm, respectively; the head model is the original Rush–Driscoll model with relative skull resistivity of 80/1; the total source volume (brain) in the spherical model is 2140 cm^3.

Table 1. Correspondence between electric and magnetic detector configurations and their sensitivity distributions with short and long baselines

EEG			MEG		
Configuration	Sensitivity with baseline		Configuration	Sensitivity with baseline	
	Short	Long		Short	Long
One-electrode		Radial	Single coil		Vortex
Two-electrode	Tangential	Radial	Planar Gradiometer	Tangential	Vortex
Three-electrode	Radial	Radial	Second order Planar Gradiometer		

9. RESULTS OF THE HSV CALCULATIONS

9.1. Original Rush–Driscoll model

Our earlier results of the HSV calculations for one-, two- and three-electrode EEG leads and single coil and planar gradiometer MEG leads in the original Rush–Driscoll model are repeated in Fig. 5 [[6]].

For EEG, HSV was calculated for bipolar and three-electrode leads with point electrodes as a function of the electrode distance (Fig. 5, a and b). These have tangential and radial sensitivities, respectively. For MEG, HSV was similarly calculated for axial and planar gradiometers as a function of the gradiometer baseline (Fig. 5, a and b). The axial gradiometer has tangential (vortex) sensitivity. The planar gradiometer has linear sensitivity. The radii of the MEG coils were 10 mm and their distance from the scalp 20 mm (Fig. 5, c and d). The baseline in Fig. 4 varied from 0 to 180 degrees.

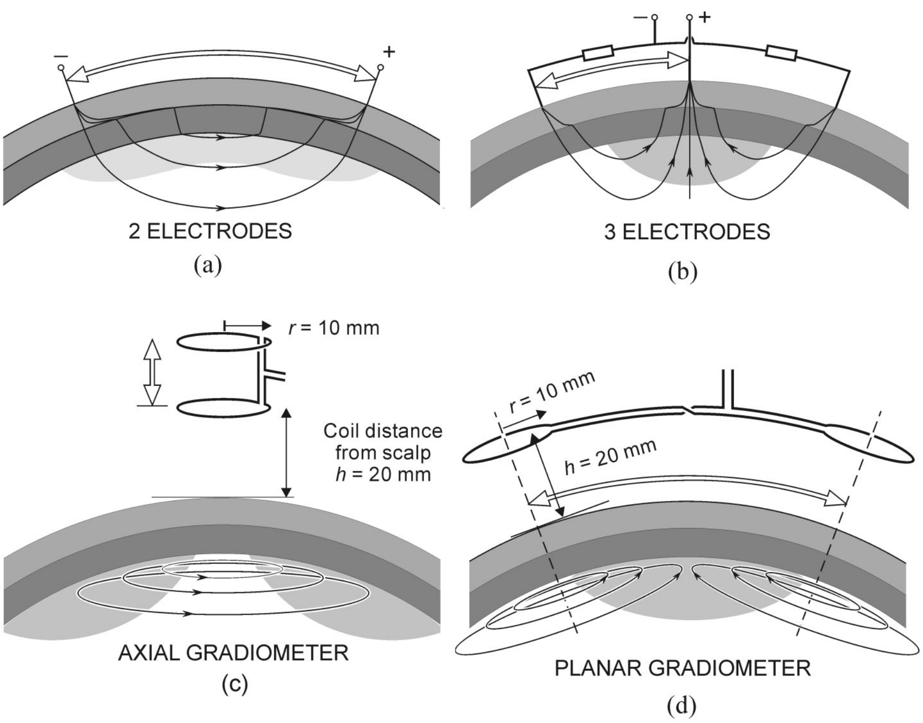
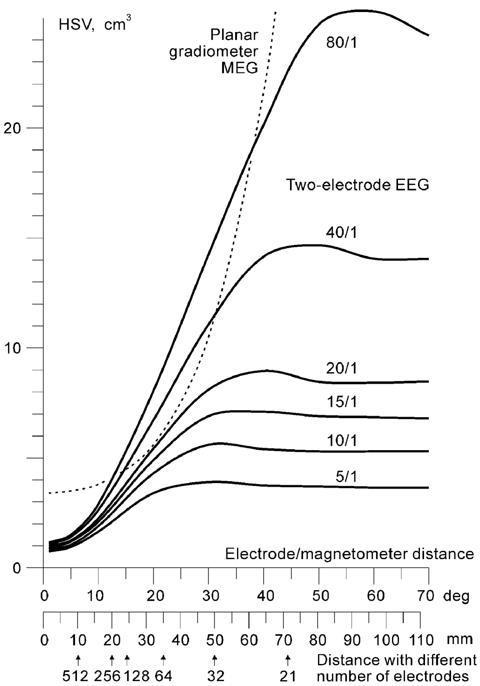


Fig. 5. Measurement configurations and dimensions for electric and magnetic leads: (a) two-electrode electric lead; (b) three-electrode electric lead; (c) axial gradiometer lead; (d) planar gradiometer magnetic lead; HSVs are shown in light gray colour; note the tangential nature of the sensitivities of the two-electrode electric lead and planar gradiometer magnetic lead; the three-electrode electric lead has radial nature.

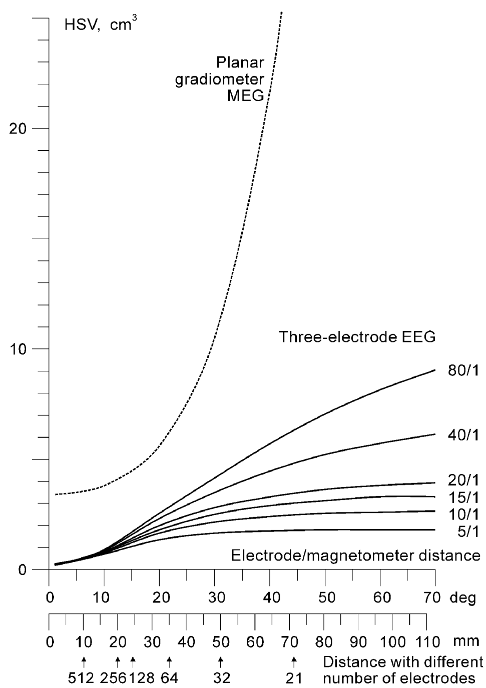
9.2. Relative resistivity of the skull as a parameter

We have calculated HSVs for two- and three-electrode EEGs with relative resistivity of the skull of 5/1, 10/1, 15/1, 20/1, 40/1 and 80/1 as a function of the electrode distance. The results are shown in Fig. 6. The skull resistivity has no effect on the HSV of the MEG. The HSV for planar gradiometer MEG is repeated here. The results are given within the interesting baseline area 0–110 mm. It can be observed that with the realistic resistivity values for the skull, 5/1, 10/1 and 15/1, the HSV of the EEG is smaller than that of the MEG with all values of electrode and magnetometer distances.

The most interesting electrode distance is 20 mm, corresponding to the 256-electrode high resolution EEG system. In this region the HSV of the bipolar EEG is about 50 % smaller than that of the planar gradiometer MEG.



(a)



(b)

Fig. 6. Half-sensitivity volumes of two- and three-electrode EEG and planar gradiometer MEG as a function of electrode/magnetometer distance, respectively; the relative skull resistivity is 5/1, 10/1, 15/1, 20/1, 40/1 and 80/1 for skull/brain and scalp; the electrode distances for EEG lead systems with different numbers of leads are also indicated.

10. MAXIMUM AMOUNT OF ELECTRODES IN EEG

As can be seen from Fig. 6, the HSV calculations indicate that the number of EEG electrodes may be increased up to 500 and more and still more information from the electric activity of the brain can be obtained.

We have studied this issue also with another method to confirm this result [11]. As the model of the head, the Rush–Driscoll model was used but with a relative skull resistivity of 15/1. Singular value decomposition was used to evaluate the spatial resolution with various measurement noise estimates. The results suggest that as the measurement noise increases, the advantage of dense electrode systems is decreased. With low realistic measurement noise a more accurate inverse cortical potential distribution can be obtained with an electrode system, where the distance between two electrodes is as small as 16 mm, corresponding to as many as 256 measurement electrodes. In optimal noise situation over 500 electrodes give more information on the electric field of the brain.

11. PRACTICAL ISSUES IN THE APPLICATION OF EEG AND MEG

The maximum number of EEG electrodes is in practice limited by the electrode cap construction and electrode positioning. Various electrode cap constructions have been developed but they all suffer more or less from contact problems. Good contact between the electrode and the scalp is obtained with careful gel insertion and sufficient cleaning of the contact region (Fig. 7). Fast development of digital cameras and the decrease of their price have made the photogrammetry an attractive solution for accurate electrode localization.

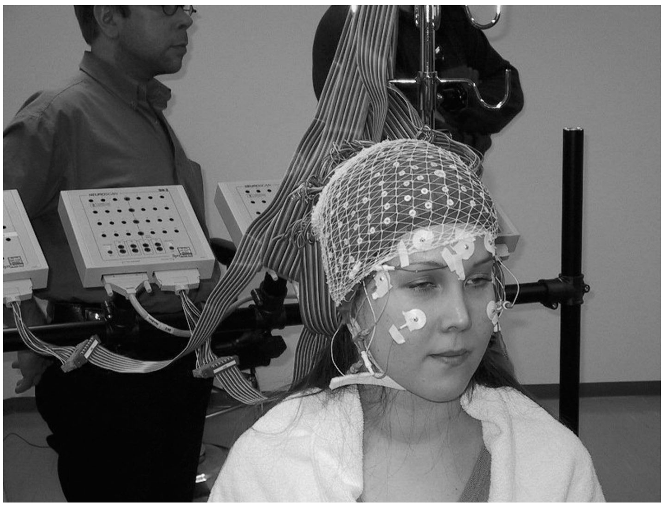


Fig. 7. Application of the 256-channel EEG; the electrodes are fixed to a cap which is quickly placed on the subject's head.

A great benefit of MEG is the ease of its application. Placing the dewar over the head of the patient is easy. The location of the detector coils is well known and their situation relative to the head can be identified with calibration coils placed on the scalp. One limiting factor in the application of MEG is that the patient must keep the head fixed in the same position during the whole measurement session. MEG is also impossible to apply simultaneously with MRI or other imaging methods. The requirement for low level of the magnetic noise limits the location of the MEG installations in hospitals. Also the price of the MEG installation is one or two orders of magnitude higher than that of the EEG installations.

12. DISCUSSION

Several studies on the comparison of the spatial resolution of EEG and MEG have been published. Liu et al. have recently made an excellent review of these studies [12]. In this paper they also published their own results on the subject. They studied the spatial resolution of EEG and MEG for a distributed-source model with the Monte Carlo method. Though they used in their realistic head model for the skull the resistivity ratio of 80/1, they found that the localization of EEG is more accurate than that of MEG.

The HSV method is one of the very few methods, used for the comparison of relative merits of EEG and MEG, which gives numerical results. How accurately they can be applied in practice is another issue. The real practice is much more complicated than the spherical model we used. However, by comparing these HSV calculations to other theoretical investigations and practical measurements, there is no doubt that these results describe the relative merits of EEG and MEG in the spatial resolution on the surface of the brain with sufficient accuracy.

13. CONCLUSIONS

EEG has better spatial resolution than MEG with all electrode and magnetometer distances. It is possible to increase the spatial resolution of EEG by increasing the number of electrodes up to 250. Even doubling the number of electrodes from this improves, in optimal noise situation, the spatial resolution. The results of this article should encourage the users of EEG to improve their measurement configurations because the theoretical limits of the spatial resolution of EEG are not reached by the standard clinical 10-20 system. The spatial resolution of the EEG method is far better than estimated until now.

ACKNOWLEDGEMENT

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EEG ja MEG võrdlemine

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Artiklis on võrreldud elektro- (EEG) ja magnetentsefalograafiat (MEG). On näidatud, et EEG ruumiline eraldusvõime on kõrgem kui MEG-l.