Measurements and a Simplified Interpretation of Magnetocardiograms from Humans

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SUMMARY

To understand and assess the magnetocardiograph, the magnetic field produced by electrical heart activity was mapped around the torsos of one abnormal and six normal heart subjects, at different times during QRS. The measurements are here presented in the form of instantaneous distributions of vector segments 6 cm from the skin, anteriorly. An electrical model is presented which explains the first-order features of these distributions. This model is a simple, rotating bipole heart source of changing strength which produces the same ion currents as measured with surface electrocardiograms and vectorcardiograms; these currents produce a time-changing magnetic field extending outside the torso. Further analysis of second-order details of the distributions can reveal information not obtainable with various forms of surface electrocardiography.

Additional Indexing Words:
Magnetic field  B-vector  Ion currents  Rotating bipole  Electrical model

RECENT ADVANCES in magnetic techniques and instrumentation have made it possible to detect the very weak magnetic fields produced by natural ion currents in living material. Examples of such ion currents in humans are the alpha-rhythm currents in the head which produce part of the electroencephalogram, currents from skeletal muscle activity which produce the electromyogram, and currents from myocardial activity which produce the electrocardiogram.

In brief review, one way to understand the magnetic field concept is by comparison with the more familiar electric field. At any point in space the electric field (or its close relative, the voltage) is a measure of the force on a test electric charge at that point; the source of this field is a distribution of charges in space, and the electric field involves only the strength of these charges and their positions. If these charges are in motion, then there is also another field of force present, called the “magnetic field.” The magnetic-field vector at any point, called the “B-vector,” is the force on a test current at that point. The magnetic field involves only the flow of charges and the positions of the flow movements. Stated otherwise, a group or distribution of source currents produces special forces distributed in space called the “magnetic field”; these forces act only on any currents located within this field. The magnetic properties of some materials can be considered to be a result of particular currents within the materials; each crystal of iron, for example, in a sense contains a large net-circulating current made up of aligned electrons spinning about their own axes.

Magnetic fields produced by ion currents in living material were first detected by Baule and McFee. They recorded, at the human chest, the fluctuating field produced by the heart, and took the first magnetocardiograms (MCGs). The MCG was found to have a maximum of \(-5 \times 10^{-7}\) gauss at the

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peak of QRS; this is \( \sim 10^{-6} \) of the earth’s steady magnetic field and \( \sim 10^{-3} \) of the fluctuating or a-c magnetic background in an urban environment; hence the major experimental difficulty was the extraction of this small signal out of the background noise. This difficulty was partly overcome by their design of the detector, which consisted of two identical side-by-side coils, each with 2,000,000 turns and an axial ferrite rod. At each coil, any small, changing magnetic field induced a small voltage across the coil terminals. The coils were connected in opposition so that the magnetic background noise, assumed to be uniform, was cancelled, while the heart induced a net voltage because its proximity produced a local magnetic gradient. This arrangement was then a gradient detector, the output of which was fed to a selected vacuum tube amplifier. The Baule-McFee experiments were later repeated and confirmed by a group from the USSR. The measurements by both groups were limited by the complexity of gradient detection, imperfect coil cancellation, and field distortion due to ferrite which was necessary to raise the heart signal above the amplifier input noise.

These limitations did not exist in our direct measuring system made possible by new magnetic shielding techniques and the low-frequency parametric amplifier. The subject and detector were housed in a triple-layer shielded room which reduced the background below the detector noise level. The detector consisted of a single coil feeding the essentially noiseless parametric amplifier; therefore, one component of the field vector was measured directly, instead of the gradient. This system was first used to verify and measure a few simple properties of the human heart’s magnetic field. We then undertook a more detailed magnetic study of the heart which was followed by the detection of the much weaker fields from the alpha-rhythm currents in the human head. The more detailed cardiac study consisted of mapping the fields around the torsos of a number of normal and abnormal subjects.

This article presents some of the measurements of this study, and some of the more obvious interpretations; a paper with extensive evaluation and interpretations will appear later.

A major reason for studying magnetic fields from various human organs, such as the heart and the brain, is the limitation of surface potential measurements. These are made on the surface of a living volume, in both research and diagnosis, to gain internal information, as with the ECG, the EEG, and the EMG. Such measurements yield information about the internal charge distribution, hence about polarized tissues such as muscles and nerves. Unfortunately, the internal charge distribution cannot uniquely be determined by instantaneous surface potential measurements. In electrocardiology this means that even with the most elaborate lead systems there will always be some indeterminacy or “error.” The combination of both potential and magnetic surface measurements also yields no unique distribution. In theory, however, surface magnetic measurements can yield internal information not possible with surface potential measurements alone. Hence, it was worth while to make some detailed magnetic measurements on a biological system which is understood and which has had extensive surface potential mapping—the human heart. Ultimately, the interpretation of these heart magnetic measurements could show whether the benefits of magnetic measurements which are theoretically possible can be conveniently realized in practice. The first step in such an interpretation is to show the proper connection between the surface magnetic and surface...
potential measurements; that is, between the MCG and the ECG. This connection involves a “first-order” electrical model by which the internal currents, which produce the surface potentials, also are involved in the production of the heart’s external magnetic field. In this report the interpretation of the experimental data proceeds only to the inter-
mediate point of showing this connection and verifying the model which explains the first-order features of the heart’s magnetic field. The second-order details of the field’s measurements presumably contain information not detectable with the ECG and are now undergoing analysis and evaluation.

MCGs, which superficially resemble ECGs are records of one component of the magnetic vector versus time at one position. MCGs were taken of the heart subjects at many front, back, and side positions around the torso. This report, which is only intended as an introduction to the technique of magnetocardiography, is restricted to presenting data from only 49 positions confined to a plane about 6 cm in front of, and parallel to, the chest and are about 5 cm apart; the data are also restricted to one normal and six normal heart subjects. It was felt that the inclusion of more data would detract from the first-order presentation and would unnecessarily complicate this report.

Methods

The detector was an air-core 200,000-turn coil, about 8 cm in diameter by 5 cm. To prevent field distortions, no flux-gathering ferrite core was used; as a result the MCG signal was smaller, and a signal-averager was used to reduce noise. Figure 1 shows the experimental arrangement. Three MCGs were taken at each of the 49 positions shown in figures 3, 4, and 5. With the center of the coil fixed at a point, an MCG was recorded for each of three perpendicular orientations of the coil axis, corresponding to

![Figure 1](image1)

Arrangement for obtaining the MCG and the ECG simultaneously. The subject, detector, and parametric amplifier (Texas Instruments RA3A) are in the shielded chamber; other units are at an external station. The noise-averaging Enhancetron made 100 additive 0.5-
sec sweeps for each measurement, triggered at QRS onset from the ECG. The magnetic channel band-pass was 0.5 to 100 hz; this introduced delay of the MCG relative to the ECG but eliminated troublesome 180-hz magnetic background.

![Figure 2](image2)

First step in processing MCG data. This MCG was taken with the coil axis normal to the chest, hence the ordinate is proportional to $B_n$, the normal component of the magnetic vector, here evaluated at 0-time; the 6-msec delay with respect to the wide-band ECG is due to the low-pass filter.

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the magnetic vector components vertical, horizontal, and normal to the chest. Simultaneous lead II ECGs were taken, on which the R-peak of QRS was defined as 0-time, thereby setting a time scale for the MCGs; the details are shown in figure 2. Magnetic amplitudes were evaluated at -16, -8, 0, +8, and +16 msec during QRS. Hence, at each of the 49 points there were three numbers at each of these times. A particular format was chosen to display this large amount of data instructively; for each subject, at each time a separate structure or model was made. The purpose of each model was to show 49 B-vectors, one at each of the 49 points in the plane of measurement. The B-vectors were made by fastening wire segments into a plastic sheet, as seen in figures 3, 4, and 5. Each wire was aligned in angle by using the three numbers to indicate its projection onto the three perpendicular axes. Experimental errors due to noise, data-handling, and others show up as angle misalignment and length error for each wire. Examination of each model qualitatively shows the extent of error from the angular and length "jitter" from wire-to-wire. In most cases the curves in the broken turnover lines are beyond experimental error and are real. The dotted heart outlines were taken from chest x-rays of each subject, who also had physical examinations, ECGs, and VCGs.

Results

The experimental results are displayed in the models of figures 3, 4, and 5. The normals, so defined because their ECGs and VCGs were within normal limits, have B-vector distributions with the same general features. The magnetic fields have a maximum at 0-time at a position somewhat below the heart of about $5 \times 10^{-7}$ gauss. The broken turnover lines are irregular and usually displaced several centimeters from the

Figure 3

Front and side photographs of the same display showing measured magnetic vector segments 6 cm from the chest of a subject with a normal heart during QRS peak. The white wires are embedded in a lined plastic sheet on the reader's side, black wires on the back side. The wire lengths are proportional to the vector amplitudes; each ring is $5 \times 10^{-4}$ gauss. The front view shows the projected heart position and a broken line where the segments turn over or reverse polarity. The side view shows an artist's construction of four B-vectors, as if measurements had been made continuously along those segments involved, at distances other than 6 cm.
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Heart center toward the left shoulder; they are generally concave-upward. As seen in figure 5, the turnover line rotates during circulation, volume XXXIX, March 1969.

QRS and the rotation is usually counterclockwise in normal subjects. The clinical details of the abnormal heart will not be discussed.

Figure 4

B-vector displays of five other normals and one diseased abnormal (W. B.). The normals show the same general, first-order features. The abnormal scale has been doubled because of weaker magnetic fields. Missing segments of some displays are due to unsatisfactory data. There are second-order differences among the normals.
Figure 6
Simple electrical model showing the heart producing a magnetic field outside the torso at 0-time. The normal heart is a rotating bipole “battery” producing ion currents (dashed ovals) in the torso volume conductor; these produce magnetic B-vectors outside and normal to the torso (solid curved lines). A segment near an end of a solid line corresponds to a black or white segment in figures 3, 4, and 5. The heavy dashed line joining turnover points always has the same direction as the bipole frontal projection and corresponds to the broken turnover lines in figures 3, 4, and 5.

Figure 5
B-vector displays of a normal subject at four different times during QRS. Turnover line rotates counterclockwise as in most MCGs, in contrast to clockwise rotation in most VCGs of its analogue, the frontal plane projection.
Discussion

There is a simple model, shown in figure 6, which connects the surface electrocardiogram with the magnetocardiogram and which explains the first-order features of the normal distributions. This model was initially suggested and used by Baule and McFee although they did not have the experimentally measured distributions to verify the validity of the model. The heart current source is a rotating dipole (a clump of positive charge separated by a finite distance from an equal clump of negative charge) with a fixed center, schematically shown as a battery; it rotates in three dimensions, changing in strength during rotation and is shown at 0-time with the negative pole toward the right shoulder. This dipole produces both positive and negative ion current loops in the torso volume conductor, where the arrows indicate the direction of negative ion flow. These are essentially the assumptions of vectorcardiography. Now let the magnetic field, shown as solid, curved lines, be produced by these same ion currents; if the heart is situated anteriorly, then this field will extend out of the torso and have roughly the same distribution as the normals of figure 4. As pointed out by Baule and McFee, the position of the heart within the torso is most important for magnetic considerations, as in this example: let there be a current source at the center of a spherical volume conductor, then even though there will be ion current loops extending to the surface, there will be zero magnetic field everywhere outside the sphere; this is because of perfect cancellation of fields from the symmetric current loops. If the heart were small and at the center of a cylindrical, homogeneous torso, the same would be true. The reason for the existence of the heart’s field outside the torso is the nonsymmetric, anterior position; the more anterior the heart, the more the situation approaches the condition of a current source near the surface of a semi-infinite volume conductor. In this case the external magnetic field is large if the current source is parallel to the surface, and zero if perpendicular, by symmetry arguments. Hence, the component of the dipole which is perpendicular to the chest gives zero external frontal magnetic field and can be ignored. The turnover line in figure 6 therefore is related only to the dipole frontal-plane projection and rotates with it. The model readily shows the connection between the ECG and MCG, at least to first order. Consider the lead II ECG, for example; in effect it samples the largest right-side current oval of figure 6 by measuring the potential differences between the top and bottom of this oval. This particular oval is heavily weighted in the magnetic field production (normal component) 6 or 7 cm above and perhaps to the right of the navel so that the lead II ECG should look like the normal MCG at that place. Furthermore, from the magnetic lines of figure 6 it is seen that the normal MCG near the left shoulder should be similar, but of reversed polarity. This turns out to be true experimentally for all normals. Four such cases are shown in figure 7; these comparisons tend to confirm the validity of the model.

Figure 7

Comparison of particular ECGs and MCGs from four normals, in agreement with predictions of the model. All 12 traces are noise-averaged outputs after 100 sweeps 0.5 sec long. The top tracing for each subject is the lead II ECG, the middle tracing is the normal component MCG somewhat above the navel and the lower tracing is the normal component MCG near the left shoulder.
First-order features of the measured B-vector distributions can be understood from the model in the following way. If the torso was a semi-infinite volume conductor, then the turnover line would be located directly over the heart, and rotate with the frontal-plane projection of the heart dipole. It can be shown by electromagnetics that at 0-time the actual torso geometry, or presence of boundaries, must shift the turnover line toward the left shoulder and curve it concavely upward, as seen in figures 3 and 4. The usual counterclockwise rotation of the turnover lines in the MCG, as opposed to the usual clockwise rotation of the frontal-plane projection in vectorcardiography, also involves the torso geometry. In vectorcardiography the normal frontal-plane loop is very narrow; hence only slight changes in the measuring or lead system could tilt the loop, converting it from counterclockwise to clockwise and vice versa. Both the inherent deviations of the vectorcardiographic lead system and slight boundary effects in magnetocardiography can easily contribute to tilting the three-dimensional loop made by the rotating dipole; only further analysis will tell which measuring system is more accurate or useful as far as the direction of rotation is concerned. Further analysis of the second-order features will also yield the eventual assessment of magnetocardiography.

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References

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