diffusion pathway contains an additional mesophyll resistance, equal in magnitude to the other resistances. This should result in transpiration being reduced more than is photosynthesis when an extra resistance—the plastic film—is placed in the diffusion pathway. This should be so even if the film has a $P_{H_2O}:P_{CO_2}$ ratio of 1.

However, under conditions of water stress, such as may be induced by low levels of soil moisture or by high evapotranspiration potential, film-type antitranspirants may in fact increase photosynthesis relative to that in untreated plants growing under the same conditions. The theoretical basis of this interaction, which has been reviewed (*1*), was demonstrated in the laboratory (Fig. 1); the experiment showed the effect of stress, induced by elevation of the air temperature, on the photosynthesis of bean leaves.

Figure 1 shows that under mild conditions (25°C and high humidity) photosynthesis in the treated leaf was 20-percent lower than that by the control. When the temperature was raised to 35°C, photosynthesis in the control fell to a lower level than that in the treated leaf. When the air temperature was returned to 25°C, photosynthesis in the control rose above that in the treated leaf even though the rate of photosynthesis did not return to its former value in either plant. These data are preliminary; more detailed but similar results will be reported.

We conclude that whenever treatment with an antitranspirant lowers the *T:Ph* ratio, the effect is not a result of the film's selective permeability to gases but is indirectly due to the treatment's ability to reduce stress.

J. GALE
A. POLJAKOFF-MAYBER
*Department of Botany, Hebrew University of Jerusalem, Jerusalem, Israel*

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# Magnetic Fields around the Torso: Production by Electrical Activity of the Human Heart

Abstract. *A search was made outside the torso for fluctuating magnetic fields produced by the heart. Detector and subject were housed in a highly shielded enclosure. Magnetic signals with amplitudes of $10^{-8}$ to $10^{-7}$ gauss were detected synchronously with the electrocardiogram, confirming previous reports. A magnetocardiographic chest map, consisting of the magnetic field plotted against time at various spatial positions, shows general QRS and T-wave structure, as in the electrocardiogram; this structure varies with spatial position.*

Techniques for measuring weak, alternating, low-frequency magnetic fields have been developed so that, under certain conditions, fields as low as $\sim 10^{-9}$ gauss can now be detected. One interesting source of such weak, fluctuating fields is the small ion currents in living material. The most detectable of such currents would supposedly be produced by large masses of excitable, synchronously firing tissue, such as heart muscle, which produces the potentials and currents easily seen with electrocardiograms (ECG). If one were to explore the fluctuating magnetic fields produced by living material, he would perhaps begin with the human heart fields outside the torso. Such measurements are important because new, needed fundamental information about the heart's electrical properties may be obtained and because such measurements might have diagnostic value; the experience gained would be instructive in the extension of the techniques to measure the smaller fields of less synchronous muscles and of the nervous system.

The detection of human heart signals has been reported by two groups (*1*, *2*). I made a search for these signals using a different and more direct technique. Positive results were obtained, verifying the previous reports; in addition, some magnetic maps were made which showed changing patterns around the chest and the same temporal sequence of events as occurs in the ECG.

The amplitudes of the heart magnetic fluctuations were usually in the range of $10^{-8}$ to $10^{-7}$ gauss at 10 cm from the torso; these are $\sim 10^{-7}$ of the earth's steady field and $\sim 10^{-4}$ of the fluctuating background of earth plus typical city "noises," in a bandwidth of $\sim$ 1 to $\sim$ 30 cy/sec (always used in this study). The major problem in detection of heart signals is therefore to somehow reduce this background so that the heart signal predominates. Baule and McFee (*1*) used the technique of gradient detection in which the detector consisted of two adjacent identical coils 30 cm long, 9 cm in diameter and each consisting of 2 million turns. The coils were connected in opposition so that the induced voltages from the spatially uniform magnetic background fluctuations were almost completely canceled; the induced voltages from the nearby heart were not canceled, and the voltage difference was then a measure of the rate of change with time of the gradient of the B-vector component parallel to the coil axes. The Russians (*2*) reported a similar gradient technique, except that the subject and detector were placed inside a shielded enclosure with 1.5-cm thick iron walls, which presumably reduced the background by a factor of perhaps 10. Both groups were successful in obtaining magnetocardiograms with the coils very close to the torso; Baule is currently interpreting many gradient measurements. My measurements were made inside a highly shielded enclosure which reduced the background by about a factor of 1000; this was sufficient so that two-coil canceling was no longer necessary, and only one coil could be used. The coil voltage was then a measure of the rate of change with time of the direct B-vector component.

The enclosure, described in detail elsewhere (*3*), consists of two nested cubical shells of 0.15-cm thick molypermalloy, 2.3 m on the inside. "Shaking" and negative feedback loops were used to enhance the shielding and bring the relatively low east-west background component down from the value of $\lesssim 5 \times 10^{-5}$ gauss r.m.s. (root mean square) obtained when there was no shielding whatsoever to $\lesssim 5 \times 10^{-8}$ gauss r.m.s. On magnetically quiet nights, the background was reduced below the resistive coil noise. The coil consisted of several units of 200,000 turns stacked in series. Each coil was about 5 cm long and 8 cm in diameter;

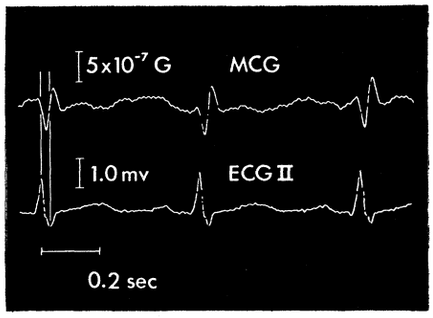



Fig. 1. Magnetocardiogram without noise averaging at same chest position of Mr. R.C. as Fig. 6, hN. The 15-msec delay is instrumental.

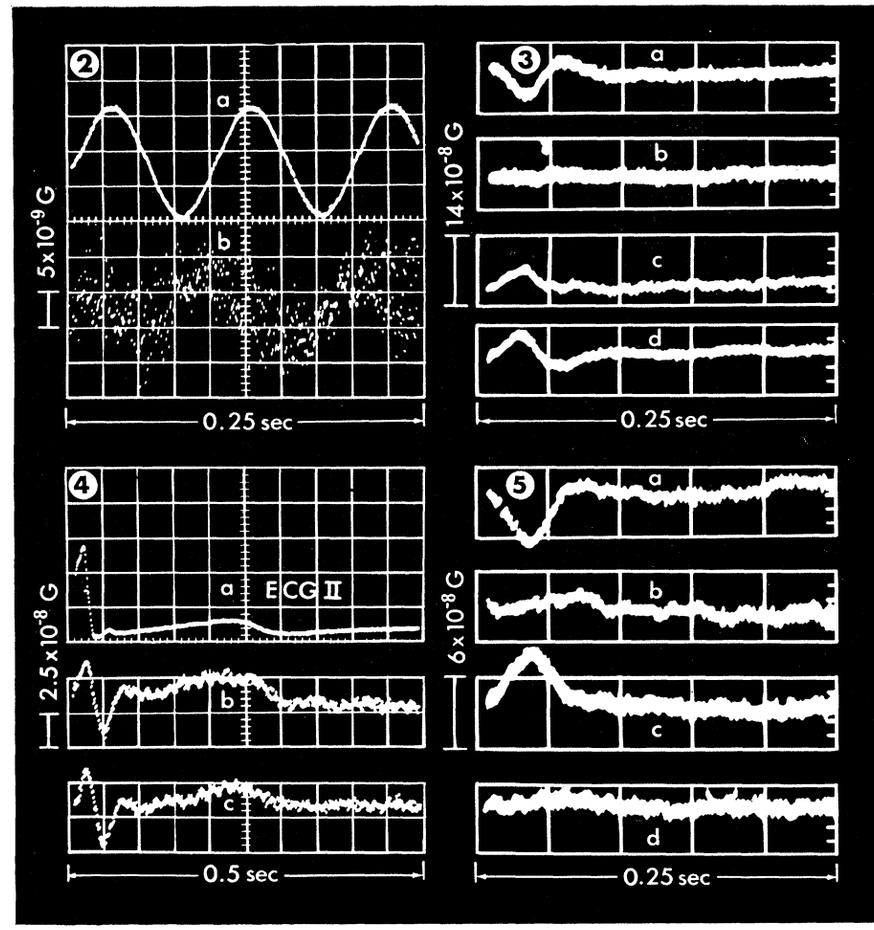
these were enclosed in an electrostatically tight brass cylinder fixed at the enclosure center in the east-west direction. Subjects were placed in various positions near the fixed cylinder. The coil output was connected to a Texas Instruments RA3 parametric amplifier several feet away which fed the display station outside the enclosure. The display station consisted of a variable-falloff band-pass filter (0 to ~30 cy/sec), an integrator, and these parallel display units: an oscilloscope, one channel of a two-channel fast chart recorder, and one of two channels of a Nuclear Data noise-averaging Enhancetron. The other channels were usually fed by a lead II electrocardiograph to allow simultaneous display with the subject's magnetocardiogram (MCG). The integrator reconverted the signal so that the output showed only the original B-vector component instead of the rate of change with time.

With a subject's chest wall several inches from the detector, a magnetic signal output was produced synchronously with the ECG. Repeated trials unambiguously showed that the signal: (i) rapidly decreased with increasing distance from the chest wall; (ii) was eliminated when a 0.036-cm molypermalloy shield was placed around the detector; (iii) was unchanged by a similar Cu shield; and (iv) was unchanged by disconnection of the ECG leads at the subject. The signal therefore must be positively identified as a B-vector produced by ion currents from somewhere within the torso; these currents must be directly or indirectly powered by the heart muscles.

Studies were made to calibrate and optimize the detector. The coils used by the other two groups were long because many turns were needed to raise the signal above the noise of conventional amplifiers, resulting in uncertainty as to "where" the gradient was measured. Their coils were wound on ferrite dumbbells; the heavy end-caps gathered and shunted flux through the coil centers, further increasing the signal, but also changing the field distribution. With my system the almost noiseless parametric amplifier permitted use of smaller coil signals and hence of a short coil and little or no ferrite. Tests with a heavy end-cap showed severe distortion of the magnetic fields, occasionally with reversal of the magnetocardiogram pattern.

As a compromise between sensitivity and spatial resolution, I used two coil units around a capless ferrite rod, for all data shown here. The heart signal could then be seen above background only close to a favorable subject, as in Fig. 1; this MCG contains unusually bad 60 cy/sec background from the shielding "shakers" superimposed on the random coil noise. Much 60 cy/sec background had already been removed by the band-pass filter which also produces the ~15-msec delay in all these data. Under the more usual conditions of weaker signals, or magnetically noisy nights, or both, signal averaging was used to reduce the noise about 12-fold, and Figs. 2–6 are Enhancetron output photographs after about 150 sweeps or superpositions.

The detector sensitivity was periodically calibrated with a small alternating current loop. Figure 2 shows a typical calibration on an average night; about half the noise is background, and half is intrinsic. The data of Figs. 3–5 show some MCG characteristics. The times of sweep, initiated on the QRS wave of the ECG, were never longer than 0.5 second because variation of the heart rate washed away




Figs. 2–5. Fig. 2. Calibration of detector at 10 cy/sec and $3 \times 10^{-9}$ gauss (r.m.s.), near sensitivity limit. (a) Calibrating loop current; (b) simultaneous detector output with 20 msec instrumental delay. Fig. 3. The MCG's of Mr. R.C. (a) Left front; (b) same as (a) except with magnetic shield; (c) right back; (d) left back. Fig. 4. The MCG's of Mr. R.D. (b) Left front; (c) same as (b) except with Cu shield. Fig. 5. The MCG's of Mr. D.S. (a) Left front; (b) same as (a) except with magnetic shield; (c) right back; (d) subject removed.

most detail, including P-waves, after about 0.25 second. The coil axis was normal to the torso which was always vertical and west of the detector for these figures. Figure 3a was taken 20 cm away, 10 cm below the left nipple. Figure 3b was run under identical conditions except the magnetic shield was slipped over the detector. Figure 3, c and d, was at 13 cm from the upper back; Fig. 3 complements Fig. 6. Figure 4, at 15 cm from the left nipple, shows no effect of the Cu foil shield; it demonstrates the similarity between ECG and MCG. Figure 5, a and b, taken at 20 cm from the left nipple, again shows the effect of a magnetic shield; Fig. 5c was taken at 20 cm from upper right back, and Fig. 5d at $> 70$ cm.

Figure 6 is a map of Mr. R.C.'s chest, showing some or all components at ten positions. No elaborate analysis has yet been made of the ion-current distributions producing these MCG's, and it is therefore too early to assess the long-range potential of the magnetocardiograph. The map is nevertheless instructive because it shows that (i) the MCG has the same origin as the ECG; that is, there is a QRS complex and a T-wave, and therefore the MCG currents must be produced by ventricular depolarization and repolarization, and so forth; (ii) the MCG has the same order of information content as the ECG in that the patterns change significantly with position. Since the MCG and ECG are powered from the same source, their patterns are thusly connected: at any instant the ECG is the difference between lead contact point potentials, where each is an integration of all charge/distance; crudely stated, the MCG is the integration of all current/(distance)$^2$, properly modified by vector directions. Because current involves moving charge, the MCG is closely connected through these integrals with the ECG rate of change. However, the MCG "samples the space" very differently and "knows the directions"; it is also insensitive to the medium between ion current and detector. Because of these fundamental differences between the ECG and MCG, the latter may eventually become a significant diagnostic or research tool after many patterns have been analyzed.

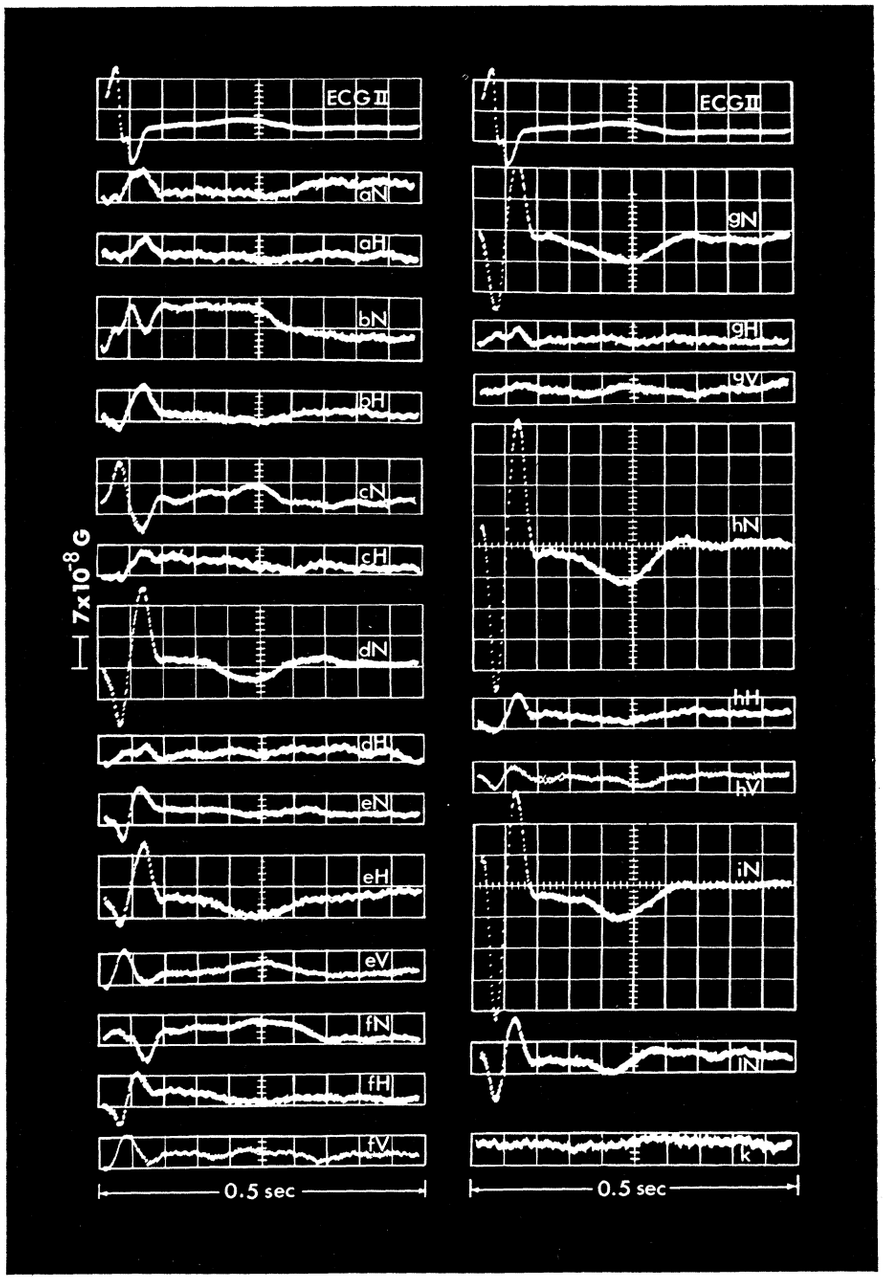



Fig. 6. Partial magnetocardiographic map with coil center always 10 cm from chest wall of Mr. R.C., except for k which was recorded at 70 cm. N, H, and V denote B-vector components pointing normally at chest, horizontally from left to right, and vertically up. Small letters denote these chest points on rectangular grid: (a) upper right corner; (b) upper center; (c) upper left corner; (d) right of right nipple; (e) mid-nipple; (f) left of left nipple; g, h, and i are ~ 7 cm below sites noted in d, e, and f; j is below g.


DAVID COHEN

*Department of Physics,*
*University of Illinois at Chicago Circle, Chicago*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