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Comparison of Cardiac and 60 Hz Magnetically Induced Electric Fields Measured in Anesthetized Rats

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Extremely low frequency magnetic fields interact with an animal by inducing internal electric fields, which are in addition to the normal endogenous fields present in living animals. Male rats weighing about 560 g each were anesthetized with ketamine and xylazine. Small incisions were made in the ventral body wall at the chest and upper abdomen to position a miniature probe for measuring internal electric fields. The calibration constant for the probe size was 5.7 mm, with a flat response from at least 12 Hz to 20 kHz. A cardiac signal, similar to the normal electrocardiogram with a heart rate of about 250 bpm, was readily obtained at the chest. Upon analysis of its spectrum, the cardiac field detected by the probe had a broad maximum at 32–95 Hz. When the rats were exposed to a 1 mT, 60 Hz magnetic field, a spike appeared in the spectrum at 60 Hz. The peak-to-peak magnitudes of electric fields associated with normal heart function were comparable to fields induced by a 1 mT magnetic field at 60 Hz for those positions measured on the body surface (where induced fields were maximal). Within the body, or in different directions relative to the applied field, the induced fields were reduced (reaching zero at the center of the animal). The cardiac field increased near the heart, becoming much larger than the induced field. Thus, the cardiac electric field, together with the other endogenous fields, combine with induced electric fields and help to provide reference levels for the induced-field dosimetry of ELF magnetic field exposures of living animals. *Bioelectromagnetics 18:317–323, 1997.*

Key words: ELF magnetic fields; dosimetry; electrocardiogram; endogenous fields; in vivo fields

INTRODUCTION

Extremely low frequency (ELF) magnetic fields interact with an animal by inducing electric fields inside the body. These induced fields represent the internal exposure or ''dose'' resulting from an external exposure [Kaune, 1992; Bracken, 1992]. We have developed and used a miniature electric-field probe to measure 60 Hz induced electric fields in homogeneous agar models [Miller, 1991]. This probe has also allowed some measurements of induced fields in rat carcasses [Miller, 1993, 1996]. The induced fields are proportional to the time derivative of the alternating current (AC) magnetic exposure field and generally decrease from a maximum at the body wall to zero at the center of the body. The induced fields are perturbed by membranes and conductivity variations within the body, producing a complex pattern of internal electric fields, which are typically less than expected from consideration of homogeneous models [Miller, 1996].

In living animals, a variety of natural endogenous electric fields also exist internally. These fields arise from normal physiological activity, and extend into adjacent tissue throughout the body. These endogenous fields are detectable at the surface of the body via skin electrodes and yield signals that are useful in medical diagnosis; for example, the electric field of the heart is detected as the electrocardiogram (ECG), of the brain as the electroencephalogram, and of muscles as electromyograms. In general, the endogenous electric fields have complex patterns in both space and time.

At any position in the body, the endogenous fields will combine by simple addition with any fields induced by external exposure to AC magnetic fields. Assessment of this combination for various exposure situations represents an integral part of the induced-field

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dosimetry of ELF magnetic field exposure. Recently, Wachtel (1992) has considered theoretically the fields of electrically active cells and concluded that these fields can be quite important for considerations of 60 Hz dosimetry. Bergeron (1993) focused on the heart as the dominant source of endogenous ELF fields, and noted that an ECG contains a significant 60 Hz component when analyzed theoretically by Fourier transform. Although the electrical signal associated with the heart function is well known, little information exists on local electric field levels for comparison to the magnetically induced field.

The purpose of this study was to extend our experimental methods to experimentally observe the complex electric fields in living (anesthetized) rats. The miniature electric-field probe detects any electric field present, and could therefore be used to simultaneously measure both endogenous fields and externally induced electric fields. The signals obtained were frequency analyzed for detailed assessment of the complex signals. These measurements allowed comparison of the relative magnitudes of induced fields to the normal endogenous fields. These results should help to provide a more complete picture of ELF induced field dosimetry.

METHODS

Large male Sprague-Dawley rats, which had been previously used in behavioral research, were obtained for this study. These were housed singly and were provided with water and food ad libitum. In preparation for measurements, a rat was anesthetized with ketamine (50 mg/kg) and xylazine (10 mg/kg) injected intraperitoneally and maintained with extra injections as needed during measurement sessions. After anesthesia was established, the animal was prepared by cutting 1-cm long holes in the skin and making small incisions (about 4 mm long and 1 mm, or less, deep) into the muscle wall of the chest and abdomen for contact with the tip of the probe. These areas were kept moist by periodic applications of 0.2 S/m saline (made with approximately 1 g/liter sodium chloride in distilled water), which roughly matches the average conductivity of animal tissue. Each of 10 animals was weighed; they averaged 560 g (57 g SD). Heart rate was determined from the cardiac signals obtained and averaged 252 bpm (41 bpm SD). This heart rate was somewhat lower than normal, which is about 330 bpm in old rats [Barnard, et al., 1974], but was consistent with the application of ketamine/xylazine anesthesia in rats [Hsu et al, 1986; Wixson, et al, 1987]. The heart rate appeared to be quite regular and stable, and there was no evidence of arrhythmias. For example, one rat had a rate of 242 bpm, which increased slightly to 253 bpm after an hour. The relatively slow heart rate was not considered to be a problem for this research because the local electric fields produced by the heart are not expected to be strongly dependent on heart rate (for example, there is little change in amplitude for heart rate reduction by anesthetic [Osborne, 1981] or increase by exercise [Bhargava and Goldberger, 1982]). A warming pad with circulating 37 °C water was used to help maintain body temperature, and rats checked with a rectal probe (RET-2 and TM-10A Clinical Thermometer, Sensortek, Saddle Brook, NJ) remained between 36 °C and 37 °C for at least an hour. All measurement procedures were accomplished within about an hour, at which time the animal was killed by injection of euthanasia solution (T-61 Euthanasia Solution, National Laboratory Corp., Sommerville, NJ).

The miniaturized probe used for measuring electric fields in conductive media has been described previously [Miller, 1991]. Briefly, the probe consisted of three silver electrode wires with chlorided tips imbedded in a 15-Ga hypodermic needle. The center electrode was grounded, and the differential voltage between the two outer electrodes was measured by using a lock-in amplifier (model 5210, Princeton Applied Research, Princeton NJ). The voltage across the probe was divided by its electrical size to obtain the electric field at the probe tip. The probe used for this study had a calibrated electrical size of 5.7 mm (actual diameter, 3.4 mm) and a flat response from at least 12 Hz to 20 kHz. As expected, the in vivo measurements obtained were somewhat variable. Part of this variation may have resulted from the conductivity differences between the tissue and the liquid surrounding the probe [Miller, 1994]. No correction was made for this potential source of variation because the tissue (primarily muscle) was bathed in 0.2 S/m saline (little blood was present), which minimizes the problem.

For measurements, the animal was placed on its side, together with a probe positioning gantry, in a magnetic field exposure system. This system consisted of four pairs of 40 by 50 cm coils of 8-Ga copper wire arranged to minimize leakage fields. Two pairs of 50-turn inner coils were spaced 15 cm apart, with two pairs of 100 turn outer coils placed symmetrically 50 cm apart. The wire turns were imbedded in potting compound and the coils electrically shielded with grounded copper foil. The coils were connected in series and energized by a power amplifier (model 7571 Techron, Elkhart, IN) using a signal from the lock-in amplifier. This system readily produced a 1 mT magnetic field at 60 Hz with no obvious heating or vibration. The field was uniform to within 2% over a 20 by 40 cm area of the specimen mounting plate. Both verti-

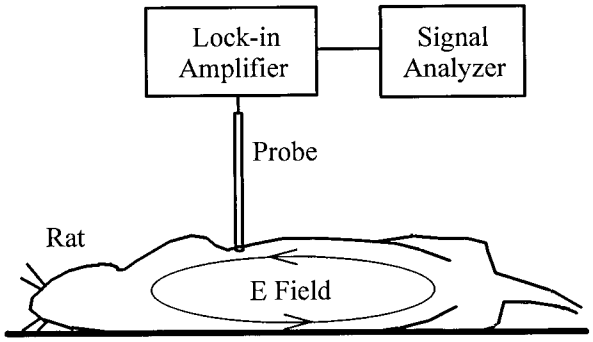

Fig. 1. An illustration of the probe placement relative to the rat lying on its side for frontal exposure with a horizontal field (i.e., oscillating in and out of the page). The three-probe electrodes within the probe tip were aligned with the induced electric field, which is shown as an elliptical current loop within the rat.

cal and horizontal orientations of the exposure field could be obtained by rotating the entire coil assembly by 180 degrees.

For a given measurement session, the probe was cleaned, chlorided, and calibrated. The exposure field was checked with an high-field field meter (Emdex-II, Enertech Consultants, Inc., Campbell, CA). The rat was prepared and placed into the exposure system. Measurements were then made at the body wall both at the mid-chest between the third and fourth ribs and at the abdomen approximately at the mid-point of the rat's body (excluding head and tail). The probe was directed vertically downward for axial and frontal exposure (horizontal field) or horizontally for lateral exposure (vertical field), approximately along a line through the body axis, and the tip was placed in the prepared incision. The electric field was measured in the expected direction of the maximum induced field; that is, in the direction of the body axis for frontal and lateral exposure, and in the perpendicular direction for axial exposure. For example, the configuration for frontal exposure is illustrated in Fig. 1. Both the endogenous and induced fields were detected simultaneously.

This measurement scheme represented a compromise that allowed consistent measurements to be made. For example, the probe placement maximized the induced field signal, while still yielding a strong heart signal. At the chest, moving the probe nearer the abdomen resulted in a very large signal related to the breathing motion, which precluded measurements of the induced and endogenous signals, while positions nearer the head were blocked by the foreleg. At some positions, the signal obtained with the probe appeared quite similar to an ECG signal (model EK-5A, Burdick, Milton, WI) obtained between the right foreleg and left hind leg (which was only useable with the exposure field off), when displayed on the dynamic signal analyzer. However, the miniature probe provides a calibrated measure of the local electric field, not simply the difference in electrical potential between the extremities, and also allows measurements with the 1 mT external exposure field applied to the animal.

For frequency analysis, an amplified signal from the lock-in amplifier (but not using the lock-in feature) was observed with a dynamic signal analyzer (Hewlett Packard model 3561A). Repetitive 1-s time traces were obtained, and frequency spectra of the time traces were analyzed over a 400 Hz frequency range with 1 Hz resolution (3.82 Hz equivalent noise bandwidth). AC coupling was used to avoid exaggerated DC signals (i.e., 0 Hz) and to reduce the baseline fluctuations resulting from the breathing motions of the animal. Ten spectra were root-mean-square (RMS) averaged to help smooth the final spectrum. This measurement arrangement was a compromise that allowed both a desirable frequency range and some resolution of the lower frequency components of the signal. The RMS magnitude of the induced 60 Hz signal was approximately constant for various time spans because it was a simple sine wave. However, with this measurement method, the 60 Hz component of the endogenous signal varied with the time span, because the signal was not coherent at 60 Hz. This variation was approximately inversely proportional to the square root of the time span; for example, if measured over a 250-ms heart beat interval, then the RMS magnitude of the 60 Hz component was about twice as large as for the 1-s time span results reported here.

Repetitive measurements were made in five rats with and without a horizontal exposure field for frontal and axial exposure, and in an additional five rats with and without a vertical exposure field for lateral exposure. After the measurements at the body wall, a small hole was made in the chest between two ribs (either the third and fourth, or the fourth and fifth) to allow the probe to be inserted near the heart (this hole tended to close, forming a seal around the probe, so that breathing was maintained). For this measurement, the probe was loosened from its holder to allow the probe tip to be maneuvered near the heart to maximize the signal. Measurements were made near the heart for frontal and axial exposure, but could not be performed for lateral exposure due to the arrangement of the rib cage and sternum. The probe could also be inserted into the abdomen, and scanned into the approximate center of the rat, but this gave poor results, as noted below. Data were gathered from the dynamic signal analyzer, which was in agreement with the lock-in amplifier display for lock-in operation (i.e., the 60 Hz RMS value of the induced field read the same on both

TABLE 1. Results for 60 Hz, 1 mT Magnetic Field Exposure in Frontal, Lateral, and Axial Directions Are Given in the Table as the Mean Electric Field in Microvolts per Centimeter (SD) for 5 Rats (6 Rats for Frontal Exposure to the Chest)

	Endogenous field		Induced field	
	Peak-Peak	60 Hz RMS	Peak-Peak	60 Hz RMS
Frontal				
Chest wall	292 (96)	6.3 (2.0)	136 (24)	48 (11)
Inside chest	25 900 (8500)	248 (63)	—	—
Abdominal wall	55 (22)	1.5 (0.6)	224 (57)	76 (11)
Lateral				
Chest wall	330 (99)	8.1 (2.5)	154 (46)	55 (5.9)
Inside chest	—	—	—	—
Abdominal wall	158 (64)	2.4 (0.9)	290 (73)	112 (16)
Axial				
Chest wall	209 (68)	4.4 (1.7)	66 (8.8)	24 (3.4)
Inside chest	19 700 (13 800)	171 (82)	—	—
Abdominal wall	24 (11)	1.0 (—)	97 (20)	33 (4.3)

*Values of the 60 Hz RMS component of the electric fields were obtained from frequency spectra of 1-s time spans.

instruments). For consistency in illustrating the waveforms, a single large rat (624 g) was used to generate all the figures. Data from this rat were similar to those from the others and were averaged with the other results (i.e., in Table 1) for frontal exposure at the chest.

RESULTS

An electrical signal which was similar to an electrocardiogram signal was readily obtained upon placing the probe tip at the chest site. Figure 2 illustrates this type of signal, obtained with the probe direction corresponding to the axis of the rat. The maximum peak-to-peak voltages were measured on the time-voltage trace and the electric field of the ''QRS'' complex averaged 268 $\mu V\ cm^{-1}$ (84 $\mu V\ cm^{-1}$ SD) peak-to-peak (PP) in the direction of the axis of the animal. When 1-s intervals of the signal were frequency-analyzed, as shown in Fig. 3, the signal at the chest wall included a normal ECG spectrum with a broad peak at around 60 Hz. The actual ''fundamental'' frequency was at the approximately 4 Hz heart-beat frequency, but the instrument was adjusted to emphasize the frequencies of interest in the ELF range. The -3 dB points of the spectrum in Fig. 3 are at 32 Hz and at 95 Hz, excluding the minor ripples in the spectrum (which are mostly artifacts of the digital signal analysis). The 60 Hz component of the cardiac electric field obtained from the frequency spectrum averaged 6.3 $\mu V\ cm^{-1}$ (2.0 $\mu V\ cm^{-1}$ SD) root-mean-square (RMS). This value was well above the background noise level of the instrumentation. At the end of some sessions, the background noise level (field off) at 60 Hz was checked after the rat had been killed, and this value averaged 0.4 $\mu V\ cm^{-1}$ for 5 measurements.

When 60 Hz magnetic field exposure was applied, the induced electric field added a 60 Hz sinusoidal signal to the cardiac signal. The magnitude of this signal was proportional to the magnetic field strength. For 1 mT frontal exposure, with the probe oriented as in Fig. 1, a combined signal appeared as shown in Fig. 4. The sinusoidal electric field averaged 136 $\mu V\ cm^{-1}$ (24 $\mu V\ cm^{-1}$ SD) PP. A spectrum of this type of signal is illustrated in Fig. 5, and includes a spike at the

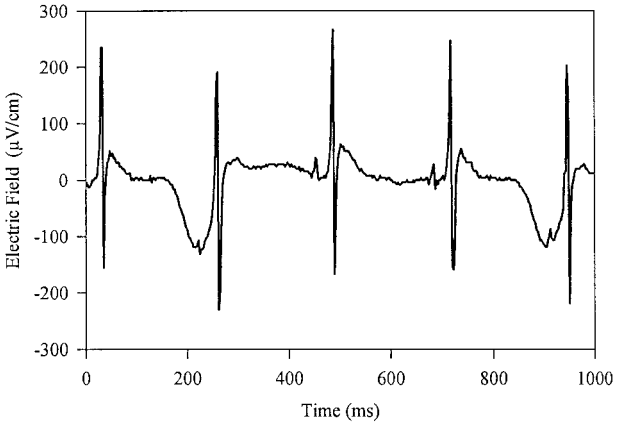

Fig. 2. A trace of the electric field versus time at the chest wall of a large rat lying on its side with the probe oriented with the longitudinal axis of the animal. The trace resembles an electrocardiogram waveform, except for the baseline fluctuations (evident at about 200 and 900 ms in this trace) which are associated with the animal's breathing motion.

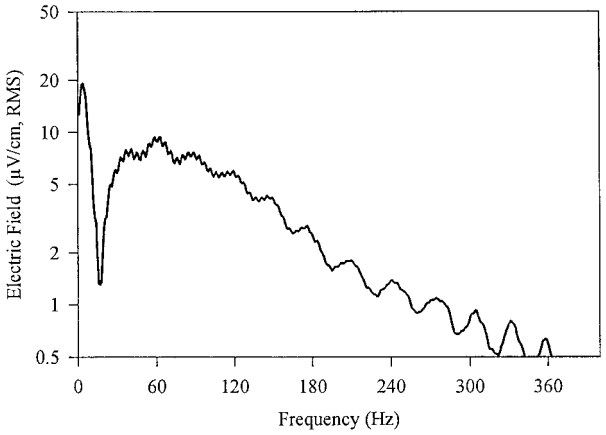

Fig. 3. Spectral analysis for the conditions used in Fig. 2, obtained by averaging 10, 1-s traces. The relatively high readings at about 4 Hz represent the fundamental heart beat frequency.

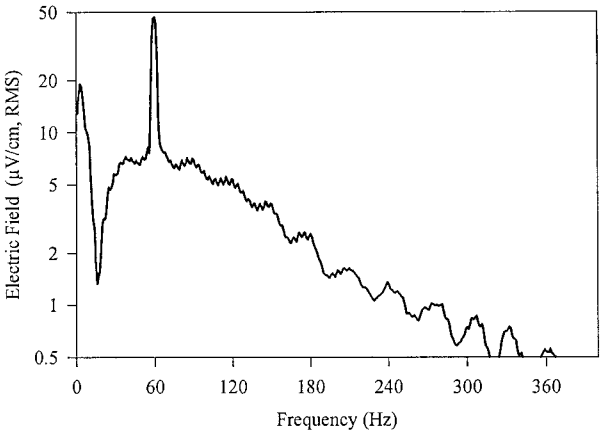

Fig. 5. Spectral analysis for the conditions used in Fig. 4, obtained by averaging 10, 1 s traces. This is very similar to the spectrum shown in Fig. 3, with an added spike at 60 Hz.

exposure frequency of 60 Hz. This component averaged 48.4 μV cm^{-1} (10.9 μV cm^{-1} SD) RMS.

Inside the chest, near the heart, relatively large endogenous fields were present, which were highly variable with probe position or orientation. An example of the signal obtained near the heart is shown in Fig. 6, with a corresponding spectrum shown in Fig. 7. This endogenous field was much larger than the induced field at the same position (at 1 mT, the added 60 Hz component could not be detected in the spectrum). The near-heart cardiac field component at 60 Hz was even larger than the maximum induced field at the abdominal wall.

The electric field measurements for the chest wall, near the heart and abdominal wall are listed in Table 1 for frontal, lateral, and axial exposure. Near-heart measurements could not be made for lateral exposure due to the arrangement of the rib cage and sternum. At the abdominal wall, the cardiac signal was reduced, while the induced field was maximized. In all cases, the probe was oriented in the direction of maximum induced field (e.g., with the longitudinal axis of the animal for frontal exposure), because induced fields were essentially zero in the orthogonal directions. Inside the abdomen, both fields were relatively small, or zero. These abdominal fields were not quantified, due to uncertainties resulting from motion of the diaphragm during breathing at about 70–80 min^{-1} (which caused relatively large baseline fluctuations in the signal).

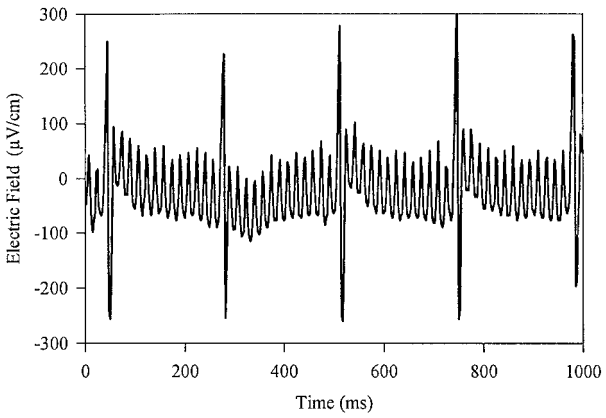

Fig. 4. A trace with the same conditions as Fig. 2, but with added 60 Hz magnetic field exposure at 1 mT from the front. The cardiac signal is still evident, with an added sinusoidal wave on the baseline.

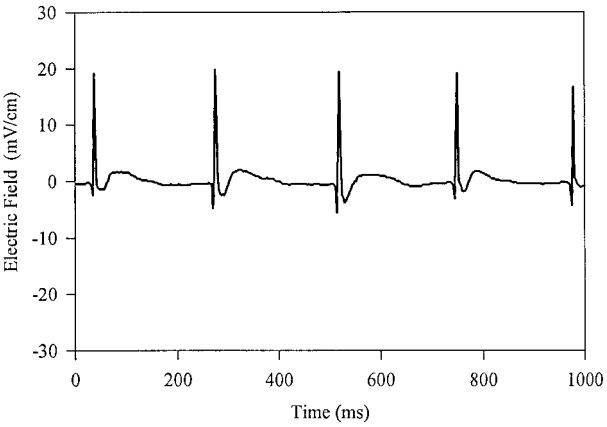

Fig. 6. A trace of the electric field versus time for the probe tip located inside the chest wall near the heart of a rat lying on its side. The trace resembles an electrocardiogram waveform, as in Fig. 2, except that the fields are much larger near the heart.

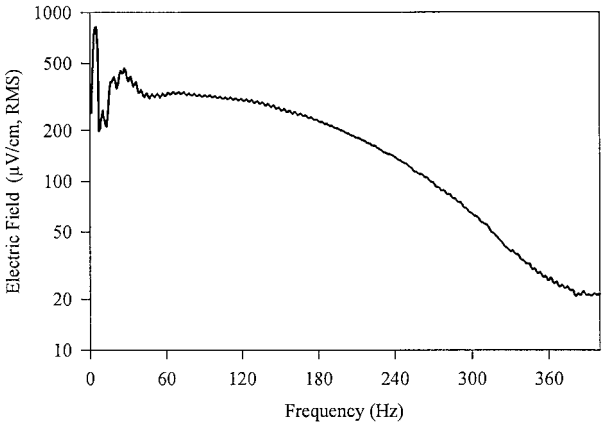

Fig. 7. Spectral analysis for the conditions used in Fig. 6, obtained by averaging 10, 1-s traces. This spectrum illustrates the presence of endogenous electric field components over the entire ELF frequency range (30 to 300 Hz).

DISCUSSION

Physiological electric fields were readily detected by the miniature-probe system. The waveform obtained from the probe at the chest wall resembled a standard ECG waveform. When an external 60 Hz magnetic field was applied, an induced electric field was added to the endogenous fields. The induced electric field increased from zero at the center of the body, to values comparable to the cardiac field at the body wall for 1 mT exposures. However, the largest endogenous field (inside the chest near the heart) was much larger than the largest 1 mT induced field (at the abdominal wall).

The frequency spectrum of the cardiac electric field at the chest (Fig. 3) contained a broad peak at 32–95 Hz, which is of interest for ELF dosimetry due to its coincidence with the 50 and 60 Hz frequencies of the electric power distribution system. This feature of the spectrum has been noted by other authors (for example, Golden et al 1973; Bergeron, 1993]. This broad peak is readily understood in terms of the intermittent nature of the ECG waveform, in which the 15–20 ms QRS complex is repeated at the heart rate [Osborne, 1981]. This wave-form intermittency results in the approximately 50-to-1 ratio of the peak-to-peak to 60 Hz RMS voltages obtained in this study for 1-s durations (e.g., see Table 1). For comparison, the theoretical ratio of the peak-to-peak to RMS voltages for a continuous sine wave is 2.8. The 60 Hz electric fields induced by the externally applied 1 mT magnetic field at the body wall were somewhat less than estimates calculated from homogeneous models, as expected from previous induced-field measurements in rat carcasses [Miller, 1996].

The endogenous fields help to provide a reference level for ELF magnetic field exposures in terms of induced field dosimetry, which is relevant to the bioeffects problem. It seems plausible to assume that endogenous fields are essentially harmless, and that induced fields would need to be substantially higher in some respect for harmful bioeffects to occur. The comparison of fields can be made in two ways. First, by simply comparing the relative magnitudes of the endogenous field to induced fields, magnetic-field exposure levels needed to exceed endogenous fields can be estimated. Fairly high 60 Hz magnetic exposure fields would be needed for equal induced and endogenous fields. For rats at the chest, peak-to-peak induced fields from frontal 1 mT exposure were only about half as large as the endogenous field, and thus would be equal at 2 mT. The 60 Hz RMS induced field (1-s interval) was 7.7 times the 60 Hz RMS endogenous field, and would be equal for 0.13 mT exposure. Of course, larger exposure fields would be needed to induce fields comparable to endogenous fields near the heart. Second, the induced and endogenous fields can be compared by considering the 60 Hz exposure as the ''signal'' with the endogenous fields considered as ''noise.'' As noted by Barnes (1992), the endogenous fields considered in this study represent additional noise in a hypothetical resonance system and would be reduced in importance when averaged over a sufficiently long time during which the exposure signal was coherent.

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