

Reprinted with permission from: JP Wikswo; JP Barach and JA Freeman, Magnetic Field of a Nerve Impulse : First Measurements, *Science*, 208 (4439), © 1980 [American Association for the Advancement of Science](#)

You may view, browse, and/or download this material for temporary copying purposes only, provided these uses are for noncommercial personal purposes. Except as provided by law, this material may not be further reproduced, distributed, transmitted, modified, adapted, performed, displayed, published, or sold in whole or in part, without prior written permission from the publisher.

### Magnetic Field of a Nerve Impulse: First Measurements

John P. Wikswo, John P. Barach, John A. Freeman

*Science*, Vol 208 (4439), pp 53-55, 1980

*Abstract. The magnetic field of the action potential from an isolated frog sciatic nerve was measured by a **SQUID** magnetometer with a novel room-temperature pickup coil. The  $1.2 \times 10^{-10}$  tesla field was measured 1.3 millimeters from the nerve with a signal-to-noise ratio of 40 to 1. © 1980 by the American Association for the Advancement of Science*

Although the electrical potentials produced by a propagating nerve action potential can be measured readily, the accompanying magnetic fields have thus far never been observed directly. The failure of previous attempts is readily understood. The nerve action potential has the form of a moving, azimuthally symmetric solitary wave (1) which can be modeled as two opposing current dipoles driven by a potential change of the order of 70 mV. The peak currents range (2) from 5 to 10  $\mu$ A. The external magnetic

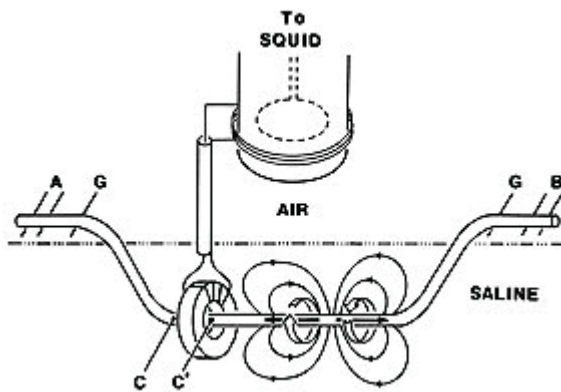


Fig. 1. Diagram of the experiment (not to scale). A nerve action potential propagates from proximal to distal (left to right in the figure). The wide and narrow arrows around the nerve represent the magnetic and electric field, respectively; the arrows on the nerve axis are equivalent dipole sources. Stimulation may be from either electrodes A or B, with the other or C as recording electrodes. The toroidal pickup coil is connected to a large transfer coil around the cylindrical Dewar vessel

that contains the SQUID magnetometer and its pickup coil (indicated by dashed lines) surrounded by liquid helium.

field B can be estimated from Ampere's law, in which I is the net axial current enclosed by a closed path of integration c

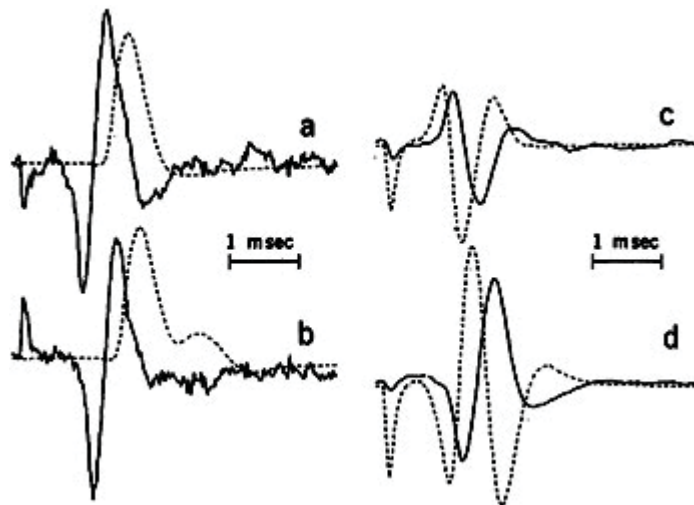
$$\oint_c \mathbf{B} \cdot d\mathbf{l} = \mu_0 I$$

where  $\mu_0$  is the magnetic permeability of free space, and the dot (inner) product between  $\mathbf{B}$  and  $d\mathbf{l}$ , the differential element that describes  $c$ , is integrated over the complete length of the path.

If the nerve is immersed in a conducting medium, the maximum magnetic field of  $10^{-10}$  T occurs at the nerve surface (radius  $r \leq 0.3$  mm), with the numbers depending upon the preparation used. As the distance from the nerve is increased, an increasing fraction of the external current returns within  $c$ , so that the field at 1 cm is a few picoteslas and decreases thereafter in proportion to the inverse cube of the distance (3). The weakness of the magnetic field, its rapid falloff with distance, and the required 1-to 2-kHz bandwidth place the signal at the limit of detectability of magnetometers used for biomagnetic measurements (4).

Two groups of investigators have used large room-temperature coils and conventional amplifiers to obtain signals interpreted as the magnetic field from the action potential of an isolated frog sciatic nerve (5). These signals did not exhibit the expected reversal of polarity upon reversal of the direction of impulse propagation, and the measurements were

Fig. 2. The magnetic field (solid lines) and electric potential (dashed lines) of an isolated frog sciatic nerve. The recordings in (a) and (b) were made with opposite polarities of the 1.6-V, 20- $\mu$ sec stimulus. As evidenced by the initial stimulus artifact, the stimulus was negative (cathode toward toroid) in (a) and positive in (b); in both cases it was applied to electrodes A (as in Fig. 1), with the electrical recordings made in air



from B (as in Fig. 1) referenced to the grounded saline bath. The magnetic signal is the average of 256 traces recorded in a 1-Hz to 5-kHz bandwidth, with the first peak having an amplitude of 70 pT; the electric signal is the average of 128 traces in a 1-Hz to 1-kHz bandwidth. The late peak in (b) is due to slow beta fiber activity. The recordings in (c) and (d) were made with opposite directions of impulse propagation. In (c), the 0.5-V negative stimulus was applied at A (Fig. 1) and the electrical recording was made in the saline Ringer solution at C (Fig. 1); in (d) the stimulus was applied at B (Fig. 1) and the electrical recording again made at C (Fig. 1). The magnetic and electric signals are the average of 1024 and 128 traces, respectively, in a 0.01-Hz to 1-kHz bandwidth. The amplitude of the first magnetic peak is 60 pT.

made with a nerve supported in air so that all of the electrical currents were

confined to the nerve bundle and the coaxial layer of moist electrolyte surrounding it (6). As a consequence, there could be no magnetic field in the air outside such a nerve. These previous measurements may have been sensitive to capacitive coupling, since three electrostatically shielded pickup coils of adequate sensitivity were unable to detect the magnetic field of a moist nerve in air (7). Our measurements avoid this difficulty in that we used a nerve immersed in a conducting medium.

Sciatic nerves from bullfrogs (*Rana catesbeiana*) were dissected and placed in a dish containing aerated Ringer solution. The nerve was gently laid over an arrangement of chlorided silver wire electrodes (Fig. 1) so it could be electrically stimulated proximally and the action potential could be recorded distally, or vice versa. The magnetic fields were recorded with a SQUID (superconducting quantum interference device) magnetometer (8) [ $1.3 \times 10^{-14}$  T (Hz)<sup>1/2</sup> sensitivity (9), 18.7 mV per flux quantum calibration]. The SQUID magnetometer detects magnetic flux changes through a superconducting pickup coil in a liquid helium environment. At the closest coil-to-nerve separation of 15 mm, the nerve magnetic field could barely be detected.

Considerable effort was thereby expended to increase the signal-to-noise ratio. The distance between the nerve and the detector coil was reduced an order of magnitude by threading the nerve through a specially constructed miniature toroidal transformer (10). The transformer consisted of four turns of wire (No. 38) wound on a 1.2-mm-thick ferrite core (minor diameter, 1.2 mm; major diameter, 2.6 mm; and effective relative permeability of 6800 at 2000 Hz). The effective cross-sectional area of this toroidal pickup coil was  $3.9 \times 10^{-2}$  m<sup>2</sup>. The toroid was inductively coupled (mutual inductance, 5.4 nH) to the magnetometer face coil by a transfer coil wrapped around the outside of the Dewar. Experiments were performed in an area isolated from large magnetic objects and magnetic transients. The 60-Hz noise was further reduced by analog cancellation. By averaging 1024 or more re-petitive signals (0.5- to 1.5-mV signal, 0.4-mV random SQUID noise, 1.5-mV power line harmonics) with a Nicolet 1170 signal averager, signal-to-noise ratios of 30 to 40 were obtained with a 1-kHz bandwidth.

Figure 2 shows the magnetic (solid lines) and electric (dashed lines) signals recorded from a frog sciatic nerve immersed in Ringer solution. The proximal end of the nerve was stimulated by means of the A electrodes (Fig. 1) 23 mm from the toroid. The action potential was recorded in air 15 mm distal to the toroidal coil by a single B electrode referenced to the grounded saline bath. The stimulus was negative (cathode nearest toroid) (Fig. 2a) and positive (Fig. 2b). The magnetic field of the stimulus current, visible at the beginning of the traces, reverses polarity when the stimulus is reversed. As expected, the magnetic and electric signals of the action potential do not reverse.

The electric trace in Fig. 2a exhibits a small stimulus artifact and has the shape of action potentials reported previously (If), with a rather steep initial portion followed by a slower region of recovery; the average asymmetry in our work was 1.9 to 1. The differing delays between the stimulus and the onset of the

electric and magnetic signals are due to the separation of the toroid and the B electrodes and to the measured (22 m/set) conduction velocity. In the electric signal shown in Fig. 2b, the positive stimulus activated a slower set of beta nerve fibers (12), and a corresponding shoulder is visible on the magnetic record.

The magnetic signal is produced predominately (see below) by the total axial current  $I$ , enclosed by the toroid while the voltage action potential in air expresses the charge on the membrane. The magnetic trace should then be proportional to the first derivative of the electric trace, as observed in Fig. 2. On exchange of the stimulating and recording electrodes, the magnetic signal must reverse sign (Fig. 1) since the leading current loop reverses direction; but the single-ended voltage trace should be similar. When this experiment was performed, the magnetic signals reversed as expected and had a common mode portion of only one part in seven while the electric signals did not reverse, verifying that the magnetometer signal represents the magnetic field of the action potential and not capacitative coupling to the electric field.

Since more detailed interpretation of the data in Fig. 2, a and b, is hindered by the separation between the toroid and the recording electrodes, the recordings in Fig. 2, c and d, were obtained with a pair of spherical electrodes (C and C' in Fig. 1) with a 5.3-mm separation placed in close proximity to the nerve as it passed through the toroid. This arrangement provided coincident magnetic and electric data but the electric measurement had to be made in the saline bath. We show records for distal and proximal stimulation in Fig. 2, c and d, respectively; the reversal of the magnetic trace (solid line) is apparent, and the crossover of the magnetic trace occurs at the peak in the electric signal. Since the electric trace is differential, it measures  $dV(z)/dz$ , which is proportional to the axial return current, in the saline adjacent to the nerve. It must therefore reverse when the direction of impulse propagation reverses, in contrast to the potential recorded by single-ended measurements in air. Also, we would expect (3, 13) precisely the derivative relationship between the electric and magnetic traces in Fig. 2, c and d. Thus the first peak of the magnetic trace indicates the forward loop of the currents in the moving action potential. By Ampere's law, the toroidal coil measures that part of the axial nerve current which is not canceled by return flow within the toroid. The current paths must be known in detail to determine precisely what fraction of the total nerve current is measured.

Further checks of the magnetic signals were performed. When the coils were tied to ground, rather than left floating electrically, much larger electrical artifacts were produced. Removal of the saline, so that only a slight amount of current could circulate around the still moist surface of the toroid, resulted in large, immediate signal reductions. Also, an artificial electric current dipole was made by applying a square voltage pulse to a twisted pair of No. 34 wires with bared, separated ends immersed in the saline. When this source threaded the toroid, a large signal was received; when it was oriented so that substantial field cancellation was expected, the signal dropped by a factor of 5 although the saline surrounding the toroid underwent an identical voltage excursion.

The toroid sensitivity was calculated and measured as a function of frequency ( $f$ ) ( $80 \text{ Hz} < f < 10 \text{ kHz}$ ). Since the coil has resistance  $R$  in addition to inductance  $L$ , the true magnetic field wave-form  $B(t)$  has an additional small component proportional to the integral of the observed signal  $B'(t)$  multiplied by  $R/L$  for the toroid (10). Thus

$$B(t) = B'(t) + R/L \int B'(t') dt' \quad (2)$$

with a calibration of 150 pT/mV. While the exact calculation of this correction requires inclusion of the effect of the low-pass filter (10), the  $R/L$  effect is small during the early part of the action potential. To show this, we approximate the first magnetic peak in Fig. 2d by a Lorentzian profile of height  $h = 4.8 \text{ mV}$  and width  $w = 89 \text{ psec}$ . Then Eq. 2 becomes

$$B(0) = B'(0) + 0.30 h/w \tan^{-1} (500/89) \quad (3)$$

resulting in a field of 125 pT at the first peak ( $t = 0$ ) of which 27 percent is due to the second, correction term. This represents a net current through the toroid of  $0.8 \pm 0.2 \mu\text{A}$  in the forward loop of the action potential, with a precision adequate for the exploratory purposes of this work. Larger currents and larger fields will, of course, exist closer to the nerve fibers.

Even at their semiquantitative level, these experiments demonstrate that magnetic measurements of nerve function can be made directly in the conducting fluid; unlike electrical signals, they are not obliterated by the high conductivity fluid. The use of a split toroid will allow measurements on living systems without requiring puncture or intrusive contact with the nerve axon. More important, this technique measures current density directly (14) and allows determination of current profiles without assumptions about conductivity and electric boundary conditions that are necessary to unfold the nerve current from voltage recordings (15). Since the magnetic trace is very close to an actual current measurement, it is therefore a particularly strong complement to the electrical record.

**John P. Wikswo, Jr.**

**John P. Barach**

*Department of Physics and Astronomy  
Vanderbilt University,  
Nashville, Tennessee 37235*

**John A. Freeman**

*Department of Anatomy,  
Vanderbilt University School of Medicine  
Nashville, Tennessee 37232*

1. A. C. Scott, *Rev. Mod. Phys.* **47**,487 (1975).
2. B. Katz, *Nerve, Muscle and Synapse* (McGraw-Hill, New York, 1966).
3. K. R. Swinney and J. P. Wikswo, Jr., *Biophys. J.* in press.
4. M. Reite and J. Zimmerman, *Annu. Rev. Biophys. Bioeng.* **7**, 167 (1978).
5. J. H. Seipel and R. D. Morrow, *J. Wash. Acad. Sci.* **50** (No. 6), 1 (1960); J. A. Geneerelli. N. J. Holter, W. R. Glasscock, *J. Psychol.* **57**, 201 (1964); *ibid.* **52**, 317 (1961).
6. A. Kolin, *Phys. Today* **21** (No. 5), 39 (1968).
7. A. M. Cook and F. M. Long, in *The Nervous System and Electric Currents*, N. L. Wulfsohn and A. Sances, Eds. (Plenum, New York, 1970), pp. 9-13
8. SHE model BMP-55; S.H.E. Corp., 4174 Sorrento Valley Blvd., San Diego, Cahf. 92121.
9. J. P. Wrkswo, Jr., *AIP Conf. Proc.* **44**, 145 (1978)
10. J. P. Barach *et al.*, in preparation.
11. R. Lorente de Nó, *Studies Rockefeller Inst. Med. Res.* **132**, 384 (1947).
12. J. Erlanger and H. S. Glasser, *Electrical Signs of Nervous Activity* (Univ. of Pennsylvania Press, Philadelphia, 1937).
13. J. Clark and R. Plonsey, *Biophys. J.* **8**, 842 (1968).
14. Properly it measures curl  $J$ , but in simple axon geometries, Stokes's theorem is used at once (3).
15. R. K. Hobbie, *Intermediate Physics for Medicine and Biology* (Wiley, New York, 1978), pp. 178-203
16. We thank J. K. Wikswo, J. Madey, E. Iufer, R. Debs, and J. Barnes for their encouragement and H. Sadisky for the ferrite cores. Supported in part by grants from the Research Corporation and the Vanderbilt University Research Council, and by NIH biomedical research support grant RRO7089-13.

24 September 1979; 6 December 1979