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Effects of coil design on delivery of focal magnetic stimulation. Technical considerations [1]

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Summary The localization of effects from magnetic coil stimulation is not immediately obvious. We measured the magnetic fields produced by several different coils and compared the results with theoretical calculations. Magnetic stimuli were delivered from a Cadwell MES-10 magnetic stimulator using 3 circular coils (one 9 cm in diameter; two with an angulated extension, 5 and 9 cm in diameter) and twin oval coils arranged in a butterfly shape (each coil approximately 4 cm in diameter) and from a Novametrix Magstim 200 using two circular flat-spiral coils (6.7 and 14 cm in diameter). Peak-induced strength of the magnetic field was recorded with a measuring loop (1 cm in diameter) at different distances from the center of the coil. When the measuring loop was moved in the same plane laterally from the center of the coil, for all coils except the butterfly-shaped coil, the field was highest in the center and fell off near the circumference of the coil. The field dropped progressively when measurements were made more distant from the plane of the coils. The electric field induced from the magnetic coil could be calculated from the coil geometry. For all coils except the butterfly-shaped coil, the largest electric field was at the circumference of the coils. The 6.7 cm flat-spiral coil induced currents similar to those induced by the larger coils but more focally. The butterfly-shaped coil induced the largest currents under its center, where the circumferences of the two component coils come together. The component of the electric field parallel to the wire in the center of this coil was the largest and most localized.

Key words: Mapping; Motor evoked potentials; Cortical stimulation; Magnetic stimulation; Magnetic coil; Motor cortex

Non-invasive techniques for transcranial electrical stimulation of human motor cortex have been developed in recent years (Merton and Morton 1980; Marsden et al. 1981; Levy et al. 1984; Hassan et al. 1985; Rossini et al. 1985). We have recently described the use of this technique for non-invasive mapping of different body part representations in the motor cortex (Cohen and Hallett 1988a,b). However, transcranial electrical stimulation may be painful. Barker et al. (1985) have described a less painful method of transcranial stimulation of the brain involving the use of a brief magnetic pulse. This technique is useful for studying central motor conduction velocities (Barker et al. 1986; Hess et al. 1986, 1987).

For magnetic stimulation, a brief strong current is passed through a coil of wire, generating a time-varying magnetic field. This coil of wire is called a magnetic coil (MC). An electric field is induced at right angles to the magnetic field and is proportional to the time-rate of change of the magnetic field. The resulting induced currents flow in closed conductive paths in planes parallel to the plane of the stimulating coil, producing action

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potentials in excitable cells. Movements of different individual fingers with different orientations of the stimulating round MC over the scalp have been reported (Amassian et al. 1987; Cohen et al. 1990). This finding shows the possibility of delivering focal magnetic stimuli, but also stresses the difficulties of obtaining reproducible results in different subjects or in the same subject at different sessions.

Several types of MC are currently used for clinical purposes. To understand better the ability of these coils to excite brain and peripheral nerves focally, we measured the magnetic fields and mathematically modeled the magnetic and electric fields induced by each MC. Our measurements and calculations were performed for a coil in air. The presence of non-homogeneous tissue near the coil will alter the electric field. However, this alteration will not affect the relative focalizing ability of different coils as determined from the field produced in air. We can only estimate what the induced currents would be in the presence of inhomogeneous conductive tissue. This information forms a theoretical basis for the choice of the geometry of an MC better able to deliver focal brain stimulation. In a separate report, we will discuss the ability of each type of MC to excite different motor representation areas focally in humans.

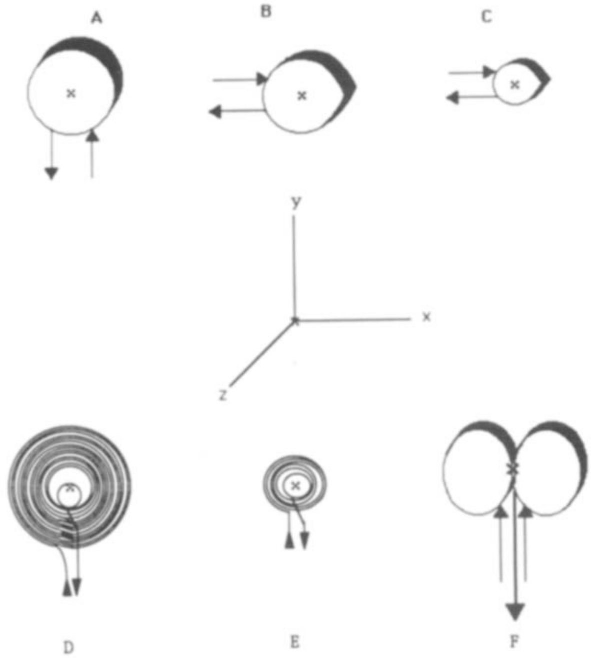

Fig. 1. The different shapes and geometries of the magnetic coils tested. A, B, C and F are tightly wound circular coils, and D and E are circular flat-spiral coils. The X in each coil marks the arbitrarily defined coil center. The origin of the coordinate system is at the defined center of each coil. See text for details.

Methods

Stimulating coils

We studied 3 tightly wound circular coils, one 9 cm in diameter (A) and two with an angulated extension, 9 cm in diameter (B) and 5 cm in diameter (C); two circular flat-spiral coils, one 14 cm in diameter (D) and one 6.7 cm in diameter (E); and tightly wound twin oval coils arranged in a butterfly shape, each coil approximately 4 cm in diameter (F) (Figs. 1 and 2). Current flowed in opposite directions in the two component 'wings' of the butterfly-shaped coil. The center of each MC was defined to facilitate reproducible positioning of the coil.

A Cadwell MES-10 magnetic stimulator was used to generate the stimuli delivered by coils A, B, C, and F, and a Novametrix Magstim 200 was used for the stimuli delivered by coils D and E. Stimulus intensities were set to be 30% of the total

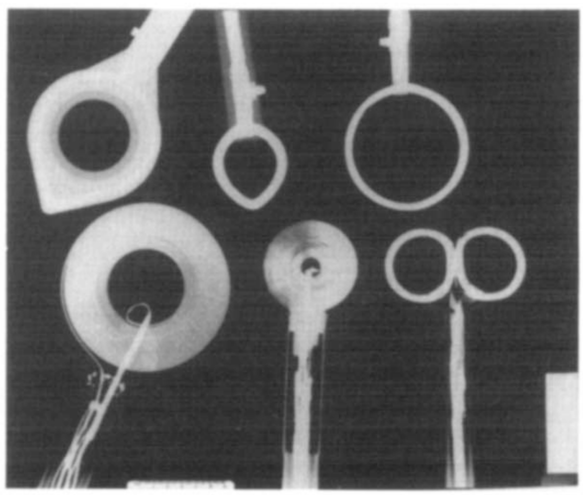

Fig. 2. X-ray picture of the different magnetic coils used. From left to right, they are B, C, and A at the top, and D, E, and F at the bottom.

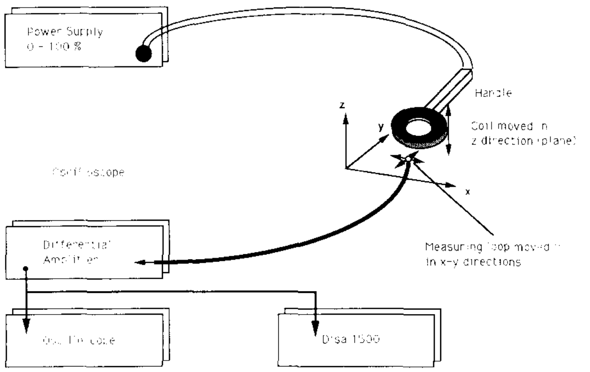

Fig. 3. Diagram of the experimental set. The power supply was Cadwell or Novametrix magnetic stimulators set at 30% of the maximal output. Voltages induced in the measuring loop were amplified and measured from an oscilloscope.

output of the stimulator (the peak magnetic field is 2.2 T for the Cadwell stimulator and 1.5 T for the Novametrix stimulator).

Experimental measurement of magnetic fields

Peak voltages proportional to the time derivative of the magnetic field were recorded with a measuring loop (1 cm in diameter). Signals were amplified by a Tektronix differential amplifier, and peak voltages were measured simultaneously on-line from a Tektronix oscilloscope and a Dantec 1500 system.

The measuring loop was attached to a plane parallel to the MC (defined by the x and y axes) and the stimulating coil was progressively more distant from it (along the z axis) (Fig. 3). The magnetic coil was separated from the loop by 1 cm layers. When testing an MC, we positioned the loop under the defined center of the MC and then moved centimeter by centimeter along the x axis (laterally) and y axis (proximal-distal in relation to the handle of the MC) until the recorded voltages were minimal. The procedure was repeated at

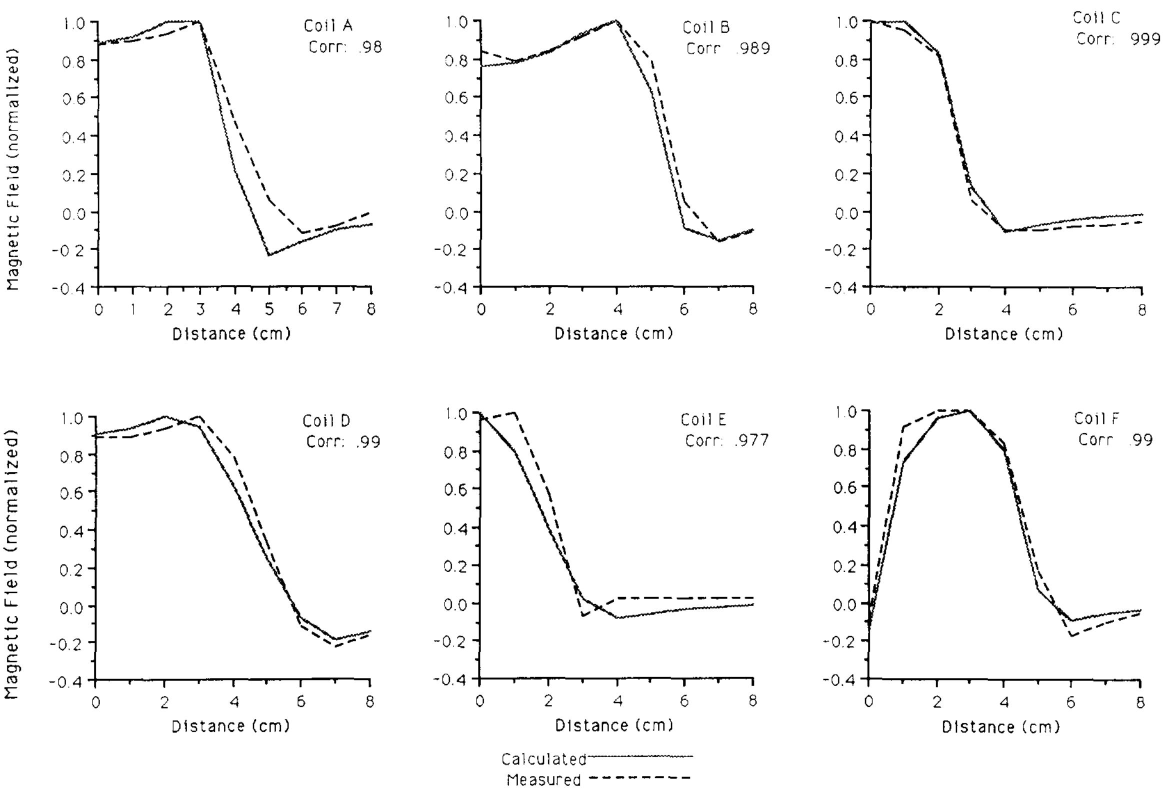

Fig. 4. Z component of the magnetic field induced by different magnetic coils according to experimental measurements (dashed lines) and mathematical modeling (solid lines). The abscissa shows the distance from the center of each coil measured along the x axis. Field magnitudes are normalized to the maximal values for each coil.

different distances from the MC along the z axis until negligible voltages were induced.

Mathematical modeling of electric and magnetic fields

The magnetic field, B, at a point r can be related to the current in the stimulating coil, I, by the law of Biot and Savart (Jackson 1975)

$$B(r) = \frac{\mu_0 IN}{4\pi} \int \frac{dl' \times (r - r')}{|r - r'|^3} \tag{1}$$

where N is the number of turns in the coil, μ_0 is the permeability of free space ($4\pi \times 10^{-7}$ T m/A), and the integral of dl′ is over the coil path and r′ is a vector indicating the position of the coil path. The induced electric field can be calculated using the vector potential, A(r), which is related to the current in the coil by the expression (Jackson 1975)

$$A(r) = \frac{\mu_0 NI}{4\pi} \int \frac{dl'}{|r - r'|}. \tag{2}$$

The vector potential is in turn related to the electric and magnetic fields by the expression (Jackson 1975)

$$B = \nabla \times A \quad E = -\frac{\partial A}{\partial t}. \tag{3}$$

The primary advantage of eqn. (2) over eqn. (1) is that it contains no cross-products. Note that the only quantity changing with time in eqn. (2) is the current I. The number of turns and the maximal rate of change of current for each coil were provided by the manufacturers and used in these calculations.

The integrals in eqn. (1) and (2) are difficult to evaluate for arbitrarily shaped coils, but we can solve these integrals analytically along a line segment and then approximate our coil as a polygon and sum the contribution from each side. The polygon approximation of the coil geometry was determined by fitting an M-sided polygon (M is an integer which varied from 20 to 42) to an X-ray photograph of the coil. For tightly wound coils, we assumed that each of the N turns lies in the same location, i.e., we neglected the cross-sectional width of the coil. This approximation was not valid for the spiral coils and in this case we digitized each turn individually.

Results

The MCs induced different magnetic fields according to their respective configurations. Experimental measurements and mathematical modeling of magnetic fields were in agreement (Fig. 4). For coils A through E, the magnetic field was large between the center and the border of each MC and reversed polarity outside the border. The magnetic field for coil F was largest 3 cm from the center and also reversed polarity outside the border.

While the agreement between the experimental and modeled magnetic fields gave us confidence in our mathematical model, the physiologically more relevant parameter for analysis is the electric field. The electric field is difficult to measure experimentally. However, the mathematical model allowed us to calculate the electric field for every

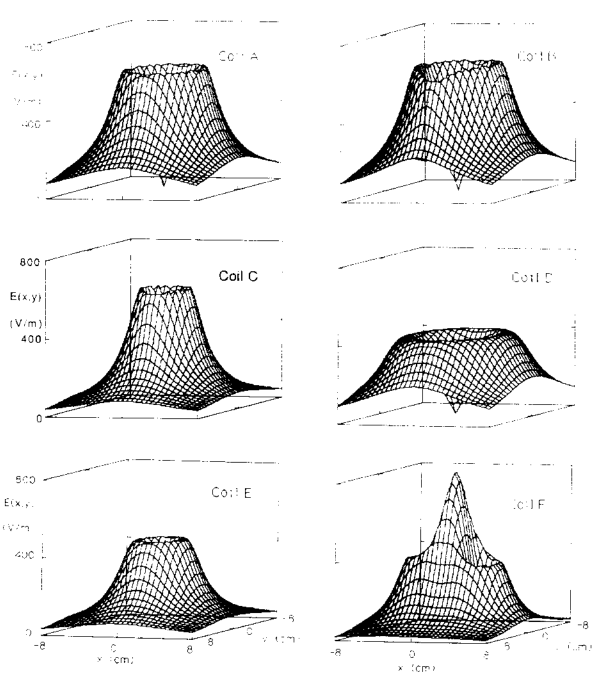

Fig. 5. Magnitude of the electric field as a function of x and y calculated on the basis of a mathematical model (eqns. (2) and (3)) 1 cm below magnetic coils A through F.

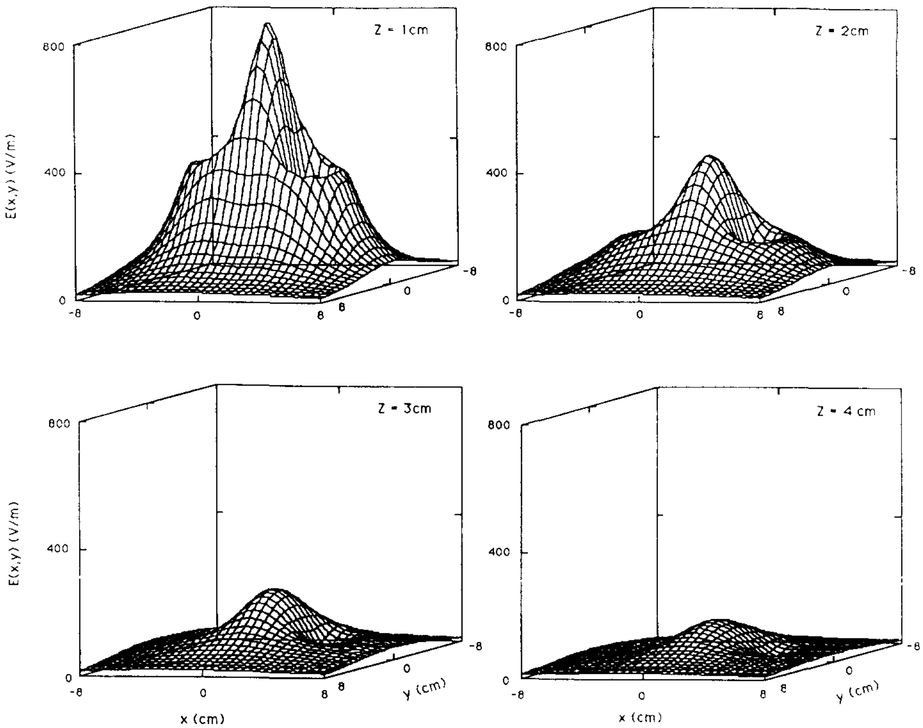

Fig. 6. Magnitude of the electric field as a function of x and y induced at $z = 1$, 2, 3, and 4 cm under magnetic coil F. Note the progressive decay of the field with increasing distance from the coil.

point in space and for any orientation of the MC. For coils A through E, the electric field was maximal under the border and minimal at the center (Fig. 5). For coil F, the electric field was maximal under the center, had a second, lower, peak under the outer edges, and was minimal under the middle of each wing. The shape and magnitude of the electric field induced by coils A and B (round and angulated, respectively) were similar. The spatial fall-off near the border was sharper for tightly wound coils (A, B, C, and F) than spiral coils (D and E). On this basis, we found coil F to be the best for inducing a focal maximal electric field under a well-defined center.

The magnitude of the electric field induced under the center of coil F dropped to 50%, 27%, and 16% at 2, 3, and 4 cm relative to its value at 1 cm under the center (Fig. 6). Not only the magnitude, but also the direction of the electric field is important for stimulating neural tissues.

The electric field is defined in space by its 3 components along the x, y, and z axes. Since the coils lie entirely on a plane defined by x and y directions, the z component of the electric field is zero. The y component of the electric field induced by coil F is larger and more localized than the x component (Fig. 7). These results are consistent with the induced electric fields being in the opposite direction to the currents in the coil during the rising edge of the pulse.

We also calculated the electric field resulting from coil F when the wings were oriented at an angle θ from the horizontal ($\theta = 0°$, $30°$, $60°$, and $90°$). When $\theta = 0°$ both wings were in the same plane, as in Fig. 1. The magnitude of the electric field 1 cm below the center of the coil was similar for the 4 different angles considered (Fig. 8). However, the direction of the electric field (not shown in Fig. 8) changed for different angles. The z component of the electric field was produced when $\theta \neq 0°$ and was largest when $\theta = 90°$. Thus, although the magnitude of the field remained the

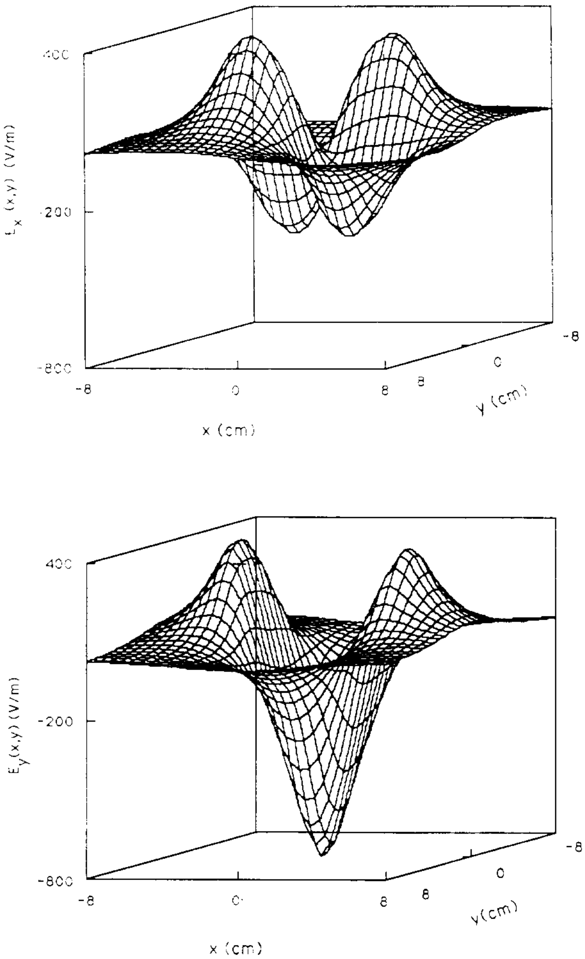

Fig. 7. The electric field is represented as a vector with both magnitude and direction. This vector has proportions along each axis. The x and y components of the electric field induced 1 cm below magnetic coil F are illustrated. Note that the largest component of the electric field is in the y direction right beneath the center of the coil.

same with the greater angle, the components in the x-y plane diminished. Also, the shape of the electric field was progressively more focal with larger θ angles. When $\theta = 90°$ the butterfly-shaped coil corresponded to a single oval coil with twice the number of turns oriented perpendicular to the x-y plane. Our calculations for circular coils were with the coil parallel to the x-y plane.

Discussion

This study evaluated the capability of available MCs to deliver focal brain and peripheral nerve stimulation. The results show that differently shaped MCs induce electric fields with different characteristics. This information is crucial when the purpose is to deliver focal brain or peripheral nerve stimulation.

The electric fields for coils A through E were maximal close to the borders and decreased toward the center of the MCs. The large region of the maximum of the electric field explains the poor ability of these devices to deliver focal stimuli when they are placed in a flat position for transcranial (Cohen et al. 1990) or peripheral nerve (Maccabee et al. 1988; Hallett et al. 1990) stimulation. It is possible, however, to deliver more focal magnetic stimuli by adjusting the angle of the circular MC so that restricted areas of the coil are in contact with the scalp (Amassian et al. 1987). Certain orientations of the MC are better for stimulating brain neurons pre- or postsynaptically (Amassian et al. 1988). It should also be taken into consideration that the currents induced by these coils run in one direction (clockwise or counterclockwise). The main drawback of using a circular coil at different angles is the large variability

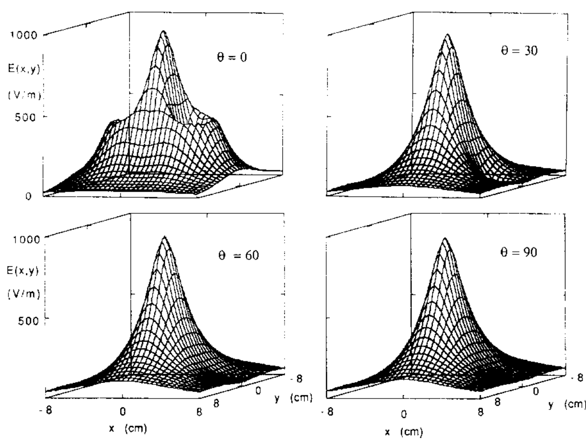

Fig. 8. Electric fields calculated for coil F with the wings at different θ angles from the horizontal (see text). Note that when $\theta = 0°$ both wings are along the sample plane (as in Fig. 1), and when $\theta = 90°$ the butterfly-shaped coil corresponds to an oval coil positioned perpendicular to the x-y plane.

in the results between different subjects and even in the same subject (Cohen et al. unpublished observations). As expected, the smaller circular MCs generated more spatially restricted electric fields than did the larger MCs. Similarly, tightly wound coils (A, B, C, and F) induced more localized electric fields than spiral coils (D and E). Forming an angulated extension, as in coil B, did not greatly improve the coil's ability to focalize or increase the magnitude of the electric field in comparison with coil A at 1 cm under the coil.

The different design of coil F represents an interesting innovation. The region of the largest electric field is under its center; a smaller peak is at the outer edge of the coil. The coil design makes the current flow clockwise in one wing and counterclockwise in the other, further enhancing the electric field at the center. The butterfly shape allowed us to define a clear center (intersection of the twin coils) and position it over different locations. This design seems particularly well suited to deliver focal stimuli over spherical surfaces like the head, because while the coil center is over the scalp location to be stimulated, the plane of the coil is tangential to the curvature of the scalp. As a result, only a restricted part of the coil (the center) is close to the brain, while the outer borders of the coil are farther from the neural tissue and less effective. Because the electric field decreases with the distance on the z axis, only neutral structures close to the center of coil F become activated. When the wings of coil F were oriented at different angles θ from the horizontal, the coil's focalizing ability was improved, but the direction of the field changed. Interactions between the electric field and the biological medium will determine a coil's effectiveness in stimulating target neural tissues.

Another factor for consideration is that magnetic stimulators produced by various manufacturers deliver pulses with different characteristics and consequently induce electric fields having different wave forms (Hallett et al. 1990, Fig. 2). Likewise, MCs with similar external shapes can evoke different electric fields if the internal configuration of the wiring is different.

In summary, our results suggest that coil F induces the most focal electric fields and may be the best suited for systematic, centimeter-by-centimeter mapping of body part representation areas in the human motor cortex. Mapping of these areas has been accomplished by transcranial electrical stimulation (Cohen and Hallett 1988a, b), but the use of magnetic stimulation would render transcranial stimulation painless, with obvious advantages.

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