The recent years have seen a large increase in both the development and use of magnetic stimulators world-wide. Commercial stimulators can achieve pulse rates up to 100Hz with stimulus intervals as low as 1msec and full computerisation. A growth in research interest has produced a marked spread into the clinical areas of diagnosis, prognosis, monitoring and therapy. As with other technologies, the world-wide web is acting as a catalyst to speed the spread of information amongst researchers and clinicians.

The ability of magnetic stimulation to induce electrical currents to flow within body tissue permits the clinician to influence or monitor many of these functions. It is able to reach deep neural structures such as the motor cortex and spinal nerve roots non-invasively and without pain. As a result the technique has become of great interest in psychiatry and is providing further aids to the clinician for the therapeutic treatment of spasticity in multiple sclerosis and stroke patients, and other forms of rehabilitation.

This guide has been written to provide an overview of the technique of magnetic stimulation from the first principle through to some of the clinical applications now feasible. Also included are details about different stimulator types and a look at more recent developments. A list of some 500 papers is provided organised by discipline. We thank our readers who have contributed helpful information or suggestions towards this edition.

Please note that this guide describes the state of the art in magnetic stimulation and is intended for a world-wide readership. Some techniques and magnetic stimulator devices described represent uses that are considered as investigational in the USA. In particular this applies to the use of cortical magnetic stimulation. Further details on the regulations governing the use of investigational devices can be obtained from the FDA (www.fda.gov).

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This section describes one of the emerging applications in magnetic stimulation, that of the use of repetitive transcranial magnetic stimulation, or rTMS, in the treatment of mood disorders.

Modest benefit in the treatment of depression was first demonstrated in 1995 by George et al. [NeuroReport, 1995, 6: 1853-1856] where sub-motor threshold magnetic stimuli were delivered cortically. The approach is radically different from electroconvulsive therapy, ECT, in that the patients are alert and do not need to be anaesthetised. There have been a number of other studies (see reference list’s section on psychiatry) with a notable one having been conducted by Pascual-Leone et al. [Lancet, 1996b, 348: 233-238] where patients receiving left dorsolateral prefrontal cortex rTMS showed a significant improvement when compared with stimulation of other areas.

Unlike single pulse magnetic stimulation, where risks of inducing a seizure are extremely low, cortically applied rTMS can induce a seizure under certain circumstances. The risk can, however, be minimised through the use of carefully selected parameters. Before the initiation of rTMS the clinician is recommended to study a milestone paper covering safety issues [Wassermann EM, Electroencephalogr Clin Neurophysiol, 1998, 108: 1-16]. As well as a comprehensive set of guidelines the paper contains a table which sets a train duration limit based on stimulating power and frequency. This is an excellent starting point for a study.

Currently there are two approaches with regards to the treatment of depression: (a) repetitive high frequency stimulation, and (b) repetitive low frequency stimulation. The two treatment modalities are shown in the two figures A and B on this page. Being more recent, repetitive low frequency stimulation is less well studied but appears to have similar beneficial effects to rTMS. If the initial data can be confirmed, the significantly lower risk of inducing a seizure will prove an undeniable plus over rTMS.

The majority of rTMS work has been done at 10Hz with the other frequencies being 1Hz, 5Hz, 15Hz and 20Hz. There appears to be little difference in the effectiveness of the different rTMS frequencies so long as the number of stimuli per train remain the same.

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**Brief History**

Electromagnetic induction, producing a current in a conductive object by using a moving or time-varying magnetic field, was first described by Michael Faraday in 1831 at the Royal Institution of Great Britain, and is probably the most relevant experimental observation for magnetic stimulation. Faraday wound two coils on an iron ring and found that when ever the coil on one side was con nected or dis con nected from a bat tery, an elec tri cal cur rent passed through the coil on the other side. The iron ring acted as a chan nel link ing the mag netic field from the first coil to the sec ond. A change in the mag netic field, re lated to the changing cur rent in the first coil, in duced a cur rent in the sec ond coil. In fact the iron ring only im proved the cou pling ef fi ciency be tween the two coils mak ing the ex peri ment more prac ti cal to per form, and given suf fi cient pri mary cur rent it could be dis pen sed with. This is the case in non-invasive mag netic stimu la tion where the stimu lating coil acts as one coil, space as the me di um for the flow of the mag netic field and the elec tri cally con duc tive liv ing body as the sec ond coil.

In 1896 d’Ar son val [C R Soc Biol; 1896, 3: 450- 51] re ported phe nes (flick er ing lights in the vis ual field) when plac ing his head in a coil driven by an alter nating 110 volt sup ply at 30 am peres. It is now known that this was due to the di rect stimu la tion of the retina. Bick ford and Frem ming in 1965 [Di gest 6th Int Conf Med Elec Biol Eng, 1965, p112] dem on strated non- invasive mag netic stimu la tion of fa cial nerves. In 1982 Pol son et al. pro duced a mag netic stimu la tor capa ble of per iphe ral stimu la tion and re corded the first mus cle evoked po ten tial [Med Biol Eng Com put, 20: 243-4]. The tech nique of mag netic stimu la tion came of age in 1985 when Barker et al. in Shef field [Lan cet, 1985, 1106- 1107] achieved mag netic stimu la tion of the hu man mo tor cor tex. For a more de tail ed his tori cal re view the reader is re fer red to a pub li ca tion by Ged des LA [J Clin Neu ro physiol, 1991, 8:1-9].

Pro gress has been rapid since 1985 with se ver al new ar eas of re search us ing new de vel op ments. These in clude coils with mul ti ple wind ings for ac cu rate stimu la tion; train of pulses for in traop era tive moni tor ing and treat ment of de pres sion and other psy chi atric dis or ders; use of fast re petitive stim uli to de term inate the lat er al ity of speech cen tres; and high en ergy stim uli to re start the fi briel lated heart.

![Figure 2](Image2.png)  **Figure 2:** Magnetic stimuli are produced by passing strong electric current pulses through a coil of wire.

![Figure 3](Image3.png)  **Figure 3:** The time varying magnetic field produces an electric field surrounding the stimulating coil which in turn induces small eddy currents in a conductive medium such as the human tissue.

![Figure 4](Image4.png)  **Figure 4:** Magstim Model 200 shown above is a magnetic stimulator with a monophasic output pulse. The unit weighs 17kg with the heaviest items being the storage capacitor, transformers and thyristor.
Principles of Magnetic Stimulation

Magnetic nerve stimulators typically consist of two distinct parts: a high current pulse generator producing discharge currents of 5,000 amps or more; and a stimulating coil producing magnetic pulses with field strengths of 1 tesla or more with a pulse duration of some 1ms.

The discharge current flowing through the stimulating coil generates the necessary magnetic pulse as is shown in Figures 2 and 3. This pulse inducts current in an electrically conductive region, such as the human body. If the induced current is of sufficient amplitude and duration it will stimulate neuromuscular tissue in the same way as with conventional electrical stimulation.

The first commercial magnetic stimulators were produced in Sheffield in 1985. The Magstim Model 200 is based on the original Sheffield design and is referred to in this document to provide detailed information on real systems used in magnetic stimulation. The Magstim in Figure 4 consists of a capacitor charge/discharge system together with the associated control and safety electronics. Using the charging circuitry the energy storage capacitor is charged to a set level determined by front panel controls up to a maximum of 2,800 volts (2.8kV). When the Magstim receives a trigger input signal the energy stored in the capacitor is discharged into the stimulating coil. The stored energy, apart from that lost in the wiring and capacitor, is transferred to the coil and then returned to the instrument to reduce coil heating. The discharge switch consists of an electronic device, which is called a thyristor, capable of switching large currents in a few microseconds. Thyristors conduct current only in

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<th>Circular 50mm Type 9993</th>
<th>Circular 70mm Type 9762</th>
<th>Circular 90mm Type 9784</th>
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<td>530</td>
<td>530</td>
<td>660</td>
<td>N/A</td>
<td>660</td>
<td>N/A</td>
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*Table 1: The physical characteristics and maximum calculated outputs of the coils used with the Magstim 200.*

*Figure 5: The above waveforms are those produced by the circular 70mm Coil (see Table 1 for winding detail) showing a fast rise and slow decay culminating in induced charge which depolarises the nerve.*
one direction and hence the Magstim Model 200 produces a monophasic discharge current with no current reversal. Monophasic discharge currents reduce heat dissipation in the coil, discharge click noise, the stimulus artefact and in increase stimulus accuracy. In addition, the stable and well-defined monophasic pulse allows for a better understanding of the mechanisms involved in magnetic nerve stimulation. The waveforms of coil current through to inducted charge are shown in figure 5 for the 70mm circular coil (Type 9762 in Table 1).

Although the figures usually quoted for the total energy stored in magnetic stimulators typically range from 500J to 10kJ or more, the important factor in the effectiveness of the peak coil energy. This can be achieved by using a large energy storage capacitor and/or by having an efficient energy transfer from the capacitor to the coil. Typically 500J of energy has to be transferred from the energy storage capacitor into the stimulating coil in around 100µs. Power, measured in watts, is equivalent to joules per second. From this, the impulse power output of a typical magnetic stimulator during the discharge phase is 5,000,000 watts (5MW). During the discharge, energy initially stored in the capacitor in the form of electrostatic charge is converted into magnetic energy in the stimulating coil in approximately 100µs. This fast rate of energy transfer is necessary to achieve a rapid rate of rise of magnetic field. The current in duced as a result of the time-varying magnetic field is in the order of 1-20mA/cm² which is the same as that used in conventional electrical stimulation.

Since the magnetic field strength falls off with distance from the stimulating coil, the stimulus strength is at its highest close to the coil surface. The stimulation characteristics of the magnetic pulse, such as depth of penetration, strength and accuracy, depend on the rise time, peak magnetic energy transferred to the coil and the spatial distribution of the field. The rise time and peak coil energy are governed by the electrical characteristics of the magnetic stimulator.

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**Figure 5a:** The above diagram shows the outputs of the three main types of magnetic stimulators. Conventional recharger and close interval pulse train units have high output capability whereas rapid-rate stimulators have medium output power at repetition rates of 10-30Hz. It must be noted that rapid-rate trains of five or more pulses applied cortically have been shown to be capable of inducing seizures. This is, of course, not surprising as ECT is also capable of inducing seizures using similar parameters. Single stimuli delivered at less than 5Hz or pulse trains containing four or fewer pulses, on the other hand, are highly unlikely to induce a seizure.

**Monophasic**
- **For:** More accurate than biphasic, lower noise, lower heat
- **Against:** Not easy to obtain bilateral cortical responses

**Biphasic**
- **For:** Short efficient pulse, suited to bilateral cortical stimulation
- **Against:** Higher noise, possibly less accurate than monophasic

**Polyphasic**
- **For:** Efficient, suited to bilateral cortical stimulation
- **Against:** Highest noise and heat; less accurate than monophasic

**Figure 5b:** The above diagram shows the three main types of magnetic field output from stimulators together with their characteristics. Only one device appears to utilise polyphasic output with the remainder using mono and biphasic output in similar numbers. Some units have multiple output types.
and stimulating coil, whereas the spatial distribution of the induced electric field depends on the coil geometry and the anatomy of the region of induced current flow.

**Stimulating Coils**

The stimulating coil, normally housed in moulded plastic covers, consists of one or more tightly wound and well insulated copper coils together with other electronic circuitry such as temperature sensors and safety switches. The physical description of the coils used with the Magstim Model 200, which was described earlier, to gether with the estimated magnetic and electrical fields for these coils are shown in Table 1. The Magstim 200 is supplied with a circular 90mm mean diameter coil as standard. This is most effective in the stimulation of the human motor cortex controlling the upper limbs with a large cortical representation, and also in the stimulation of spinal nerve roots. To date circular coils with a mean diameter of 80-100mm have remained the most widely used in magnetic stimulation.

A 3D representation of the magnetic field produced on the surface of a 90mm circular coil (Type 9784 in Table 1) is shown in Figure 6. In the case of circular coils it is important to note that the induced tissue current is zero or near zero on the central axis of the coil and in creases to a maximum in a ring approximately under the mean diameter of the coil. Stimulation, therefore, is most likely to occur under the winding and not under the coil centre. During the stimulating phase, when the magnetic field is increasing from zero to its maximum, the induced tissue current flows in the opposite direction to the coil current. In the case of the Magstim 200, all single circular coils are marked with Side A and Side B. With the coil placed on the body and Side A visible, the induced tissue current flows in the clockwise direction. With Side B visible, induced tissue current flows in the anti-clockwise direction. The use of the correct coil side is particularly important in cortical stimulation as the human motor cortex appears to be more sensitive when the induced current is flowing from posterior to anterior. Hence, with the coil placed centrally on the vertex and Side A visible, the induced current predominantly stimulates the left motor cortex. With Side B visible the effect is reversed.

Although the 90mm circular coil is a very useful general purpose coil, the site of stimulation is not well defined. The most notable coil advance has been that of the double coil (also termed butterfly or figure of eight coil). Double coils utilise two windings normally placed side by side. A 3D representation of the magnetic field produced on the surface of a 70mm double coil (Type 9925 in Table 1) is shown in Figure 7. Typically double coils range from very small flat coils for brain mapping work to large contoured versions...
to stimulate deeper neural structures in the brain. The main advantage of double coils over circular coils is that the induced tissue current is at its maximum directly under its centre, where the two windings meet.

Two commercial coils, a single circular coil with a 90mm mean diameter winding (Type 9784 in Table 1) and a double coil with two 70mm mean diameter windings (Type 9925 in Table 1), have been used to illustrate the main differences between circular and double coils. In Figure 8 the calculated induced electric field 10mm below the coil surface is plotted against distance from the coil centre. This shows that in the case of the circular coil the two peaks are symmetrical about the centre and are of the same amplitude. In the case of the double coil, however, there is a central peak with over twice the amplitude of its peripheral peaks. That enables the coil to predominantly stimulate neural structures under its centre.

However, the hypothesis that the double coil only stimulates under its centre should be viewed with caution. As shown in Figure 8, there are also smaller peripheral peaks of approximately half the amplitude of the central peak on either side of the winding. These calculations refer to an available commercial coil where practical restrictions in construction have been taken into account. The chances of only stimulating under the central axis can be increased by stimulating at, or just above, the threshold level, but this cannot be guaranteed. In the case of cortical stimulation, many neural fibres have to be stimulated before a sufficient descending volley is generated. The readiness, and hence the stimulation threshold, of individual nerve fibres in the motor cortex varies significantly from one region to the next. Thus nerve fibres under other parts of the coil winding exposed to lower induced currents may also be stimulated. With the double coil placed flat, however, the natural curvature of the head helps keep the outer edges of the windings away from other areas of the cortex further improving the accuracy of the double coil.

The Double Cone Coil (Type 9902 in Table 1; also see Figure 9) is a development where two large cupshaped windings are positioned side by side with a flat central section and angled sides closely fitting the patient’s head. The advantage of this coil is that its geometry allows for better magnetic coupling, giving significantly higher induced current in the central fissure (70% higher than with the 90mm circular coil). This coil is especially useful in the stimulation of the motor cortex areas controlling the muscles of the lower torso and limbs.

**Stimulating coil construction:** The stimulating coil is the only part of a magnetic nerve stimulator which needs to come close to, or into contact with, the pa-
During the discharge of the magnetic pulse the coil winding is subjected to high voltages and currents. Although the pulse generally lasts for less than 1 ms, the forces acting on the coil winding are substantial and depend on the coil size, peak energy and construction. Careful coil design is, therefore, very important in the construction of a magnetic stimulator. Large coils utilise more copper mass than a small coil and generally have a lower electrical resistance. As a result less heat is dissipated in their windings and because of their higher heat capacity they remain usable for much longer periods of time without becoming warm.

Copper has been used as the high current conductor in coil windings due to its low electrical resistance, high heat capacity, availability and a fair tensile strength. Silver has a lower resistance than copper and hence there would be less heat dissipated within the coil during the discharge phase as a result of $I^2R$ resistive losses. The benefits are, however, offset by significantly lower heat capacity of silver due to its high conductivity.

**Magnetic Field Strength vs. Stimulus Strength**

The most widely quoted figure regarding the output of magnetic stimulators is the magnetic field strength. Although it is an important parameter, magnetic field strength alone is a poor measure of magnetic stimulator performance. Magnetic field strength is defined as the magnetic flux density and does not reflect the total magnetic flux produced by the stimulating coil over its total area. In a small coil where the magnetic flux is concentrated in a small area, the magnetic field strength is much higher than in a larger coil, but the field falls off much more rapidly with distance.

Hence a small coil is somewhat more powerful in the stimulation of superficial nerves and a large coil is more suitable for structures at depth. The amplitude, waveform and spatial characteristics of the induced current all play a role in magnetic nerve stimulation.

A more accurate indicator of the stimulating power output is the induced charge density per phase, defined as the integral of the induced current density during the rise time of the magnetic field. It provides a better measure of the actual value of the induced charge density per phase, and takes into account the effects of the nodal time constant of the myelinated nerve fibre. Unfortunately, the actual value of the induced charge density per phase is difficult to calculate accurately due to the complex nature of the structure being stimulated. Different areas, such as bone, fat, grey matter and white matter, have different conductivities affecting the induced current and its path.

As well as basing a choice of coil on magnetic field strength and induced current, the suitability of a coil for its intended application must also be taken into account. As examples, the 90 mm coil is very effective in bilateral stimulation of the phrenic nerve roots, the double 70 mm coil is used for monolateral stimulation, the circular 40 mm coil is well suited to stimulation at the Erb’s point, and the double cone coil is most powerful in cortical stimulation of the lower extremities.
Magnetic stimulation’s main ability is safely and easily to stimulate most neural structures, unim peded by fat and bone and with out dis com fort. The major ity of the clinical applications for the technique are for the non-invasive stimulation of the periph eral and cen tral mo tor path ways. Other uses in clude stimu la tion of the left and right prefrontal cortex, visual cortex, lan guage centre, cerebellum and periph eral sen sory nerves. These applications cover uses for diag no sis, prog no sis, moni tor ing and therapy. The fol low ing sec tions give an out line of how re sponses are ob tained and how their char ac ter is tics are meas ured, to gether with sev eral sam ple ap pli ca tions from op er ating room moni tor ing to urology. There is also a sec tion on safety with a con clud ing sec tion of ref er en ces.

**Motor Evoked Potentials (MEPs)**

Magne tically evoked mo tor po ten tials can be ob tained by stimu lat ing neural tis sue such as the mo tor cor tex, spi nal nerve roots and periph er al nerves. Re sponses can be record ed in the normal man ner using EMG or evoked po ten tial equip ment. In the case of mus cle ac tion po ten tial the size of the response and de pen ding on the re sponse size, nerve ac tion po ten tials at re sponses may re quire av er aging. Mag netic stimu la tion of a periph eral nerve, in this case the ul nar nerve, was pre vi ously shown in Fig ure 1. Us ing a more ac cu rate fig ure of eight coil the re sponses in Fig ure 10 de mon strate the use of the Mag stim 200 to mea sure the periph eral mo tor nerve con duc tion ve loc ity (NCV).

It is the stimu la tion of deeper and less ac cess i ble nerves, how ever, where the tech nique of mag netic stimu la tion excels over con ven tional elec trical stimu la tion. Mo tor evoked po ten tial re sponses were record ed from the ul nar nerve at the elbow and wrist using a double 30mm coil giving con duc tion latencies of 7.36ms and 3.20ms re spec tively. The dis tance be tween the two sites of stimu la tion was 24.5cm giving a healthy nerve con duc tion ve loc ity of 59m/s. The lower traces show ex panded wave forms (note scales) to highlight the take-off points.

**Facilitation:** In the case of the cen tral ner vous system it is pos si ble to reduce the stimu la tion threshold by ap prox imately 25%, increase the re sponse am pli tude 2-5 times and re duce re sponse la tency by some 1-3ms through pre- ac ti vation of the tar get mus cle (see Fig ure 11 not ing the changes in scales). This tech nique, re fer red to as fa ci li ta tion, has been de scribed in con sid er able de tail by oth ers [e.g. Roth well et al. Review ar ti cle. Exp Physiol 1991, 76: 159- 200]. Where the pa tient is able to con tract the target, the clinician has the choice of re duc ing the power level re sulting in in creased pa tient com fort. Ta bles of nor mal data are, how ever, avai lable for both re lax ed and fa ci li tated mus cles. The mea sure ment of con duc tion la tency with facilitation re quires sev eral su per im posed re sponses to al low the de termina tion of exact take-off point.
MEP variability: From the very early days of cortical magnetic stimulation a considerable amount of response variability has been noted, unlike in peripheral stimulation where responses are accurately repeatable. Initially these were thought to be as a result of changes in coil positioning and power level variation from pulse to pulse. Careful experiments have, however, shown the variability to be a real neurological phenomenon as a result of continually changing excitability of the cortex. In fact the measurement of this variability may also prove useful clinically in certain disorders.

Central Motor Conduction Time (CMCT): An estimation of the central motor conduction latency can be made by subtracting the cortical to muscle conduction latency from the peripheral conduction latency. The peripheral conduction latency can be measured either by magnetic stimulation of the spinal nerve roots or by using conventional F-wave techniques where applicable (see Figures 11 and 19 for example wave forms). The CMCT is abnormal in many disorders of the nervous system and, together with the other parameters measured, forms the basis for diagnosis and assessment. There are numerous normative and patient data available for many human

**Figure 11:** The set-up shown below allows the recording of motor evoked potentials. The example responses shown on the right have all been recorded over the left dorsal interosseous (FDI). From top to bottom: (a) Stimulation at Erb’s point using the circular 40mm coil (see Table 1); (b) Stimulation at the neck with the 90mm circular coil centred over C7; (c), (d) and (e) Stimulation of the motor cortex with the 90mm circular coil placed centrally on the vertex - three superimposed responses each. Responses (d) and (e) are facilitated by slight preactivation of the target muscle. Note that there is a silent period after the compound motor potential lasting over 100ms.
muscles or, if required, users can obtain their own tables of data to suit their circumstances. Typically if the CMCT is outside mean plus two standard deviations the conduction is considered abnormal.

Threshold: The threshold of stimulation (normally described as a percentage of the maximum power level of a particular stimulator manufacturer and model) is gaining popularity. It is also a sensitive indicator of abnormality in certain disorders, especially where CMCT can be normal, for example in stroke (tests for this disorder are described later). Threshold can be defined as the power level at which a response can be detected 50% of the time and it can be measured for both facilitated and relaxed muscles. Note that due to the variations explained earlier it is necessary to repeat the stimulation a few times - three responses out of six stimuli has been suggested as a standard. Measurement of the threshold of stimulation in the disorders of the central nervous system has the advantage of requiring low power levels and no peripheral stimulation. Its repeatability and comfort eases longer term assessment, for example in rehabilitation and drug therapy monitoring.

Response amplitude: Whereas in peripheral stimulation a supramaximal stimulus is generally a requirement, a different approach is taken for the stimulation of the central nervous system. Since it is not possible to obtain supramaximal responses evoked by cortical stimulation, the amplitude tends to be noted either on its own or as a ratio of supramaximal peripheral response. In normal subjects this ratio is above 50% for facilitated hand muscles but can be 5% or less in several disorders such as stroke and multiple sclerosis.

**Figure 12:** The central motor conduction time (CMCT) is typically measured by first stimulating the motor cortex (top right) and then the relevant peripheral nerve as it exits the spine (right). The difference is the time it takes for the impulse to travel from the motor cortex to the spinal vertebral foramen. This time is commonly referred to as the CMCT. The responses shown above were obtained from the first dorsal interosseous (FDI) muscle in the hand.
**Limitations:** It should be noted that the technique of magnetic stimulation has certain limitations. It has not yet proved possible directly to stimulate the spinal cord although this may be possible through the development of new specialised coils and pulsing arrangements. It is thought that the bone and cartilage surrounding the cord impedes current flow from the outside and that the spinal cavity itself does not have a large enough area to allow sufficient induced current. Cortical and spinal root stimulation, however, overcome this limitation in most clinical uses (as an example see section on cervical spondylosis).

While the Double 25mm Coil does not replace all the standard MCV and SN CV tests, magnetic stimulation has become as reliable as conventional electroneurography in the assessment of demyelinating neuropathies, with less discomfort and a shorter examination time. Magnetic stimulation is of particular use in the non-invasive stimulation of deep nerves and especially spinal nerve roots such as the phrenic nerve where electrical stimulation is both very painful and sometimes unreliable.

**Magnetic Pulse Pairs** with a timed stimulus interval can be effected either through a single coil or through two independent coils. There are a number of options provided to match most needs.

A Magstim Model 250 unit can deliver two individual pulses through one coil at a predetermined interstimulus interval from 1.0ms to 99.9ms in 0.1ms steps, with both pulses at the same power level. The resulting MEP’s will show a facilitated response. The pulses can be summed to provide a single high-power pulse.

The BiStim set-up provides the flexibility to evoke either an inhibited or facilitated response, or to deliver two pulses through one or two coils. The set-up utilises two Magstim Model 200 units and a BiStim Module.

Delivering two pulses through one coil main tains the same induced current flow (see Figure 13). Using the BiStim Module it is possible to adjust both the power level and the timing of each pulse in dependent of each other. In the case of cortical stimulation the site of stimulation may not remain the same, as the first pulse can activate both inhibitory and facilitatory mechanisms and modify the threshold and readiness of other sites to stimulation by the second pulse. For this reason most studies using magnetic pulse pairs have used double coils for selective stimulation. The BiStim Module is allowing the detailed study of the inhibitory and facilitatory mechanisms of the brain and spinal cord.
**Sensory Evoked Potentials (SEPs)**

Magnetic stimulation can also be used to stimulate sensory nerve fibres. The response can be recorded over sensory nerves and averaged in the normal manner. The advantage of magnetic stimulation over electrical stimulation is that coils may warm up before the end of the data acquisition. Magnetic sensory evoked potentials, so far relatively uncommon, are gaining popularity as coils improve and pulse repetition rates increase. A typical sensory response to magnetic stimulation at the wrist is shown in Figure 14. Owing to the small signal size careful arrangement of recording electrodes is necessary to avoid stimulus artefact interference.

**Sample Applications**

The following sections provide a variety of sample applications covering peripheral, spinal nerve root and cortical stimulation. For further information on the sample applications and other areas not covered in this guide the reader is referred to the reference section starting on Page 27.

**Operating Room Monitoring**

The standard method of operating room (O.R.) monitoring, where the central nervous system or spinal nerve roots are at risk, has so far been that of sensory evoked potentials (SEPs). The technique of recording SEPs cannot, however, be used to monitor the central motor pathways directly. As a result damage to the motor system may go unnoticed leaving the patient at a higher risk of post-operative disability. O.R. monitoring of motor pathways has, therefore, become an important goal.

With the pain aspects of stimuli being largely unimportant during surgery, where possible electrical stimulation has been used to stimulate motor nerves. This technique has proved quite successful in the case of some peripheral nerves and is already in widespread use during appropriate surgical procedures. One example is facial nerve monitoring during acoustic neuroma brain tumour surgery.

Direct electrical stimulation of the central nervous system, however, poses important problems because of bone and the depth of tissue to be stimulated. These obstacles mean that large currents are required to stimulate with potential for electrode burns. In addition the need for direct electrical contact with the patient in increases the risk of electric shock through the malfunction of the stimulator and/or other equipment in contact with the patient. Magnetic stimuli on the other hand have no

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**Why Take the Magnetic Train Route in O. R. Monitoring?**

- A single pulse train is all that is needed for bilateral lower limb responses; no averaging is required. Responses can be obtained regularly.

- Unlike with SEPs the motor tract is monitored. The process is non-invasive and can be carried out at regular intervals.

- The setup is simple and fast using conventional surface pick-up electrodes. Epidural electrodes, prone to misplacement, are not required.

- The stimulating coil, unlike electrical stimulation, makes no electrical contact with the patient minimizing risk of electric shock or burns.

- Additionally magnetic stimulation can be used pre-operatively for baseline data and diagnosis and post-operatively for prognosis.

- When used together with SEPs the spinal cord is monitored in both directions by two independent systems increasing reliability.
**O.R. Monitoring**

Figure 15: The above drawing shows one of the many possible O.R. monitoring set-ups during spinal surgery. The Double Cone Coil, specifically designed to obtain good lower limb responses, is centred over the motor cortex and is used to deliver the magnetic pulse. The coil can either be fixed into position using a variety of clamping arrangements or brought into contact as and when monitoring is required. In this case recordings are made from Tibialis Anterior where clear signals of up to 1mV p/p may be expected. Hand responses can also be recorded to act as controls. Although the painless aspects of magnetic stimulation are no longer important with the patient under anaesthesia, it is possible to directly compare responses with pre- and post-operative data obtained using the magnetic stimulator. Additionally the lack of direct contact with the patient eliminates the risk of electrode burns possible with high power electrical stimulators.

Figure 16: Example Responses obtained using the Magstim QuadroPulse, Train of Four Magnetic Stimulator, together with the set-up shown in the previous figure. Single or double pulses produced no responses (when using propofol based anaesthesia) whereas three and four pulses produced large amplitude motor evoked potentials. The advantage of three pulses over two are clear-cut whereas the advantage of four pulses over three becomes apparent only in the more difficult cases. 

Waveform reproduced with the kind permission of Dr. Lavern Gugino.
Figure 16a: The above drawing shows a possible set-up for monitoring upper limbs during cervical surgery. In contrast to the monitoring of lower limbs where the double cone coil was used, the most suitable coil for recording upper limb responses is the curved round coil. The use of epidural electrodes is not necessary but is shown here to demonstrate alternative recording techniques. Responses can be recorded using conventional evoked potential equipment.

Figure 16b: Responses obtained using the Magstim QuadroPulse, Train of Four Magnetic Stimulator, together with the set-up shown in the previous figure. Responses produced with the kind permission of Dr. Lavern Gugino. The two sequental traces show the stability of responses from one stimulus train to the next and over a period of 30 minutes. The interlaced traces show alternate responses obtained with the polarity of the magnetic pulse reversed from one pulse to the next.
difficulty passing through bone at very much lower current levels of around 20mA/cm², instead of the 1A/cm² of electrical cortical stimulators. Additionally, magnetic stimulation’s non-invasive and more comfortable characteristics give the advantage of allowing a pre-operative baseline to be established and a post-operative assessment to be carried out. This enables meaningful comparisons to be made before, during and after surgery.

Cortical magnetic stimulation in the awake subject typically produces a descending volley consisting of several direct waves (I-waves) with 1-3ms separation. The descending volley summates and triggers the spinal alpha motoneurone with the impulse leading to the muscle where the muscle action potential (MAP) is recorded in the normal manner. With most anaesthesia, however, the I-waves are significantly reduced in number or are eliminated completely. With the advent of short interpulse magnetic pulse trains it has become possible to mimic the natural descending volley even during depressive anaesthesia which would normally reduce or eliminate the I-waves. The technique utilises temporal summation at the spinal alpha motoneurone to produce the MAP in distal muscles. This enables the use of more common anaesthetic agents, such as propofol, and possibly even with inhalation agents.

Work by Taylor et al. [J. Neurol, Neurol surg & Psychiat, 1993, 56: 104-106] demonstrated the benefits of using electrical pulse pairs for spinal MEPs. Temporal summation at the spinal level significantly enhanced the MEP response allowing the use of “good surgical anaesthesia”. This technique was then extended successfully to the use of the BiStim Module with a Double Cone.

Option for Operating Room Monitoring

- Conventional single pulse magnetic stimulators can only be used for reliable motor tract monitoring if special anaesthetic regimen are used (e.g., based on etomidate and/or ketamine). More common anaesthetics such as propofol tend to obliterate lower limb responses. Experimental shows that the D-wave remains intact but the I-waves are reduced in number and amplitude. Epidural electrodes can, therefore, be used to record the D-wave directly.

- Temporal summation can be used with trains of two or more pulses to get muscular responses with more common anaesthesia. The Magstim QuadroPulse is unique in that it is capable of very closely spaced trains (1ms+). Using a train of four, interpulse spacings of around 3ms appear to be optimal giving robust MEPs with propofol based anesthesia. Large amplitude bilateral responses are obtainable with single stimuli trains without averaging.
A pulse separation of 2ms was used with propofol as the main anesthetic. Large motor evoked potential responses (~1mV/p/p) were present when using magnetic pulse pairs and absent with a single pulse, clearly demonstrating the benefit of using paired stimuli. The stimulation and recording sites were over the vertex and tibialis anterior muscles respectively. An example of the facilitation possible in an awake subject was shown in Figure 13.

As a result of the work with the BiS stim module, a new generation of magnetic stimulator capable of longer trains with higher output power was developed. This instrument, Magstim Quadro Pulse, is capable of very close interval (1ms+) trains of up to four pulses. With the O.R. set-up shown in Figure 15, it has proven possible to obtain clear, stable and large responses before, during and after spinal surgery. Examples are shown in Figure 16. The block diagram is shown in Figure 17.

Magnetic stimulators are also used for the purpose of operating room peripheral nerve monitoring of certain cranial and peripheral nerves (e.g. facial nerve), where anesthetic effects are less important [Thumfart WF et al. Ann. Otol. Rhin. Laryng. 1992, 101: 629-634]. The additional depth of penetration in magnetic stimulation of ten allows the stimulation of nerves not readily accessible to conventional electrical stimulation. Further information regarding the stimulation of facial nerve can be found on Page 21.

**Cervical Spondylosis**

Magnetic stimulation is also used in the early diagnosis and assessment of spinal disorders such as Cervical Spondylosis. The testing procedure is simple, takes less than 45 minutes to carry out in the majority of cases, and is virtually painless.

The tests are based on the fact that the muscles in the shoulders and arms are fed from different cervical nerve roots and that the anatomical connections are well understood. Hence, by looking at the responses after cortical and peripheral stimulation from carefully chosen muscle groups, the progress of the impulse can be monitored through the brain, specific nerve roots and nerve trunks.

The muscles used in this example are the Biceps, typically fed by C5, C6, and C7 nerve roots, and First Dorsal Interosseous (FDI), typically fed by C8 and T1 nerve roots (see Figure 18). This choice allows the differentiation between upper and lower cervical disorders.

The stimulating coil is used cortically and responses are obtained from the left and right Biceps and FDI. The coil is then positioned over the cervical nerve roots and at the Erb’s point and responses are once again recorded from the left and right muscles. In the case of the FDI muscles conventional F-wave recordings can also be used to differentiate between central and spinal nerve root lesions. Example response waveforms are shown in Figure 19.

With cortical stimulation the latencies from the Biceps and FDI comprise the Central, Nerve Root and Peripheral conduction times (C+R+P). With the coil placed over the cervical roots the measured latencies relate to
the Peripheral segment of the motor pathway (P) excluding the nerve root.

Subtraction of the second latency (P) from the first (C+R+P) gives the time taken for the impulse to travel from the motor cortex to the exit of the motor roots from the intervertebral foramen (C+R). This is referred to as the central motor conduction time.

F-wave measurements which include the spinal nerve roots and the reflex centre can be combined with magnetic stimulation to calculate the actual central motor conduction time and also to separate out conduction times in the spinal nerve roots. As an example, subtraction of twice the distal conduction latency, obtained using magnetic stimulation at the neck, from the sum of the F-wave and M-wave latencies gives the time taken for the impulse to travel through the spinal nerve roots and the reflex centre.

The latency, amplitude and waveform measurements are sensitive indicators of location and severity of cervical spine disorders. In addition, comparison of the responses from left and right hand sides allows the determination of the laterality of the abnormality. The test is used to indicate, quantify and monitor the progress of the spinal disorder, confirm radiological and clinical findings, and also to indicate the level of involvement of motor pathways in patients with soft tissue injury.

**Phrenic Nerve Stimulation**

The Magstim is used for bilateral or unilateral phrenic nerve stimulation. Its advantages over electrical stimulation are ease of positioning, repeatability and much improved patient tolerance. It is used reliably to assess the function of respiratory muscles by measuring the diaphragmatic response and strength. The ease of application and patient comfort makes it possible to also carry out longitudinal monitoring studies. Magnetic stimulation can be used to obtain responses even when the phrenic nerve cannot be located using conventional electrical stimulation.

The technique described here has been used by the Kings College and Brompton Hospitals in London. With the neck flexed the standard stimulating coil is positioned over the spinous processes of C6/7 and the magnetic stimulus is given with the patient at FRC (see Figure 20). This allows for bilateral phrenic nerve stimulation. The coil is moved up or down the midline until the maximum response is obtained. Oesophageal and gastric pressures are recorded from latex balloon catheters. Twitch diaphragmatic pressure is then computed by the subtraction of oesophageal from gastric pressure, using $P_{di}$ at FRC as the reference point.

Unilateral phrenic nerve stimulation can also be achieved by using a single coil placed at the side of the neck. Diaphragmatic muscle action potentials can then be recorded using surface electrodes placed in the 7th and 8th intercostal spaces. Responses obtained when using a double small coil are shown in Figure 20 demonstrating supramaximal stimulation.

A development is with two individual figure of eight coils connected to a BiStim set-up allowing almost simultaneous discharge for bilateral stimulation. The advantage of this approach over a single coil placed at the back of the neck is less current spread and better patient comfort. The slight disadvantage is that the two coils
need to be positioned individually as is the case with electrical stimulation.

Twitch $P_{di}$ measurement may be used to confirm or refute the diagnosis of bilateral or unilateral diaphragm weakness. In patients with neuromuscular disease Twitch $P_{di}$ may permit accurate clinical assessment of disease progression. Ongoing developments include the adaptation of magnetic stimulation for use with neonates or children. Additionally, by measuring pressure changes in the mouth (Twitch $P_{mo}$) it is now possible to obtain a non-volitional measure of diaphragm strength.

**Urology**

Recent advances in the diagnosis and monitoring of urogenital tract dysfunction have become of interest to the Urologist, Neuro-Urologist and Urosurgeon alike. The contribution of the somatic fibres to the innervation of the lower genitourinary tract is marginal compared to that of the autonomic nerves. Magnetic stimulation allows the investigation of central motor pathways and autonomic nerve function alongside the more traditional techniques which look at somatic pathways only.

Pelvic floor motor evoked potentials (MEPs), in response to either cortical or peripheral stimulation, can be recorded on standard EMG or evoked potential equipment. Responses can be picked up by concentric needle electrodes from the muscles concerned (e.g. anal or periurethral sphincter, bulbocavernosus muscles, or detrusor muscle), or by sphincter plugs containing silver plated electrodes. Stimulation of the motor cortex gives an overall response which includes conduction along both the central and peripheral portions of the motor pathway. As an example, stimulation of lumbar and sacral nerves and roots, as shown in Figure 21, is also possible.

By subtracting the response latency to peripheral stimulation from the response latency to cortical stimulation,
the central motor conduction and spinal root conduction time is obtained. The results can then be compared to tables of normal responses for interpretation. Both congenital and acquired disorders such as menigomyeloceles, multiple sclerosis, Parkinson’s disease, pelvic floor disorders including lesions of the sacral cord, corda equina, the motor branch of the pudendal nerve, and fractures of the pelvic bones, can be evaluated by this method. Recent research is revealing exciting prospects for non-invasive control over bladder emptying and bladder training.

Ad deionally magnetic stimulation of the lumbar spine can be used to record sensory evoked potentials, for example from the sensorimotor cortex. The main advantage is that of patient comfort where conventional electrical stimulation may be unbearable [Tsuji S, et al. Cortical Somatosensory Potentials Evoked by Magnetic Stimulation of Thoracic and Lumbar Roots. Neurology, 1993, 43: 391-396].

Stroke

The technique of magnetic stimulation is being used for the prognosis and monitoring of stroke affecting the central nervous system. When applied soon after the onset of stroke - within 7 days - it provides important early data regarding the prospects of recovery, especially movement. It is a quick test to conduct and its non-invasive nature allows for repeated use to monitor progress without patient discomfort.

A accurate diagnosis of stroke is currently achieved through clinical examination and scanning techniques. It is, however, difficult to determine the prospects for recovery. Such improvement in monitoring is important to determine patient need, arrange physiotherapy where appropriate, and give relatives a more accurate indicator of recovery.

In its simplest form motor evoked responses present on the parietic side precede good recovery and an absence of any response in the parietal cortex.

The test is simple and quick to carry out and when needed other parameters such as response amplitude, stimulation thresholds and conduction latency are also determined to allow finer prognostic distinctions to be performed. These measurements also form the baseline data for long term monitoring.

**Stimulation:** The initial examination should be carried out soon after the onset of stroke, preferably within 7 days. Cortical stimulation can be achieved through the use of a variety of coils. A circular coil is generally used to record upper limb responses and the Double Cone Coil (see Table 1 for details) for lower limb responses. Typically coils are placed centrally over the vertex and the stimulating power is gradually increased until a response is obtained. Typically two or more muscles are studied in the hands and the legs for both the parietal and non-parietal sides.

**Threshold:** With the coil placed over the vertex and the target muscle relaxed, the threshold of stimulation is noted. The results are compared either with responses from the non-parietal side or tables of normal data or both. If it is not possible to obtain a response from the target muscle at high power levels the patient is asked to attempt to contract the target muscle to facilitate a response. Where cooperation is difficult to obtain, facilitation can be achieved by using the contralateral side, the facial muscles, or by using paired magnetic stimuli to include a facilitory pulse. In a number of patients no response will be obtainable even at maximum power level. In such cases prognosis is poor.

**Response Amplitude:** Once the presence of a response has been established response amplitude, which is also a sensitive indicator, is measured. Largest responses are obtained with slight contraction of the target muscle at around 20% above threshold. Response amplitudes, however, are quite variable and depend on other factors such as electrode position. Paired magnetic stimuli can be used to halve the coefficient of variation.
tion of amplitude, particularly in conditions where no collaboration is possible. As before, this measurement can be compared either with responses from the non-paretic side or tables of normal data or both.

The central motor conduction time (CMCT) is calculated by subtracting peripheral latency from the cortical to muscle conduction latency. Shortest CMCTs are measured with facilitated muscles. There are numerous published tables of CMCT available for many muscles through out the body. The CMCT is typically normal in the case of cerebral stroke but can be abnormal with non-cerebral stroke.

Interpretation of Results: A normal response predicts good and complete recovery. Absence of a response indicates poor recovery. Although in some 5-10% of patients without an initial response some degree of recovery is possible, complete recovery remains unlikely. Abnormal thresholds, small response amplitudes or a long CMCT all indicate motor nerve involvement. The prognosis is variable to good depending on the number of parameters which are abnormal. Where magnetic stimulation is carried out in the first few days, the results form the basis of a very early prognostic indicator of motor recovery.

Other related applications: The ability of magnetic stimulation to non-invasively stimulate the motor cortex allows it to be used for a variety of other central motor disorders and injuries to complement other techniques.

Coma: Similar measurement techniques to that described in this section also apply in comatose patients, with the exception of voluntary effort to facilitate responses. Several facilitation techniques have been shown to also work with unconscious patients. These include the use of tendon vibrator and magnetic pulse pairs and pulse trains to produce temporal summation. In a case study it proved possible to obtain distinct responses from the patient in full coma and sedated with flunitrazepam.

Facial Nerve

Magnetic stimulation has been used for across-the-lesion testing of the facial nerve function. It complements MRI by looking at the function instead of the anatomy of the nerve. Both cortical and peripheral motor areas as well as the intracranial part of the facial nerve can be assessed. Coil positioning for peripheral stimulation of the facial nerve and typical responses are shown in Figure 22 using the circular 50mm coil (see Table 1). The actual site of peripheral nerve stimulation was shown to be in the labyrinthine segment of the facial canal [Rösler et al. EEG, 1991, Suppl 43: 362-368]. Cortical stimulation can also be achieved through the use of the same coil placed over the contralateral motor cortex allowing the calculation of the central motor conduction time. This method makes it possible to assess the motor routes to the facial muscles over distinct segments and to provide evidence of facial nerve lesions located in the facial canal at an early stage, such as in Bell’s Palsy. Transcranial central stimulation of the trigeminal, hypoglossal and accessory motor pathways and extracranial stimulation is also possible. Clinical uses for these techniques include the pre-operative assessment of facial nerve involvement in patients with acoustic neuromas which can be detected by transcranial mag-

Figure 22: Peripheral stimulation of the facial nerve can be achieved at low power levels by the placing the coil centred above the ear. This technique can be used for diagnosis as well as for intra-operative monitoring purposes where anaesthetic effects are less important than with the stimulation of the central nervous system. Typical responses recorded over the frontal muscle using the Small 50mm Coil are shown below.
Magnetic stimulation even in patients with small and medium sized tumours but with clinically normal facial function.

**Rapid-Rate Magnetic Stimulation**


Following on from earlier research with repetitive transcranial magnetic stimulation (rTMS) in 1990, one of the first papers to draw at attention to the potential of this technique was "Induction of speech arrest and counting errors with rapid transcranial stimulation", Pascual-Leone et al. Neurology, 1991;41:697-712. Trains of stimuli of 25Hz lasting up to 10 seconds were used to determine the laterality of the speech centres prior to neurosurgery as an alternative to the Wada test. More recent pro tocols have evolved using much slower repetition rates. The use of rapid-rate stimulation has been extended to other cognitive studies and treatment procedures for psychiatric disorders such as depression, schizophrenia, epilepsy, Parkinson's disease and pain relief. Repetitive stimulators are also providing a role for peripheral stimulation for the relief of spasticity in both stroke and multiple sclerosis, in muscle stimulation to simulate a cough in spina!ly injured patients, to look at muscle strength, fatigue and recovery, and aiding in the maintenance of muscle bulk during a period of incapacity. With Urology it may be an aid in the assessment and monitoring of urinary tract dysfunction with the faster rates reducing data acquisition times, allowing fatigue studies, and performing muscle therapies. New clinical papers are emerging which point the way to advanced applications for repetitive stimulation both in diagnosis and therapy.

At this stage it is important to remember that the effects of stimulating the cortex with repetitive stimulation may have important safety considerations and results may not be as anticipated even if established protocols are being used (Pascual-Leone et al. Lancet, 1992, 997). Considerable work is, however, being put into quanti-
fying the risk and developing treatment strategies which minimise the risk of seizures and other possible side-effects. (See also section on Safety and Precautions)

Of course seizures are induced during treatment of depression using ECT (electroconvulsive therapy) and many departments have the means of coping with a seizure. It has always remained possible that, given a sufficient number of magnetic stimuli at a high stimulus strength and repetition rates, a seizure can be induced. Thus, if the potential benefit of a particular treatment outweighs the risk, then the risk of inducing a seizure can be reduced by each or a combination of the following measures:

(a) Stimulate at 90% or less than motor threshold,
(b) Stimulate at the lowest frequency possible,
(c) Stimulate away from motor cortex areas,
(d) Use a focal coil for less current spread,
(e) Stimulate for as brief a period as is possible,
(f) Allow sufficient recovery time after each train.

Repetitive stimulation equipment is now available from all major magnetic stimulation manufacturers some of whom have safety features built into their control programmes, and allow accurate dosage built into their control programmes, and allow accurate dosage to be determined, with the three main parameters being: power level, frequency and train duration (or pulse number). A computer screen from one of the rapid-rate stimulators is shown in Figure 23. Further developments in the interest of operational use and safety permit protocols to be installed or written to specification and precisely delivered. These protocols will automatically be updated with safety parameters built into the programme. The information about subject, date, time, frequency, power levels, train duration etc. is automatically logged onto hard disc for later recall and analysis, or can be printed to hard copy if required. A control screen of the Session Software protocol controller is shown in figure 24.

The high level of interest now shown in TMS and rTMS has led to the formation of the International Society for Transcranial Stimulation, ISTS, which will become a representative and supportive body, co-ordinating developments at both the scientific and regulatory level. The International Federation of Clinical Neurophysiology has appointed a Special Commission to make recommendations for the worldwide use of TMS and the American Academy of Neurology has appointed a Commission to evaluate the clinical usefulness of TMS and to develop a position statement for the AAN on the issue of recommendations for the technology.
<table>
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<th></th>
<th>Sheffield Magstim 90mm Coil</th>
<th>Circular 50mm Type 9999</th>
<th>Circular 70mm Type 9762</th>
<th>Circular 90mm Type 9784</th>
<th>Double 70mm Type 9925</th>
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<tr>
<td>Magnetic Field Strength (T)</td>
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<td>Energy Deposited Per Phase (µJ/cm³)</td>
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<td>2.3</td>
<td>3.0</td>
<td>3.5</td>
<td>5.3</td>
</tr>
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Table 2: Estimated maximum exposure of the human brain to magnetic and electric fields, induced current and charge
Reza Jalinous

Guide to Magnetic Stimulation

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Related Web Sites

http://www.visits.unibezh
http://www.uts.edu/tmn/minorni/TMSsearch.html
http://www.syme.mtu/dpesemilatranscri.fun.html
http://www.psy.helsinki.fi128.214.75.169/magstim.html
http://members.aol.com/magstim/
http://www.mageal.com/ (Starts June 1998)

Safety Precautions & Issues

The presence of pacemakers and other electronic implants must be considered as a contraindication in case of interference due to the induced electric fields and currents resulting from the magnetic pulse. Although the stimulating coil does not have a magnetic effect on small metallic objects, such as stainless steel aneurysm clips placed a few inches away, it could exert a more significant physical force on larger metal objects within 20 cm of the coil. Much depends on the size of the object, its conductivity and whether or not it is ferromagnetic.

Studies with very powerful magnetic stimulators with energy densities of 10 kJ or more, giving an order of magnitude increase in the stimulating power over conventional stimulators, have shown that it is possible to induce an electric and magnetic field in a dog’s heart and also the relatively fast pulse rise time of the magnetic field waveform. Clinical magnetic defibrillators, however, remain a possible safety concern for the future.

Kindling is a phenomenon whereby a permanent epileptic focus is induced by very many repetitive stimuli given to the brain using implanted electrodes. Goddard et al. [Goddard et al. Exp Neurol, 1969, 25: 295-330] were unable to induce kindling in animals at frequencies of less than 10 Hz, irrespective of the number of stimuli given. Using magnetic stimulators, with a maximum discharge repetition rate of less than 1 Hz, no risk of kindling should exist. In addition, the number of stimuli given to any one subject is very low compared to the minimum required to cause kindling in animals. The risk of kindling in the case of fast repetitive magnetic stimuli (10 Hz) remains unknown.

In Table 2 the calculated figures of magnetic and electric field strengths, induced current, charge density, and deposited tissue energy are provided for the Magstim 200 together with the original Sheffield Magstim. In the calculation of these parameters, used for considering the physiological effects of the magnetic stimulator on humans, it has been assumed that the brain does not lie closer than 5 mm to the coil surface. Other than in the case of exposed brain this should not normally be the case. A uniformity of 0.35% for magnetic and electric fields of the order of 2 T can have any harmful effects. The current U.K. guidelines for whole body exposure to static magnetic fields during magnetic resonance imaging is 2.5 T [NRPB, Radio graph, 1984, 50:220]. In addition, it should be remembered that, in most cases, the output from a magnetic stimulator lasts only 1 ms and there is no obvious reason why purely magnetic effects from such a pulse should be greater than from a static field.

The figures in Table 2 show that the maximum electric field and induced current density are 540 V/m and 19 mA/cm² respectively. As expected these levels of exposure are similar to conventional electrical stimulation using surface or needle electrodes which have proven quite safe. In addition the induced stimulus from the magnetic stimulator is in the ently charged balanced, eliminating the possibility of electrolytic cell damage in cases of prolonged stimulation. The figure of charge density per phase of 1.4 μC/cm² is well below the minimum figure of 40 μC/cm² at which evidence of neural damage has been found when stimulating for long periods at 50 Hz [Agnew et al. Neurosurg, 1987, 20: 143-147]. In addition the total charge delivered is less than 0.1% of that used for...
ECT. The maximum calculated figure of 5.3 µJ/cm³ for the energy density per pulse dissipated in tissue is extremely small, giving a temperature rise of only 10°C in the tissue. At a repetition rate of 1 Hz, the total power dissipation of less than 1 mW for the whole brain is more than four orders of magnitude lower than the adult brain base met a bolic rate.

It has been suggested by Counter et al. [Neurology, 1990, 40: 1159-1162] that the discharge click noise produced by a 5 cm stimulating coil causes hearing loss in albino and chinchilla rabbits with the coil placed over the external auditory meatus. The sound output from the coil was measured to be as much as 157 dB peak SPL. Barker and Stevens [Physiology Soc, London, 1991, 19: 14P] measured sound output from the standard commercial coil (Type 9784 in Table 1) supplemented with the Magstim 200 and found it to be a maximum of 124 dB(A) on the coil surface falling to 117 dB(A) 50 mm away from the coil surface. These values are within that required by the U.K. Noise at Work Regulations (1989) as long as the number of discharges at the maximum power level does not exceed 4000 stimuli per day - clearly an unlikely event. It should be noted that the discharge click noise depends on the coil size (small coils are louder than larger versions), power level and most importantly the manufacturing method.

**Low Frequency Stimulation**

Since 1985 many tens of thousands of subjects have been examined using low repetition rate (<1 Hz) magnetic stimulators to assess motor function of the peripheral and central nervous system. There is now a considerable volume of data supporting the safety of magnetic stimulation. There have been no ill effects reported with magnetic stimulation of the peripheral nervous system and in the case of cortical stimulation the incidence of side effects has been extremely low and well within that expected by available statistics for various patient groups [Kandler R, Lancet, 1990, 335, 1: 469-70]. An area of concern has been the triggering of epileptiform activity in inviduals at a high risk to epilepsy [Hömberg & Netz, 1989, 2: 113]. Nevertheless one of the areas where magnetic stimulators have been successfully used has been in the study of epilepsy and the determination of the site of the epileptic focus [Hufnagel et al. Ann Neurol, 1990, 27: 49-60]. Over all magnetic stimulation has proved to be a very safe and effective clinical tool.

**High Frequency Stimulation Guidelines**


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Reza Jalinous, Ph.D.
E-mail: rjalinous@compuserve.com
The Magstim Company Limited, U.K.
Tel: +44-1994-240798
Fax: +44-1994-240061
Revision: March 1, 1998
<table>
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What's New!

The following is a list of selected new papers added from August 1996 through to December 1997. The same papers are also organised by discipline in some 69 sections together with earlier selected papers. Where more recent papers have been available, the majority of papers published before 1995 have been removed.


Curt A, Diez V. Prognosis of spinal cord injury. The meaning of clinical and electrophysiological findings. Nervenarzt, 1997; 68: 485-495. (German)


Davey NJ, Pur i BK, Lewis HS, Lewis SW, Ell away PH. Effects of antipsychotic medication on electromyographic responses to transcranial magnetic stimulation of the motor cortex in schizophrenia. Journal of Neuropsychiatry and Psychiatry, 1997; 154: 120-129.


Detsch C, Kochs E. Effects of Ketamine on central nervous system function. A naesthetist, 1997; 46: S20-S29. (German)


Feistner H, A wis zu s F, Sa iler M, H innrichs H, H einze HJ. A method for rapid response estimation of single human motorneurones to transcranial magnetic and peripheral electrical stimulation. Z. EEG-EMG, 1996; 27: 80-84. (German)


Schubert M. Clinical and Experimental aspects of magnetic stimulation. Z. EEG-EMG, 1997;28:114-118. (German)


References Organised by Discipline

1. Anaesthesia

Detsch C, Kochs E. Effects of Ketamine on central nervous system function. A aesthesis, 1997;46:S20-S29. (German)


2. Anaesthesia, Primate Studies


3. A.L.S.

See section on Motor Neurone Disease.

4. Ataxias


5. Auidiology


6. Basic P rinciples, T echnical A spects


W assermann EM, Grafman J, Combining transcranial magnetic stimulation and neuroimaging to map the brain T rends in Cognitiv e S ciences, 1997;6:199-201.

7. B asic P rinciples, C linical A spects


8. B ends ( D e compression S ickness)


9. B rain M apping


10. Brainstem Stimulation


11. Cerebellum


12. Cerebral Palsy

See section on Stroke.

13. Cerebral Palsy

Aiso see section on Paediatrics.


14. Clinical Applications and Reviews


R oricht S, Irlbacher K, Petrow E, M eyer BU. Normative data for callasally and corticospinally mediated electromyographic effects in hand muscles following a hemisphere-selective magnetic cortex stimulation in man. Z. EEG-EMG, 1997;28:34-38. (German)


Schubert M. Clinical and Experimental aspects of magnetic stimulation. Z. EEG-EMG, 1997;28:114-118. (German)


15. Coil Configuration


16. Colorectal Disease

Aiso see section on Urology.


17. Coma


18. Computer Modelling


19. Diabetes


20. Dystonia


21. Electrical versus Magnetic Stimulation


22. Electroconvulsive Therapy

A disclosure on Psychiatry.


Kirkcaldie M, Pridmore S, Reid P. Bridging the skull: Electroconvulsive therapy (ECT) and repetitive transcranial magnetic stimulation (rTMS) in psychiatry. Convulsive Therapy, 1997;13:83-91.


23. Epilepsy


24. Experimental Research


Schubert M. Clinical and Experimental aspects of magnetic stimulation. Z. EEG-EMG, 1997;28:114-118. (German)

25. Facial Nerve


26. Facilitation

A also see section on Silent Period.


27. Gastroenterology

Also see Stroke.


28. Guillain-Barré Syndrome


29. History and Background


31. Huntingdon’s Disease


32. Hysteria


33. Inhibition

See also section on Silent Period.


34. Kallman's Syndrome


35. Locked-in Syndrome


36. Magnetic Pulse Pairs

37. Mirror Movements


38. Miscellaneous Diseases


39. Motor Cortical Inhibition

See section on Inhibition.

40. Motor Evoked Potentials


41. Motor Neurone Disease


42. **Multiple Sclerosis**


43. **Myoclonus**

A see section on Epilepsy.


44. **Neuropathy**

A see section on Peripheral Stimulation.


45. Operating Room Monitoring

A Iso see section on A naesthesia.


46. Paediatrics


47. Pain


48. Paraplegia

See section on Spinal Disorders and Urology.

49. Parkinson’s Disease

See also section on Silent Period.


50. Peripheral Stimulation

See also sections on Facial nerve, Thoracic M edicine and U rology.


51. Plasticity


52. Psychiatry

A lso see sections on Parkinson’s Disease, Safety and Speech.


H aag C, Padberg F, M oller H J. T ranscranial magnetic stimulation (TMS). A diagnostic tool from neurology as a therapy in psychiatry? N erv enarzt, 1997;68:274-278. (German)


53. Pulmonary Medicine

See section on Thoracic Medicine.

54. Rehabilitation and Therapy


55. Repetitive Transcranial Magnetic Stimulation

A Iso see section on Safety.


56. Research
See sections on Experimental Research, Computer Modelling, Anaesthesia; Primate Studies, Coil Configuration, etc.

57. Respiratory Medicine
See section on Thoracic Medicine.

58. Rett’s Syndrome

59. Safety


60. Silent Period

A Iso see section on Inhibition.


Haug BA, Kukowski B. Latency and Duration of the Muscle Silent Period Following Transcranial Magnetic Stimulation in Multiple Sclerosis, Cerebral Ischemia, and Other Upper Motoneuron Lesions. Neurology, 1994, 44:936-940.


61. Sleep


62. Somatosensory Evoked Potentials


63. Speech - Wada Test


64. Spinal Disorders and Injuries


Curt A, Diez V. Prognosis of spinal cord injury. The meaning of clinical and electrophysiological findings. Nervenarzt, 1997;68:485-495. (German)


65. Stroke


66. Thoracic Medicine


Travaline JM, Sudarshan S, Criner GJ. Recovery of PdiTwitch following the introduction of diaphragm fatigue in normal subjects. Am J. of Respiratory and Critical Care Medicine, 1997;156:1562-1566.


Zifko UA. Electrophysiology of respiration. Nervenarzt, 1997;68:945-955. (German)

67. Urology

A Iso see section on Colorectal Diseases.


68. Visual Cortex/Ophthalmics


69. Wilson's Disease


