Reduction of Peripheral Nerve Stimulation via the use of Combined Gradient and Uniform Field Coils

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Introduction

The need to avoid peripheral nerve stimulation (PNS) due to rapidly switched magnetic fields sets an upper limit on the magnetic field gradient strengths that can be employed in MRI. PNS results from the electric fields induced in the conducting tissue of the body by temporally varying magnetic fields. In a complex heterogeneous conducting object such as the human body, the pattern of induced electric field can not be simply related to the imposed magnetic field variation. However it is generally the case that the stronger the magnetic field experienced by the body, the larger the peak induced electric field. Consequently limiting the largest magnetic field which the body experiences for a given gradient strength is a sensible approach to producing larger switched gradients at the PNS threshold. It is particularly important to limit the magnitude of the magnetic field, |B|, in body regions that have large cross-sectional area, since high fields in such regions lead to large rates of change of the flux linked and consequently large induced electric fields. In assessing the field magnitude |B|, it is necessary to consider the 'concomitant fields', B_x and B_y, as well as the axial component of the field, B_z, that is important for the evolution of the NMR signal; analysis of Maxwell's equations indicates that if B_z varies with position, at least one of the comcomitant fields must also vary spatially. This behaviour means that for all three types of gradient, one component of the field varies linearly with axial position. Since the height of the human body is considerably greater than its breadth or width, it is generally the axially varying component of the field that produces the largest field magnitude in the body. In designing a gradient coil to allow higher rates of change of gradient with time, d*G*/d*t*, to be produced at PNS threshold, it is therefore common practice to reduce the axial extent of the coil's region of linearity (1, 2). For axial gradient coils, this works by limiting the peak magnitude of B_z in the body, while in the case of *x*- or *y*-gradient coils it is the peak magnitude of the concomitant field that is limited. This approach however has the disadvantage of reducing the extent of the region over which imaging can be carried out. An alternative method is to add a uniform field varying synchronously with the applied field gradient (3, 4). This can reduce the magnitude of the largest field component in sensitive regions of the body, without affecting the extent of the region over which a uniform gradient is produced. For example with the head (or hips) centred in the gradient coil, it is possible to reduce the peak field produced in the torso by adding a uniform field, at the expense of increasing the field in the head (or legs). Since the larger cross section of the torso generally leads to the induction of larger electric fields for a given rate of change of magnetic field than in the head or legs, this approach has promise for allowing larger values of dG/dt to be achieved before stimulation occurs. In the case of a transverse gradient, implementation of this method involves adding a uniform concomitant field, while for a *z*-gradient it is a uniform field B_z which must be synchronously applied. To evaluate the efficacy of this approach, we have therefore designed and constructed coil pairs that produce: (i) a *y*-gradient and a uniform B_y-field; (ii)a *z*-gradient and uniform B_z-field. These have been tested in ethically approved volunteer studies to evaluate the gains in d*G*/d*t* that can be achieved at PNS threshold.

Method

Unscreened, *y*- and *z*-gradient coils of approximately 62 cm diameter were designed using conventional methods (4) so as to have low inductance while producing a field which deviated from linearity by less than 5 % within a 40 cm diameter spherical volume. This process yielded a *y*-gradient coil with an efficiency of 93 $\mu Tm^{-1}A^{-1}$ and an inductance of 620 µH (4) and a z-gradient coil with 62 $\mu Tm^{-1}A^{-1}$ efficiency and 141 µH inductance. Low inductance coils that produce a uniform field, B_y or B_z were designed using a similar approach. In designing these coils it was ensured that addition of the field from the uniform field coil, did not compromise the size of the region of gradient linearity. Two uniform field coils were produced for each axis, to allow the effect of coarsely varying the strength of the synchronously applied uniform field to be evaluated. The coil properties, including the axial position where the field on axis is nulled when the uniform field coil is driven in conjunction with the appropriate gradient coil are detailed in Table 1.

Table 1

Coil Type	B_y-strong	B_y-weak	B_z-strong	B_z-weak
Efficiency	23 μTA^{-1}	11.5 μTA^{-1}	12.7 μTA^{-1}	8.2 μTA^{-1}
Inductance	330 µH	85 µH	78 µH	34 µH
Axial null	25 cm	12.5 cm	20.5 cm	13.2 cm

Coils were constructed by winding 3 mm diameter copper wire onto the surface of a fibre-glass cylinder and then overlaying with further glass-fibre and epoxy resin. Volunteer stimulation experiments were carried out by driving the gradient coils with a sinusoidal current waveform of 2.4 kHz frequency. The gradient and uniform field coils were connected in series, to ensure that the currents in the two coils were in phase. Each coil arrangement was also connected in series with an appropriately chosen capacitance to form a series LCR circuit with a 2.4 kHz resonant frequency. This allowed access to large values of dG/dt (350 $Tm^{-1}s^{-1}$ and 600 $Tm^{-1}s^{-1}$ for the *y*- and *z*-coil arrangements respectively) with the limited available drive voltage of 300 V. Experiments were carried out with subjects located at four different axial positions within the coil set, such that the: (a) head [13 ± 2cm]; (b) heart [44 ± 4 cm]; (c) hips [-82 ± 7cm]; (d) knees [-124 ± 7cm] were located at the iso-centre of the coil arrangement. Here the numbers in square brackets represent the average (± standard deviation) of the distance from the top of the head to the coil centre. Three different coil arrangements ((i) gradient alone; (ii) gradient plus weak uniform field; gradient plus strong uniform field) were evaluated for each axis. In each case a burst of 32 cycles of the current waveform was applied approximately once a second with the strength gradually increasing until the subject reported the onset of stimulation. At this point the subject indicated the site of stimulation and the form of the sensation. 20 and 15 volunteers have been studied for the *y*- and *z*-coil arrangements respectively. Two subjects were tested at a wider range of axial locations within the *y*-coil arrangement.

Results and Discussion

The results of the volunteer studies for the various coil arrangements are summarised in Table 2. For each configuration the average and standard deviation of the value of dG/dt at the onset of stimulation are detailed. The number of subjects who could not be stimulated at the largest accessible value of dG/dt_{max} is indicated in brackets where relevant. In such cases the value used in forming the average was taken to be dG/dt_{max}. The results show that as expected the threshold for stimulation is considerably lower for the *y*-gradient coil arrangements than for the *z*-coil. This is a consequence of the greater body cross section in the *x*-*z* plane. Table 2 also indicates that for three out of the four body positions studied (head-, hips- and knees-centred), significantly larger rates of change of gradient can be achieved when a uniform field is applied in conjunction with the gradient field. The achievable gains in dG/dt are greater for the *y*-coil arrangement (head ×1.7, hips ×1.5, knees ×1.3) than for the *z*-coil (head ×1.4, hips ×1.3, knees ×1.2). The inability to stimulate large numbers of volunteers in the head- and knees-centred positions means these gain factors are likely to be significantly underestimated. Larger gains were generally achieved using the weaker uniform field coils. The more detailed study of two volunteers indicated that with the available B_y coils there is a range of only ~20-30 cm of body position relative to the centre of the coil's homogeneous region over which the gradient coil alone provides a higher achievable value of dG/dt. The ability to vary the relative strength of the uniform field more widely and thus find an optimum value for each body location (e.g. by using a separate amplifier for each coil) should provide further gains.

Table 2

Body Position/ Coil Type	Head-Centred Threshold ($Tm^{-1}s^{-1}$)	Heart-Centred Threshold ($Tm^{-1}s^{-1}$)	Hips-Centred Threshold ($Tm^{-1}s^{-1}$)	Knees-Centred Threshold ($Tm^{-1}s^{-1}$)
G_y	200 ± 33	255 ± 58	195 ± 30	264 ± 41
$G_y + B_y$-weak	297 ± 43	239 ± 42	283 ± 44	347 ± 31 (11)
$G_y + B_y$-strong	330 ± 46 (16)	169 ± 23	286 ± 47 (4)	335 ± 30 (16)
G_z	395 ± 78	477 ± 80 (1)	398 ± 52	490 ± 98 (4)
$G_z + B_z$-weak	539 ± 102 (9)	371 ± 64	525 ± 75 (5)	564 ± 72 (11)
$G_z + B_z$-strong	540 ± 76 (10)	340 ± 54	501 ± 93 (4)	553 ± 76 (11)

References

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