Shim Coil Design, Limitations and Implications

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Introduction

The term "shimming" originates from the small pieces of magnetic material that were used in the early days of NMR in order to improve the homogeneity of the magnetic field produced by pole magnets. The "shims" of metal acted as local dipole field corrections to the main field, in analogy to the metal "shims" that are applied to automobile wheels in order to balance them by correcting the moment of inertia of the wheel. In MRI one typically wishes to correct the field to be homogeneous to better than 1 part in 10^6–10^7 in order to perform imaging using demanding pulse sequences, such as echo planar imaging, or for performing *in-vivo* spectroscopy.

The central equation governing the magnetic field distribution within a volume through which no current passes is the Laplace equation $\nabla^2\Phi=0$, in which Φ is the scalar magnetic potential. This is most conveniently solved using the spherical polar coordinate system. This framework provides a simple basis set for expressing the magnetic field inhomogeneities in the magnet, and also for generating correction terms using either passive or active shim approaches. By expressing the scalar magnetic potential in this way it then follows that $\mathbf{H}=-\nabla\Phi$, which allows the magnetic field to be calculated. Likewise, Φ can be solved from $\mathbf{H}$ by integration.

Assessing Magnetic Field Homogeneity

Upon installation of a new magnet, one of the first and most important procedures is to establish the homogeneneity of the magnet *in situ*. Often the magnetic field homogeneity is degraded either by inevitable inaccuracies in fabrication, or by local site issues, such as the presence of steel in the building itself, or by asymmetries in the geometry of any magnetic shielding that may be incorporated into the scan room. The most common approach used to map the magnetic field is to use a small NMR coil and sample, that are contained within a plotting rig and that can be moved around the magnet in (typically) spherical coordinates. The magnetic field can be measured simply by recording the Larmor frequency at which the sample resonates at a given position in the magnet. By plotting the field at regular azimuthal angles, radii, and at a number of axial coordinates, it is possible to calculate the field distribution within the magnet bore. This can then be fed into an algorithm that will calculate the required amount and location of passive shim material that must be placed about the magnet bore in order to homogenize the magnetic field. In the case of active shimming an algorithm can be used that calculates the currents that must be applied in superconducting shims in order to remove the appropriate harmonics.

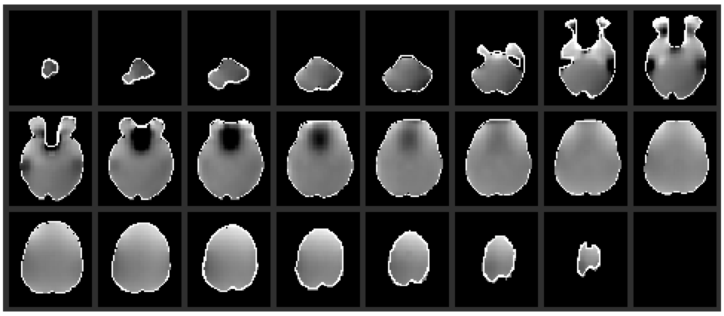

Fig. 1: Field plot obtained in a human brain. Greyscale extremes are +0.6ppm to −0.6ppm.

Once the bare magnet has been shimmed it is often necessary to assess the homogeneity of the magnetic field within a sample placed in the bore. This is because typically each sample will inherently distort even a perfect magnet field, either due to its overall geometry, or to internal compartments of differing magnetic susceptibility. The easiest way to determine the field within the sample is to collect two images, each with different gradient echo weighting. The difference in phase between the two images is then given by $\Delta\phi = \gamma\Delta B_0\Delta TE$, where ΔB_0 is the desired magnetic field inhomogeneity. Fig. 1 shows an example of such a measurement performed in human head, demonstrating the severe intrinsic magnetic field inhomogeneities in the frontal and temporal lobes of the brain.

Passive Shim Approach

Passive shimming of the magnet involves the introduction of small amounts of magnetic material that can be positioned in such a way as to compensate for unwanted terms in the spherical harmonic basis set that describes the magnet field within the bore (Roméo and Hoult, 1984; Hoult and Lee, 1985). This is the most common method for shimming the "bare" magnet once it is installed in the MRI facility. Hoult and Lee have shown that the contribution to the scalar magnetic field potential at point P arising from a dipole positioned at point Q is given by $d\Phi=-(d\mathbf{m}/4\pi).\mathrm{grad}_Q(1/\nu)$, where $d\mathbf{m}=\chi H_z dV\mathbf{k}$ and $(1/\nu)$ is a Green's function described in Hoult and Lee. By solving this equation for a collection of elemental passive shim dipoles, or a set of continuous passive shim rods, it is possible to identify the locations for positioning the passive shim material such that specific unwanted spherical harmonic terms in the plotted field can be eliminated. The advantage of passive shimming is that very high order spherical harmonics can be realised. The disadvantage is that the shim material is temperature sensitive, causing B_0 shifts as the bore heats (e.g. during gradient-intensive sequences).

Active Shim Coil Approach

Super-conducting active shim coils are used in some magnet designs to allow the bare magnet to be shimmed to specification, although most vendors now use passive shimming for this purpose. However, active shim coils are commonly used to provide subject-by-subject corrections to the field via room temperature electrical windings that are either wound on a specific cylindrical former that constitutes the outer structure within the room temperature bore of the magnet, or are incorporated into the gradient coil structure. The coils are wound in such a way as to attempt to reproduce the basis set that is the spherical harmonic solution to Laplace's equation.

Table 1 shows, in a Cartesian reference frame, the basis functions that are commonly encountered in MRI. The basis functions for all degrees, *m* ($m\leq n$), and for orders, *n*, up to three are shown, plus the fourth order term n=4, m=0. Few commercial human scanners have room temperature shim correction terms beyond 2nd order, although some specialist (mostly ultra high field) scanners have terms beyond.

Clearly, the easiest way to generate the n=1 terms is to use small offsets in the main gradient coils themselves. The higher order correction terms may be generated by winding specialist shim coils, usually on a cylindrical former. Various coil design approaches have been described in the literature to provide both the axial (m=0) and transverse ($m\neq0$) harmonics (Roméo and Hoult, 1984).

Shorthand	n	m	Function
Z	1	0	z
X	1	1	x
Y	1	-1	y
Z2	2	0	$z^2 - (x^2 + y^2)/2$
ZX	2	1	$3\,z\,x$
ZY	2	-1	$3\,z\,y$
X2-Y2	2	2	$3\,(x^2 - y^2)$
XY	2	-2	$6\,x\,y$
Z3	3	0	$z^3 - 3\,z\,(x^2 + y^2)/2$
Z2X	3	1	$6\,z^2x - (3/2)\,x\,(x^2 + y^2)$
Z2Y	3	-1	$6\,z^2y - (3/2)\,y\,(x^2 + y^2)$
Z(X2-Y2)	3	2	$15\,z\,(x^2 - y^2)$
ZXY	3	-2	$30\,z\,x\,y$
X3	3	3	$15\,x^3 - 45\,y^2x$
Y3	3	-3	$-15\,y^3 + 45\,x^2y$
Z4	4	0	$z^4 - 3\,z^2\,(x^2 + y^2) + (3/8)\,(x^2 + y^2)^2$

Table 1. List of spherical harmonic orders and spatial functions together with the shorthand commonly encountered. The *z*-axis is defined as superior to inferior.

Auto Shimming Methods

Many MRI systems now incorporate procedures that utilize the available room temperature shim terms in a subject-by-subject fashion. A number of approaches to mapping the magnetic field distribution within the subject can be used. But typically the mapping is done either by collecting gradient echo images at differing echo time, and then by calculating the field from the phase difference between the images (as described above), or by collecting “pencil” profiles along different directions, again using two gradient echo acquisitions at differing echo times (Gruetter, 1993). The benefit of the latter approach is the potential for a speedier acquisition.

Once the magnetic field distribution within the subject is known, an optimization algorithm can be used to optimally adjust the currents in each available room temperature shim coil in order to be able to homogenize the field (Webb and Macovski, 1991; Wen and Jaffer, 1995). Often, such a procedure is further improved by excluding from the fitting algorithm regions of the sample that do not need to be optimally shimmed. For example, in the case of functional MRI regions of the head that lie outside the brain can be excluded, yielding improved correction of the brain region itself (Wilson *et al.* 2002a). One can even weight the shim solution to specific areas within the brain, when there is preferential interest in a particular structure. Such an approach is clearly better achieved using an image based field mapping approach rather than a pencil profile approach.

Dynamic Shimming

In many body regions the spatial frequency of the available room temperature shim terms is not well matched to the desired spatial frequency of the required shim correction. Examples of this include the heart, where local field inhomogeneities from the lungs create field distortions that have a high order spatial dependence, and also in the brain, where distortions induced in the vicinity of the frontal sinuses and mastoid air cells have a very high spatial frequency. In such cases one can use dynamic shimming in order to improve the apparent spatial frequency correction ability of the room temperature shim coils. The procedure involves dynamically altering the room temperature shim coil settings on a per-slice basis. Early attempts at dynamic shimming updated the linear (n=1) shim terms dynamically such that any slice-dependent linear field gradients could be removed (Blamire *et al.* 1996; Morrell and Spielman 1997). Clearly such an approach is only of use in multi-slice imaging, rather than in 3D imaging, but significant improvements can be realised in rapid multi-slice techniques such as echo planar imaging. More recent efforts are attempting to include the higher order shim terms. However, most commercial shim power supplies do not currently offer the high bandwidth that is required in order to actively drive the 2nd and 3rd order room temperature shim terms.

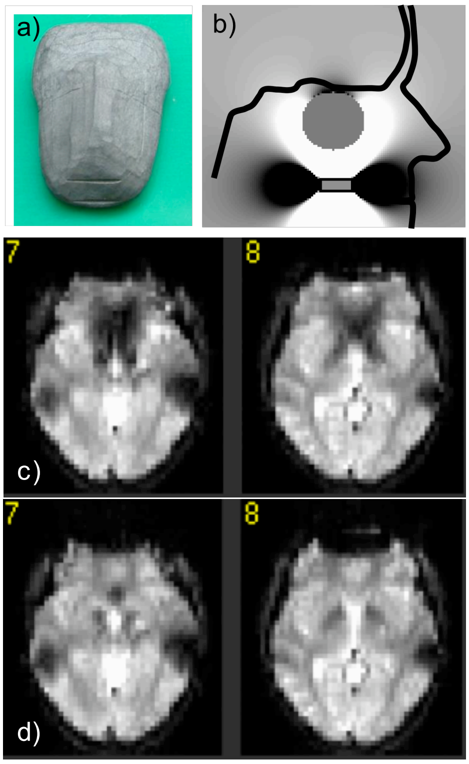

Fig. 2: Example of highly diamagnetic CNPG mouth shim (a), and simulation of correcting effect of the shim when placed in the mouth (b). (c) shows gradient echo EPI images at two slice location without the mouth shim, and (d) shows the same slices with the mouth shim present.

Local Shimming

Another method for overcoming the problem of high spatial frequencies encountered in many practical magnetic field inhomogeneities, versus the low spatial frequency corrections possible using conventional shim coil designs, is to move the shimming technology closer to the subject. A traditional method for improving the magnetic field homogeneity in certain structures is to improve the geometry of the object by, for example, simply placing saline bags over the structure in order to relocate a surface tissue/air interface away from the structure under study. Another proposal for imaging the frontal areas of the brain (notoriously problematic due to various local air pockets) is to use a volume of highly diamagnetic material that is placed inside the mouth (Wilson *et al*., 2002b; Wilson *et al*. 2003). This has the effect of producing a field shift in the relevant region that counters the deleterious effect of the sphenoid and ethmoid sinuses. Significant recovery of signal can be realised using this simple method, as shown in Fig 2.

A related interesting approach is to use an active electrical coil in the mouth cavity that generates a field extending into the inferior frontal lobes that again counters the intrinsic field inhomogeneities (Hsu and Glover, 2005). Yet another approach was proposed by Jesmanowicz *et al*. (2001) who proposed designing a tailored pattern of ferromagnetic photocopier toner on a printed sheet that can be inserted into the head coil around the subject.

Acquisition Based Methods

One of the particular problems of gradient echo imaging is the signal drop out that occurs when there is significant magnetic field inhomogeneity, particularly when this inhomogeneity has a component in the slice direction. This leads to the sort of signal loss shown in Fig. 3a. One method for addressing this problem, albeit with a penalty in scan acquisition time, is to repeat the image acquisition a number of times, each time with a different value for the slice-refocusing gradient pulse. This so-called Z shimming approach was first introduced by Frahm *et al*. (1988), but has since been used by numerous authors in ever more sophisticated variants (Yang *et al*., 1998; Glover, 1999; Constable *et al*. 1999). Fig. 3b shows an image that has been reconstructed using the GESEPI approach (Yang *et al*. 1998) that effectively Fourier encodes in the slice dimension to recover the k_z-shifted signal caused by slice-direction gradients.

Another acquisition method that has been attempted to recover signal that has been lost by intravoxel dephasing in gradient echo experiments is to use hybrid RF pulses that selectively pre-set the excitation phase of the spins in different parts of the slice such that the spins are all in phase at the time of the gradient echo (Stenger *et al*., 2000; Stenger *et al*. 2002). In principle the required pre-phase can be set based on knowledge of the field inhomogeneity profile within the slice. The disadvantage of the method is that it requires very long complex RF excitation pulses, or multi-shot excitation, in order to achieve the possibility of pre-setting the phase of the spins to different values through the slice direction.

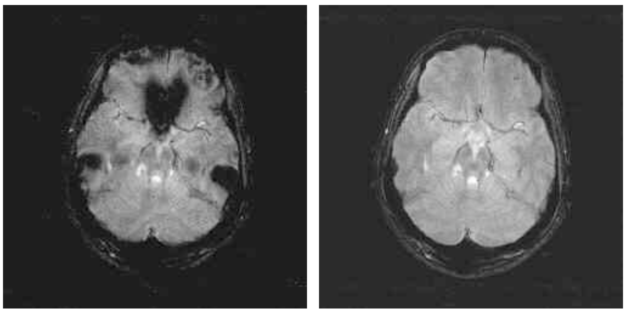

Fig. 3: Images showing the signal loss from a 5mm thick axial slice in human brain (left) and the recovered signal from the GESEPI technique (right). Reproduced with permission from Yang et al. (1998).

Echo Planar Imaging

Echo planar imaging is a pulse sequence that is particularly affected by magnetic field inhomogeneity and so a more detailed treatment of the effects is provided here. The main artefacts introduced into EPI by B_0 inhomogeneities can be broadly divided into geometric distortions and signal loss artefacts, the latter occurring in gradient echo sequences. Geometric distortion in the echo planar image is most prominent in the phase encode direction, due to the low pixel bandwidth in this dimension. In the presence of a field inhomogeneity $\Delta B_0(x,y)$, the pixel at location (x,y) is mis-located according to the equation:

$$\Delta r_{pe}(x,y) = \frac{\gamma}{2\pi}\Delta B_0(x,y) N \tau_{pe} FOV_{pe}$$

where τ_{pe} is the echo-to-echo time in the EPI readout train and N is the matrix size of the image. If a field map is collected then it is possible to correct the geometric distortion via this equation (Jezzard and Balaban, 1995). An alternative approach that can be used for spin echo data is to collect two images with blips in opposing directions, and then reconstruct an undistorted image without recourse to a field map (Andersson *et al*., 2003; Morgan *et al*. 2004).

The signal loss problem in the case of gradient echo EPI is quite complex, and the details of its manifestation depend on the direction of the field inhomogeneities with respect to the slice, phase encode and readout directions. Shim gradients that exist in the phase encode direction will locally alter the effective echo time of the EPI sequence, according to $TE_{eff}=TE/Q$, where TE is the intended echo time of the center of *k*-space and Q is a dimensionless constant given by (Deichmann *et al*., 2002):

$$Q(x,y,z) = 1 \pm \frac{\gamma}{2\pi}\tau_{pe}FOV_{pe}g_{pe}(x,y,z)$$

where g_{pe} is the shim gradient. It is interesting to note that the value of Q depends (via the +/− sign) on whether *k*-space is scanned bottom-to-top or top-to-bottom, and hence the effective echo time is non-linearly affected by this choice (see Fig. 4). Slice direction field gradients lead to intra-slice dephasing and signal loss, that for a notionally rectangular slice profile is given by $I=I_0\text{sinc}(\Delta\Phi_z(x,y,z))/Q$ (for $\Delta\Phi_z(x,y,z) < 2\pi$) and $I=0$ (for $\Delta\Phi_z(x,y,z) \geq 2\pi$). The phase spread parameter, $\Delta\Phi_z$, is in turn given by:

$$\Delta\Phi_z(x,y,z) = \gamma g_z \Delta z \mathrm{TE}_{\mathrm{eff}}$$

where Δz is the slice thickness. Note that if the local TE_{eff} falls outside the acquired data window, or if the phase spread is greater than 2π, then no signal intensity is recorded from that location.

In the context of BOLD fMRI Deichmann *et al*. (2002) have introduced a useful parameter known as the BOLD sensitivity, given by the product of the effective echo time TE_{eff} (affected by phase encode direction shim gradients as described above) and the local image intensity (affected by shim gradients in any direction). Deichmann *et al*. (2003) further show that the BOLD sensitivity can be maximised in the difficult orbitofrontal cortex by optimizing the orientation of the slice and by applying a modest constant Z-shim.

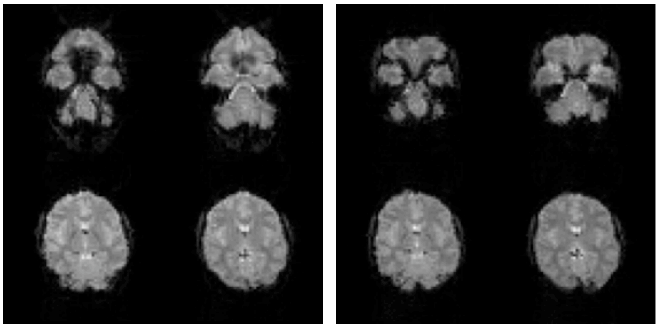

Fig. 4: Example of the different effects on geometric distortion depending upon whether k-space is traversed top-bottom or bottom-top. All other parameters are identical.

Summary

Magnetic field inhomogeneities can have a dramatic effect on the quality of the data that can be obtained in an MRI experiment. Problems that can be encountered include substantial signal loss, geometric distortion of the image, incomplete fat suppression, and poor linewidth in spectroscopic measurements. To counter these problems scanner hardware is increasingly being provided with the capability for subject-specific optimization of the room-temperature shim currents. However, the conventional terms (usually only available up to 2nd order spatial harmonics) are unable to correct many practically encountered magnetic field inhomogeneities. As such, there is increasing interest in developing other methods that can provide a higher spatial frequency of correction. These include alternative acquisition approaches, local passive and active shimming, and dynamic shimming.

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