Sources of Distortion in Functional MRI Data

Peter Jezzard* and Stuart Clare

FMRIB Centre, Department of Clinical Neurology, University of Oxford, Oxford, UK

Abstract: Functional magnetic resonance image (fMRI) experiments rely on the ability to detect subtle signal changes in magnetic resonance image time series. Any areas of signal change that correlate with the neurological stimulus can then be identified and compared with a corresponding high-resolution anatomical scan. This report reviews some of the several artefacts that are frequently present in fMRI data, degrading their quality and hence their interpretation. In particular, the effects of magnetic field inhomogeneities are described, both on echo planar imaging (EPI) data and on spiral imaging data. The modulation of these distortions as the subject moves in the magnet is described. The effects of gradient coil nonlinearities and EPI ghost correction schemes are also discussed. *Hum. Brain Mapping* 8:80–85, 1999. © 1999 Wiley-Liss, Inc.

Key words: functional MRI; distortions; echo planar imaging

INTRODUCTION

The majority of functional magnetic resonance imaging (fMRI) studies to date have used the echo planar imaging (EPI) pulse sequence [Mansfield, 1977], or the spiral imaging pulse sequence [Ahn et al., 1986; Meyer and Macovski, 1987]. The reason is the impressive ability of both to collect data rapidly, often at a rate of five or more slices per second. The tradeoff for this impressive acquisition speed is that a relatively low spatial resolution must be employed (often 64×64 pixels). Additionally, both EPI and spiral pulse sequences suffer from a variety of distortions that degrade the quality of the resulting images. Aside from the familiar Nyquist ghost artefact in EPI, the most prominent source of distortion for both sequences is the effect of magnetic field inhomogeneities, although their manifestation is different for EPI and spiral data. Echo planar images also can be quite sensitive to the

presence of “Maxwell” fields, which are inherent in all gradient coil designs. All magnetic resonance pulse sequences will additionally suffer some amount of spatial distortion from the imperfect ability of the gradient coil to generate a linear field gradient.

The above sources of artefact are reviewed briefly here, as are several strategies to correct for their effects. Other less obvious, but equally important, effects also will be discussed. These include the ramifications on the subsequent image distortions in the choice of EPI Nyquist ghost correction method. Also, the effects on EPI data of subject motion over the course of the scan session are analysed. This leads to a nonlinear local modulation of the image distortions, which is apparent as increased “noise” in the data.

EFFECTS OF GRADIENT COIL NONLINEARITIES

The perfect field gradient coil for use in an fMRI experiment will have the following properties: high gradient strength (>30 mT/m), fast switching (>200 T/m/s), low nonlinearity ($<1\%$), and low acoustic noise (<80 dBA). In practice it is very difficult to

*Correspondence to: Peter Jezzard, FMRIB Centre, John Radcliffe Hospital, Headington, Oxford OX3 9DU, UK.

E-mail: peterj@fmrib.ox.ac.uk

Received for publication 30 March 1999; accepted 20 May 1999.

achieve these parameters simultaneously. Often, the linearity specification of the coil is compromised in order to achieve high gradient strength and fast switching. This affects all MRI sequences, whether conventional or ultrafast. The resulting effect distorts the shape of the image and causes the selection of slices to occur over a slightly curved surface, rather than over rectilinear slices. Most body gradient coils have specifications of <1% linearity over a 40 cm sphere (meaning that the gradient error is within 1% of its nominal value over this volume). It should be noted, however, that a 1% gradient error can translate into a much more significant positional error at the extremes of this volume, since the positional error is equal to $\int G_x dx$ (G_x being the field gradient in the x direction). In the case of head insert gradient coils, the distortions may be significantly higher, given the inherently lower linearity specifications of these coils. Correction of in-plane distortion can be achieved by unwarping the precalibrated positional error. Many manufacturers perform this operation during image reconstruction. Correction of slice selection over a curvilinear surface is, however, rarely made.

A related problem that manifests itself in EPI data is the effect of “Maxwell terms” in the magnetic field generated by the coils. These are error gradient terms that are a consequence of the fundamental laws of electromagnetism described in Maxwell’s equations. In particular, Maxwell’s equations require the following conditions to be satisfied:

$$\frac{\partial B_z}{\partial z} = - \left(\frac{\partial B_x}{\partial x} + \frac{\partial B_y}{\partial y} \right) \quad (1a)$$

$$\frac{\partial B_z}{\partial x} - \frac{\partial B_x}{\partial z} = 0 \quad (1b)$$

where $\partial B_z / \partial z$ is the desired z-direction field gradient and $\partial B_z / \partial x$ is the desired x-direction field gradient. The other terms induce second-order modulations in the desired field gradients. Whereas, for most imaging sequences the error terms produce a negligible effect, for EPI scanning the error terms can lead to noticeable geometric distortions in the data. Weisskoff et al. [1993] have described the form of these distortions in detail. The effects are most prominent at low static magnetic field strength or high field gradient strength.

EFFECT OF MAGNETIC FIELD INHOMOGENEITIES

A significant source of artefact for EPI and spiral data is the effect of inhomogeneities in the static

magnetic field on the resulting image. To a certain extent the static magnetic field may be made homogeneous throughout the sample (in this case, the head) by putting offset currents in the room temperature “shim” coils. These coils compensate for low-order polynomial deviations in the static field profile. Nevertheless, inhomogeneities remain, in particular close to tissue-air and tissue-bone interfaces such as around the frontal sinuses and petrous bone. The way that the field inhomogeneities manifest in EPI and spiral images is different and is dealt with separately.

EPI data

The echo planar technique suffers from a very low bandwidth in the phase encode direction. Typically, the bandwidth per pixel is <20 Hz, which implies that a local shim inhomogeneity of 100 Hz (as is quite typical close to the frontal sinuses at 3.0 Tesla) can lead to a mis-location of the signal in that region by 5 pixels. Additionally, when the field homogeneity varies rapidly over a short distance, there may be loss of signal (gradient echo sequences) or hyperintense regions (spin echo sequences). Although the gross signal intensity changes are difficult to compensate for (making EPI imaging of the frontal and temporal lobes problematic), it is possible to correct substantially for the pixel mis-location [Jezzard and Balaban, 1995]. This is accomplished by noting that pixel mis-location in the read-out direction of the image is negligible, and so the problem may be regarded as a series of one-dimension pixel shifts in the phase encode direction given by:

$$x = x_0 + \Delta v(x) / BW_{pe} \quad (2)$$

where x is the mis-located position, x_0 is the correct position, $\Delta v(x)$ is the spatial distribution of the static magnetic field in Hz, and BW_{pe} is the bandwidth per pixel (also in Hz) in the phase encode direction. Note that the static magnetic field can be expressed in frequency units (Hz) or magnetic field (Tesla) via the Larmor equation $v = \gamma B_0$, where γ is the gyromagnetic ratio for hydrogen (42.575 MHz/Tesla).

Figure 1 shows the effects of applying this principle to an axially collected slice obtained at 3.0 Tesla. Figure 1a shows the spin-echo image in a spin-echo/asymmetric spin-echo pair that was collected prior to the fMRI study in order to calculate the field map distribution (the $\Delta v(x)$ term). The field map is calculated via the phase difference of these two high resolution images, i.e., through the relationship: $\Delta v(x,y) = \Delta \phi(x,y) / (2\pi \Delta TE)$, where $\Delta \phi(x,y)$ is the spatial distribution in phase and ΔTE is the echo time difference of the

Figure 1.

(a) Spin-echo image in a spin-echo/asymmetric spin-echo pair used to calculate $\Delta v(x,y)$. (b) Phase difference of the two high resolution images. (c) Uncorrected 64×64 pixel EPI scan. (d) EPI image after application of a geometric correction. The overlaid box is provided for reference. The phase encode direction is horizontal.

spin-echo/asymmetric spin-echo pair. Figure 1b shows the resulting phase map distribution in the axial slice. Figure 1c shows an uncorrected 64×64 pixel echo planar scan. Areas where the image is distorted and does not align correctly with the high resolution spin-echo image are evident from the overlaid box. Figure 1d shows the same EPI image after application of a geometric correction based on Eqn. 2.

In practice it is not necessary to unwarped every image in the EPI fMRI data set. Rather, it is possible to compute the statistical maps in the warped frame and then simply unwarped the statistical map itself in order to make accurate identification of activated areas. Note

that if the amount of warping is substantial, it may be necessary to apply a correction to the statistical confidence to account for the altered voxel dimensions.

Spiral image data

Spiral image data are affected by static field inhomogeneity in a different way to EPI data. Specifically, pixels in any off-resonance (poor shim) regions of the image will be locally blurred rather than relocated. As in the case of EPI data, it is possible to correct for the effects of field inhomogeneity if the field map distribution is known [Noll et al., 1991] or can be deduced to

Figure 2.

(a) Field map collected from a subject at an arbitrary Position 1. (b) Field map collected after ~5° flexion (to Position 2). The same arbitrary greyscale is used for (a) and (b). (c) Field difference map following realignment of Position 2 data to Position 1 and subtraction. Field excursions of up to 50 Hz are evident.

some low polynomial order [Irrazabal et al., 1996]. Also, as with EPI, the degree of distortion will scale linearly with static magnetic field strength, all other factors remaining equal. This suggests that it is increasingly important to account for the effects of imperfect shim as researchers work at ever higher magnetic field strengths.

Head motion in an inhomogeneous field

Head motion is a substantial problem in functional MRI data sets. Slight movements of the head over the course of the fMRI study can lead to large signal changes in the image time series, and these will obscure the subtle signal changes that are being studied. A growing literature exists that addresses the problem of realigning images when motion of the subject has taken place. Most researchers use a rigid body (6-parameter) or affine (12-parameter) realignment. Neither of these transformations will account for the sort of local nonlinear warping of the image that can occur with EPI data when the subject moves through an inhomogeneous magnetic field. An example of this problem is shown in Figure 2a, which shows a field map collected from a subject at an arbitrary Position 1. Figure 2b shows the field map collected from the subject after moving through ~5° flexion to Position 2. The images from Position 2 were

then registered to Position 1 using a rigid body AIR transformation [Woods et al. 1992]. The transformation results were subsequently applied to the two field maps and a difference map was generated (Fig. 2c). In a well-shimmed sample, the field map difference between Position 1 and Position 2 should be zero. However, in general, the shim field distribution throughout the sample will be position dependent and, therefore, the local geometric distortions in the sample will be modulated by head motion. Figure 2c shows that field excursions of up to 50 Hz (leading to 2–3 pixels local distortion in the echo planar image) can result from a 5° flexion. This amount of motion would be unusual in a typical fMRI study; however, 1° rotation (0.4–0.6 pixel distortion) would be quite common. This will manifest as apparent noise in the fMRI time series.

It would be possible to correct the local nonlinear distortions by applying Eqn 2 to the individual time points in the fMRI series. To do this, it is necessary to know the field map deviation as a function of time. This information generally can be obtained directly from the phase information of the gradient-echo EPI data itself. Most scanners discard the phase information in the data during image reconstruction (retaining only the magnitude information). If the phase information is retained, the field map can be calculated from the phase by noting that $\Delta v(x,y) = \Delta \phi(x,y)/(2\pi TE)$,

Figure 3.

Simulated EPI data from an ideal grid phantom. An $x^2 + y^2 - xy$ shim inhomogeneity term has also been simulated. (a) shows the reconstructed data before any ghost correction has been applied (phase encoding direction vertical). (b) shows the image after a simple single reference line correction has been applied. Note that

the ghost has been suppressed without alteration of the geometric distortion. (c) Image after a line-by-line phase correction has been applied. The ghost is suppressed and the geometric distortion is partially removed, but some blurring is also induced by the nonlinear nature of the phase correction strategy.

where $\Delta\phi(x,y)$ is the phase difference between time point n and time point 1 and TE is the gradient-echo time. The nonlinear modulations to the local distortions should be eliminated by applying the standard motion correction transformation parameters to these field deviations maps, followed by application of Eqn 2 to the EPI data.

EFFECT OF CHOICE OF NYQUIST GHOST CORRECTION SCHEME

The Nyquist ghost, a well-recognised artefact in echo planar images, results from differences between the echoes acquired under the positive and negative read-out gradients. For snapshot EPI, it manifests as a ghost image shifted by half the field of view in the phase encoding direction of the image (Fig. 3a). A common method used to correct for the ghosting is to acquire a reference scan, which is identical to the EPI scan except that no phase encoding gradient is applied. Phase variations between lines in the reference scan are due only to errors in the acquisition and can be used to correct the phase encoded data and suppress the Nyquist ghost (to $<5\%$ of the intensity of the true image).

Less well known is that the choice of phase correction algorithm will also affect the geometric distortion in the final image.

The simplest form of phase correction is to apply a single phase correction (to some polynomial order) to all even k-space lines and a single phase correction to

all odd k-space lines. Application of this principle will correct the ghost and will leave intact the geometric distortion caused by the poor shim (Fig. 3b). If, however, the phase correction values vary with the phase encoding line, as is the case when the reference scan phase correction is applied on a point-by-point or line-by-line basis, then the geometric distortion in the image is affected (Fig. 3c).

In general, the application of a phase correction strategy that is allowed to vary with increasing k-space line index will cause shifts of points in each column in the image. If the shim inhomogeneity contains terms in the read direction (e.g., x , x^2 etc), then a line-by-line or point-by-point ghost correction will remove those contributions to the geometric distortion. Note that terms in y will not be removed. Additionally, if the phase correction varies nonlinearly with k-space line index, then the points in a column will not only be shifted, but also will be spread (blurred). This spreading does not improve the distortion in the image, but does compromise the point spread function in the phase encode direction.

Therefore, whereas a line-by-line or point-by-point phase correction will correct both the ghost and x^n terms in the shim, they cannot correct the y^n terms and, indeed, may introduce additional blurring. Also, their effect must be accounted for in any subsequent geometric distortion correction. Conversely, the application of a phase correction strategy that is based on a single pair of lines in the reference scan does not affect the geometric distortion and does not introduce blurring.

For this reason, the latter approach should result in a better overall artefact reduction in the image once both ghose and geometric distortion corrections have been applied.

CONCLUSIONS

There are a number of sources of distortion to MRI data, some of which affect all types of pulse sequence and some of which affect only specific pulse sequences. As echo planar imaging is widely used for collecting functional MRI data, it is the main focus of this article. When using the EPI pulse sequence, researchers should take great care in attempting to register their functional data with scans collected using other pulse sequences. If correction schemes are applied, as described above, it should be possible to overlay EPI data on to structural scans reliably. Other more sophisticated algorithms also may improve the statistical power of the experiment by decreasing other sources of noise, such as the subtle modulations in geometric distortion that are associated with head motion.

REFERENCES

- Ahn CB, Kim JH, Cho ZH. 1986. High speed spiral-scan echo planar NMR imaging—I. *IEEE Trans Med Imag* MI-5:2–7.
- Irrarrazabal P, Meyer CH, Nishimura DG, Macovski A. 1996. Inhomogeneity correction using an estimated linear field map. *Magn Reson Med* 35:278–282.
- Jezzard P, Balaban RS. 1995. Correction for geometric distortion in echo planar images from B0 field variations. *Magn Reson Med* 34:65–73.
- Mansfield P. 1977. Multi-planar image formation using NMR spin echoes. *J Phys C* 10:L55–L58.
- Meyer CH, Macovski A. 1987. Square spiral fast imaging: interleaving and off-resonance effects. *Proc 6th Soc Magn Reson Med* 1:230.
- Noll DC, Meyer CH, Pauly JM, Nishimura DG. 1991. A homogeneity correction method for magnetic resonance imaging with time varying gradients. *IEEE Trans Med Im* 10:629–637.
- Weisskoff RM, Cohen MS, Rzedzian RR. 1993. Nonaxial whole-body instant imaging. *Magn Reson Med* 29:796–803.
- Woods RP, Cherry SR, Mazziotta JC. 1992. Rapid automated algorithm for aligning and reslicing PET images. *J Comput Assist Tomogr* 16:620–633.