Use of Spherical Harmonic Deconvolution Methods to Compensate for Non-linear Gradients Effects on MRI Images

H. Zhao[1], A. Janke[1], G. J. Cowin[1], G. J. Galloway[1], D. Doddrell[1]

[1]Centre for Magnetic Resonance, The University of Queensland, Brisbane, Queensland, Australia

Introduction

The quality of a magnetic resonance image is dependent upon the accuracy by which physical position is spatially encoded. As MRI data is now used routinely for stereotaxy, longitudinal studies of atrophy and functional studies, ensuring images have no distortion and inhomogeneity is critical. The principal machine dependant sources of this inhomogeneity are eddy currents, gradient non-linearity, B_0- and B_1 inhomogeneity [1]. Here, we present an analytical approach to calculating and removing the effects of non-linear gradients only. The primary reason for a gradient only solution is the recent interest in short-bore high-speed gradients. Although peripheral nerve stimulation is a limiting feature of short rise times, such gradients have found use in high-speed echo planar imaging (EPI) of the heart and diffusion tensor imaging of the brain. To achieve short rise times and avoid peripheral nerve stimulation, designers have restricted the length and limited the number of turns in gradients. These constraints, although suitable for the implementation of pulse sequences having the desired speed, have the undesired consequence of increased non-linearity.

Non-linear pulsed field gradients induce image distortions due to incorrect spatial encoding of the signal. If we assume field gradients are linear, it follows that k-space is sampled linearly and thus the FFT is suitable for reconstruction. However, any deviation from linearity in the gradients results in non-linear data sampling and subsequent errors in image spatial encoding. A non-linear Fourier transformation would allow this data to be correctly transformed to an image. Unfortunately, non-linear Fourier transformation greatly increases computation time by $N/\log_2 N$, relative to a FFT, making real-time image generation computationally prohibitive. Here, we present a general analytical solution to correct image distortions induced by gradient non-linearity. The method is applicable to any gradient configuration. It is robust and more importantly; is based upon an approach where the FFT is maintained for image reconstruction.

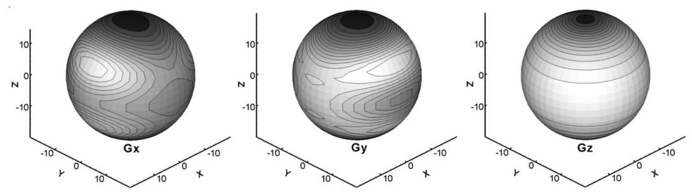

Figure 1. Gradient distribution.

Theory

In standard MR imaging, k-space is sampled at a single time rate and under the assumption that linear gradient fields are applied. If the gradient field is non-linear, there will be a geometric distortion of these images. Knowing the exact gradient field profiles is the key to solving this problem. Accurately describing the gradient field distribution is not a simple task. The most general approach is to expand the field using spherical harmonics as the basis function [2]. The spherical harmonics expansion of order n and degree m of each component of the gradient field has the form: $B_{\mathrm{V}(n,m)}(r,\theta,\phi) = r^n\left[a_{\mathrm{V}(n,m)}\cos(m\phi) + b_{\mathrm{V}(n,m)}\sin(m\phi)\right]P_{(n,m)}(\cos\theta)$. Once knowledge of the gradient field $B_Z(\mathbf{r})$ is established, it can be defined as $G_{\mathrm{V}}(\mathbf{r}) \equiv \frac{\partial Bz_{\mathrm{V}}(\mathbf{r})}{\partial v} \equiv \frac{\partial Bz_{\mathrm{V}}^{\mathrm{L}}(v)}{\partial v} + \frac{\partial Bz_{\mathrm{V}}^{\mathrm{N}}(\mathbf{r})}{\partial v} \equiv G_{\mathrm{V}}^{\mathrm{L}} + G_{\mathrm{V}}^{\mathrm{N}}(\mathbf{r})$. Where the subscript v is used to denote a spatial dimension (x, y, or z), $Bz_{\mathrm{V}}^{\mathrm{L}}$ is the linear gradient field that has only the desired first order harmonic (v), and $Bz_{\mathrm{V}}^{\mathrm{N}}$ is the non-linear gradient field defined by the higher order harmonics, that is $\left(G_{\mathrm{X}}^{\mathrm{L}} = a_{\mathrm{X}(1,1)},\ G_{\mathrm{Y}}^{\mathrm{L}} = b_{\mathrm{Y}(1,1)},\ G_{\mathrm{Z}}^{\mathrm{L}} = a_{\mathrm{Z}(1,0)}\right)$, $B_{\mathrm{X}}^{\mathrm{N}}(r,\theta,\phi) = \sum_{n=2}\sum_{m} B_{\mathrm{X}(n,m)}(r,\theta,\phi)$, $B_{\mathrm{Y}}^{\mathrm{N}}(r,\theta,\phi) = \sum_{n=2}\sum_{m} B_{\mathrm{Y}(n,m)}(r,\theta,\phi)$ and $B_{\mathrm{Z}}^{\mathrm{N}}(r,\theta,\phi) = B_{\mathrm{Z}(1,1)}(r,\theta,\phi) + \sum_{n=2}\sum_{m} B_{\mathrm{Z}(n,m)}(r,\theta,\phi)$. By employing the following notation: $\eta_{\mathrm{X}}(x,y,z) = B_{\mathrm{X}}^{\mathrm{N}}(r,\theta,\phi)/G_{\mathrm{X}}^{\mathrm{L}}$, $\eta_{\mathrm{Y}}(x,y,z) = B_{\mathrm{Y}}^{\mathrm{N}}(r,\theta,\phi)/G_{\mathrm{Y}}^{\mathrm{L}}$, $\eta_{\mathrm{Z}}(x,y,z) = B_{\mathrm{Z}}^{\mathrm{N}}(r,\theta,\phi)/G_{\mathrm{Z}}^{\mathrm{L}}$, then the mapping from the distorted to undistorted image space can be written as $x' = x - \eta_{\mathrm{X}}(x',y',z')$, $y' = y - \eta_{\mathrm{Y}}(x',y',z')$, $z' = z - \eta_{\mathrm{Z}}(x',y',z')$, where (x,y,z) is original distorted imaging and (x',y',z') is corrected imaging.

Results

The gradient set analyzed in this paper is the high-speed Sonata gradients manufactured by Siemens Aktiengesellschaft in Germany. The experimental gradient distributions for the X, Y, and Z gradients are shown in Figure 1.A phantom manufactured by Bruker Medical in Germany for assessing gradient linearity was employed to validate the image space correction method described above. This phantom consists of a cylinder having perfectly flat ends. The diameter of the phantom is 18.5cm and the depth 3cm. Within the phantom there are point markers the spacing between which is a known distance being 2cm for the main grid. If such a phantom is placed in the XY plane of the magnet the resultant image should appear as a perfect circle. As well, the point markers can be used to map the imaging volume of the gradients by moving the phantom fixed distances along the X, Y, and Z-axes. Figure 2(a) shows a 2D image of the phantom sliced in the XY plane with the centre of the phantom placed at the magnet isocentre. Note that the grid markers are visible when this phantom is imaged. A white circle has been superimposed on this image to demonstrate distortions. The image has been corrected using the harmonics to the 5th order. The resultant image is shown in figure 4(b). The background of the image has also been corrected illustrating that the non-linear properties of the gradients increases for points further removed from the magnet isocentre.

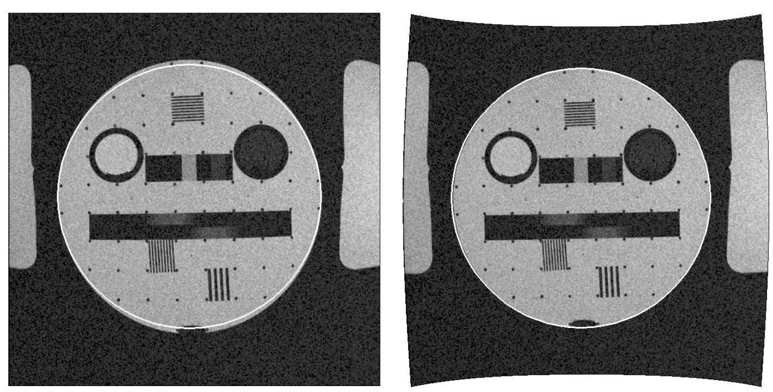

Figure 2. (a) Original imaging, (b) Corrected imaging.

Conclusions

Image distortions induced by gradient non-linearity are a serious problem for the generation of meaningful images using some high-speed gradient sets. Although it might be possible to correct such distortions using translation, warping, and twisting matrices any such attempt will always be empirical. In this presentation we have demonstrated that if the gradient field is quantified using an appropriate expansion then the exact characteristics of the gradient set can be used to correct an image.

Reference

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