

A method for spatial mapping of gradient field nonlinearity in magnetic resonance imaging

D. Wang¹, G. Cowin¹, K. McMahon¹, D. M. Doddrell¹

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

Introduction

Gradient coils in magnetic resonance imaging (MRI) play an increasingly important role as many advanced imaging sequences developed in recent years normally require faster and stronger gradients. To meet this demand, gradient designers employed different design strategies using shorter gradient coil structure with a compromise in gradient field linearity. Increased gradient field nonlinearity in MRI scanners equipped with shorter gradient coils has been shown to cause a range of undesirable imaging errors such as geometric distortions [1], errors in diffusion-weighted imaging and diffusion tensor imaging [2] and in phase contrast MRI [3]. Recently, a comprehensive approach employing a 3-dimensional phantom has been developed for investigating the geometric distortions principally caused by the gradient field nonlinearity [4]. In this presentation, we describe a method that provides a spatial mapping of the gradient field nonlinearity derived indirectly from the measured geometric distortion.

Method

The spatial characteristics of the gradient fields generated by a MR gradient sub-system can be compactly described by the so-called gradient coil tensor, $\mathbf{L}(\mathbf{r})$ [2]. It can be shown that the gradient coil tensor is related to the positional deviations as follows

$$\begin{pmatrix} L_{xx}(\mathbf{r}) & L_{xy}(\mathbf{r}) & L_{xz}(\mathbf{r}) \\ L_{yx}(\mathbf{r}) & L_{yy}(\mathbf{r}) & L_{yz}(\mathbf{r}) \\ L_{zx}(\mathbf{r}) & L_{zy}(\mathbf{r}) & L_{zz}(\mathbf{r}) \end{pmatrix} = \begin{pmatrix} 1 + \frac{\partial(dx(\mathbf{r}))}{\partial x} & \frac{\partial(dy(\mathbf{r}))}{\partial x} & \frac{\partial(dz(\mathbf{r}))}{\partial x} \\ \frac{\partial(dx(\mathbf{r}))}{\partial y} & 1 + \frac{\partial(dy(\mathbf{r}))}{\partial y} & \frac{\partial(dz(\mathbf{r}))}{\partial y} \\ \frac{\partial(dx(\mathbf{r}))}{\partial z} & \frac{\partial(dy(\mathbf{r}))}{\partial z} & 1 + \frac{\partial(dz(\mathbf{r}))}{\partial z} \end{pmatrix}$$

where $dx(\mathbf{r})$, $dy(\mathbf{r})$ and $dz(\mathbf{r})$ are the positional deviations ($dx(\mathbf{r})=x'(\mathbf{r})-x$ where $x'(\mathbf{r})$ is the measured displaced position at \mathbf{r} along x axis and x is its true value, etc..) In the proposed method using a 3D phantom, as the positional deviations are only measured at the discrete array points, an interpolation is, therefore, required for the calculation of the derivatives. The thin-plate spline interpolation was employed for such purpose. Also, in this phantom-based approach, the measured geometric distortion normally contains contributions from other sources such as the static field inhomogeneity and susceptibility difference. In order to derive the gradient field nonlinearity from the measured geometric distortion, it is necessary to separate these other contributions. This can be conveniently done through the acquisition of a pair of 3D scans with the readout gradient reversed [5].

An illustrative example

Using the proposed method, the gradient field nonlinearity in a Siemens 1.5 T MRI system was investigated. This system was equipped with the Siemens Sonata gradient system. The statistical data for the nine components of the gradient coil tensor, $\mathbf{L}(\mathbf{r})$, are presented in the following Table. The statistical data were obtained by sampling at 10,830 array points defined by the geometry of the 3D phantom [4] that spans an effective volume of $257.04 \times 259.02 \times 261.0 \text{ mm}^3$ centred at the isocentre of the system. The value used for the smoothing parameter λ in the thin-plate spline interpolation was set to 0.10. Investigation was also made by using different values (0.05 to 0.15) for the smoothing parameter, λ , and no appreciable differences were observed.

	$ L_{xx} $	$ L_{yx} $	$ L_{zx} $	$ L_{xy} $	$ L_{yy} $	$ L_{zy} $	$ L_{xz} $	$ L_{yz} $	$ L_{zz} $
μ (mm)	1.041	0.023	0.065	0.022	1.042	0.059	0.023	0.024	1.033
σ	0.033	0.023	0.063	0.024	0.034	0.058	0.026	0.026	0.026
max (mm)	1.165	0.159	0.300	0.165	1.197	0.270	0.126	0.129	1.185

Discussion and conclusion

A practical method has been presented for the mapping of the spatial dependence of the gradient field nonlinearity. Unlike the previous methods in which the gradient field nonlinearity is either calculated from the known geometry of the gradient coils or derived from the mapped magnetic fields, the gradient field nonlinearity in the proposed method is indirectly derived from the measured geometric distortion. The method is based on a 3D phantom with which a comprehensive and complete measurement of the geometric distortion along all three orthogonal axes (the axes of the gradients X, Y and Z) can be carried out. The geometric distortion that is caused by the gradient field nonlinearity can be measured using two sets of phantom MR images acquired with different readout directions. It should be pointed out that when acquiring the phantom scans, it is necessary to "switch off" the vendor's correction method. In other words, the phantom images should be acquired with the full effect of the gradient field nonlinearity. The phantom scans should also be acquired using a true 3D sequence, i.e. phase-encoding the other two directions. One attractive feature of the proposed method is that it provides a convenient way for mapping the gradient field nonlinearity as it only requires 2 sets of simple 3D scans. Although in the proposed method the gradient coil tensor is evaluated only at the discrete points of a 3D array, its components can be calculated at any given point within the effective volume through interpolation. The accurately mapped gradient field nonlinearity can be useful for correction of the related errors in diffusion-weighted imaging, diffusion tensor imaging or in phase contrast MRI.

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