Metallic distortion correction in spin-echo imaging using 3D cubic B-splines

S. Skare[1], J. L. Andersson[1]

[1]Karolinska MR Center, Stockholm, Sweden

Introduction

In 2D spin-echo (SE) imaging, a metallic object distorts the anatomy in both the frequency encoding and slice selection direction, leading to an effective distortion in the direction **v** (in voxel units), depending on the values of excitation and receiver bandwidths

$$\mathbf{v} = \left[1/\omega_{recv} \quad 0 \quad 1/\omega_{exc}\right]^T / \left\|\left[1/\omega_{recv} \quad 0 \quad 1/\omega_{exc}\right]\right\|$$

This abstract is based on a post-processing method proposed by Chang & Fitzpatrick[1] (CF) in 1992. From two magnitude SE images (denoted here as I_+ and I_-), differing only in the polarity of the frequency and slice encoding gradients, the displacement field, proportional to the underlying B_0 field inhomogeneity, can be estimated from the images themselves without involving phase map measurements. In this work, some of the problems with the CF method have been addressed. The most important improvements over the original method are

1) Modeling of a single smooth and continuous 3D displacement field built by 3D cubic B-splines as its basis functions. Thereby avoiding former problems with discontinuities perpendicular to the distortion direction.
2) No need for defining common landmarks in the images
3) Incorporation of a 3D rigid body motion correction in the model to allow for subject movements between the two scans.
4) Distortion correction is applied in the actual angle of distortion defined by **v**, rather than in the frequency encoding direction only. This is important since, for typical clinical scan parameters, this angle can seldom be neglected. E.g. commonly used receiver bandwidths of 125 Hz/pixel and slice thicknesses of 4 mm results in distortion angles (in metric units) of about 30 degrees.

In essence, the task is to minimize the difference between the I_+ and the I_- image volume in an iterative manner by changing the B-spline coefficients, contained in vector **c**, building the displacement field and the six motion parameters, $\mathbf{m} = [\Delta x, \Delta y, \Delta z, pitch, roll, yaw]$. The objective function $O(\mathbf{p})$ becomes

$$O(\mathbf{p}) = \sum_{\mathbf{x}\in V}\left[I_+(\mathbf{T}(\mathbf{m})\mathbf{x} + d(\mathbf{c})\mathbf{v})\cdot(1+D_{\mathbf{v}}(d)) \;-\; I_-(\mathbf{x}-d(\mathbf{c})\mathbf{v})\cdot(1-D_{\mathbf{v}}(d))\right]^2$$

where $D_{\mathbf{v}}(d) = \langle \nabla d, \mathbf{v}\rangle$ is the directional derivative of the field along **v**, $\mathbf{T}(\mathbf{m})$ is the rigid body transformation matrix and $\mathbf{p} = [\mathbf{m}\ \mathbf{c}]^T$ is the parameter vector of motion and B-splines coefficient to be estimated. As the number of elements in vector **p** is in the order of tens of thousands, a derivative based search algorithm has been used to achieve realistic convergence times. For this work, the variable metric algorithm was chosen.

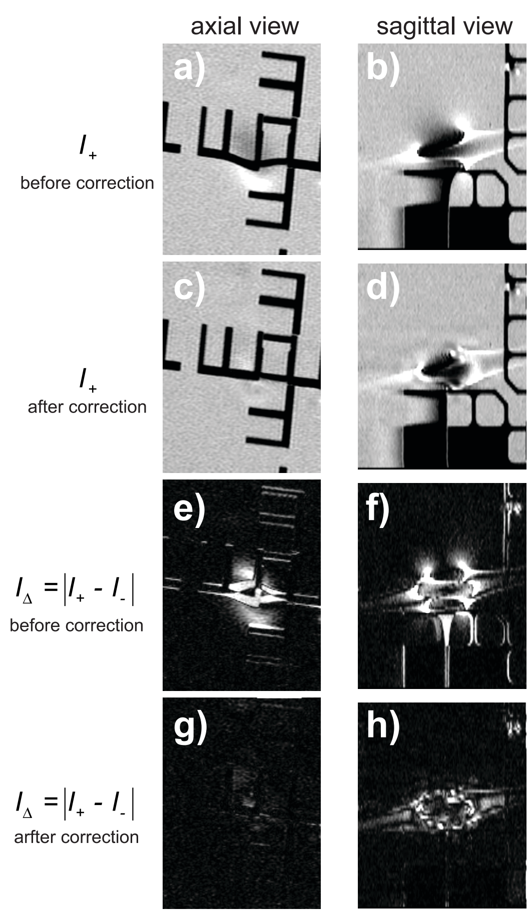

Figure 1. Axial and sagittal views of the part of the phantom nearest the aneurysm clip. a-b) I_+ image before correction. c-d) I_+ image after correction. e-f) Mismatch between I_+ and I_- before correction. g-h) Corresponding images after correction

Materials & Methods

A gel phantom containing plastic structures and an embedded aneurysm clip was scanned on a GE 1.5T Twinspeed system. 80 axial contiguous 1 mm slices were scanned two times using a SE sequence with "+x+z" and "-x-z" gradient polarity. Relevant scan parameters include FOV = 24×24 cm, resolution = 512×256, 4 averages, and receiver bandwidth ω_{recv} = 61.05 Hz/pixel (±16kHz/FOV). The direction of distortion was **v**=±[0.998 0 -0.071] (voxel units). The image volume was cropped, with the clip centered before correction to obtain a clinically relevant total number of voxels. In the post-processing step, the "knot spacing", defining the distance between the B-splines building $d(\mathbf{c})$ was chosen to be 3.75×3.75×4 mm.

Results

In Figure 1, axial and reformatted sagittal views of the phantom are shown. The I_+ image volume is shown before (a-b) and after (c-d) correction. Corresponding difference images, $I_\Delta = |I_+ - I_-|$, are presented below, before (e-f) and after (g-h) correction. The axial slice is located only a few millimeters below the aneurysm clip, just outside the core area where the pixels shifts are so large that the Jacobian ($1 \pm D(d)$), i.e. the intensity modulation, becomes less than zero and where this method cannot restore the true data. Outside this region, in both the axial and sagittal views, the rectangular plastic objects in the I_+ image appear straight and the I_Δ image shows little residual mismatch.

Discussion

The method proposed here is more robust to noise by working with a continuous 3D distortion field instead of about $n_y \times n_z$ individual 1D fields. The latter may introduce signal discontinuities in the corrected image (Fig. 4 in ref [2]). The use of basis functions for the displacement field, together with a 3D rigid body motion correction included in the model, eliminates the need of defining anatomical landmarks or edges on the same time as it allows scanning of patients without stereotactic frames or other types of fixation devices. After correction the I_+ and I_- image volumes can be averaged to gain SNR. As e.g. T_1-weighted images are often scanned with two averages (NEX/NSA), one may instead use two single average scans with opposite gradient polarity without increasing the scan time, why it may be an attractive solution in the clinical practice.

References

1. Chang H, Fitzpatrick JM. A Technique For Accurate Magnetic-Resonance-Imaging In The Presence Of Field Inhomogeneities. IEEE Transactions On Medical Imaging 1992;11(3):319-329.
2. Andersson JL, Skare S, Ashburner J. How to correct susceptibility distortions in spin-echo echo-planar images: application to diffusion tensor imaging. Neuroimage 2003;20(2):870-888.