Improving the Registration of B_0-distorted EPI Images using Calculated Cost Function Weights

Mark Jenkinson

Oxford Centre for Functional Magnetic Resonance Imaging of the Brain (FMRIB)[†]
email: mark@fmrib.ox.ac.uk

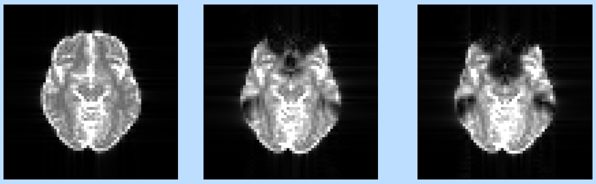
Abstract

A method is proposed that reduces the effects of B_0-inhomogeneity induced signal loss and geometric distortion on the accuracy of EPI to structural image registration. It does this by using (1) pixel-shift unwarping, (2) intensity correction and (3) cost function weighting (in the registration) based on an estimate of the signal loss from a field map acquisition. Results show an improvement in global registration accuracy (and hence localisation of functional activations) by using this method.

Introduction

Accurate localisation of fMRI activations depends on the quality of the registration between EPI and structural images which is a difficult problem due to B_0 inhomogeneities which distort EPI images. The two main sources of B_0 distortion (shown in Figure 1) are:

- Geometric distortion due to average field offset
- Signal loss due to through-slice dephasing.



Undistorted EPI — EPI with Geometric Distortion — EPI with Signal Loss and Geometric Distortion

Figure 1: Simulated EPI images (at 3T) illustrating the effects of signal loss and geometric distortion due to B_0 inhomogeneities.

Correction Methods

Field maps measure the B_0 inhomogeneity and allow corrections to be applied to the EPI, including:

- Pixel-shift unwarping to compensate for geometric distortion [1][2]
- Intensity correction to compensate for signal loss: $I_{corr} = I/S$

However, intensity correction is sensitive to errors in estimating the signal loss factor, S, correctly. Hence in areas where the signal loss estimation is known to be poor, cost function weighting can be applied to minimise these effects.

This work seeks to show how these weights can be calculated from field map values and used to improve registration quality.

Signal Loss

Accurate estimation of signal loss is difficult because it depends on the field gradients which cannot be easily measured from the acquired field maps [2] due to:

- Noise in field maps
- Partial volume effects

The partial volume effect is quite serious because the signal loss is a non-linear function of field gradient and so average gradients do not give accurate estimates of the amount of signal loss. Specifically, the signal loss factor, S, is given by:

$$S = \left| \int \exp\left(i\gamma T_E\, B(z)\right)\, dz \right| \tag{1}$$

where γ is the gyromagnetic ratio, T_E is the echo-time and $B(z)$ is the field at position z. This is not linear in B, so that S cannot be simply expressed in terms of the average change in B through a slice.

Simulations

To quantify the effects of noise and partial volume on the estimate of the signal loss factor, S, simulated images were generated of:

- Field maps: symmetric-asymmetric spin-echo pair (64x64x25 voxels)
- EPI (64x64x25 voxels) - as shown in Figure 1

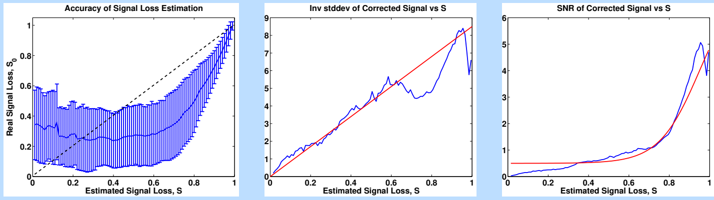
The simulations used an anatomical model [5], a calculated high-resolution B_0 fiel [2] and an MRI simulator [3][4]. All images included noise with SNR=100.

Signal loss factors were derived from these as:

$$S_0 = \frac{I_{\text{distorted EPI}}}{I_{\text{undistorted EPI}}} \qquad\qquad S = \operatorname{sinc}\left(\gamma T_E\, \Delta B\right) \tag{2}$$

where S_0 is the real signal loss experienced in the distorted EPI, S is the estimated signal loss factor from the field map, and ΔB is the change in field across the low-resolution slice (from the field map).

Figure 2(a) shows the difference between signal loss experienced in the EPI, S_0, and the signal loss estimated from the field maps, S, due mainly to partial volume effects.



(a) (b) (c)

Figure 2: (a) estimated (field map) and real (EPI) signal loss in simulations; inverse standard deviation (b) and SNR (c) of the corrected image (image divided by estimated signal loss) as a function of signal loss. The red lines represent fits that are to be used as weighting functions.

Weighting Functions

To reduce the effect of inaccurate signal loss estimation on the intensity corrected images, I_{corr}, cost function weighting can be used to assign relative values (weights) to each voxel reflecting their importance in the overall cost function calculation. Thus voxels where the intensity varies greatly due to poor signal loss estimation should be given small weight values. The effective variation for the corrected images is shown by the (inverse) standard deviation and SNR, which are plotted as a function of S in Figures 2(b) and 2(c).

Two candidate weighting functions were chosen, based on fits to the inverse standard deviation and SNR of I_{corr}, as shown by the red lines in Figure 2. The weighting functions W_1 and W_2, respectively, are given (up to a scaling constant) by:

$$W_1 = S \quad \text{and} \quad W_2 = \frac{1}{1 + \exp(-10(S - 0.95))}. \tag{3}$$

Registration Tests

Registration of the simulated EPI with a simulated T1 image from the same phantom data [5] was performed using FLIRT [6] with 6 Degrees Of Freedom under the following conditions:

- with and without field map pixel-shift correction (using FUGUE [6])
- with and without intensity correction, based on the estimated signal loss, S
- with and without applying the weighting functions W_1 and W_2 during registration

Qualitative results are shown in Figure 3. It can be seen that there is an improvement of several millimetres between the first case (no weighting or unwarping) and the last (both sigmoidal weighting, W_2, and unwarping).

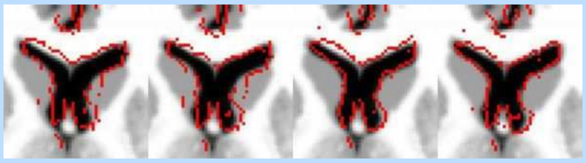
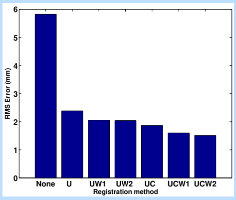


Figure 3: Results of registration near ventricles with (from left to right): (a) no unwarping or weighting; (b) unwarping but no weighting; (c) weighting but no unwarping; (d) both unwarping and weighting (W_2). Red lines show edges from the EPI overlayed on the T1 structural.

Quantitative registration accuracy (using known ground truth) was measured by the rms error, in mm. This is shown in Figure 4 for each registration case and demonstrates the largest improvement by using unwarping, as well as additional improvement from using intensity correction and cost function weighting.



Key	Description
None	No unwarping, correction or weighting
U	Pixel-shift Unwarping
C	Intensity Correction
W1	Weighting with W_1
W2	Weighting with W_2

Figure 4: Quantitative registration results showing RMS error, in mm, using different registration schemes.

Conclusion

Signal loss corrections are difficult due to noise and partial volume effects. By using pixel-shift unwarping, intensity correction and a weighting function (W_2) based on estimated signal loss from a field map, an improvement of several millimetres in global registration accuracy can be achieved.

References

[1] Cusak et al, NeuroImage 18(1), 2003
[2] Jenkinson, HBM 2001
[3] Drobnjak and Jenkinson, HBM 2004
[4] MIDAS consortium, HBM 2004
[5] Collins et al, IEEE TMI 1998
[6] FSL - www.fmrib.ox.ac.uk/fsl

Acknowledgements

The author wishes to thank the UK-EPSRC (MIAS-IRC) for funding this research.

[†] **Address for correspondence: Oxford Centre for Functional Magnetic Resonance Imaging of the Brain (FMRIB), Department of Clinical Neurology, University of Oxford, John Radcliffe Hospital, Headington, Oxford OX3 9DU, UK**