Quantitative Simulation of Affine Registration for Correction of Eddy Current Distortions in Diffusion-Weighted Images

R. G. Nunes[1], I. Drobnjak[1], S. Clare[1], P. Jezzard[1], M. Jenkinson[1]

[1]Centre for Functional MRI of the Brain, University of Oxford, Oxford, United Kingdom

Introduction: Diffusion-weighted (DW) imaging has shown great potential for mapping white matter tracts in the brain [1]. In order to be able to infer on fibre structure, DW images need to be collected under a set of different diffusion gradient directions. The presence of the eddy current fields induced by the diffusion gradients during the readout window can lead to significant geometric distortions in snapshot echo planar images (EPI) [2]. Misalignment of the DW images will in turn lead to errors in the estimates of the diffusivities and fibre orientations [2]. While a doubly-refocused sequence has been presented which enables the minimisation of the level of eddy currents at source [3], the geometric distortions should nevertheless be correctable using affine registration methods, provided that the eddy current effects can be considered time invariant throughout the acquisition window. This assumption together with concerns regarding potential signal-to-noise loss due to the presence of an extra imperfect 180° pulse have prevented the wider implementation of the doubly-refocused sequence. The standard single spin-echo sequence still remains therefore the most commonly used. While the presence of higher order eddy current terms has been experimentally demonstrated [4], the intrinsically low signal-to-noise of DW images, together with the presence of subject's motion complicates their quantification. A recently developed scanner simulator [5] has made it possible to quantitatively evaluate the extent to which affine registration can correct for eddy current distortions.

Methods: The method proposed is based on the application of POSSUM, an MRI simulator software [5,6]. The algorithm is based on the fundamental Bloch equations to model the behavior of the magnetization vector for each small volume element of the brain and for each tissue type independently. A 3D digital brain phantom from the MNI BrainWeb database is used. RF excitations are modeled with a simple rotation of the magnetization vector about the *x* axis. The RF timings and the EPI gradient waveforms are specified from an input file. The signal coming from one voxel is obtained by analytical integration over each voxel and the total signal formed by summing the contributions from all the voxels. A gradient-echo EPI was generated. The matrix size was of 128×128 corresponding to a resolution of 2×2×3 mm^3. The bandwidth was 100 kHz, the readout window 163 ms and the echo time 90 ms. The effect of the eddy currents produced by each of the two sequences (single spin-echo and doubly-refocused) was simulated by superimposing a sum of exponentially decaying terms to the EPI readout gradients: $\sum(\pm)G_{diff}(\pm)\varepsilon \exp[-(t-t_i)/\tau]$. Each of these terms correspond to the switching either on (-) or off (+) each of the positive or negative diffusion gradients at time t_i. Two diffusion gradients were considered for the standard sequence each with a duration of 28 ms. For the doubly-refocused sequence four diffusion gradients with a duration of 14 ms were considered. The amplitude of the diffusion gradients G_{diff} was 21.3 mT/m while the maximum amplitudes achieved by the read and phase-encode EPI gradients were 9.1 mT/m and 4.0 mT/m respectively. A set of eddy currents were simulated with different relative amplitudes ε and time constant τ.The diffusion gradients could also have been applied either along the read (xx) axis, producing skews in the images or along the phase-encode direction (yy) producing scaling [2]. Although the difference in contrast displayed by both the T_2-weighted image and the DW images should also have an influence in the efficiency of the registration step, here we did not attempt to model anisotropic diffusion. To provide the distorted images with a realistic contrast, we opted to simply consider the mean attenuation produced in each tissue. The K-space data corresponding to CSF, white and gray matter were separately generated and an appropriate attenuation factor applied to each of them before summation. These were based on a b-value of 1000 smm^{-2} and a simple exponential decay was assumed: $S/S_0=\exp[-b\langle\mathbf{D}\rangle]$. The mean diffusivities $\langle\mathbf{D}\rangle$ for each tissue were: 3.2×10^{-3} mm^2/s (CSF), 0.8×10^{-3} mm^2/s (gray matter), 0.7×10^{-3} mm^2/s (white matter) [7].The distorted attenuated images were then registered to the undistorted T_2-weighted image using an affine registration method [8]. Five degrees of freedom were allowed for: translation along xx and yy, rotations around the zz axis, skews in xy and scaling along yy. No difference should exist between the registered images and the undistorted attenuated image. To evaluate the ability to correct for eddy-current-induced distortions, the sum of squares of the difference image between each registered image and the undistorted DW image was therefore calculated.

Results and Discussion: The undistorted T_2-weighted a) and DW b) images are shown in Figure 1. Images generated for the single spin-echo sequence with eddy currents along the read c) and phase encode direction f) are also displayed. The distortions are clearly visible when looking at the difference images d) and g), where b) was subtracted from each of the distorted images. Although registration did enable to reduce these differences, residual distortions are still visible in both e) and h).

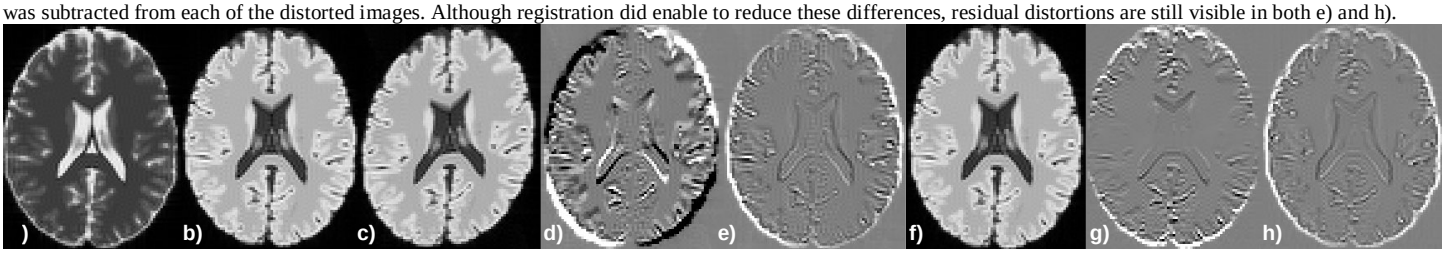

Figure 1 – **a)** T_2-weighted image; **b)** DW image; **c)** DW image with an eddy current along the read direction (standard sequence, ε=0.01%, τ=10ms); **d)** Subtraction of image b) from c); **e)** Subtraction of the image obtained after registering c) to a); **f)** DW image with an eddy current along the phase-encode direction (standard sequence, ε=0.01%, τ=10ms); **g)** Subtraction of image b) from f); **h)** Subtraction of the image obtained after registering g) to a).

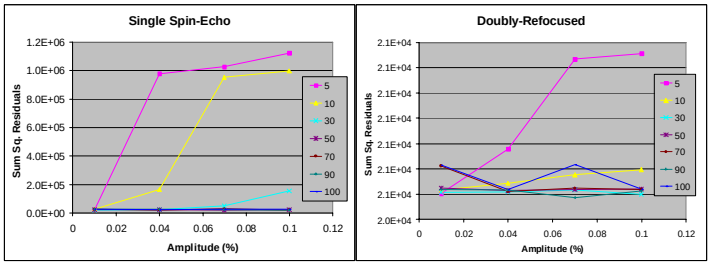

Figure 2 – Sum of squares of the residuals for both sequences, for eddy currents along the read axis. Each curve corresponds to a different time constant in ms.

From the residuals plotted in Figure 2, it can be noted that the affine registration method does not seem to be able to cope effectively with eddy currents of large amplitudes and time constants much lower that the duration of the gradients. Unless a more effective correction method is used, minimising eddy currents at source should therefore be preferable. The simulation results also confirm previous experimental observations [3] as the residuals corresponding to the doubly-refocused sequence are significantly lower. There can be several causes for the residuals observed. One factor can be the presence of higher order eddy currents, not corrected using affine registration. On the other hand, the presence of either skews or scaling, even if affine, in the distorted images would distribute the overall signal over a wider number of pixels. This signal cannot be totally recovered by performing registration even if the images become perfectly re-aligned and should lead to errors when estimating diffusivities and/or fibre orientations. Another factor that can influence the quality of the registration is the presence of eddy-current-induced ghosting.

Conclusion: Using POSSUM, we have demonstrated that using an affine registration method is not sufficient to correct for strong eddy current effects. By using the doubly-refocused sequence, the level of eddy currents is substantially reduced and the registration step seems to perform equally well over a wide range of amplitudes and time constants. In the future we plan to extend this investigation even further by including the effect of motion throughout the acquisition time.

References: [1] MORI et al., *NMR in Biomed*, 2002; **15**:468. [2] JEZZARD et al., *MRM*, 1998; **39**:801. [3] REESE et al., *MRM*, 2003; **49**:177. [4] SHEN et al.,*MRM*,2004; **52**: 1184. [5] DROBNJAK et al., ISMRM 2004; 496. [6] MIDAS consortium, HBM 2004; 496. [7] LE BIHAN et al., JMRI, 2001; **13** :534 [8] JENKINSON et al., *NeuroImag*, 2002; **17**: 825