Rapid Isotropic Diffusion Weighted Imaging using PROPELLER

I. Teh[1,2], B. Zheng[2], J. V. Hajnal[1], X. Golay[2,3], and D. J. Larkman[1]

[1]Imaging Sciences Department, MRC Clinical Sciences Centre, Hammersmith Hospital, Imperial College London, London, United Kingdom, [2]Laboratory of Molecular Imaging, Singapore Bioimaging Consortium, Singapore, Singapore, [3]Department of Neuroradiology, National Neuroscience Institute, Singapore

Introduction

Diffusion-weighted imaging (DWI) has been established as an important tool for studying tissue architecture and for monitoring acute ischemic change and other pathologies in clinical and pre-clinical research. In many applications it is sufficient only to measure the mean apparent diffusion coefficient (ADC) of the tissue which requires sampling of only the trace of the diffusion tensor. Typically single shot techniques like echo planar imaging (EPI) or turbo spin echo (TSE) are used due to their insensitivity to motion. Three scans with diffusion weighting applied in orthogonal directions and a reference scan need to be acquired. Single shot methods achieve efficiency, but usually suffer from poor resolution. At high field such as used for murine imaging, image distortion and short T2* circumscribe the use of EPI and make multi-shot methods more attractive. However, motion is then a problem. TSE methods with 2-D navigator correction schemes have been devised to address these issues [1]. Periodically Rotated Overlapping ParallEL Lines with Enhanced Reconstruction (PROPELLER) [2] also enables multishot DWI, exploiting the inherent self navigating properties of the PROPELLER acquisition. The success of PROPELLER motion correction and the rotational symmetry of the acquisition lead us to question if this symmetry could be exploited in the measurement of isotropic diffusion. This study investigates the impact of PROPELLER acquisition parameters on the image fidelity for isotropic DW imaging in the presence of isotropic and anisotropic tissues.

Theory

Radially-oriented acquisitions allow rotational averaging of diffusion weighting. We can thus obtain average in-plane DW data from one rather than two acquisitions [3]. Rotating the diffusion weighting gradients with the PROPELLER blades samples both in-plane components without additional gradients. The z direction DW can be acquired via a second acquisition or combined with the x-y acquisition at the expense of TE. In the former, the inherent reduced efficiency in PROPELLER (typically by ⅓ depending on the degree of oversampling in k-space) compared to standard multishot cartesian acquisition is compensated for by requiring two instead of three acquisitions. In addition, because each blade samples the centre of k-space, PROPELLER reconstructions can be fully shot by shot corrected for motion. Rotating the DW with the blades results in x-y isotropic DW at the centre region of k-space where all blades overlap and minimizes the change in signal between blades. DW in the outer regions of k-space is not isotropic, resulting in anisotropy dependent modification of the point spread function (PSF).

Methods

A two acquisition DW PROPELLER scheme has been simulated and implemented. The simulations use a multi-compartment phantom with typical anisotropy values for the brain in different compartments [4]. Images were produced and the PSF calculated for a range of acquisition parameters. For pilot data, spin echo (SE) acquisitions were acquired on a Varian 9.4T scanner with G_{max}=20G/cm and Slew $Rate_{max}$=125G/cm/ms. Diffusion-weighted images of a rat were acquired post mortem, to exclude the effects of motion initially, using reference Cartesian-sampled SE and SE-PROPELLER DW methods at 4 b-values ranging from 0-1000s/mm^2. Diffusion weighting was applied using Stejkal-Tanner trapezoidal DW pulses. The Trace image was generated from 3 SE acquisitions with orthogonally oriented diffusion weighting, while the isotropic-DW image from the PROPELLER sequence combined one image with in-plane diffusion weighting and another with through-slice diffusion weighting. Imaging parameters were: TR=1000ms, TE=28ms, Matrix=256 x 256 (Cartesian) & 256 x 32 x 12 (PROPELLER), FOV=40mm x 40mm, Ave=2, δ=5.8ms, Δ=12.315ms, G=0 - 20G/cm.

Results and Discussion

Two acquisition multishot DW PROPELLER takes a similar total acquisition time as three acquisition multishot Cartesian DWI provided the rotational symmetry of the acquisition is exploited. Importantly, PROPELLER provides robust motion compensation which is essential in DWI in living subjects. Initial simulations suggest that the penalty in PSF broadening (Figures 1 and 2) is tolerable even in regions where the anisotropy is very strong providing that a sufficiently wide blade is chosen. This needs to be weighed against the increase in T2 decay due to a wider blade. The mean ADC measured by both methods in ex vivo brain agree. In the thalamus for example, $ADC_{Cartesian}$ = 0.254 +/- 0.0087 * 10^{-3} mm^2/s and $ADC_{PROPELLER}$ = 0.249 +/- 0.0095 * 10^{-3} mm^2/s (Figure 3). Applying diffusion weighting in the phase encode as opposed to the readout direction along each rotated blade improves image consistency as it negates the influence of eddy currents induced by the diffusion weighting on the readout gradient. The rotation of the diffusion weighting smoothes the transition between blades improving the reconstruction quality, in the presence of both anisotropy and eddy current effects. The method can be made single acquisition by adding a z gradient to the sequence and using bipolar pulses to mitigate off-diagonal terms in the b value matrix. However this extends the TE.

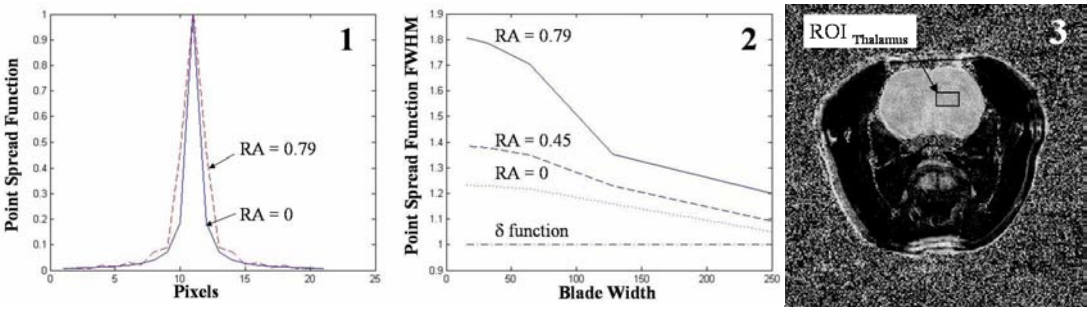

Fig 1. Point spread function of physiologic simulation data improves as relative anisotropy (RA) tends to zero but is tolerable even at high RA; **Fig 2.** Full width half maximum of the point spread function decreases with blade width and increases with relative anisotropy; optimal blade widths appear to be between 32 and 64 phase encode lines; **Fig 3.** Mean ADC image from isotropic DW-PROPELLER acquisition

References

[1] M. Mengershausen *et al*, MAGMA, 2005, 18, 206-216

[2] J.G. Pipe *et al*, MRM, 2002, 47, 42-52

[3] O. Dietrich *et al*, MAGMA, 2001, 12, 23-31

[4] N.G. Papadakis *et al*, MRI, 1999, 17, 881-892

The authors wish to acknowledge the Agency for Science, Technology and Research, Singapore for funding and the Biological Imaging Centre, Imaging Sciences Department, Clinical Sciences Centre, Imperial College London for use of its facilities and expert advice.