Measuring isotropic diffusion with rotating diffusion gradients

I. Teh[1,2], X. Golay[1,3], and D. Larkman[2]
[1]Lab of Molecular Imaging, Singapore Bioimaging Consortium, Singapore, Singapore, [2]Imaging Sciences Department, Imperial College London, London, United Kingdom, [3]Institute of Neurology, University College London, London, United Kingdom

Introduction

Measuring the mean apparent diffusion coefficient (ADC) provides valuable contrast in studying disease processes such as stroke and tumours. ADC is usually measured by acquiring a reference image and 3 images with diffusion weighting (DW) applied in 3 orthogonal directions. ADC is then defined as the mean of the 3 diffusivity maps. This can be time consuming, and single-shot methods for isotropic diffusion weighting have been developed. Mori and van Zijl proposed using a series of bipolar DW gradients [1], while Wong, et al. achieved this by numerically optimizing DW schemes [2]. These multipolar methods reduce the total acquisition time at the expense of an extended DW preparation period, leading to greater T_2 or T_2^* decay. This is problematic at high field strengths such as 9.4T.

An alternative approach by Dietrich, et al. and Cheryauka, et al. explored the effects of rotating diffusion gradients in the presence of anisotropic tissues in 2D diffusion tensor simulations using radial [3] and PROPELLER trajectories [4] respectively. Other work proposed combining in-plane rotating DW gradients with PROPELLER in ex-vivo rat data [5]. These methods assume that the effective DW is the net average DW across all rotated lines/blades as the centre of k-space is oversampled. However, they fail to account for the significant impact of cross-terms when DW is applied simultaneously along multiple axes in 3D. In addition, the rotating DW was applied in a fixed direction with respect to each rotating line/blade (eg. always along the readout (RO) or phase encode (PE) direction), leading to significant anisotropy dependent variations in the measured ADC.

Methods and Materials

To overcome these limitations, we propose an FSE-PROPELLER acquisition [6] using concurrent Stejskal-Tanner DW gradients on multiple axes for time-efficient DW that alternate in direction with successive blades. Eg. In blade 1, DW was applied along the RO and slice select (SS) (+) directions; in blade 2, DW was applied along the PE and SS (-) directions, and so forth, where the blades rotate about the SS axis. Fig. 1 concatenates the applied DW gradients and effective field gradients for a 12 blade PROPELLER scan, excluding other sequence objects.

A 2D phantom of tumors of varying geometries (ADC=0.85x10^{-3}mm^2/s, FA=0.4) embedded in grey matter (GM) (ADC=0.59x10^{-3}mm^2/s, FA=0.38) was simulated, where tumor and GM anisotropy were oriented left-right and through plane respectively. DW was applied in standard orthogonal (DW_{fix}), rotating RO (DW_{ro}), rotating PE (DW_{pe}), and alternating between rotating RO and PE directions (DW_{alt}), in conjunction with a PROPELLER acquisition of 48 blades, with blades of 256x8 (See Fig. 2). T_2, motion and noise were neglected for clarity. Two regions-of-interest (ROIs) were identified and the ADC was measured using the four DW methods, over a range of tumor FAs from 0 to 1 (See Fig. 3). Finally, in-vivo mouse brain data were acquired at 9.4T (Varian, Palo Alto, CA) using the four DW-PROPELLER methods. TR=4000ms, TE=41ms, esp=6.3ms, FOV=25.6x25.6mm, blade matrix = 192x8, N_{blades} = 38, reconstructed matrix=192x192, NEX=3 and b-value=800s/mm^2.

Results and Discussion

Figures 2 and 3 show that the measured ADC values are dependent on the lesion geometry and FA respectively, particularly in the rotating schemes DW_{ro} and DW_{pe}. This arises from the relationship between the signal attenuation due to the applied DW in relation to the tissue anisotropy, and the signal contribution of variously shaped tissues to different regions of k-space. In DW_{alt}, the alternating in-plane and through-plane DW across blades result in isotropic DW and cross terms that cancel out in the central region of k-space, and a pseudo-isotropic DW in the periphery of k-space due to the rapidly oscillating DW. DW_{alt} removed the dependence of the ADC on tissue geometry, and reduced the measured ADC error in ROI 1 from +76% using DW_{ro} to -14% for physiologic FAs of up to 0.8. The in-vivo results (See Fig. 4) were consistent with the simulated data. For example, the corpus callosum, with high anisotropy in the left-right direction, had an ADC that was hypointense (p=1.9e-4) and hyperintense (p=1.0e-9) with DW_{ro} and DW_{pe} respectively. DW_{alt} was resistant to tissue geometry and FA effects, and yielded an ADC that was not significantly different from DW_{fix} (p=0.54), while halving the time for acquisition.

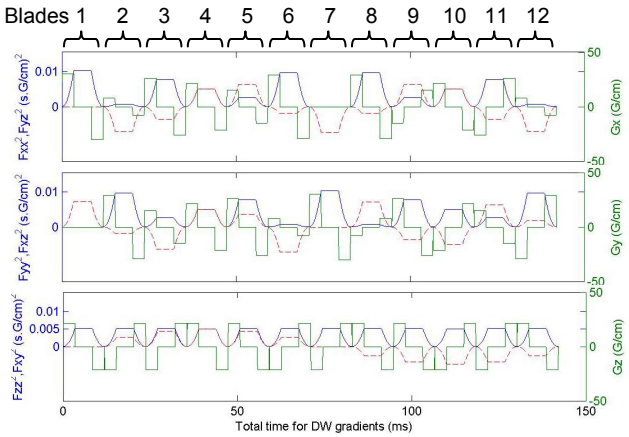

Fig. 1. Concatenation of applied DW gradients (green), on-diagonal (blue) and off-diagonal (red) effective field gradients over 12 PROPELLER blades along 3 orthogonal axes. In isotropic DW, the total on-diagonal terms are equal across all three axes, as the total off-diagonal terms sum to zero.

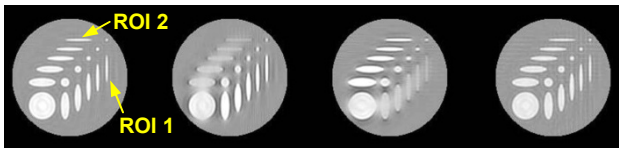

Fig. 2. Simulation of PROPELLER reconstruction of tumors embedded in GM using (from left to right) DW_{fix}, DW_{ro}, DW_{pe} and DW_{alt}. Selected ROIs for measurement of ADC are highlighted.

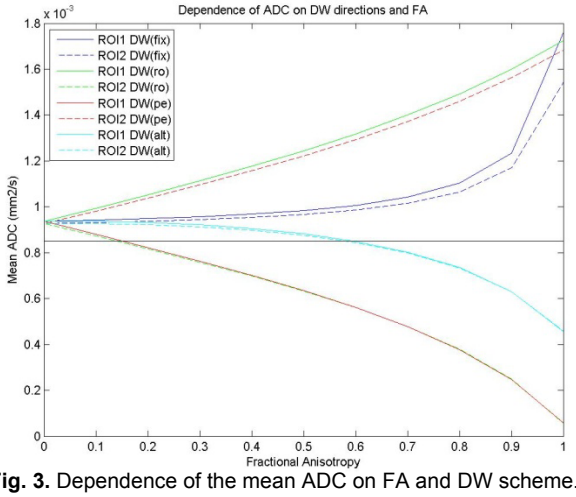

Fig. 3. Dependence of the mean ADC on FA and DW scheme.

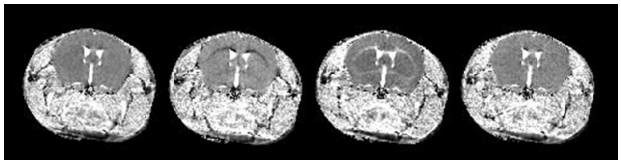

Fig. 4. In-vivo mouse brain ADC data acquired using (from left to right) DW_{fix}, DW_{ro}, DW_{pe} and DW_{alt}. $ADC_{corpus\ callosum}$ = [0.64±0.2, 0.47±0.2, 0.95±0.2, 0.67±0.2] * 10^{-3} mm^2/s respectively.

References [1] Mori S., et al, MRM, 1995;33:41-52; [2] Wong E.C., et al, MRM, 1995;34(2):139-143; [3] Dietrich O., et al, Proc ESMRMB 2002; [4] Cheryauka A.B., et al, MRI, 2004;22(2):139-148; [5] Teh I., et al. Proc ISMRM 2007:1481; [6] Pipe J.G. MRM, 1999;42(5):963-969. The authors wish to acknowledge the Agency for Science, Technology and Research, Singapore for funding.