Quantification of Perfusion Contributions using DWI using low b-values

J. Chia[1], Z. J. Wang[2], D-J. Wang[3], and N. K. Rollins[2]
[1]Philips Healthcare, Cleveland, Ohio, United States, [2]Radiology, Children's Medical Center of Dallas, Dallas, Texas, United States, [3]The Children's Hospital of Philadelphia

Introduction

Diffusion weighted imaging was initially named "intravoxel incoherent motion imaging" due to its sensitivity to the types of motions of spins including diffusion and perfusion. As early as 1988, Le Behan et al showed that perfusion and diffusion can be separately obtained using this technique [1]. The spins in the flowing blood account for a small fraction of all protons in the brain; however, these protons move at a much faster rate than other spins. Although it is safe to ignore the perfusion component in quantification of diffusion, simultaneous perfusion measurement may still be possible by inclusion of the sampling of low b-values. In this study, we look to quantify and characterize perfusion contributions of diffusion weighted imaging (DWI) at 1.5T and 3T.

Materials and Methods

DWI data were acquired on Philips Achieva 1.5T and 3T clinical scanners with R2.6.1 software using a single-shot echo planar imaging (EPI) technique. The DWI parameters were as follows: FOV = 220/220 mm, number of slices= 30, 2.5 x 2.5 x 4 mm (1.5T) and 2 x 2 x 4 mm (3T) voxel size, TR = 4000 ms, TE = 90 ms (1.5T) and 60 ms (3T), and 3 signal averages. Total scan time was 6 minutes and 23 seconds. In order to acquire the rapid initial signal decay, multiple sampling points with low b values (as low as 1 s/mm^2) were included in the data acquisition, in addition to a high b-value of 1000 s/mm^2. The b-values used were b = 1, 5, 25, 100, 1000 s/mm^2. The DWI signal is modeled as two components as shown below:

$$S(b)=S_0 \{f \cdot \exp(-b \cdot D_{perf}) + (1-f) \cdot \exp(-b \cdot D_{diff})\}$$

Here, f is interpreted as the fractional volume of protons flowing in the blood. From D_{perf} and D_{diff}, we can derive two apparent diffusion coefficients (ADC) values related to perfusion and diffusion:

$$ADC_{perf} = Tr\{D_{perf}\} \text{ and } ADC_{diff} = Tr\{D_{diff}\}$$

ADC_{perf} contains contribution from spins resulting from diffusion; however, ADC_{perf} is dominated by the higher flow effects in perfusion. A relative cerebral perfusion measurement (rCBF) can be obtained from the product $f \cdot ADC_{perf}$. Region of interest (ROI) based analysis was performed on an offline computer using internally developed software in order to characterize the signal in multiple areas of the brain. ROI's were drawn and analyzed in eight different white and eight different grey matter areas on four individual slices located above the ventricles. The values for f, ADC_{diff}, ADC_{perf}, and rCBF were recorded. In addition, maps for the different components were calculated and displayed.

Results

Figure 1 shows the signal decay as a function of b in three orthogonal gradients (x, y, z) directions in both white and gray matter volume at 1.5T and 3T. We can see the differences in signal decay between grey and white matter as well as differences in signal decay characteristics between low b-values and high b-values. From the areas analyzed, the mean f is 6.3% in white matter and 10.7% in grey matter for 1.5T. At 3T the mean f is 4.0% for white matter and 8.5% for grey matter. In both white matter and grey matter for both field strengths, ADC_{perf} is much greater than ADC_{diff}. ADC_{perf} of grey matter is on average 1.3 times greater than that of white matter at 1.5T and 2.5 times greater at 3T. The derived rCBF is significantly greater in grey matter compared to that of white matter at both field strengths. Figure 2 shows maps for fraction blood volume, ADC_{perf}, and rCBF.

Discussion

The ROI analysis performed shows b value $\leq$ 5 sec/mm^2 is recommended to capture the fast initial decay in grey and white matter. An rCBF map can be derived from the initial perfusion component. There are differences in the results derived from 1.5T verses 3T. The 3T white matter perfusion values appear to be underestimated. This could possibly be due to the following reasons: difference in inherent SNR, magnetic transfer effects, T2 value differences, as well as different blood saturation effects in a multi-slice acquisition. These factors need to be better understood and minimized in order for the technique to be used clinically. We have successfully quantified and characterized perfusion contributions in DWI. Additional data collection and analysis is ongoing to further investigate how to optimize this method.

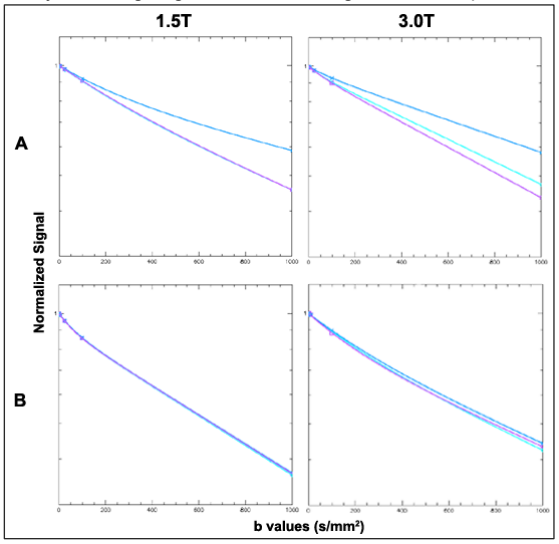

Figure 1: (A) White Matter, (B) Grey Matter

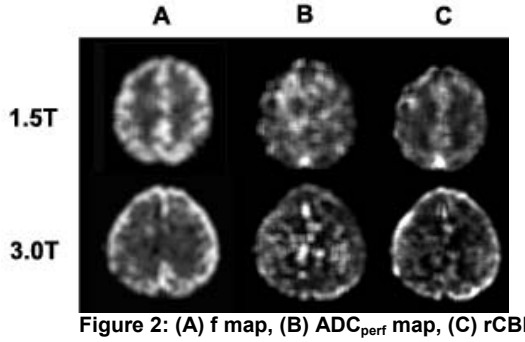

Figure 2: (A) f map, (B) ADC_{perf} map, (C) rCBF map

	f	ADC_{diff} (x 10^{-3} mm^2/s)	ADC_{perf} (x 10^{-3} mm^2/s)	rCBV (a.u.)
1.5T WM	6.3 ± 3.7%	0.62 ± 0.04	9.01 ± 3.53	46.3 ± 4.2
1.5T GM	10.7 ± 2.5%	0.96 ± 0.10	12.0 ± 2.8	124 ± 18
3.0T WM	4.0 ± 1.0%	0.62 ± 0.01	3.26 ± 1.40	13.6 ± 7.8
3.0T GM	8.5 ± 2.4%	0.81 ± 0.07	18.0 ± 8.6	142 ± 76

Table 1: Mean ROI results with standard error of mean

Reference : Le Bihan D et al, Radiology 1988: 168(2); 497-505