Partial Volume Averaging and Contrast-to-Noise Ratios in Diffusion Tensor MRI: Effects of using multiple b-values

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Introduction

Partial volume averaging (PVA) can significantly influence the accuracy of diffusion tensor (DT) measurements. In 2001 Alexander *et al* [1] used a two-tensor compartment model to describe simple partial volume effects, and compared the performance of two acquisition schemes each employing six gradient directions and only 1 b-value. The aim of this study is to analyse how the number of gradient directions (N_d) and the number of b-values (N_b) can contribute to increase the accuracy in the estimated apparent diffusion coefficient (ADC), fractional anisotropy (FA), and the principal direction of diffusion.

Methods

Simulations were computed for $6 \le N_d \le 60$ and $1 \le N_b \le 5$. The highest b-value used was b_{max}=1570 s/mm^2, and the other b-values were equally spaced in the interval $0 \le$ b-value $\le b_{max}$. The first fibre was aligned with the gradient frame of reference. The DT for the second fibre, D_2, was obtained by rotating D_1 by an angle θ around the y-axis. The noise-free diffusion weighted signals were calculated according to the two-tensor compartment model (1) [1]. Rician noise was then added to this data, and scaled to different values of signal to noise ratio (SNR). The values of FA, ADC and α (the angle between the simulated direction for the first fibre (D_1) and the estimated direction) were then computed by fitting this data to the traditional tensor model (2) [2]. To evaluate the performance of each scheme in the presence of noise, the procedure of noise generation and determination of ADC, FA and α, was repeated 2^{13} times. The simulations were repeated for different values of simulated FA and ADC of the individual fibres, using typical values for white matter (WM), grey matter (GM) and cerebrospinal fluid (CSF). The different combinations WM+WM, WM+GM and WM+CSF were considered, as well as different values of θ and f. Simulated ADC for GM was 0.7×10^{-3} mm^2/s, and 3.0×10^{-3} mm^2/s for CSF. When we have PVA in a voxel, there is no correct answer to what the real values of ADC and FA should be, and therefore the best we can expect from any acquisition scheme is that it is able to differentiate between different tissue types. For this reason, we also calculated the contrast-to-noise ratios (CNR) (3) [3] for ADC and FA, as a function of N_d and N_b. For the calculation of CNRs, we simulated the signal from voxels with 2 crossing fibres, 1 single fibre, and isotropic diffusion. These results were compared with the contrast-to-scatter ratios (CSR) (4) [3] calculated for FA and ADC for two datasets obtained for the same volunteer with a 3T Siemens Trio, with 2 different acquisition schemes: 63 directions×1 b-value and 12 directions×5 b-values. The CSRs were calculated across 6 different regions of interest (ROI): grey matter (GM), thalamus (TH), parasaggital white matter (PWM), pons (P), internal capsule (IC) and splenium (S).

$$S_{D_1+D_2}(b,\vec{r}) = S_0\left(f\, e^{-b\vec{r}^T D_1 \vec{r}} + (1-f)\, e^{-b\vec{r}^T D_2 \vec{r}}\right) \quad (1)$$

$$S_D(b,\vec{r}) = S_0 e^{-b\vec{r}^T D \vec{r}} \quad (2)$$

$$CNR = \frac{\overline{metric(fibre_1)} - \overline{metric(fibre_2)}}{\left(\sigma_1^2 + \sigma_2^2\right)^{1/2}} \quad (3)$$

$$CSR(metric) = \frac{\overline{metric(ROI_x)} - \overline{metric(ROI_y)}}{\left(\sigma_x^2 + \sigma_y^2\right)^{1/2}} \quad (4)$$

Results and Discussion

Simulations: For WM+WM simulations, the estimated values of ADC and FA decrease as both the angular separation θ and the value of f increase, up to a maximum of 44% decrease for FA and 13% for ADC (60 directions×1 b-value, SNR=10). Increasing N_b for the same number of total acquisitions reduces the decrease of FA and ADC to 40% and 10% of the single compartment values (12 directions×5 b-value, SNR=10). The estimated direction is between the two simulated fibres, converging to ½θ , for the case f=0.5. This is independent of the number of b-values used. For WM+GM and WM+CSF simulations, the estimated ADC converges to a value between the simulated ADCs for the two compartments, depending on the value of f. FA shows a significant decrease relatively to the single compartment values: for WM+GM FA decreases up to 44% for 60 directions×1 b-value, and 40% for 12 directions×5 b-values, while for WM+CSF FA decreases up to 46% for 60 directions×1 b-value, and 36% for 12 directions×5 b-values. For WM+GM, α shows an increase of 105% relatively to the single compartment value for f=0.5, indicating that the contamination of GM in a voxel significantly affects the ability of the DT model to identify the direction of anisotropy. This increase is even more significant for WM+CSF (124%), suggesting that the uncertainty associated to the estimated fibre direction increases with the mean diffusivity of the isotropic compartment.
The standard deviation calculated from the 2^{13} repetitions of the same experiment generally decrease with N_b, for all simulated parameters.
Fig.1 shows the results obtained for CNR(ADC) and CNR(FA) for the different simulated compartments. CNR(ADC) increases with N_b, while keeping the total number of acquisitions less or equal to 60 (the maximum number of acquisitions used in clinical studies, due to time limitations). CNR(FA) generally increases as we increase N_b. However, when we restrain the total number of acquisitions to a maximum of 60, the best CNR(FA) result is obtained for 30 directions×2 b-values. This is because increasing the number of b-values with a constant number of total acquisitions corresponds to a decrease in N_d, which results in a less robust estimation of FA when we use less than 20 directions (e.g. [4]). But even so, the CNR results obtained with 12 directions×5 b-values are better than the ones obtained for 60 directions and only 1 b-value.

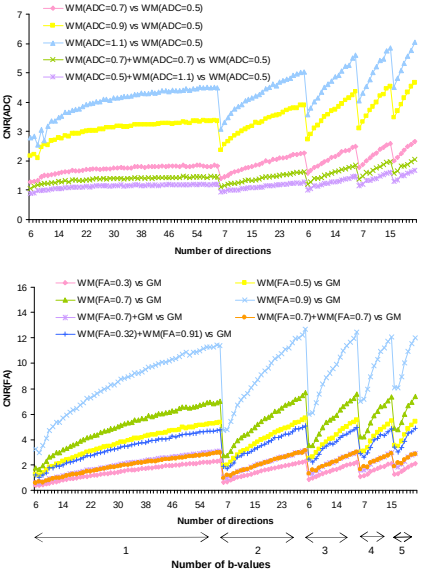

Figure 1 – CNR(ADC) and CNR(FA) obtained for the simulated data. For CNR(ADC) all fibres used had the same simulated FA=0.7071. For CNR(FA) all fibres used had the same simulated ADC=0.7 mm^2/s.

Experimental data: The standard deviations calculated for all ROIs were smaller for the acquisition scheme using 12 directions×5 b-values. In addition, this acquisition scheme also produced higher CSR values for ADC and FA, for all the 15 possible combinations of the six ROIs analysed (CSR(FA) results shown in Fig. 2). These results confirm that an acquisition scheme employing multiple b-values has a greater power to differentiate between tissue types, than one employing a larger number of unique directions but only one b-value.

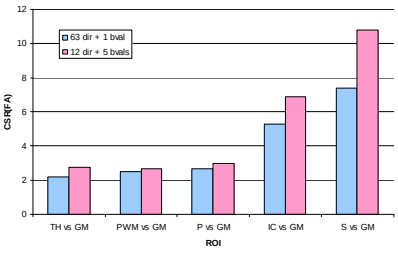

Figure 2 – CSR(FA) between GM and tissues with different anisotropies.

Conclusion

Simulations show that, in the presence of PVA, the use of multiple b-values produces less accentuated decreases of ADC and FA, and also results in less variability due to noise. Simulations and experimental data also show that the ability to differentiate between tissue types increases with N_b. While estimating fibre orientation, however, even though increasing N_b reduces the variability of the estimated direction, this estimate still does not converge to either of the simulated fibres. For this reason, the applications of tractography methods based on the DT model are very limited, and other methods, which incorporate uncertainty due to noise and PVA, should be considered.

References: 1. Alexander *et al.* Magn Reson Med 2001; 45:770-780. 2. Basser *et al.* J Magn Reson 1994; 103:247-254. 3. Basser *et al.* J Magn Reson 1994; 103:247-254. 4. Papadakis *et al.* Magn Reson Imag 2000; 18 : 671-679.