Evaluation of the effect of exchange on the diffusion weighted MR-signal in brain white matter

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Introduction

The origin of the anisotropic diffusion weighted (DW) MR-signal observed in brain white matter (WM) is still not completely understood [1]. In this study, Monte Carlo simulations of diffusion are presented in a geometry with intra- and extracellular compartments imitating WM. The effect of exchange between the compartments on the DW MR signal has been evaluated.

Methods

Software phantom construction: The diffusion is modeled in a raster with a square cross-section filled with infinitely long parallel aligned rigid cylinders. The diameters (10 μm ± 3.5 μm) and density (79.5%) of the cylinder packing were chosen similar to those observed in brain WM [1] (see Figure 1).

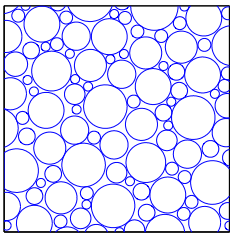

Figure 1: Cross-section of 80 μm x 80 μm through the generated phantom.

Diffusion MR simulation: The diffusion process was modeled by Monte Carlo (MC) simulation of 3.0×10^5 random walkers as described in [2]. The trajectory of random walker was generated by moving the particle during each time step t of 0.07 ms over a distance $(6Dt)^{1/2}$ in a randomly chosen radial direction. The diffusion coefficients D were chosen similar to those described in literature for the intra- and extracellular space in brain white matter [3]: 1.0×10^{-9} m2/s inside the cylinders and 2.5×10^{-9} m^2/s outside the cylinders. Exchange between the intracellular and extracellular space has been enabled by the method described in [4]. The phase ϕ of the particle was updated during each time step t according to the current position and the presence of diffusion gradients. Varying diffusion gradients were applied during a time δ of 0.7 ms. The diffusion time Δ was chosen to be 50 ms with corresponding b-factors, defined by $\gamma^2\delta^2G^2(\Delta-\delta/3)$, ranging from 0 up to 2500 s/mm^2. The diffusion weighted MR-signal S was derived as the sum of the phases of all spins $\sum e^{i\phi}$.

Data fitting: The simulated data S(b) sets were fitted to the diffusion kurtosis model [5]: $\ln\left(\frac{S(b)}{S(0)}\right) = -ADCb + \frac{1}{6}ADKADC^2b^2 + O(b^3)$,

where ADC is the apparent diffusion coefficient and ADK is the apparent diffusion kurtosis which is derived from a cumulant expansion.

Results

Figure 2 shows the natural logarithm of the simulated DW MR signal S(b) for increasing permeability P. In the limit of very high permeability, the diffusion is Gaussian with mono-exponential signal attenuation and a kurtosis equal to zero. However, when the exchange across the cells slows down, the attenuation curve becomes non mono-exponential and the kurtosis becomes positive. The corresponding fitted ADC- and ADK-values are shown in figure 2 as a function of the permeability. The relative change in ADK when changing the permeability from 0 up to 100 mm/s, is about 36 %, while the relative change in ADC is only 9 %.

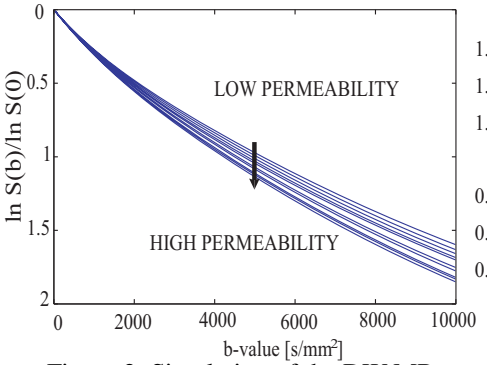

Figure 2: Simulation of the DW MR signal. The arrow stands for increasing permeability P from 0 up to 100 μm/s.

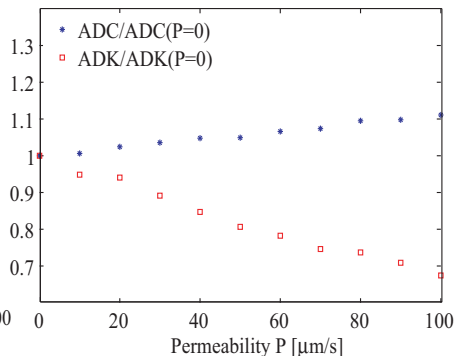

Figure 3: relative changes in the ADC- and ADK-values as a function of the permeability P.

Discussion and Conclusion

The cumulant expansion form turns is useful for characterizing the diffusion process in a multi-compartment system with or without mutual exchange. The results indicate that the kurtosis is a good probe for the presence of membranes and is sensitive to changes in permeability. Fitting the kurtosis could reveal new insights in the physiology of cells during pathological states. As an example, because of the remarkable correlation with membrane depolarization and cell swelling induced by ischemia, it has been assumed for a long time that the decrease in ADC observed in WM during stroke is caused by an increase of intracellular water. Alternatively, the decrease in ADC could be explained by a sudden drop in membrane permeability resulting in an increase in kurtosis as shown in [6,7].

References

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