Dynamic displacement in human brain studied using q-Space Diffusion MRI at a 3T clinical scanner

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Introduction

Since the early 1990's, q-space NMR has been used to investigate microstructural parameters [1]. Studies have been conducted on biological systems to quantify the sizes of cellular structures [2, 3], but these studies have primarily been made on NMR spectrometers. At a clinical MRI scanner, the short gradient pulse (SGP) approximation is harder to fulfil due to the lower gradient amplitudes. This limits the accuracy of q-space analysis as a tool for quantification [4] and only a few studies have been performed with such limitations. In this study initial results are presented on the effects of restricted diffusion on the diffusion propagator, measured *in vivo* with a 3T clinical MRI scanner.

Theory

The q-space analysis provides a framework for calculating the diffusion propagator. In the limit where the SGP condition is fulfilled and the diffusion time allows the molecules to sample the entire confining pore, the diffusion propagator becomes an autocorrelation function of the spin density [1]. In this case, the full width at half maximum (FWHM) of the propagator gives the size of the pore [2]. However, it is known that violating the SGP condition results in a systematic underestimation of the FWHM for restricted diffusion [4]. Another limitation is that for long diffusion times (Td) the FWHM does not reflect the size of the pores *in vivo* due to the semipermeability of the pore membranes [5]. For free diffusion the FWHM increases with Td as $FWHM = 4\sqrt{(D \cdot Td \ln 2)}$. In this study we explore the effects on FWHM in vivo when Td is varied.

Method

Measurements were performed on a Siemens Allegra 3T system with a 40 mT/m gradient system. In total three volunteers were scanned using a locally developed stimulated echo sequence (STEAM), similar to the design presented by Steidle et al. [6]. The imaging parameters were: TR = 2500 ms, slice thickness = 5.0 mm, in plane resolution = 1.81x1.81 mm^2 and NEX = 4. The complex images were denoised by filtering in the wavelet domain [7]. The signal curve S(q) was sampled in 6 diffusion encoding directions at 12 equidistant points (q = 10-906 cm^{-1}) for five different diffusion times (Td = 64, 81, 100, 121 and 144 ms) with a constant TE = 114 ms, but with a varying mixing time (TM). The diffusion gradient duration was δ = 43 ms. Regions of interest (ROI) were placed in gray matter (GM), posterior limb of the internal capsule (plic) and in genu of corpus callosum (gcc), see figure 1. The obtained signal curve S(q) for each ROI was mirrored so that $S(-q) = S(q)$ [8] and then re-sampled to yield equidistant steps in q. No zero filling was made. This yielded a q-space resolution of $1/2q_{max} = 5{,}5\mu m$ [10]. Signal values below than a noise cut off level were set to zero. Finally the probability distribution was obtained from the FFT of S(q), from which the FWHM was measured. To obtain rotationally invariant parameters, a tensor model was applied to the data as described in [3]. The largest and the smallest eigenvalue, corresponding to diffusion perpendicular and parallel to the fibres (λ_1 and λ_3 respectively), were analysed.

Results

The results are shown in table 1 and figure 2.

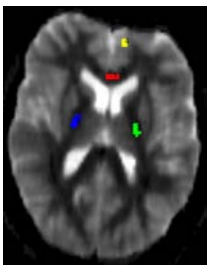

Fig 1. The employed ROI, placed in the parametric FWHM(λ_1) map.

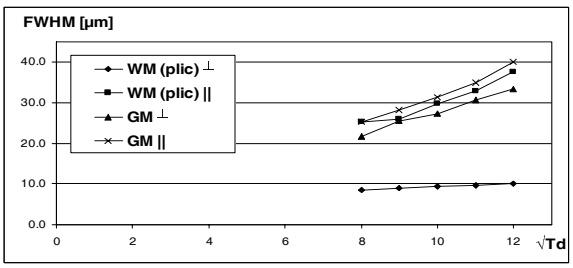

Fig 2. The FWHM values obtained from λ_1 ($\perp$) and λ_3 ($\parallel$) as a function of √Td. It can be seen how λ_1 increases linearly with √Td whereas λ_3 does not.

√Td	WM (plic)		WM (gcc)		GM	
√ms	λ_1	λ_3	λ_1	λ_3	λ_1	λ_3
8	8.5	25.2	8.8	30.8	21.8	25.3
9	8.9	26.0	6.4	35.0	25.6	28.3
10	9.4	29.9	10.1	39.4	27.3	31.4
11	9.6	32.8	10.0	45.6	30.7	34.9
12	10.1	37.6	10.9	47.6	33.4	40.0

Table 1 . The leftmost column shows the √Td and the following columns the volunteer average of the smallest (λ_1) and the largest (λ_3) eigenvalue in units of µm for three different ROIs.

Discussion

The results show only a slight increase of FWHM in the direction perpendicular to the fibres in WM, whereas FWHM increases approximately linearly with √Td in the direction parallel to the fibres, as for freely diffusing water. In GM, the FWHM generally increases linearly with √Td and shows approximately the same value for both λ_1 and λ_3. This indicates that the water molecules move more freely in GM than in WM, and without strong directional preference. The slight increase of λ_1 in WM might be explained by permeability, i.e. for longer diffusion times the probability of water molecules migrating between the compartments increases [5]. It should be noted that with δ = 43 ms the SGP approximation is violated and hence the FWHM values are expected to be underestimated. Also note that the contrast between WM and GM in terms of λ_1 increases with Td. The present results are in agreement with the NMR spectrometer results of Nossin-Manor [9]. With this study we conclude that it is possible to estimate structural brain dimensions in vivo the by the q-space analysis using a clinical MRI scanner, provided that the imaging protocol is optimized for this purpose.

References

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