Preliminary experiences with implementing high resolution CSF-suppressed Optic Nerve DTI at 3.0T with and without parallel imaging

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Introduction: The optic nerve (ON) is affected in neurological diseases, e.g. in multiple sclerosis (MS) where optic neuritis is often an initial manifestation of the disease. The ON is useful for the study of the pathophysiology of individual demyelinating lesions since it is a discrete white matter tract and MRI-detected lesions can be directly related to measures of visual function. Other techniques which can be correlated with ON MR data include Optical Coherence Tomography (OCT) for measuring retinal nerve fibre layer thickness (related to axonal loss) and visual evoked potential (VEP) latency which reflects myelination. The ON is very small in size (~4mm in diameter), therefore high spatial resolution is required using MRI. Data acquisition must also be fast to minimise problems associated with the involuntary motion of the nerve. The ON is surrounded by Cerebrospinal Fluid (CSF) and fat, which can lead to contamination of signal, therefore both fat and CSF suppression are required. The susceptibility difference between the bone cavity of the optic canal and sinuses may also lead to image distortions with EPI, magnified at 3.0T field strength. Diffusion Tensor Imaging (DTI) provides quantitative information about the microstructure of tissue *in vivo* and the diffusion behaviour of water molecules can be modified by pathology. In particular oedema and inflammation (as in acute optic neuritis) and demyelination and gliosis (observed in the ON following optic neuritis) may result in changes in the diffusion behaviour of water molecules in the ON. We present our preliminary experiences with implementing ON DTI on two different manufacturer's 3.0T scanners using different data acquisition methods including the use of parallel imaging.

Methods: We tested a number of protocols with parallel imaging using ASSET (GE) or GRAPPA (Siemens) and different speed up factors. The most promising options are presented here and can be implemented on either scanner with equivalent results:

(i) Parallel imaging – speed up factor 2: A healthy volunteer (female, age 39) was scanned coronally using a 3.0T GE Signa scanner (General Electric, Milwaukee, WI, USA) and an 8-channel receive head coil. ASSET was used with a speed-up factor of 2 in the R/L direction. FLAIR and a spectral spatial excitation pulse were used to suppress high signal from the CSF and fat surrounding the ON. Data were acquired using a diffusion scheme with 36 diffusion-weighted (dw) directions and 5 b=0. The 36 dw directions consisted of the same 6 directions repeated 6 times (b=600s mm^{-2}), and each of these was obtained by simultaneously applying 2 gradients with maximum amplitude (e.g. 1,1,0) giving the requested b-value for the minimum TE value possible. This whole procedure was repeated 7 times. Ten 4mm slices were acquired, with FOV=15cm, acquisition matrix 120x120 (pixel size 1.25mmx1.25mm), TR=8s, TI=2s, TE=61.7ms. Total scan time was approximately 40 minutes.

(ii) Reduced FOV – standard acquisition: A healthy volunteer (female, age 27) was scanned coronally using a 3.0T Siemens Magnetom TIM Trio scanner (Siemens Medical Solutions, Erlangen, Germany) and a 12-channel receive head coil. Sixteen 4mm slices with FLAIR and fat saturation on were acquired with the following parameters: acquisition matrix 128x64, FOV=15cm; phase FOV=50% (pixel size=1.17mmx1.17mm), TR=6s, TI=1s, phase encoding in the S/I direction and signal saturation bands applied both superior and inferior to the prescribed imaging area to minimise wrap-around artefacts. 40 averages were acquired, with 6 dw directions plus one b=0 image per average, taking approximately 30 minutes in total to run.
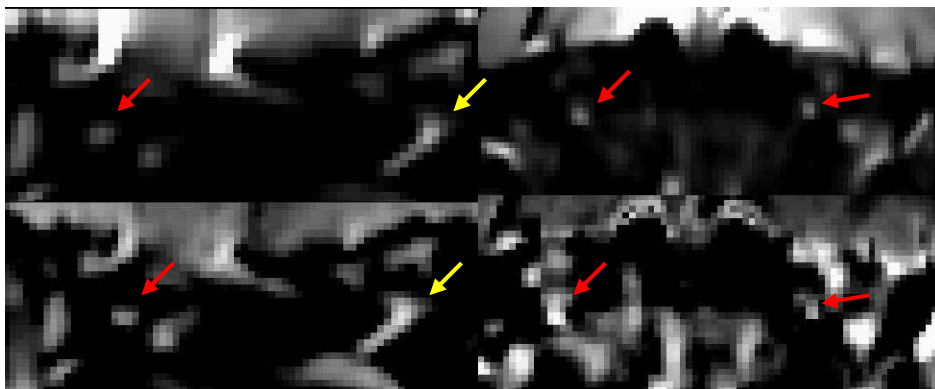
Analysis: For both methods the same analysis procedure was followed; Rayleigh noise correction was performed (with data excluded from averaging where noise correction yielded a negative number i.e. the noise correction value was larger than the signal value from which it was to be subtracted). The diffusion data were then averaged to 7 diffusion-weighted directions (including the b=0 images), non-linear smoothing (1) was performed to reduce systematic and random errors in the data, then the data were eddy corrected using the eddy correct function within FSL (2) and fitted using Camino (3) to yield mean diffusivity (MD) and fractional anisotropy (FA) maps.

Results: The figure consists of an averaged b=0 image (top) and MD image (bottom) acquired using the method with (left) and without (right) parallel imaging. A MD value of 964 (±106) x 10^{-6} mm^{2}s^{-1} was measured with parallel imaging (i) (over 4 consecutive slices in the left ON), and 1146 (±209) x 10^{-6} mm^{2}s^{-1} using the method without parallel imaging (ii) (over 4 consecutive slices and in both nerves). These values are comparable with literature: at 1.5T Wheeler-Kingshott *et al* (4) measured a MD value of 1058 (±101) x 10^{-6} mm^{2}s^{-1} over the whole ON (9 3mm slices), and the ON MD was measured as 1100 (±200) x 10^{-6}mm^{2}s^{-1} by Chabert *et al* (5).

The right/left (R/L) distortions present in the images acquired with parallel imaging result in the right ON being obscured by an artefact (yellow arrow); this artefact was also observed when data were acquired using a very similar protocol on the Siemens 3.0T scanner. When the phase encoding direction was reversed a similar artefact appeared on the other ON, therefore acquiring twice the amount of data with the phase encoding direction reversed in the second acquisition could be one possible solution to this problem, although this would double the scan time. Unfortunately, when entering the middle portion of the ON the images were greatly distorted by the presence of the temporal lobes and the proximity of the sinuses and optic canal and it was impossible to determine the position of the ON. It was also found that using parallel imaging speed-up factors>2 (on both scanners) led to reconstruction artefacts probably due to the low SNR of the images, especially for slices near the optic chiasm, i.e. deeper in the brain and further away from the coil elements.

In the images acquired without parallel imaging, a lot of wraparound artefact is present in the S/I direction, but the images are much less distorted in the R/L direction since phase encoding was applied in the S/I direction. By carefully positioning the slices, it was possible to ensure that the wrap-around falls below the level of the ON, therefore it was possible to measure the MD in both left and right ON, however the middle portion of the ON could not be identified using this protocol either.

Conclusions: High resolution coronal images of the ON were successfully acquired at 3.0T, incorporating CSF and fat suppression to aid identification and delineation of the nerve. Averaging of magnitude images (i.e. insensitive to random phase changes caused by motion) was used to compensate for the low SNR in the acquired images and for averaging the position of the centre of the ON, which can therefore be estimated with some confidence (6). The use of parallel imaging in both scanners meant that phase encoding had to be applied in the R/L direction because of the geometry of the coil elements. This resulted in R/L image distortions, rendering it impossible to successfully image both nerves simultaneously. ON MD values measured without parallel imaging were consistent with previous measurements made in the ON at 1.5T. Possible improvements to the technique might include the use of PROPELLER (7) (a k-space filling technique that corrects for large amounts of motion). With the advent of 32 channel coils parallel imaging methods with a higher speed-up factor may reopen the potential of parallel imaging methods for minimisation of R/L distortions and faster acquisition times for ON DTI.

References: [1] Parker GJM *et al* JMRI 11: 702-11 (2000), [2] Nunes RG *et al* Proc 13th Annual Meeting ISMRM, Miami, USA, p2279, 2005, [3] Cook PA *et al* Proc 14th Annual Meeting ISMRM, Seattle, USA, p. 2759, 2006, [4] Wheeler-Kingshott CAM *et al* MRM 47: 24-31 (2002), [5] Chabert S *et al* JMRI 22:307-10 (2005), [6] Wheeler-Kingshott CAM *et al* Proc 9th Annual Meeting ESMRMB, p443, 2000, [7] Pipe, J *et al* MRM 42, 963-969 (1999)

Figure: Averaged b=0 (top) & MD images (bottom) for protocols (i)-left and (ii)-right. Arrows indicate the position of the ON (red) & artefact (yellow).