Full-brain Q-Ball Imaging in a Clinically Acceptable Time: Application to White Matter Fibre Tractography

J. S. Campbell[1,2], K. Siddiqi[2], G. B. Pike[1]

[1]McConnell Brain Imaging Centre, Montreal Neurological Institute, Montreal, Quebec, Canada, [2]Centre for Intelligent Machines, McGill University, Montreal, Quebec, Canada

Introduction Q-Ball Imaging (QBI) has been presented by Tuch *et al*. as a model-free method for measuring the diffusion orientation distribution function (ODF) [5]. QBI yields a surface $\Psi(\theta,\phi)$ that reflects the underlying fibre structure, even in the presence of subvoxel partial volume averaging of fibre directions. This surface can potentially be utilized for fibre tract reconstruction without imposing a limit on the number of fibre directions in a voxel, performing peak extraction, or imposing a model for the shape of the diffusion displacement pdf. The downfall of the protocol suggested by Tuch *et al*. is that it requires measurement of 484 diffusion-weighted images, and therefore requires a discouragingly long acquisition time for the full brain (approximately one hour). In this study we investigate the feasibility of using fewer (90) diffusion weighted images to obtain full-brain high angular resolution images of the function Ψ, and use this surface to perform fibre tracking via a surface evolution-based tractography algorithm [1].

Methods MRI data were acquired on a Siemens 1.5T Sonata MR scanner (Siemens Medical Systems, Erlangen, Germany) using an 8-channel phased-array head coil. Diffusion encoding was achieved using a single-shot spin-echo echo planar sequence with twice-refocused balanced gradients. Four coregistered datasets were acquired, each consisting of 90 diffusion weighted images using isotropically spaced gradient directions, 2.8mm isotropic resolution, b=3000 s/mm^2, q=0.35 μm^{-1}, TR=8s, TE=110ms, and 30 slices. All diffusion scans were cardiac gated to reduce pulsatile motion artifacts. Each acquisition of 90 diffusion weighted images (DWIs) took approximately 12 minutes, for a total scan time of 48 minutes for the four datasets. The angular resolution achieved with this protocol is approximately 19^o. A 1mm isotropic resolution T1 weighted anatomical scan was also acquired (TR=22ms, TE=9.2ms, α=30^o). For each separate 90-DWI dataset, we calculated $\Psi(\theta,\phi)$ using the Funk-Radon transform, as described in [5]. For visualization and for fibre tracking, we normalized the values of Ψ to have a mean of unity in each voxel.

We defined an anisotropy index based on the fractional anisotropy (FA) [4] commonly used in diffusion tensor (DT) experiments. Our generalized fractional anisotropy (GFA) is the normalized standard deviation of the squared diffusion ODF measurements,

$$\mathrm{GFA} = \sqrt{\frac{\sum_{n=0}^{N} (\Psi^2(\theta_n,\phi_n) - <\Psi^2>)^2}{(<\Psi^2>)^2}},$$

where N is the number of directions $(1,\theta,\phi)$ for which $\Psi(\theta,\phi)$ has been calculated, which we set equal to the number of directions for which base DWIs were measured (90 in this study), and $<\Psi^2>$ is the mean square value of all the $\Psi(\theta_n,\phi_n)$. This index is similar in spirit to the spherical diffusion variance defined by Frank [2] to describe anisotropy in high angular resolution ADC profiles.

We performed white matter fibre tracking using a surface evolution algorithm, where the speed of surface evolution in direction (θ,ϕ) is driven by $\Psi(\theta,\phi)$. Our algorithm, Tracking Using arbitrary Fibre Orientation Likelihood Distribution (TUFOLD), is a modification and extension of fast-marching tractography (FMT) [3]. It yields a scalar map of the likelihood of connection of each voxel to a user-defined seed, as well as explicit tracts given by 3D curves.

Results Fig.1 shows the GFA plot for the full brain, the surfaces $\Psi(\theta,\phi)$ for a small ROI containing sub-voxel partial volume averaging of fibre directions, and the tracking results. The reproducibility, as indicated by the standard deviation over the 4 individual coregistered datasets, was 0.2±0.1 for the GFA (averaged over all voxels), and 0.05±0.03 for the normalized $\Psi(\theta,\phi)$ (averaged over all voxels and all (θ,ϕ)).

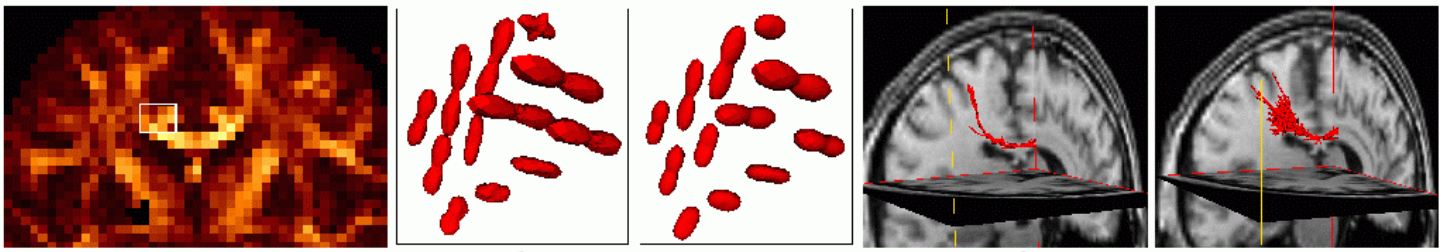

Fig. 1. (a): Generalized anisotropy index (GFA) calculated using the QBI ODF Ψ. (b-c): The ODF surfaces in each voxel of an ROI (shown in rectangle in (a)) containing partial volume averaging of fibres from the corpus callosum and the superior longitudinal fasciculus. The insets have been rotated and the ODF values squared in order to clearly show the directions. (b): QBI reconstruction; (c) DTI reconstruction from the same data. (d): Tracking results using QBI reconstruction and (e) tracking results using DTI reconstruction, overlaid on T1-weighted anatomical image. The seed point was in the centre of the corpus callosum.

Discussion As can be seen in Fig.1b, the 90-direction QBI protocol is useful for detecting sub-voxel partial volume averaging of fibre directions in the major white matter fibre systems. Fig.1d and Fig.1e show the tracking results using TUFOLD with QBI and DTI reconstructions of the ODF, respectively. Visual inspection indicates that the QBI tracking yields cleaner results, where the DTI tracking is more prone to go off track due to partial volume averaging. More investigation is required to determine the optimal balance between high angular resolution and high spatial resolution (i.e., with DTI reconstruction) for any given scan time. The optimal protocol for diffusion measurements for the purpose of fibre tracking will depend on the time the subject can comfortably stay in the MR scanner, and the size and complexity of the fibre structures being studied.

Conclusion We have shown that QBI reconstruction can be used to calculate scalar maps of anisotropy, maps of the diffusion ODF at each voxel, and 3D curves representing fibre tracts through any given seed voxel or region. We have shown results from a clinically feasible diffusion acquisition for the full brain. This method could prove to be superior to the diffusion tensor model for extracting information in the presence of partial volume averaging of white matter fibre tracts.

References [1] Campbell, J.S.W. *et al. Proc. ISMRM* 2002:1130. [2] Frank, L.R. *Magn. Reson. Med.*, 45(**6**):935-39, 2001. [3] Parker, G.J.M. *et al. IEEE Trans. Med. Imag.*, 21(5):505-512, 2002. [4] Pierpaoli, C. *et al. Magn. Reson. Med.*, 36:893-906, 1996. [5] Tuch, D.S. PhD. Thesis, Harvard-MIT, 2002.