A data acquisition strategy to acquire DTI data of the brain with sub-millimeter isotropic voxel size

T-Q. LI[1], M. Noseworthy[2]

[1]Lab of Functional and Molecular Imaging, NINDS/NIH, Bethesda, MD, United States, [2]IDepartments of Radiology & Medical Physics, McMaster University, Hamilton, Ontario, Canada

Introduction: With the development of parallel imaging techniques, diffusion tensor imaging (DTI) of the human brain with sub-millimeter in-plane resolution can be readily achieved at 3T using single-shot echo-planar imaging (EPI) techniques. However, it is still a serious challenge to achieve sub-millimeter resolution in the slice-selection direction due to the hardware and signal-to-noise ratio limitations. In this study, a data sampling strategy was investigated to obtain DTI data of the human brain with sub-millimeter isotropic voxel size using SESE EPI at 3T.

Methods: The basic idea is the following: three separate DTI data sets were acquired in three orthogonal planes using a 0.9×0.9 mm^2 in-plane resolution and a slice thickness of 2.7 mm, which was three times of the in-plane pixel size. Combining the three data sets, for each $2.7\times2.7\times2.7$ mm^3 volume composed of 27 isotropic voxels with a resolution of $0.9\times0.9\times0.9$ mm^3 there were 27 measurements available. In principle, the DTI parameters for the 27 isotropic voxels could be directly derived from the 27 measurements. However, the linear equation system for this problem is ill-conditioned, because the sum from each set of the measurement is redundant. Our approach to solve this problem was to incorporate additional prior knowledge from a 3D T_1-weighted anatomical scan at he resolution of 0.9 mm isotropic voxel size, which can be used to classify most of the voxels as CSF, gray matter, or white matter. After including prior knowledge on the diffusion characteristics for CSF, gray and white matter as constrains, stable solution was obtained for the linear equation system. This can be posed mathematically as follows: As shown in fig. 1A, if we assume that the fraction anisotropy (FA) for the individual isotropic voxels is X_i (i=1,2,…,27) and the measured FA results along axial, sagital, and coronal planes are A_k (k=1,2,…9), S_k, and C_k, respectively, then X_i are related to measured data by a linear equation AX=Y. Here X is X_i presented n a column vector; Y=(S_k; C_k; A_k); and A is a 27×27 square matrix as shown in fig. 1B. Using the schematic drawing shown in fig. 1A, the derivation of matrix A is quite trivial. For example, S1=X1+X2+X3, C1=X1+X4+X7; and A1=X1+X10+X19.

Experimental: All MRI data were acquired on a GE 3T MRI Excite system ruing Epic 11.M3 using a 8 channel receiver head coil. The DTI data acquisition method was a single-shot EPI with a SENSE acceleration factor of 2. The main acquisition parameters were: slice thickness=2.7mm (which was the limit on the 3T when a spectral spatial excitation pulse was used), FOV=23 mm, matrix size=256×256, TE=80.7 ms, TR=15 s, full k-space sampling, 20 diffusion-weighting gradient directions at b=1000 s/mm^2, and 3 samplings at b=0. 8 repeated measurements were done for each orthogonal plane direction to achieve appropriate SNR. The acquisition time for each orthogonal plane direction was therefore about 46 minutes. The entire DTI measurements lasted for more than 2 hours for each subject. In additional to DTI measurements along three orthogonal planes, T_1-weighted anatomical scan was performed using a 3D spoiled gradient recalled echo (SPGR) sequence to achieve isotropic voxel size of 0.9 mm. Each DTI data set was analyzed using a global minimization algorithm that takes into consideration the following three models simultaneously: 1) eddy current artifacts correction by estimating the whole brain based shearing, scaling, and translation effects; 2) motion correction based on 3D rigid-body motions; 3) second order self-diffusion tensor model. The SPGR data were segmented into CSF, gray, and white mater. For CSF voxels, FA=0 was assigned. After incorporating empirical FA constrains for gray and white matter voxels, the linear equation system was solved globally for all brain voxels simultaneously.

Results and Discussion: Fig. 1C shows a typical sample set of the derived FA maps. Despite of the high condition for matrix A, For the FA parameter, the solution of the liner equation system was very stable and insensitive to noise level, after applying the prior knowledge constrains. Two noise levels were tested by using averages of 4 or 8 repeated measurements. It should be pointed out that there is no need for complete classification of all voxels, For example, if a cube contains 3 CSF voxels, this information alone is sufficient to solve the equation system vigorously. With this approach, whole brain FA data with an isotropic voxel size of 0.9 mm were obtained using a 3T clinical scanner.. Extension of this approach to extract eigenvector data at the same resolution is currently undergoing.

Support gratefully acknowledged from NIH (R211-63455) and the Whitaker Foundation.

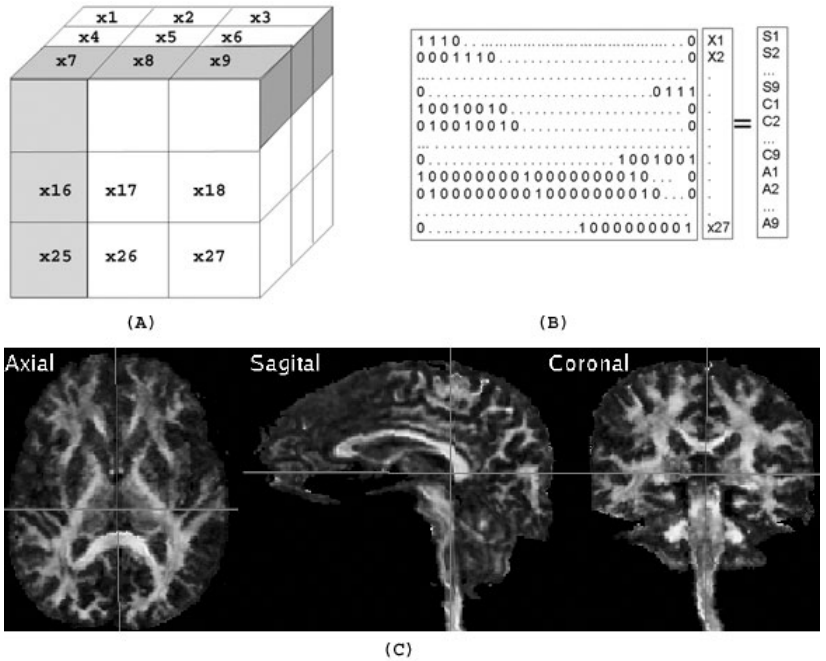

Fig 1. Each $2.7\times2.7\times2.7$ mm^3 cube was sampled three times along axial, sagittal, and coronal planes using a resolution of $0.9\times0.9\times2.7$ mm^3 (A). The diffusion anisotropy for each isotropic voxle, X_i (i=1, 2, …27) , is related to the sagitall (S1, S2, …S9), coronal (C1, C2, … C9), and axial (A1, A2, …A9) measurements by an ill-posed linear equation system (B). After incorporating additional prior knowledge from the high-resolution anatomical scan, whole FA data with an isotropic voxel size of 0.9 mm was obtained (C). The crossing lines indicate the positions of the depicted cross sections.