MR imaging of fiber-tract direction and diffusion in anisotropic tissues

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Purpose
In brain white matter1-3 and skeletal muscle4, the apparent diffusivity of water is largest when the diffusion gradient and fiber-tract directions are parallel; it is smallest when they are perpendicular.1-4. We exploit this phenomenon to image fiber-tract orientation noninvasively, using NMR. More generally, we image mean diffusion distances and other functional quantities derivable from the effective diffusion tensor.

Theory
We previously presented an equation for estimating the effective diffusion tensor, D_eff in each voxel, from spin-echo, pulsed-gradient experiments5:

\[ \ln \left( \frac{A(TE)}{A(0)} \right) = -\sum_{i=1}^{3} \sum_{j=1}^{3} b_{ij} D_{ij} \],

where A(0) and A(TE) are magnetization amplitudes, TE is the echo time, and b_{ij} are elements of the b-matrix5.

The tissue's three orthotropic axes coincide with the eigenvectors of D_eff. The effective diffusivities along these principal directions are the eigenvalues of D_eff, i.e., \( \lambda_1, \lambda_2, \) and \( \lambda_3 \). The tissue's fiber-tract direction is given by the eigenvector with the largest principal diffusivity.

Also associated with D_eff are three scalar invariants6, I_1, I_2, and I_3:

\[ I_1 = \lambda_1 + \lambda_2 + \lambda_3 = \text{Tr}(D_{\text{eff}}) \],
\[ I_2 = \lambda_1\lambda_2 + \lambda_1\lambda_3 + \lambda_2\lambda_3 \], and
\[ I_3 = \lambda_1\lambda_2\lambda_3 = |D_{\text{eff}}| \],

which are independent of the sample's orientation with respect to the lab frame of reference and depend on the tissue's microstructure5.

We propose three dimensionless anisotropy ratios. The first, \( \lambda_2/\lambda_3 \), indicates the degree of rotational symmetry about the longest (fiber-tract) axis; \( \lambda_1/\lambda_2 \) and \( \lambda_1/\lambda_3 \) are the relative diffusivities parallel and perpendicular to the fiber-tract axis.

Finally, we can construct an effective diffusion ellipsoid6 from D_eff in each voxel. The material's local orthotropic directions coincide with the ellipsoid's principal axes, with the longest axis parallel to the fiber-tract direction. The mean diffusion distances (at time t) along these orthotropic directions are the lengths of the ellipsoid's principal axes.

Materials and Methods
Using a 4.7-T imaging system, we acquired 135 diffusion-weighted sagittal images of a *in vivo* cat brain by applying combinations of diffusion gradients in the read, phase, and slice directions5. We measured A(TE) for each voxel in the image, and analytically calculated a b-matrix from each (diffusion and imaging) gradient sequence. Using Eq. (1), D_eff was optimally estimated in each voxel by weighted multivariate linear regression5.

**Results and Discussion**
From the estimated D_eff in each voxel, we imaged I_1=Tr(D_eff) for *ex vivo* cat brain (Fig. 1). Tr(D_eff), a measure of water mobility independent of tissue orientation, is significantly different in gray and white matter, fissures, ventricles, and the corpus callosum. Tr(D_eff) and the other scalar invariants will be useful for accurate assessment of tissue microstructure and its physiological state (e.g., in stroke monitoring).

We also constructed a diffusion ellipsoid image from D_eff in each voxel (Fig. 2). In a ROI near the corpus callosum, voxels in a CSF-filled ventricle appear as large spheres, indicating high isotropic diffusion, while voxels in white matter appear as prolate ellipsoids, indicating anisotropic diffusion with their longest axes correctly aligned with known fiber-tract direction.

**Conclusion**
MR diffusion tensor imaging is feasible. It reveals intravoxel microstructural information (that scalars such as proton density, T_1, and T_2 do not), embodied in images of diffusion ellipsoids, scalar invariants, and anisotropy ratios constructed from D_eff.

![Fig 1: Tr(D_eff) image of cat brain (sagittal section) with ROI.](image1)

![Fig 2: Diffusion ellipsoid image of ROI (rotated c.w. by 90°).](image2)

**References**