A SPATIALLY CONSTRAINED PROBABILITY DISTRIBUTION MODEL OF INCOHERENT MOTION (SPIM) IN QUANTITATIVE DIFFUSION WEIGHTED MRI

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Purpose: Quantitative diffusion-weighted MRI (DW-MRI) has a well-established role in the detection and characterization of a variety of abdominal abnormalities including liver fibrosis, tumors and active inflammation1-3. DW-MRI enables characterization of the tissue microstructure through measurement of the mobility of water molecules and the variations in their mobility due to changes in cell density, vascularity and membrane permeability. When imaging tissues with DW-MRI, mobility of molecules leads to attenuation in the diffusion signal, which is measured at multiple b-values. This attenuation is due to two components of diffusion: the slow component associated with water molecule diffusion and the fast component associated with bulk motion of intravascular molecules in the capillary network. The intravoxel incoherent motion (IVIM) model4 represents this two-component signal decay with a bi-exponential function that has two decay rate parameters, one for the slow (D) and one for the fast diffusion (D*) and a parameter representing the fraction of fast diffusion (f). However, in biological systems, fast diffusion due to microcirculation can occur over a large range of length and time scales due to widely varying vessel sizes and flow rates, and the single decay rate parameter for fast diffusion in IVIM model cannot represent this heterogeneity. Therefore, we introduce a probability distribution model of incoherent motion (PIM), a two-component mixture model of diffusion. Another limitation of IVIM parameter estimation is that fitting the IVIM model to the signal measured at a single voxel without utilizing information from neighboring voxels results in parameter maps, especially of fast diffusion, that are too noisy and that have high uncertainty, which becomes a hindrance for their clinical utility. We therefore introduce a spatially constrained PIM model (SPIM) that uses a spatially smoothing prior and estimates parameters for all voxels simultaneously, rather than solving for parameters of each voxel independently. The SPIM model allows for the precise characterization of the diffusion components even from relatively low SNR DW-MRI images. In this work, we introduce the SPIM model and compare it with the IVIM model in 68 abdominal DW-MRI data from five healthy volunteers and 24 Crohn’s disease patients.

Methods: We introduce a probability distribution model of the signal (p(D)) of Gamma probability density function (pdf) (p(D)) for fast diffusion and the other for slow diffusion: p(D) = f(D)p(D) + (1-f)q(D). The exponentially decaying signal for a voxel measured at b-value i can be formulated as the integral of this pdf of diffusion within that voxel: s_i = s_i exp(-b_i D). The integral in this equation is the Laplace transform of p(D). Using the analytical expression of the Laplace transform of Gamma distribution for each Gamma in p(D), the following model of the signal can be obtained:

s_i = s_i exp(-k_i log(1 + b_i D)) + (1 - f) exp(-k_i log(1 + b_i D))

The parameters (k_i, f, k_f, k_s, k_I) at each voxel can be estimated by solving a maximum-likelihood estimation problem. However, due to low SNR of the measured DWI signals, instead of solving the least-squares optimization for each voxel independently, we integrate the spatial homogeneity prior model proposed by Freiman et al. into our formulation. We estimate the parameters of the spatially constrained PIM (SPIM) model by maximizing the posterior distribution given by the product of likelihood and prior terms. This optimization problem can be formulated as a continuous Markov Random Fields problem where the spatial homogeneity prior is defined as the L1 norm of the difference between parameters of neighboring voxels. To efficiently estimate the model parameters, we used the “fusion bootstrap moves” solver.

DW-MRI data was acquired with a 1.5T scanner (Magnetom Avanto, Siemens Medical Solutions) using free-breathing single-shot echo-planar imaging with parameters: repetition/echo time (TR/TE) = 7500/77ms; matrix size = 192x156; field of view = 300x260mm; slice thickness/gap = 5mm/0mm; 40 axial slices; 7 b-values=0,50,100,200,400,600,800 s/mm2 with 1 excitation; acquisition time = 5.5min. Five healthy volunteers scanned [6,4,4,3,3] times respectively. Clinical DW-MRIs including liver fibrosis, tumors and active inflammation1-3. DW-MRI enables characterization of the tissue microstructure through measurement of the mobility of water molecules and the variations in their mobility due to changes in cell density, vascularity and membrane permeability. When imaging tissues with DW-MRI, mobility of molecules leads to attenuation in the diffusion signal, which is measured at multiple b-values. This attenuation is due to two components of diffusion: the slow component associated with water molecule diffusion and the fast component associated with bulk motion of intravascular molecules in the capillary network. The intravoxel incoherent motion (IVIM) model4 represents this two-component signal decay with a bi-exponential function that has two decay rate parameters, one for the slow (D) and one for the fast diffusion (D*) and a parameter representing the fraction of fast diffusion (f). However, in biological systems, fast diffusion due to microcirculation can occur over a large range of length and time scales due to widely varying vessel sizes and flow rates, and the single decay rate parameter for fast diffusion in IVIM model cannot represent this heterogeneity. Therefore, we introduce a probability distribution model of incoherent motion (PIM), a two-component mixture model of diffusion. Another limitation of IVIM parameter estimation is that fitting the IVIM model to the signal measured at a single voxel without utilizing information from neighboring voxels results in parameter maps, especially of fast diffusion, that are too noisy and that have high uncertainty, which becomes a hindrance for their clinical utility. We therefore introduce a spatially constrained PIM model (SPIM) that uses a spatially smoothing prior and estimates parameters for all voxels simultaneously, rather than solving for parameters of each voxel independently. The SPIM model allows for the precise characterization of the diffusion components even from relatively low SNR DW-MRI images. In this work, we introduce the SPIM model and compare it with the IVIM model in 68 abdominal DW-MRI data from five healthy volunteers and 24 Crohn’s disease patients.

Results: Fig. 3 shows the bar plot of mean CV% with error bars showing the variations. When the SPIM model is used instead of IVIM, the CV% in liver was reduced from 20% to 6% for D* and 70% to 24% for f parameter. The CV% in spleen was reduced from 17% to 5% for D* and 17% to 5% for f parameter. Statistical analysis with paired t-test indicated that the CV% was significantly lower (p<0.0001) for SPIM compared to both SCIM and IVIM.

Discussion and Conclusion: Results showed that the CV% of the diffusion parameters estimates obtained using the SPIM model were significantly lower in both the liver and spleen regions compared to the IVIM and SCIM models. D* had the largest CV% among the three parameters. The SPIM model also improved CV% compared to SCIM, which might be related to using a probabilistic model of diffusion that better represents heterogeneity in fast diffusion. Using the SPIM model, the robustness of parameter estimates was increased without acquiring additional data, while utilizing a more realistic model of heterogeneous diffusion components.