

Online ODF Reconstruction in Constant Solid Angle and its Application to Motion Detection in HARDI



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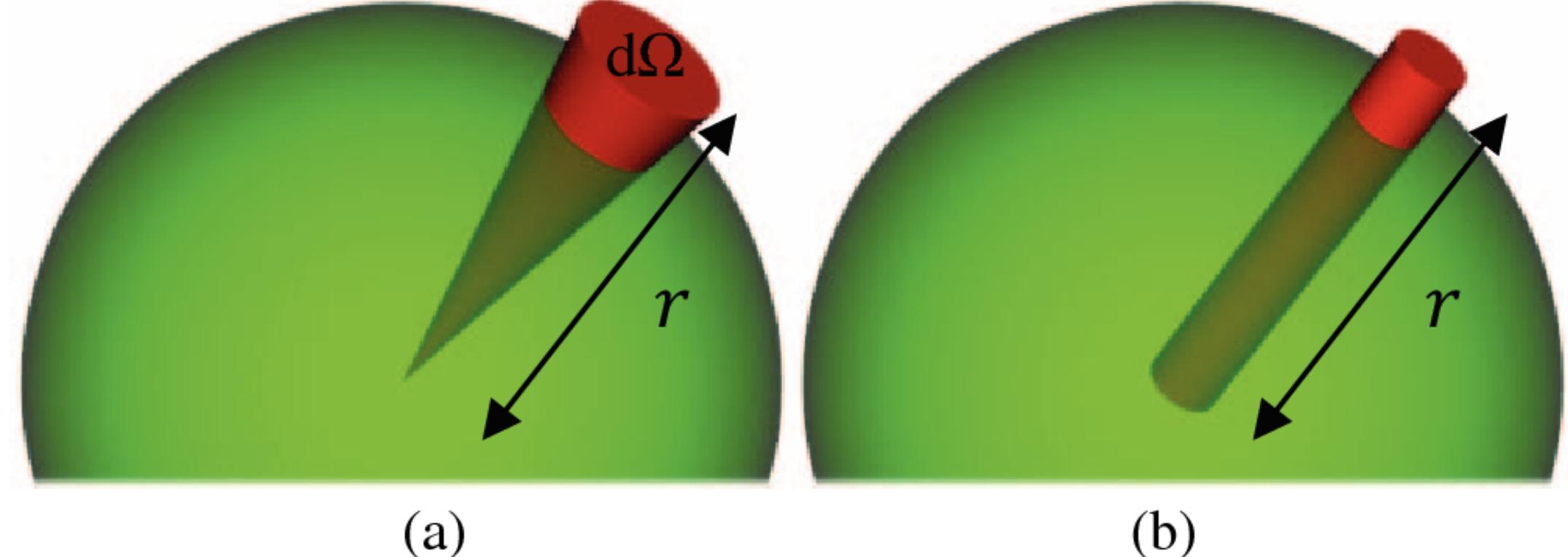
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The diffusion orientation distribution function (ODF) can be reconstructed from q-ball imaging (QBI) to map the complex intravoxel structure of water diffusion. As acquisition time is particularly large for high angular resolution diffusion imaging (HARDI), fast estimation algorithms have recently been proposed, as an online feedback on the reconstruction accuracy. Thus the acquisition could be stopped or continued on demand. We adapt these real-time algorithms to the mathematically correct definition of ODF in constant solid angle (CSA), and develop a motion detection algorithm upon this reconstruction. Results of improved fiber crossing detection by CSA ODF are shown, and motion detection was implemented and tested in vivo.

ODF in Constant Solid Angle

- By considering the Jacobian factor, we can define the OD as the actual marginal PDF of diffusion in a given direction in constant solid angle ([Aganj et al., MRM 2009], [Tristan-Vega et al., Neuroimage 2009]).



$$\begin{aligned} \text{ODF}_{\text{CSA}}(\hat{u}) &:= \int_0^\infty P(r\hat{u})r^2 dr \\ &\approx \frac{1}{4\pi} + \frac{1}{16\pi^2} \text{FRT} \left\{ \nabla_b^2 \ln(-\ln E(\hat{u})) \right\} \end{aligned}$$

Motion Detection

- There is a unique Kalman filter in each voxel
- After a few iterations, the Kalman filter can accurately predict the next signal outcome: the prediction error $\epsilon_k = s_k - \mathbf{X}_k \hat{\mathbf{c}}_{k-1}$ is expected to decrease.
- However in case of patient motion, pixels will no longer match and this will result in an abnormal increase of the prediction error over the whole volume.

Reconstruction with Kalman Filter

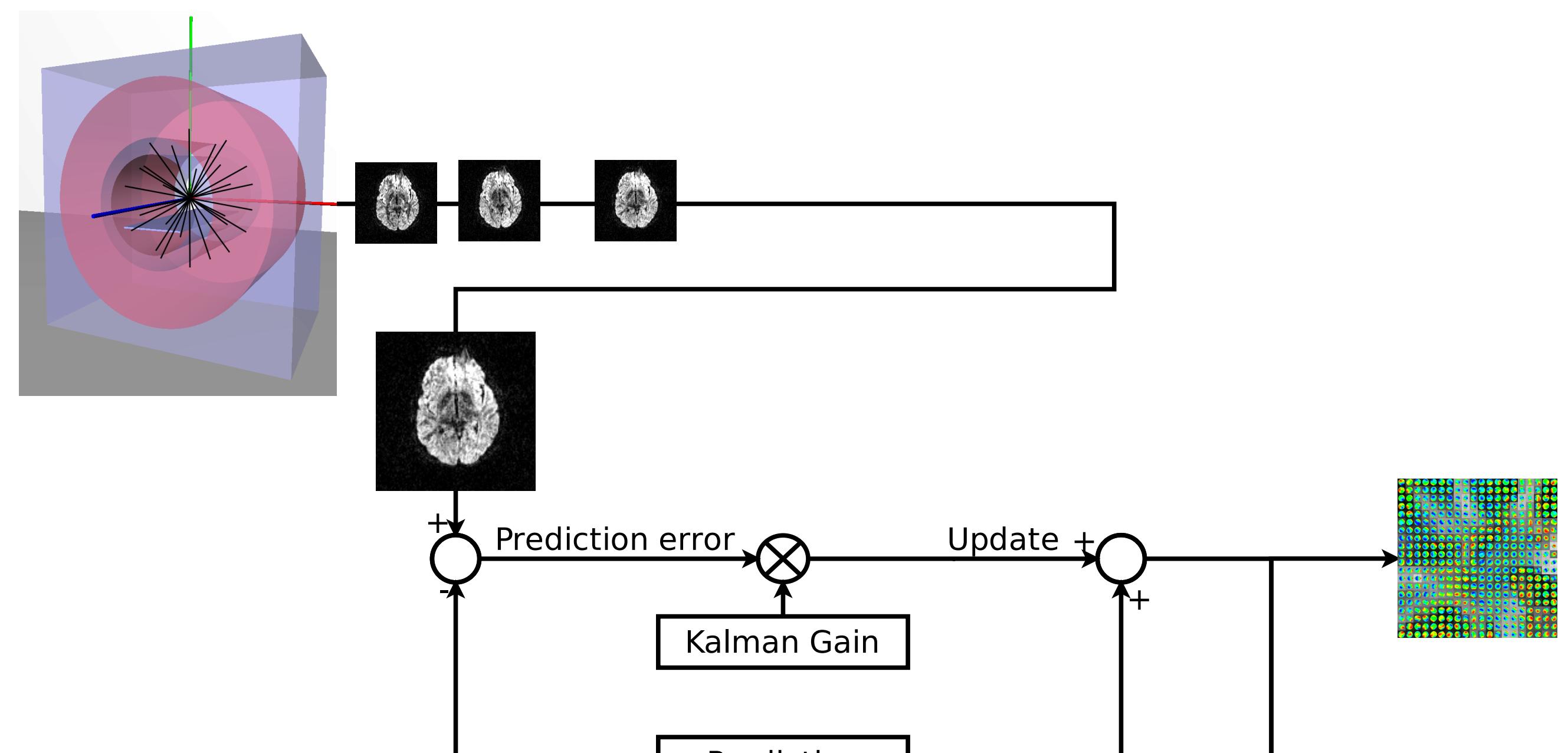
- The double logarithm of the signal is represented in the Spherical Harmonics basis, with a Laplace-Beltrami regularization.

$$s(\hat{u}) = \ln(-\ln E(\hat{u})) \approx \sum_{j=1}^R \hat{c}_j Y_j(\hat{u})$$

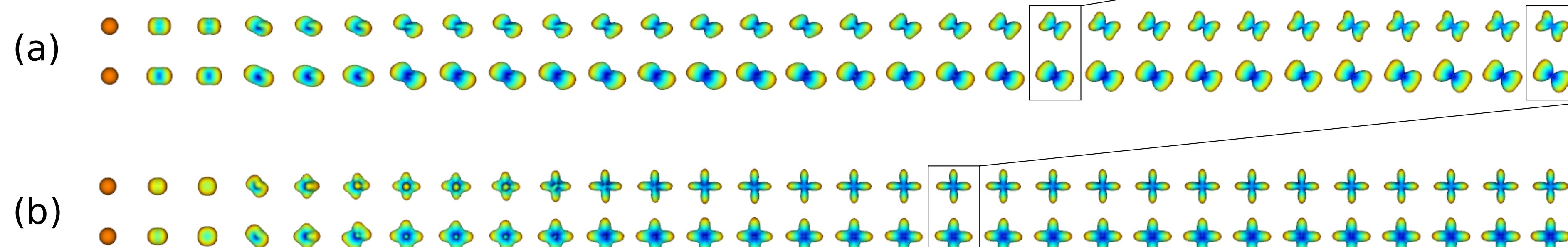
- The Kalman filter solution actually minimizes the functional [Deriche et al., Med. Image Anal. 2009]:

$$\hat{\mathbf{c}}_k = \arg \min_{\mathbf{c}} (\mathbf{s}_k - \mathbf{X}_k \mathbf{c})^T \Sigma_k (\mathbf{s}_k - \mathbf{X}_k \mathbf{c})$$

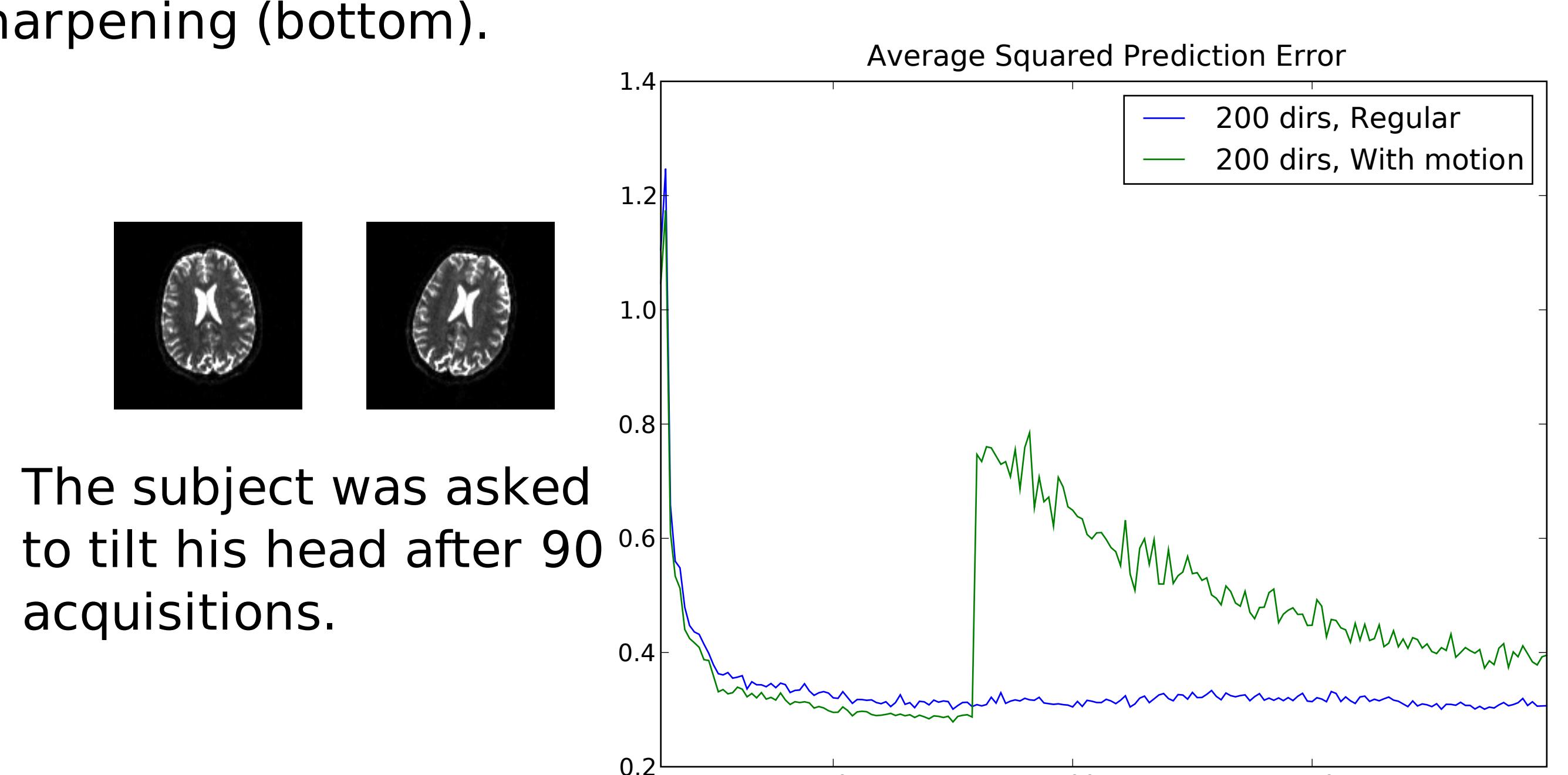
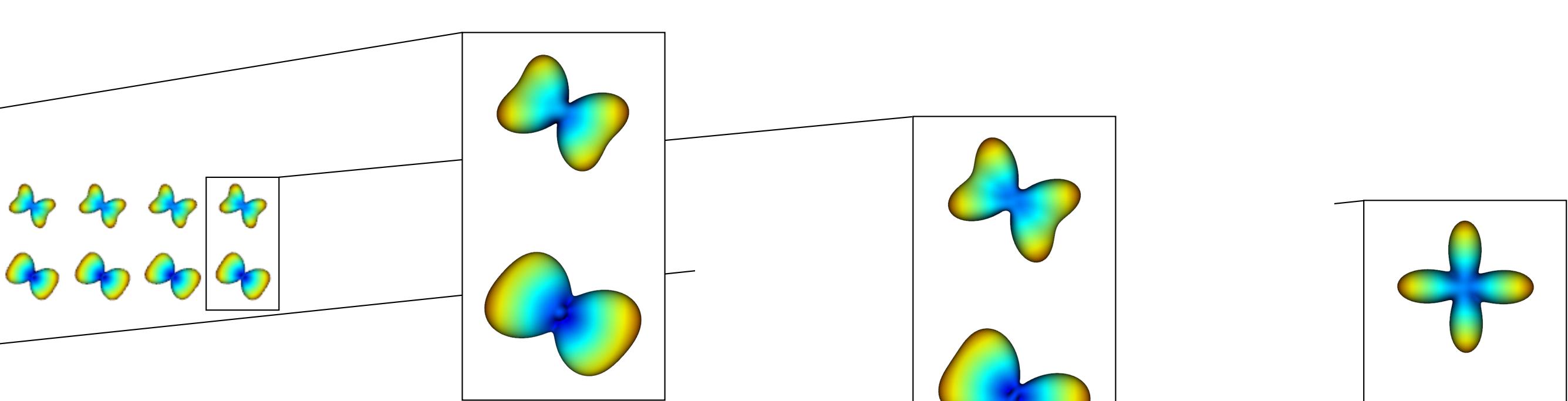
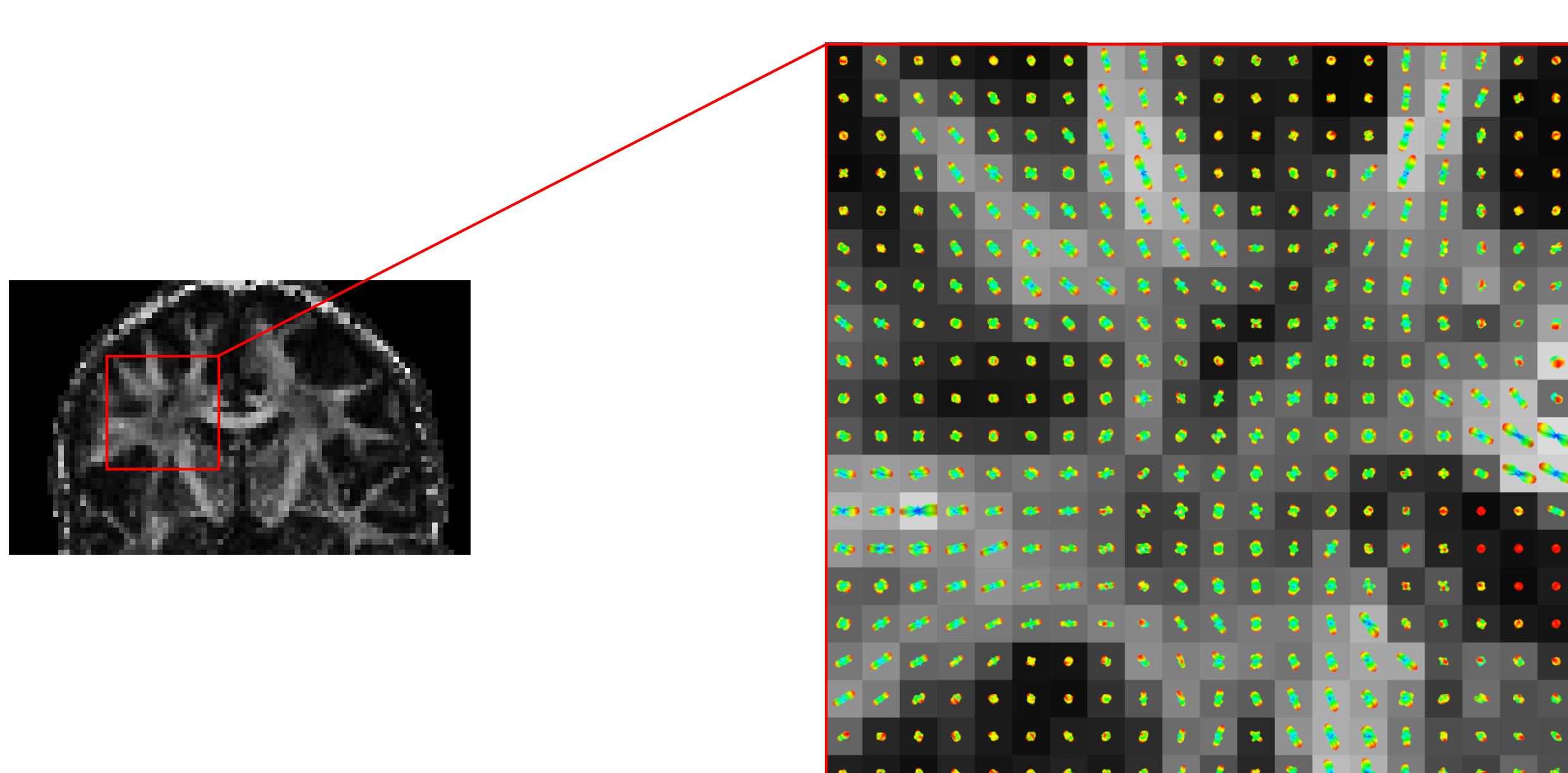
$$\begin{cases} \hat{\mathbf{c}}_0 &= \mathbb{E}[\mathbf{c}_0] \\ \tilde{\mathbf{P}}_0 &= \mathbb{E}[(\mathbf{c} - \hat{\mathbf{c}}_0)(\mathbf{c} - \hat{\mathbf{c}}_0)^T] \\ \mathbf{P}_0 &= (\tilde{\mathbf{P}}_0^{-1} + \lambda \mathbf{L})^{-1} \\ \mathbf{g}_k &= \mathbf{P}_{k-1} \mathbf{X}_k^T (\mathbf{X}_k \mathbf{P}_{k-1} \mathbf{X}_k^T + \sigma_k^2)^{-1} \\ \mathbf{P}_k &= (\mathbf{I} - \mathbf{g}_k \mathbf{X}_k) \mathbf{P}_{k-1} \\ \hat{\mathbf{c}}_k &= \hat{\mathbf{c}}_{k-1} + \mathbf{g}_k (s_k - \mathbf{X}_k \hat{\mathbf{c}}_{k-1}) \end{cases}$$



Experiments and Results



Synthetic dataset: Incrementally reconstructed ODFs from simulated fiber crossing for an angle of (a) 60 and (b) 90°, for iteration 1 to 30. Compared CSA ODF (top) and original ODF with Laplace-Beltrami sharpening (bottom).



Diffusion images of a healthy subject, acquired at the Center for Magnetic Resonance Research, University of Minnesota, on a 3T Siemens magnet. 200 encoding directions, computed following the optimal sampling scheme of [Deriche et al., Med. Image Anal. 2009], $b = 1000 \text{ s/mm}^2$, 64 slices with 2mm thickness, 25 $b = 0$ images, 128×128 image matrix, TE=90ms and TR=8500ms.

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