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# Regularizing the ODF estimate with the Laplace-Beltrami operator in 3D Polarized Light Imaging

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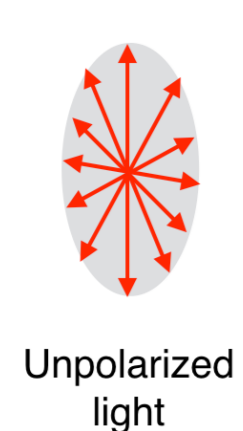
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<http://team.inria.fr/athena/>

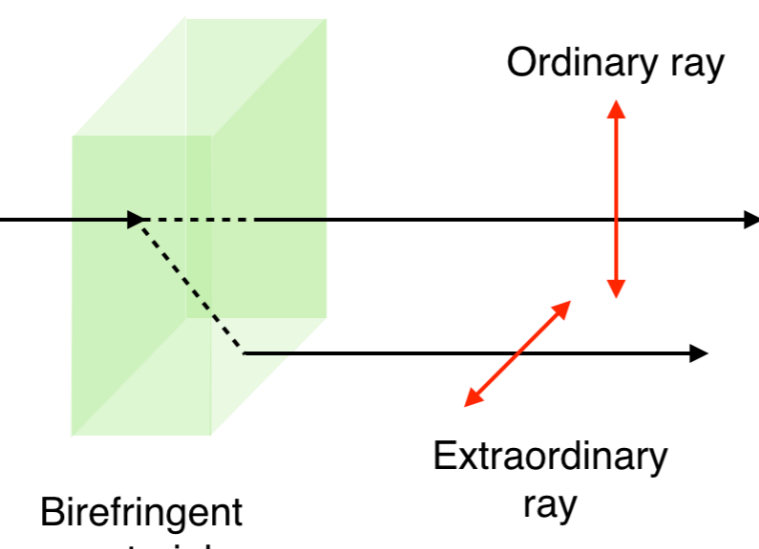
**Abstract:** 3D-PLI is an optical imaging approach that utilizes the birefringence in postmortem tissue to map its spatial fiber structure at a submillimeter resolution [1]. Recently Axer et al. proposed an estimate of the fiber orientation distribution functions (pliODF) from high-resolution PLI vector data. However, this technique introduces angular errors which, in turn, constrains the spherical harmonics (SH) expansion order. Here, we overcome this limitation by making use of the Laplace-Beltrami operator to analytically regularize the SH coefficients which define the fiber ODF at higher orders.

## 1 3D Polarized Light Imaging

### Polarization



### Birefringence



### Image acquisition and analysis

Linearly polarized light passes through the tissue section and allows to detect its birefringence. A set of intensity images is analysed through Fourier series to get fiber orientation  $(\alpha, \phi)$  in each voxel.

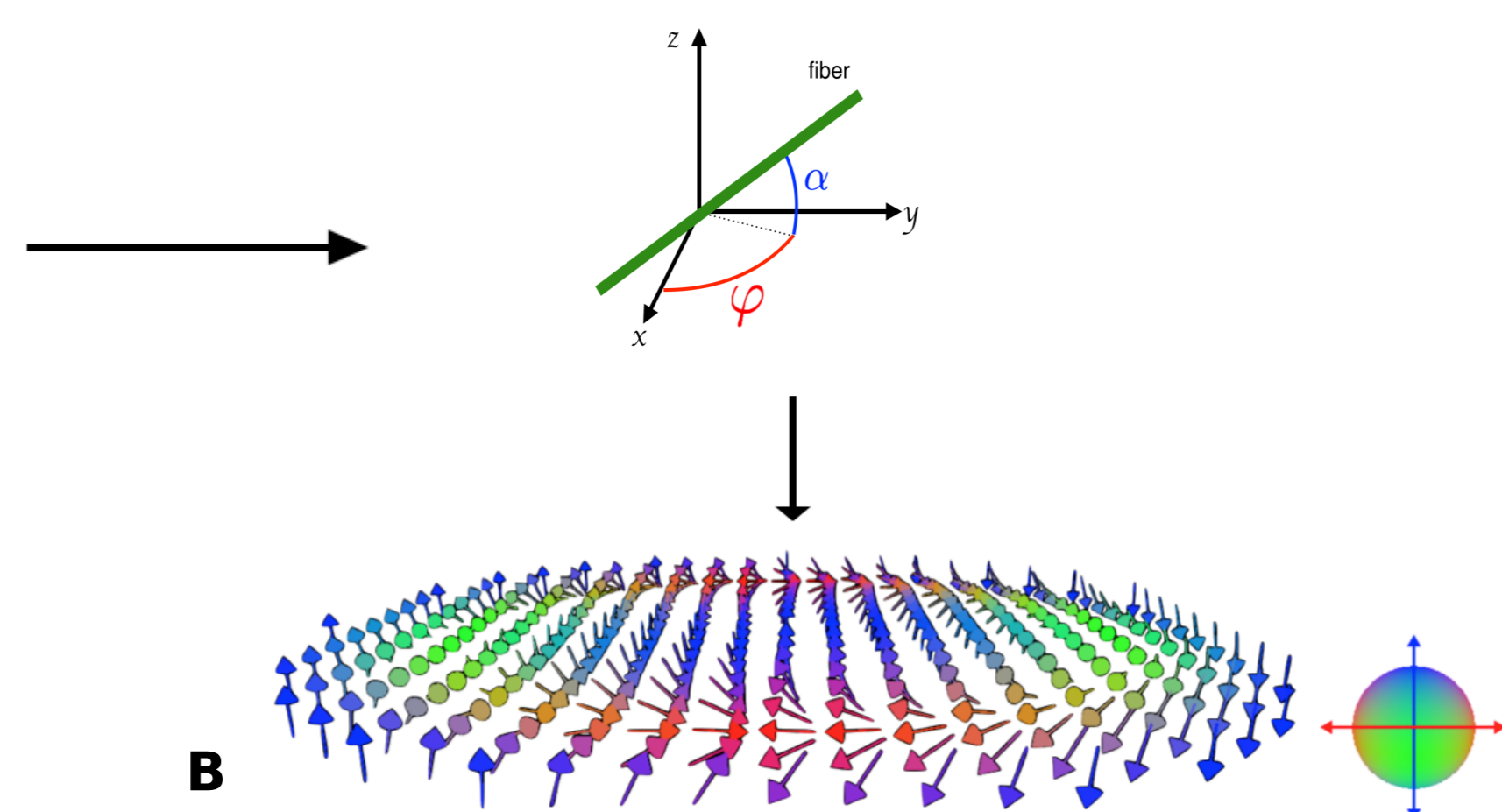
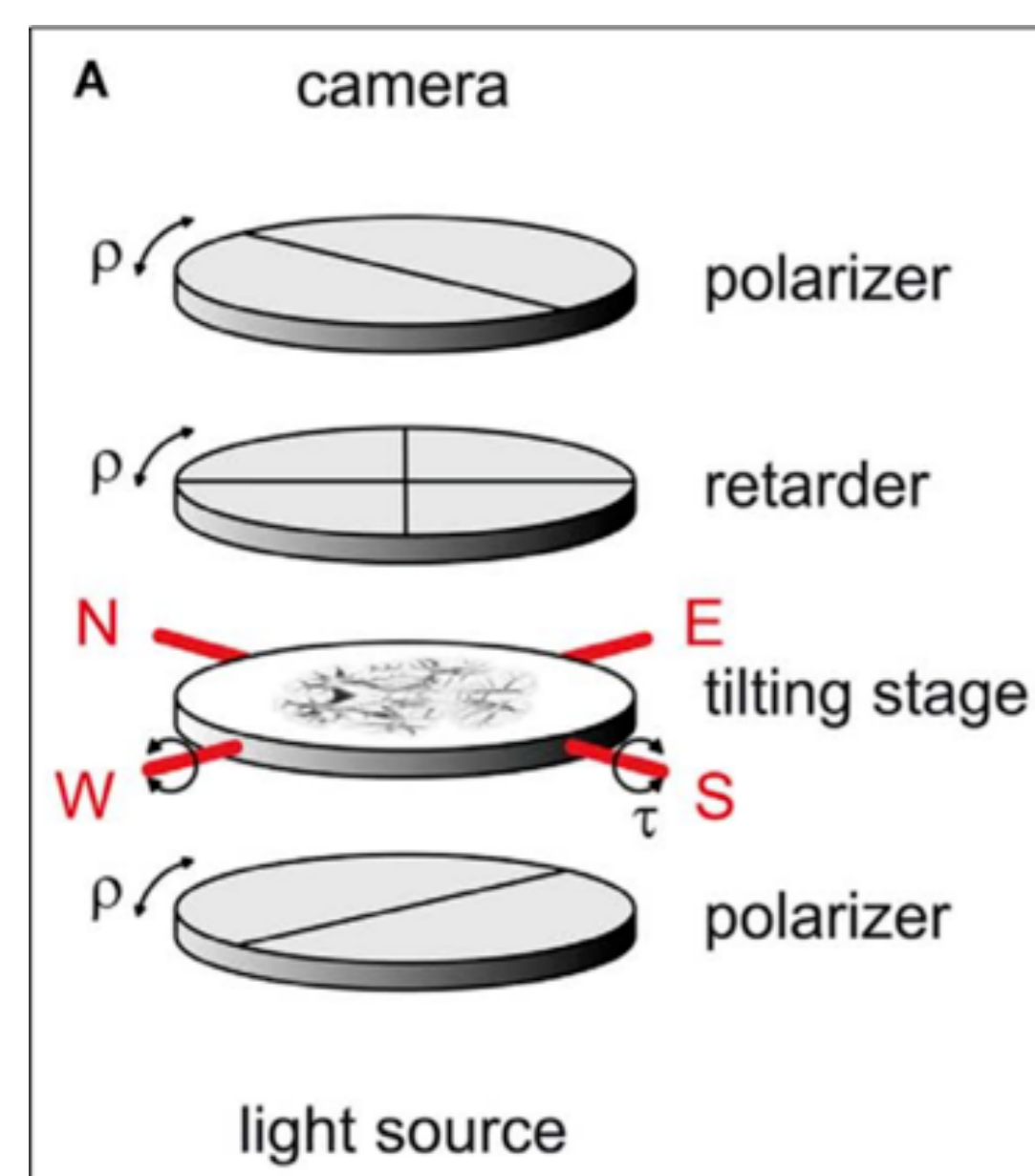


Image acquisition setup with tilting stage and different optic filters (A), from [1]. Each unstained histological section is characterized by a 3D vector field (B) visualized as a Fiber Orientation Map (FOM) at sub-millimeter resolution. The FOM is adapted from [2].

## 2 ODF reconstruction

### pliODF [3]:

1. super-voxel for down-sampling high resolution FOM
2. normalized directional histogram (DH) on unit sphere
3. spherical harmonics expansion to approximate each DH

$$h(\theta, \phi) = \sum_{l=0}^{L_{max}} \sum_{m=-l}^l c_{lm} Y_l^m(\theta, \phi)$$

- + bridging spatial scales: micro to macro
- low odf angular resolution

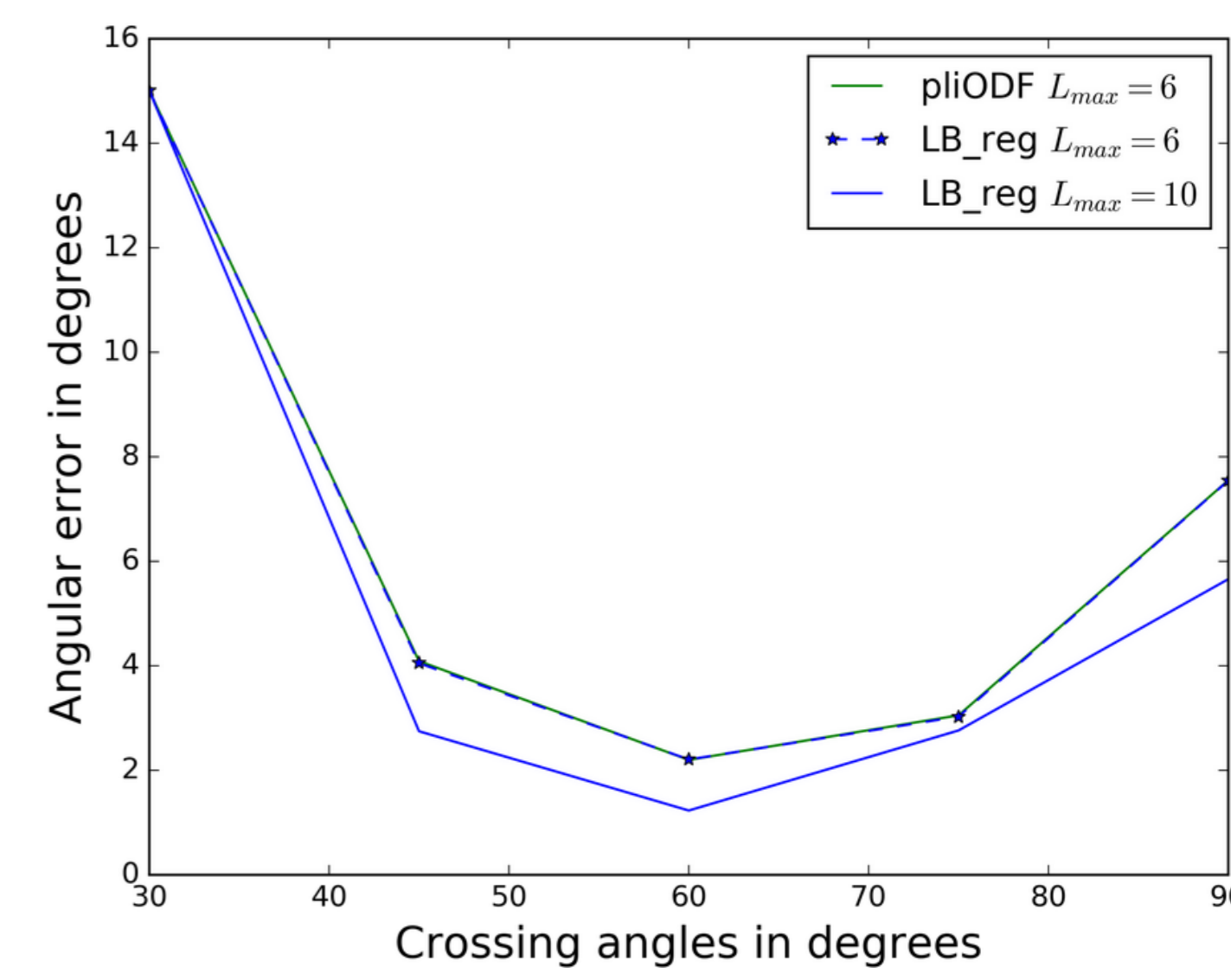
### Laplace-Beltrami regularization:

$$\Delta_b Y_l^m = -l(l+1)Y_l^m$$

The SH bandlimit  $L_{max}$  is increased to sharpen the fiber ODF [4] and the regularized estimation of SH coefficients analytically writes

$$C = (B^T B + \lambda L)^{-1} B^T H$$

- + bridging spatial scales
- + higher angular resolution
- + analytical solution

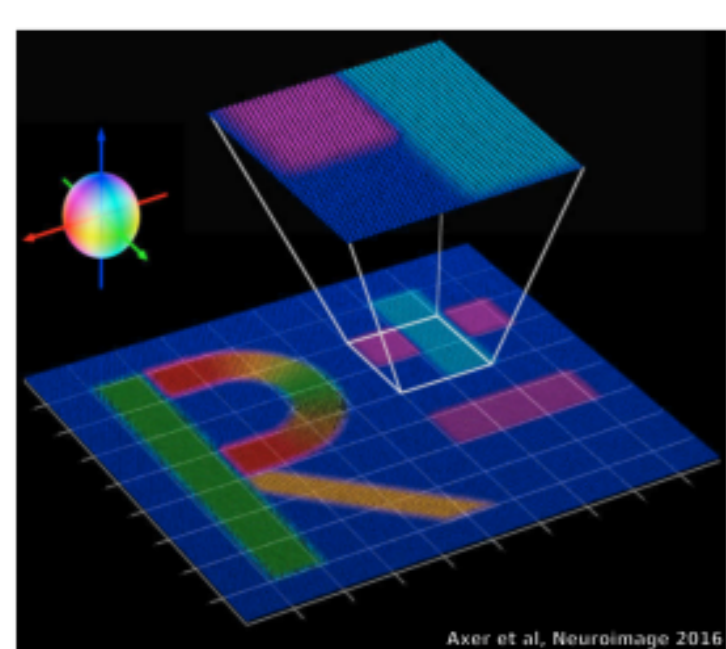


At  $L_{max} = 10$  our regularized ODF produces less angular error than pliODF (error above  $40^\circ$  due to ill-conditionned SH basis matrix, not plotted)

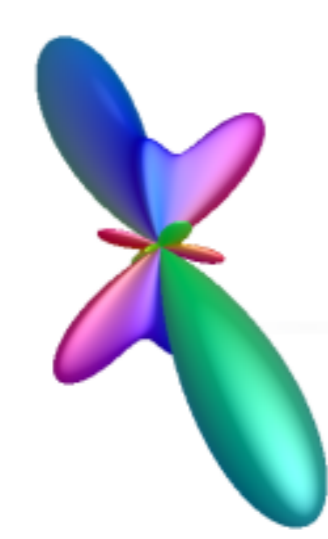
## 3 Synthetic experiment

The performance of our approach is evaluated on the simulated dataset [3]. We choose a super-voxel (zoomed compartment) composed of  $40 \times 40 \times 1$  native voxels. 25% (in blue) oriented in  $(\theta, \phi) = (90^\circ, 0^\circ)$ , 25% (magenta) in  $(-45^\circ, 0^\circ)$  and 50% (cyan) in  $(45^\circ, 90^\circ)$ .

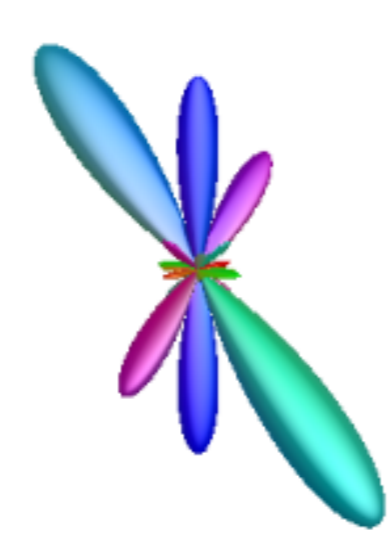
- Fiber ODF accurately reconstructed
- Angular resolution improved
- Volume fraction correctly estimated



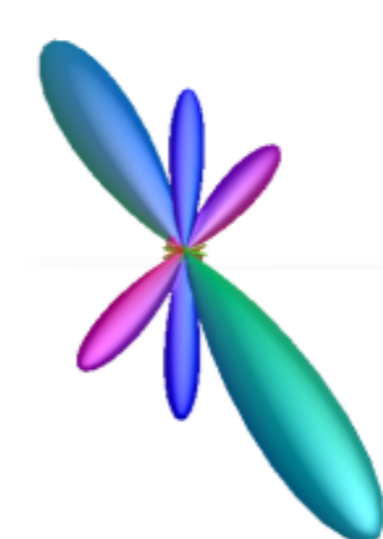
(a)



(b)



(c)



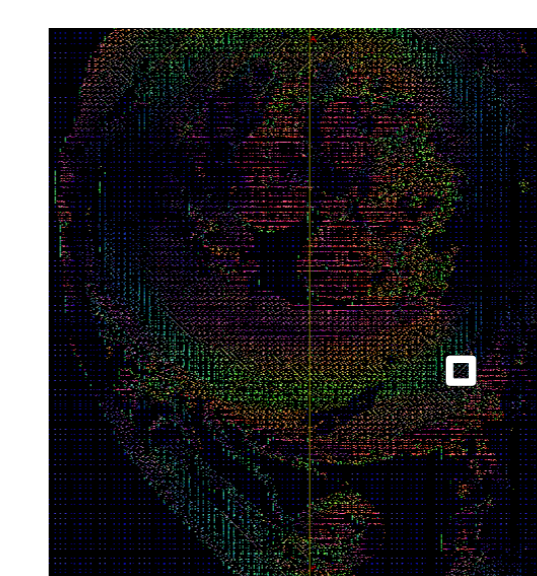
(d)

Fiber ODF generation: (a) simulated super-voxel from [3]. (b) pliODF at SH  $L_{max}=6$ , (c) and (d) Laplace-Beltrami regularized fiber odf at  $L_{max}=8$  and  $10$ , respectively. The three prevailing fiber orientation populations are preserved.

## 4 Ex-vivo human heart Data

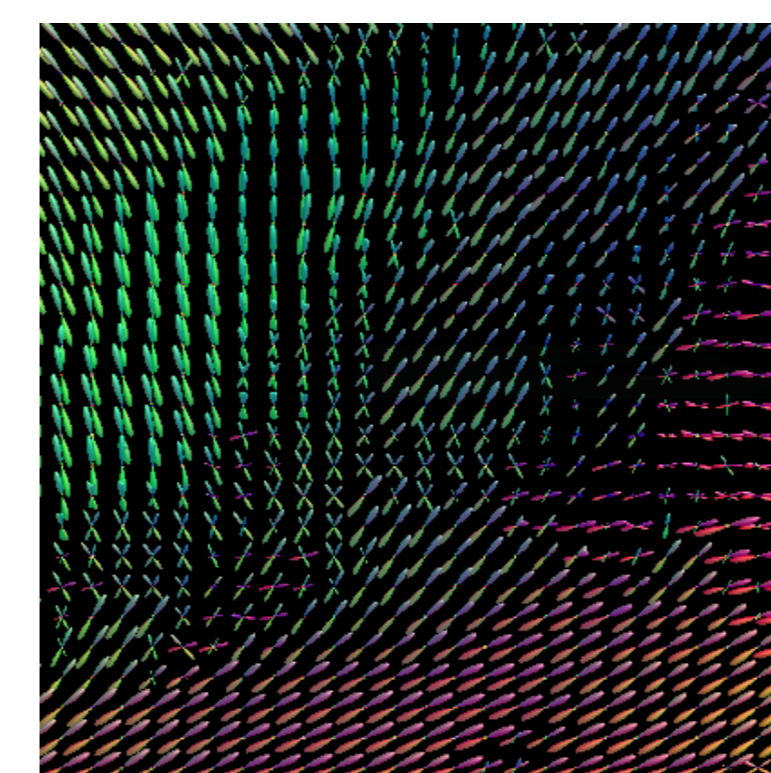


(A)

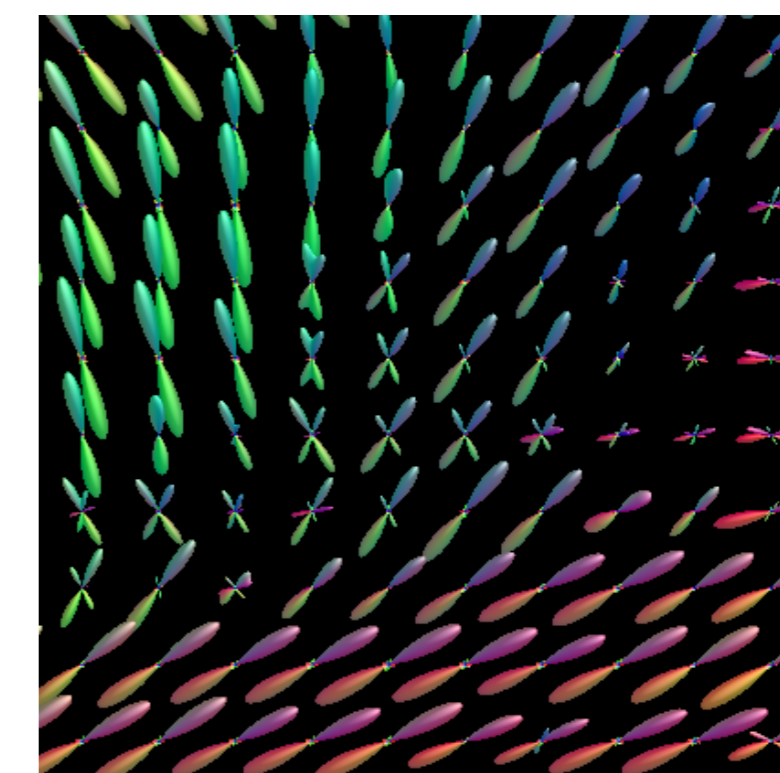


(B)

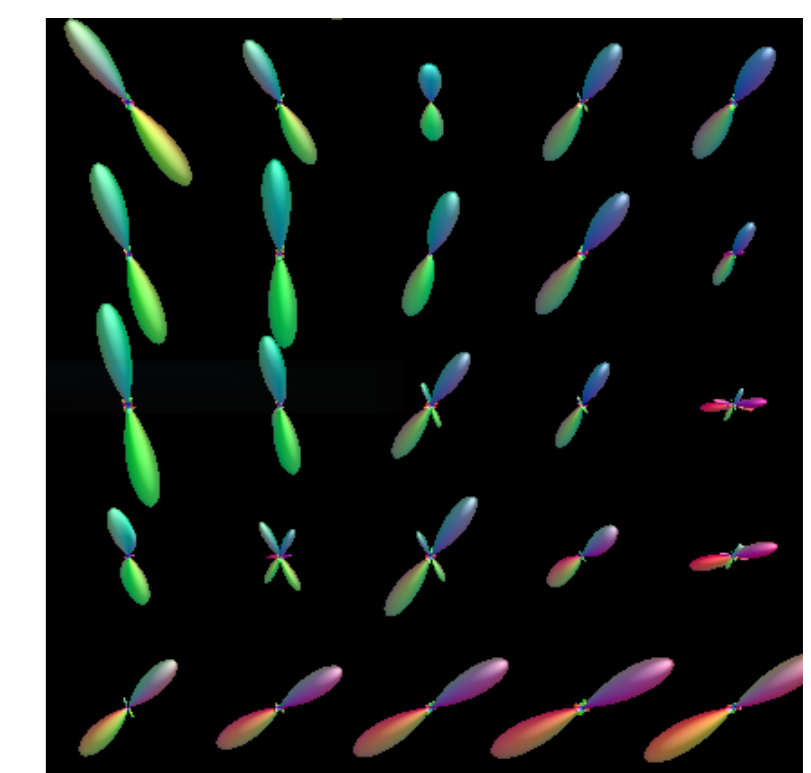
Organ is frozen, sliced and imaged with no staining. Each FOM consists of high-resolution data of  $440 \times 620$  voxels.



(C)



(D)



(E)

Coronal ventricular slice (A) with correspondng high-resolution FOM (B). Fiber ODFs from the white rectangular RIO are reconstructed at different spatial scales defined by the super-voxel size: (C)  $2 \times 2 \times 2$ , (D)  $5 \times 5 \times 2$  and (E)  $10 \times 10 \times 2$  native voxels.

## 5 Discussion and Conclusions

- The Laplace-Beltrami regularization provides an analytical solution for the SH coefficients which define the fiber ODF.
- Compared to pliODF, our regularized ODF enhances the underlying angular resolution and considerably improves the extraction of fibers, eventually for tractography.

• This method is, however, still limited by the arbitrary discretization of the directional histogram.

• The concept of super-voxel allows data integration across scales: voxel-level comparison with diffusion MRI should now be possible.

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## References

[1] Axer M et al, Neuroimage, 2011. [2] Dohmen M, PhD thesis, 2014. [3] Axer M et al, Frontiers in neuroanatomy, 2016. [4] Descoteaux M et al, Magnetic Resonance in Medicine, 2007.