PERFUSION FRACTION TENSOR IMAGING OF THE KIDNEY

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Target Audience: Radiologists and scientists who are interested in DTI, IVIM or diffusion imaging in general.

Purpose:
Intravoxel incoherent motion (IVIM) is a well-known model for determination of diffusion and perfusion parameters from diffusion weighted MR images. The standard IVIM-model implicitly assumes isotropic diffusion and perfusion properties in the examined tissue. However, diffusion tensor imaging (DTI) reveals anisotropic diffusion in human kidneys[1]. Other research disclosed that IVIM perfusion parameters in kidneys are also anisotropic[2]. Purpose of the presented study was to propose a perfusion fraction tensor for the combination of IVIM-DTI and to compare the stated model to a combination of IVIM-DTI assuming isotropic perfusion in kidneys.

Methods:
Perfusion fraction tensor
We suggest describing the perfusion fraction f as a tensor that has its primary axis aligned parallel to the primary axis of the diffusion tensor. f_a denotes the axial eigenvalue, i.e. the eigenvalue corresponding to the primary axis. Assuming circular symmetry the perfusion fraction tensor has a uniform radial eigenvalue f_r orthogonal to the principal axis. Hence, the apparent perfusion fraction of a measurement is $f(\alpha) = f_a \cos^2(\alpha) + f_r \sin^2(\alpha)$; where α denotes the angle between principal diffusion axis and applied diffusion gradient. Combination of IVIM and DTI with the perfusion fraction tensor yields the signal at high b-values as: $S = S_0 (1\text{-}f(\alpha)) e^{-b\,\mathbf{D}}$; where S_0 is the signal without diffusion weighting and **D** is the diffusion tensor. In this equation f(α) simplifies to a constant f, when isotropic perfusion is assumed. We chose Akaike's information criterion for small sample size[3] (AICc) for comparison of the isotropic perfusion fraction model with the tensor model.

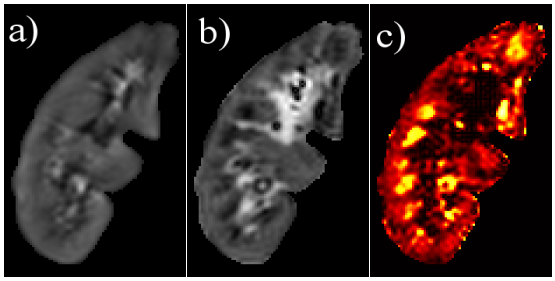

Figure 1: a) Intensity image without diffusion weighting (b=0). **b)** Isotropic perfusion fraction f. **c)** Fractional Anisotropy FA of perfusion tensor.

	Medulla	Cortex
$f_{isotropic}$	(11.2 ± 4.8) %	(24.2 ± 4.0) %
FA(f)	0.33 ± 0.08	0.09 ± 0.02

Table 1: Isotropic perfusion fraction f and FA of perfusion tensor in the anisotropic model.

The study was approved by our institute's ethics committee. Six healthy volunteers have been examined on a PRISMA 3T-scanner (Siemens Healthcare Sector, Erlangen, Germany). Imaging of coronal slices was performed with a bipolar diffusion weighted spin-echo Echo-Planar-Imaging sequence using the following parameters: TR 3.000ms; TE 82 ms; FOV 400 x 400 mm²; matrix 196 x 196 with 5/8 phase partial Fourier; slice thickness 6 mm; seven b-values (0, 200, 250, 700, 750, 780, 800 s/mm²); 30 diffusion directions each acquired twice with opposite gradient signs. All acquisitions were ECG-triggered during free breathing. Image registration was performed to compensate for respiratory motion[4]. Diffusion and perfusion parameters were calculated in every voxel in two steps: First, diffusion tensor **D** and a scalar perfusion fraction f were fitted. Second, the angle α between the applied diffusion gradient and the primary axis of the diffusion tensor was calculated for every voxel. Then, f_a, f_r and again **D** were fitted. ΔAICc was calculated as the difference of AICc in the isotropic and the perfusion fraction tensor model. Finally the fractional anisotropy of the perfusion tensor FA(f) was determined. Regions of interest (ROI) were drawn manually for cortex and medulla.

Results:
From all volunteers a total of 472 voxels for medulla and 471 voxels for cortex have been segmented. Table 2 shows the percentage of voxels where the perfusion fraction tensor model is superior (ΔAICc <-2), inferior (2< ΔAICc) or equivalent (-2< ΔAICc <2) to the isotropic model. Image quality has a severe impact on which model is preferred by AICc. In maps of one volunteer with excellent image quality more than 95% of all voxels in the entire kidney showed superiority of the anisotropic perfusion fraction model. FA(f) is significantly higher in the medulla than in the cortex (Table 1 and Figure 1). Furthermore, 62% of medullary voxels and 47% of cortical voxels are better described by the tensor model.

	Medulla	Cortex
ΔAICc < -2	62 %	47 %
-2 < ΔAICc < 2	32 %	42 %
2 < ΔAICc	6 %	11 %

Table 2: ΔAICc <-2 indicates that the proposed tensor model is superior to the isotropic model. ΔAICc >2 indicates that the tensor model is inferior. In between, neither model is clearly preferred.

Discussion:
An anisotropic diffusion tensor **D** indicates a directionally dependent velocity of moving molecules. In contrast, an anisotropic perfusion fraction tensor **f** can be interpreted as volume of blood vessels and tubules which are orientated in a certain direction. The higher FA(f), the more vessels and tubules are aligned parallel to the major direction of the diffusion tensor. Predominance of the **f** tensor model is more pronounced in the medulla where FA(f) is higher than in cortex. Our results show that the suggested perfusion fraction tensor model is equivalent or superior to an isotropic perfusion model in 94% of all voxels in medulla and 89% in cortex.

Conclusion:
We present a new mathematical model to describe anisotropic perfusion fraction. AICc values confirm that this model is beneficial in describing perfusion in kidney tissue. The degree of perfusion fraction anisotropy is higher in renal medulla than in renal cortex. The model may also be applied to other tissues with structured microperfusion, such as skeletal muscles or the heart.

References: **1.** Ries M et al. Diffusion tensor MRI of the human kidney, JMRI 2001; 14(1):42-49, **2.** Notohamiprodjo M et al. Combined IVIM and DTI for simultaneous assessment of diffusion and flow anisotropy of the kidney, Proc. ISMRM 2012. **3.** Burnham K. And David R. Anderson. 2002. Model Selection and Multimodel Inference: A Practical Information-Theoretical Approach. 2nd ed. New York: Springer-Verlag. **4.** Myronenko et al. Intensity-based image registration by minimizing residual complexity, IEEE T MED IMAGING 2010; 29(10):1882-1891