Investigations on the Non-Mono Exponential Signal Decay in Presence of a Single Vessel; Simulations, Phantom and *in vivo* Measurements

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Introduction: The BOLD effect [1] is the underlying principle of fMRI [2] and susceptibility weighted imaging (SWI) [3]. The influence of single cylindrical blood vessels or a capillary network on signal formation has been theoretically described [4]. It has been verified in phantom studies for a capillary network [5] and was used *in vivo* to estimate tissue oxygenation fractions (OEF) [6]. However, the single vessel model has not been investigated systematically in phantom measurements nor with multi-echo information to estimate the blood oxygenation level in small veins *in vivo*. The aim of this work was to simulate the signal decay of voxels traversed by a single vessel and to verify it experimentally by using a multi-echo gradient-echo sequence.

Theory: The analytical solution describing the voxel signal in the presence of a single vessel is given in Eq. 1, where λ denotes the volume fraction of the vessel within the voxel. This solution is only valid for cylindrical voxel shapes [4]. The parameter η describes the extravascular signal loss due to spin dephasing in the local field inhomogeneities generated by the vessel. The resulting local magnetic field is given in Eq. 2 and 3, where the discrete nature of the spins is taken into account by the Sphere of Lorentz [7]. The difference in the magnetic susceptibility Δχ between the intra- and extravasular compartment is given in Eq. 4 with $\Delta\chi_{do}$=0.18 ppm [8]. Due to the influence on the external field of the glass tube used in the phantom measurements Eq. 3 has to be modified accordingly (see Eq. 5), where $\Delta\chi_{tube}$ denotes the magnetic susceptibility of the tube material, $\Delta\chi_{ext/int}$ the susceptibility of the external and internal compartment, $a_{int/out}$ the tube inner and outer radius and θ the orientation angle between the cylinder axis and B_0.

$$\vec{S}_{vox}(t) = \lambda \cdot S_{0,int} \cdot e^{-i\gamma\Delta B \cdot t/T_{2,int}} + (1-\lambda) \cdot S_{0,ext} \cdot e^{-t/T_{2,ext}} \cdot \eta \quad (1)$$

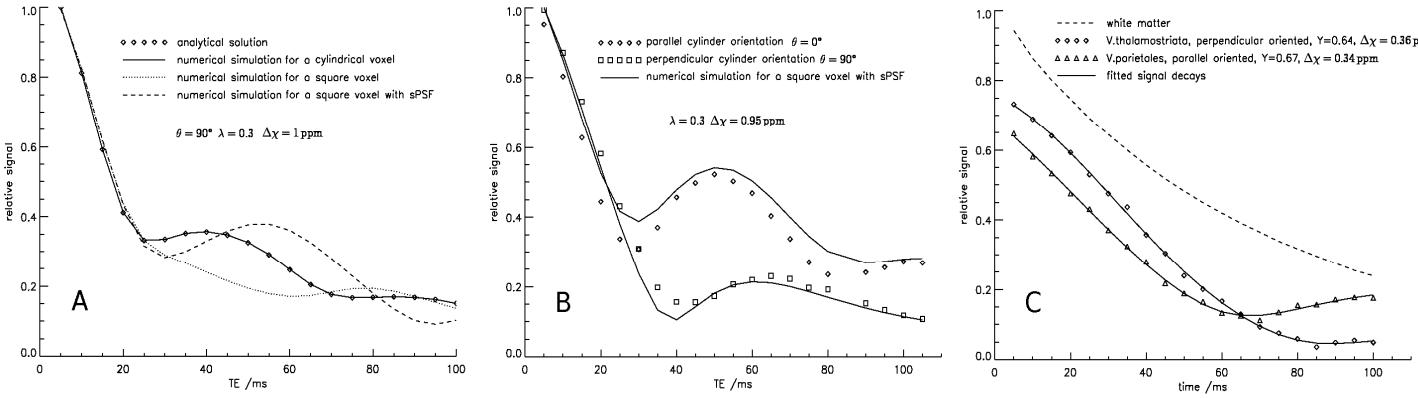
$$\vec{B}_{int} = \left\{1 + \frac{\chi_{ext}}{3} + \frac{\Delta\chi}{6} \cdot \left(3\cos^2\theta - 1\right)\right\} \cdot \vec{B}_0 \quad \forall\, r < a \quad (2)$$

$$\vec{B}_{ext} = \left\{1 + \frac{\chi_{ext}}{3} + \frac{\Delta\chi}{2} \cdot \sin^2\theta \cdot a^2 \frac{\cos 2\Phi}{r^2}\right\} \cdot \vec{B}_0 \quad \forall\, r > a \quad (3)$$

$$\Delta\chi = 4\pi \cdot \Delta\chi_{do} \cdot Hct \cdot (1-Y) \quad (4)$$

$$\vec{B}_{ext} = \left\{1 + \frac{\chi_{ext}}{3} + \frac{1}{2}\sin^2\theta \cdot \frac{\cos 2\Phi}{r^2}\left[a_{out}^2 \cdot (\chi_{tube} - \chi_{ext}) + a_{int}^2 (\chi_{int} - \chi_{tube})\right]\right\} \cdot \vec{B}_0 \quad \forall\, r > a_{out} \quad (5)$$

Methods: A 3D multi-echo gradient-echo (MEGE) sequence, velocity compensated in slice and readout direction was used at 1.5 T to obtain T_2* weighed images at echo times of TE = 5 - 100 ms in steps of ΔTE = 5 ms. The voxel size was varied in phantom measurements to obtain volume fractions of λ = 0.1, 0.2 and 0.3 for the intracapillary compartment. For the *in vivo* experiments the voxel size was about 1 x 1 x 1.5 mm³. The phantom consists of a pivotable glass tube (1 mm in diameter) inside a water bath. It was filled with an aqueous Gd-DTPA solution to adjust the magnetic susceptibility difference between the internal and external compartment. Numerical simulation based on the field (Eq. 2 and 5) was performed for different parameter settings and voxel shapes.



Results/Discussion: Figure **A** shows the simulated signal behavior of a vessel oriented perpendicular to B_0 for different voxel shapes. The analytical solution (Eq.1) is consistent with the numerical simulation of the cylindrical voxel, whereas a square shaped voxel generates a quite different signal decay. Due to the finite and discrete sampling, the shape of the sampling point spread function (sPSF) has to be considered, which changes the signal decay of the square voxel considerably. Including the sPSF into the simulation yielded good agreement between theory and experimental phantom data (Fig. **B**). *In vivo* measurements with lower Δχ and no vessel wall of a different material as in the phantom experiment demonstrate the ability of the analytical single vessel approach to obtain quantitative values of the blood oxygenation level Y (Fig. **C**), which are in good agreement with physiology and consistent for different vessel orientations.

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