Does Attenuation of Perfusion in Diffusion Weighted-MRI Behave as Exponential Decay or as Damped Oscillation?

Paul Martin Murphy[1], Nikolaus Szeverenyi[1], Claude Sirlin[1], and Mark Bydder[1]
[1]*Radiology, University of California - San Diego, San Diego, CA, United States*

Introduction: Diffusion-weighted MR imaging has been used to characterize diffuse and focal liver disease (1,2). Disease processes such as fibrosis and malignancy have been shown to affect diffusion parameters estimated from two models of signal intensity vs. b-value: the mono-exponential model, representing apparent diffusion, and the bi-exponential model (below left), reflecting the separate attenuation of perfusing spins. For isotropic, plug-like blood flow, early diffusion theory predicted a third model in which signal decays not as an exponential, but instead as a sinc function (below right) of the velocity of perfusing spins (3,4). This damped oscillation arises from odd-echo dephasing to an average phase that is either parallel or opposite to the average phase of diffusing spins. Such behavior of the diffusion-weighted signal was not empirically demonstrated in phantoms or in vivo in these studies, but, if present, would be important to recognize for accurate quantification of diffusion parameters. In this study, we demonstrate the existence of damped oscillation of the diffusion-weighted signal in a flow phantom and investigate whether the same effect can be observed in the liver of human volunteers.

$$\frac{S_b}{S_0} = e^{-b \cdot D_{slow}} \cdot \left(1 - F + F \cdot e^{-b \cdot D_{fast}}\right) \qquad \frac{S_b}{S_0} = e^{-b \cdot D_{slow}} \cdot \left(1 - F + F \cdot sinc\left(\frac{\Delta}{\sqrt{\Delta - \delta/3}} \cdot v \cdot \sqrt{b}\right)\right)$$

Methods: Following (4), a flow phantom comprised of 10 meters of 1.6 mm diameter tubing was randomly coiled into a 7 cm ball and submerged in water. Water was pumped through the tube at different velocities on the order of several millimeters per second. Spin-echo diffusion-weighted echo planar imaging was performed with 40 different b-values quadratically spaced between 0 and 600 s/mm2 TE=49.3 ms, TR=3 or 6 sec, matrix 64x64, for a voxel size of 3.4x3.4x5 mm. For each voxel, velocity estimates were obtained by fitting signal intensity vs. b-value with the sinc model. Volunteers were scanned with a similar protocol using 40 different b-values quadratically spaced between 0 and 450 s/mm2 TE=45 ms, TR=2 RR intervals, peripherally gated with TD=800 ms, matrix 64x64, for a voxel size of 6.3x6.3x10 mm, through a sagittal slice through the right lobe of the liver, during an exhalation breathhold. Data were fit with mono-, bi-exponential, and sinc models, with explicitly modeled T1 weighting, and an image was created with voxel hue corresponding to the best fitting model by the Bayesian information criterion.

Results and Discussion: In the phantom, damped oscillation of the diffusion-weighted signal was observed in voxels that contained both water flowing within the tubing and surrounding water subject only to diffusion (Fig 1). Flow velocity estimates from the sinc model were linearly related to pump speed setting (Fig 2) and did not change when TR was doubled. The original study examining this phantom (4) showed monotonic decay in a large voxel containing the entire phantom, in which flow was isotropic but laminar instead of plug-like. In this study, we show damped oscillation because flow within smaller voxels is anisotropic and laminar. The velocity magnitude profile of such anisotropic, laminar flow is parabolic and resembles the spherical profile of isotropic, plug-like flow that gives rise to sinc attenuation. In vivo, diffusion-weighted signal decay in some regions of liver was better fit by the sinc model than by the mono- or bi-exponential models (Fig 3) according to the Bayesian information criterion. Damped oscillation is observable in such regions (Fig 4). This may arise from isotropic, plug-like flow in small vessels of the liver, but may also arise from temporal confounders that occurred over the course of the scan. TR variation from heart-rate changes during gating was not responsible for the oscillation, since the T1 weighting was explicitly modeled. Velocity estimates derived from the sinc model may allow more robust quantification of the effects of hepatic disease processes on blood flow. Furthermore, diffusion coefficients estimated with mono- or bi-exponential models may be systematically biased if calculated using b-values at the peak or trough of damped oscillation. Further work to exclude other temporal confounders as the source of oscillation is ongoing, and will be informative as to whether the sinc model should be used to characterize the velocity of small vessel blood flow in hepatic disease.

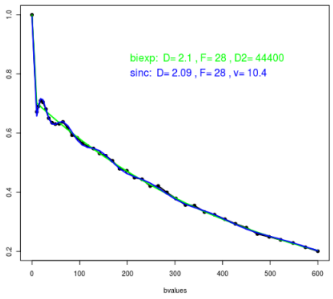

Figure 1 - Signal intensity as function of b-value in an ROI of flow phantom. Fits are shown for bi-exponential (green) and sinc (blue) models.

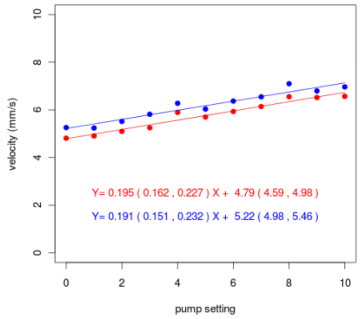

Figure 2 - Velocity of flow in phantom estimated from sinc model, as function of pump setting. Red and blue indicate different slices through phantom. Linear fits and 95% confidence intervals are shown.

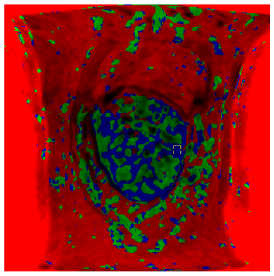

Figure 3 - Information map specifying best fitting model (mono: red, bi-exponential: green, sinc: blue) for diffusion attenuation in sagittal slice through liver of human volunteer.

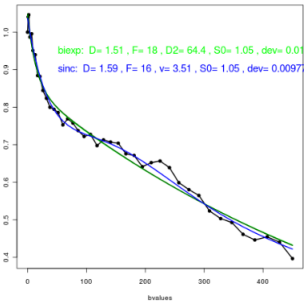

Figure 4 - TR-scaled signal intensity as function of b-value in ROI of volunteer liver indicated in Figure 3. Fits are shown for biexponential (green) and sinc (blue) models.

References:

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