Advanced diffusion weighted imaging techniques

Aims:

1) The basic principle of how magnetic resonance imaging is sensitized to free diffusion or Browinan motion of water protons
2) The introduction of the commonly used apparent diffusion coefficient (ADC) based on an increasing, mono-exponential signal decay when diffusion weighting is augmented
3) The perfusion induced deviation from monoexponential signal decay when weak diffusion sensitizing is applied, as introduced in the intravoxel-incoherent motion (IVIM) theory (1).
4) The restriction induced deviation from monoexponential signal when strong diffusion sensitizing is applied as captures using diffusion kurtosis imaging (2)
5) To show clinical examples of the application of advanced diffusion techniques

Diffusion weighted imaging and the apparent diffusion constant

Figure 1 describes the basic principle of diffusion weighting in NMR. The red arrows in circles symbolize the magnetic moment in the rotating reference frame at different spatial positions (lower rows of pictures). On top of that, the spatial profile of the magnetic field is depicted, which consists of the main magnetic field B_0 and the spatially varying gradient field. The 90° radio frequency pulse generates a transversal magnetization which is dephased by the first diffusion gradient. If no diffusion is present, the particles do not move, and the second diffusion gradient rephases the magnetization. However, if diffusion is present, the particles change their spatial position, and the second gradient does not rephase the magnetization. The signal can in some sense be regarded as the sum of the red arrows. If diffusion is present, the arrows do not point coherently along the same direction, and the signal (or the sum of arrows, respectively) drops. This signal drop is the stronger, the stronger and longer the magnetic field gradients are applied, and the farther the particles translate. Since the gradients are controlled by the measurement sequence, information about the diffusion process can be inferred by measuring the signal with and without diffusion gradients: a strong signal drop indicates strong diffusion. In biological tissue using clinical scanners with finite gradients, the signal decay can be described by

$$S = S_0 \exp(-bADC)$$

Where S_0 is the signal without diffusion weighting, S the signal with diffusion weighting and b so called b-values and describes the strength of the diffusion weighting defined as

$$b = \gamma^2 \delta^2 \mathrm{G}^2 (\Delta - \delta/3)$$

where δ is the gradient duration, G the gradient strength, Δ the mixing time and γ the gyromagnetic constant at the given field strength. Since S and S_0 can be measures and b is defined by machine setting, the only free variable it the apparent diffusion coefficient, ADC.

Intravoxel incoherent motion

The blood flowing in the capillary bed is not moving coherently, but along different directions. This incoherent motion resembles diffusional motion, and in fact, it is observed experimentally that this capillary blood motion has a similar effect on the diffusion weighted signal as a true diffusion process. This is most visible in organs of short tissue T_2 relaxation time, like in the liver and pancreas, due to the increased relative weight of the blood compartment (3). The typical distances that particles translate due to the capillary flow are much larger than due to diffusion at the typical time scales of NMR diffusion experiments at clinical scanners. Therefore, a blood flow related pseudo-diffusion coefficient D^* is measured, which is larger by more than a factor of 10 than the apparent diffusion coefficient. The most widely-used model, as originally proposed by le Bihan, assumes that two well separated compartments – one blood compartment and one tissue compartment – are present (1,3):

$$S = S_0\left(f e^{-bD^*} + (1-f) e^{-bD_{app}}\right)$$

Here, f is the perfusion fraction. Fig. 2 illustrates the IVIM-effect at the example of the healthy pancreas (4). Note that the IVIM effect occurs at very small b-values (<50 s/mm²).

Kurtosis

At larger b-values the signal decays as follows:

$$S = S_0 \exp\left(-bD + \frac{1}{6} b^2 (D)^2 K + \cdots\right)$$

for details see (2). If b is small, the b^2-term is not of importance, however for large b-values, it becomes dominant. The kurtosis K is an additional measurable parameter which is zero for free diffusion. If diffusion restrictions are present, K does in general deviate from zero, and can be either larger or smaller than zero, depending on the geometry of the restrictions. K can be regarded as a

parameter describing the deviation from Gaussian diffusion. To determine the kurtosis, the signal must be measured with several *b*-values ranging typically from 0 s/mm² to more than b=2000 s/mm².

Figures:

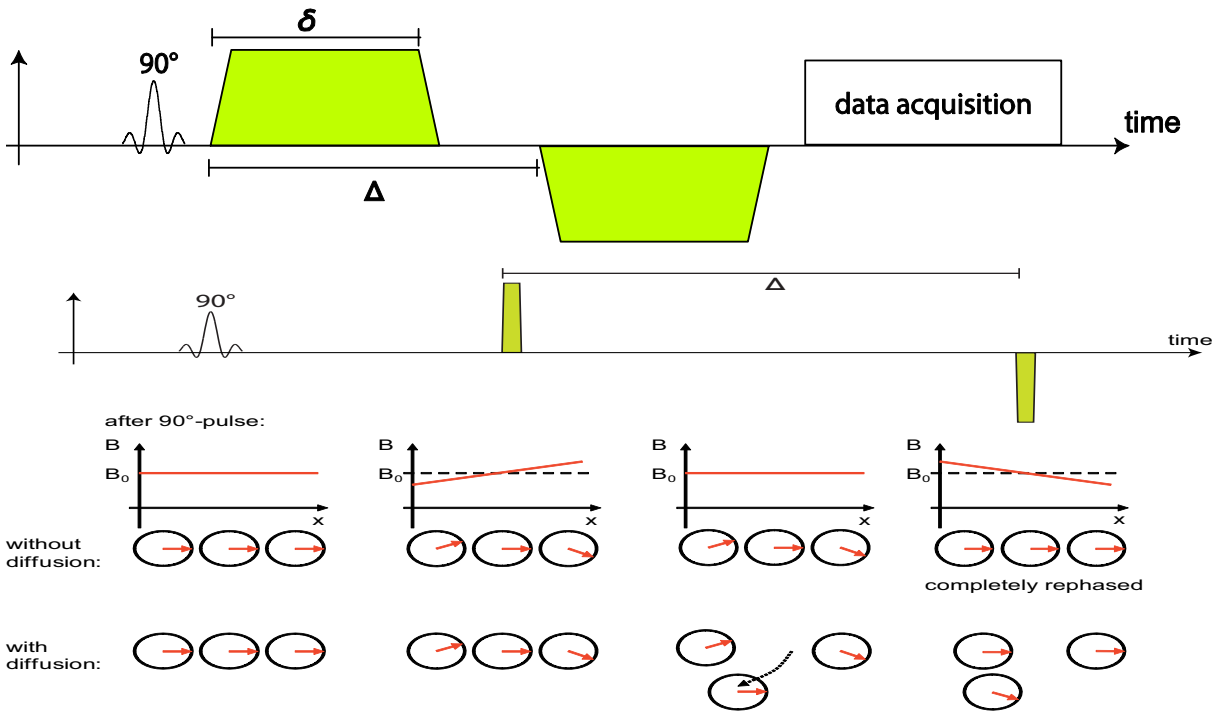

Figure 1: Diffusion weighting in NMR. In the top row, a typical diffusion sequence is shown. Gradient length (δ) and mixing time (Δ) are indicated. In the two bottom rows, the typical spin behavior without (middle row) and with diffusion (bottom row) is represented.

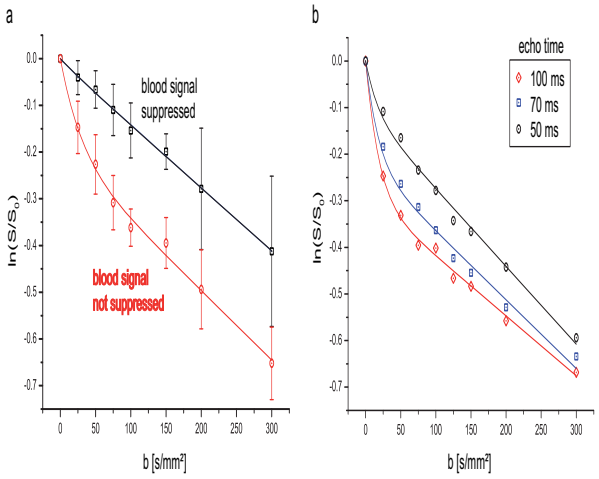

Figure 2: Logarithmic signal decay in the pancreas of six healthy volunteers with echo time TE=50 ms. Due to the incoherent blood flow, the signal decays bi-exponentially. After suppression of the blood signal by means of an inversion recovery pulse, the signal decays mono-exponentially. This indicates that the blood compartment plays an important role.

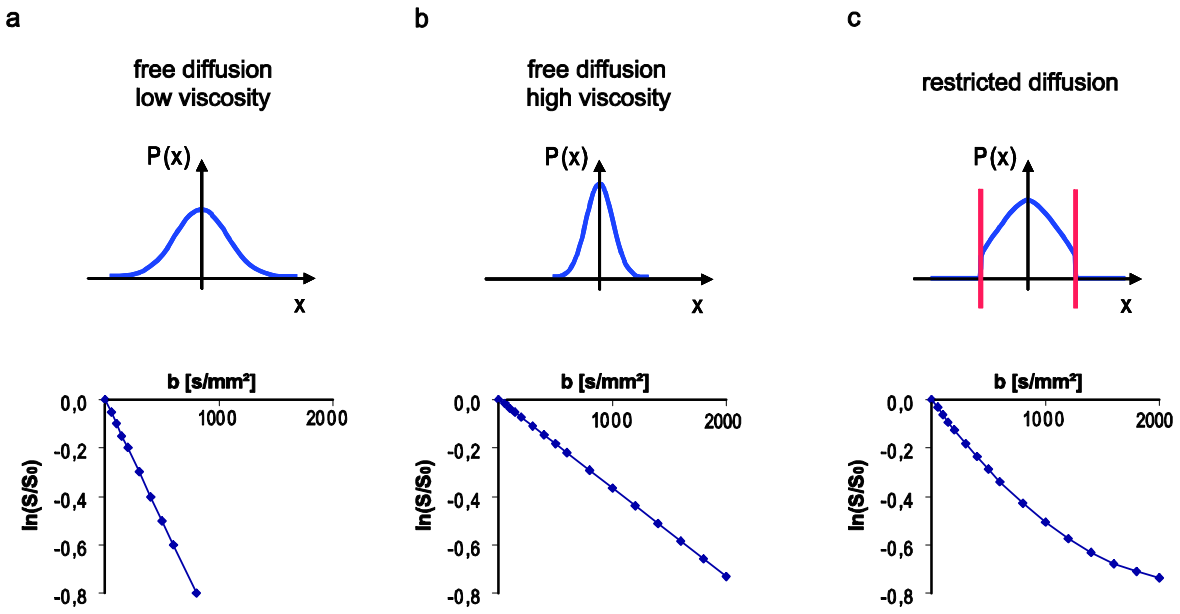

Figure 3: For free diffusion, the propagator is a Gaussian function and the diffusion weighted signal decays exponentially. Different diffusion constants due to different viscosities result in different slopes of the signal decay. c) If diffusion restrictions are present, then the propagator deviates from a Gaussian function and the signal does not decay exponentially. The occurring deviation can be quantified using the kurtosis parameter, which can be interpreted as a measure of the deviation from Gaussianity of the diffusion propagator.

References

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3) Le Bihan D, Breton E, Lallemand D, Aubin ML, Vignaud J, Laval-Jeantet M. Separation of diffusion and perfusion in intravoxel incoherent motion MR imaging. Radiology 1988;168(2):497-505.

4) Lemke A, Laun FB, Simon D, Stieltjes B, Schad LR. An in vivo verification of the intravoxel incoherent motion effect in diffusion-weighted imaging of the abdomen. Magn Reson Med 2010;64(6):1580-1585.