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Physical Principles of Diffusion Weighted Imaging

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Abstract.Статията обяснява основните физични принципи на РЧП от тип спин-ехо. Разгледана е хронологията и техниката за елиминиране на собствените нехомогенности на магнитното поле и достигане до Т2 образи чрез метод на спин-ехо. Базирайки се на този метод, е анализирана техниката на планарно ехо с имплементиране на дифузни радиочестотни градиенти, която е в основата на получаване на МР образи, акцентирани в дифузия. В допълнение са описани и характеристиките на прилаганите градиенти, тяхната роля при получаването на дифузен образ и метода на изчисление на дифузен коефициент като количествена оценка на наблюдавана дифузия в дадена област.

Introduction:

The diffusion technique is intended to depict the microscopic movements of the water in the tissues. In the biological environment water molecules are in constant motion, called Brownian (R. Brown, 1827). These molecular movements are promiscuous, random and more or less intense, depending on the biological environment. The more water is free in a given area, the greater the movement of the molecules and facilitates the diffusion is higher. Conversely, there are areas where the diffusion is reduced or restricted due to the many obstacles on the movement of water molecules.

As well as other techniques based on magnetic resonance imaging, the diffusion is a complex radiofrequency (RF) sequence. Nowadays, almost all diffuse sequences descent of the so-called RF gradient spin echo (PGSE - pulsed gradient spin echo) technique, discovered and first described in 1965 by Edward Steshkal and John Tanner.

1 Spin-Echo Pulsed Sequence

After the application of 90° RF pulse, the spins of protons start to dephase due to the variability of the magnetic field. If after a certain time TE/2 (half of echo time) after the 90° pulse, a RF pulse of 180° is applied, it will turn phases, without changing the direction of rotation. The spins

of protons, which precession is fastest will be located behind the slowest and since the rate of precession and direction are constant, the end of time TE (echo time) the spins will be back in phase, generating a signal called an echo of spin . The chronology of spin-echo is as follows (Figure 1):

- At the moment $t = 0$ the RF pulse was applied at 90°, which knocks the total net magnetization from longitudinal into a transverse plane. The spins are in phase and the transverse magnetization is maximum
- After a short time the spins are out of phase due to the variability in the static main magnetic field (inhomogeneity)
- At the moment $t =$ TE/2, a 180° RF pulse was applied
- At the moment $t =$ TE, the spins are in phase again and the signal reappears in the form of echo and can be measured

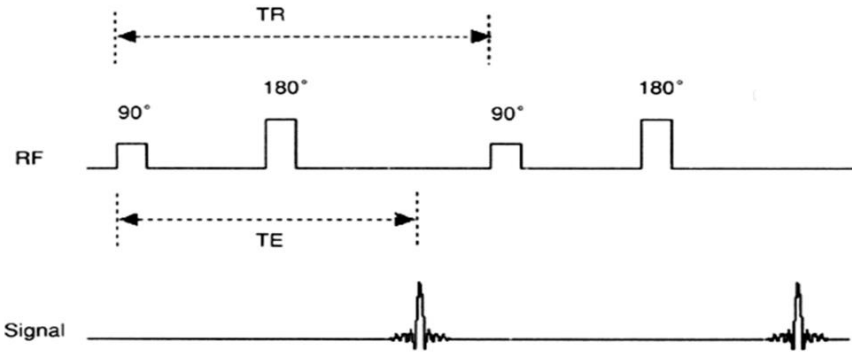

Figure 1: Spin echo pulsed sequence.

Spin-echo is one of the most commonly used sequence in MRI imaging, and it is routinely used in all kinds of imaging protocols. It allows to obtain images in T1 and T2 relaxation times, by proper selection of repetition and echo time. The main disadvantage of spin-echo sequence is relatively long time duration which is proportionable to the repetition times necessary to replenish all lines of the chosen matrix (k-space).

2 Diffusion-Weighted Sequence

In all standard sequences, including spin-echo, the loss of signal from moving protons in a given voxel is difficult to measure. To enable the visualization of microscopic molecular motions and to obtain images

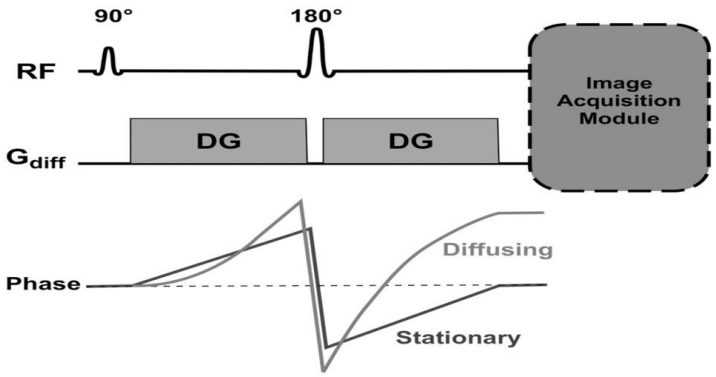

Figure 2: Diffusion sensitizing gradients applies on other side of 180° pulsa.

weighted in diffusion is necessary to apply two additional radio frequency bipolar gradients (Figure 2).

These diffusion gradients (DGs) are applied in the three axes – x, y and z on both sides of the 180° RF pulse in a sequence type called echo planar imaging – spin echo with a single excitation (single-shot spin-echo echo-planar imaging).

In practice, diffusion sequence lies in the consistent application of three planar sequences containing DGs, respectively, in the axis of slice selection, phase and frequency encoding. The obtained images in the three axes are reconstructed in fourth one which is averaged and storage only the hyper signal part. The phases of stationary spins are unaffected by the DG pair since any phase difference from the first gradient lobe is compensated by the second one. Diffusion spins move into different directions between the first and the second lobe and that's why they falling out of phase and losing signal.

The resulting image show hyper signal in the areas of reduced molecular diffusion and the weaker signal, the more molecules with higher diffusion contain the observed area.

Considering the effect of diffusion, Bloch equation acquired the following modified form:

$$\frac{dM}{dt} = \gamma(M \times B_0) + \begin{pmatrix} \frac{M_x}{\mathrm{T2}} \\ \frac{M_y}{\mathrm{T2}} \\ \frac{M_0 - M_z}{\mathrm{T1}} \end{pmatrix} + D\nabla^2 M \,,$$

where M is the net magnetization placed in a static field with magnetic

induction B_0, γ is the gyromagnetic hydrogen ratio, M_x, M_y, M_z are the components of the net magnetization and D is the diffusion coefficient. The factor that characterize the strength and timing of DGs is called b-value and is expressed as follows:

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

where G is the magnitude of the gradients, duration (δ), separated by time interval (Δ). Pulsed-gradient diffusion method is shown in Figure 3.

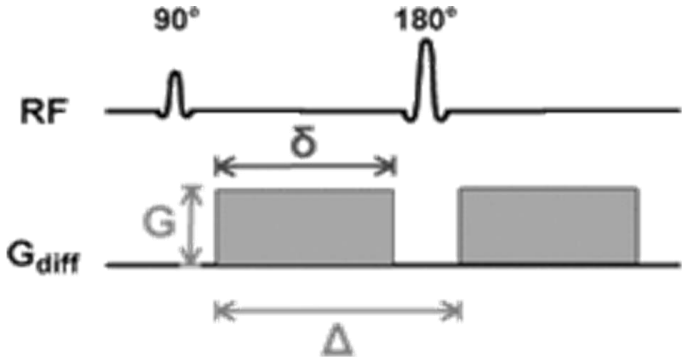

Figure 3: Pulsed-gradient diffusion method.

The b-value is an operator-selected parameter and determine the degree of diffusion weighting. A larger b-value is achieved with increasing the gradient amplitude and duration by widening the interval between gradient pulses. The optimal choice of b-value depends on the magnitude of the main magnetic field, the number of selected imaging directions and specific pathology. When we have a diffusion weighted image, an apparent diffusion coefficient map (ADC map) can be created by dividing the signal from the DWI image (S_{DWI}) by the signal S_0 (without b-value)

$$\text{ADC} = -\frac{1}{b} \ln \left(\frac{S_{\text{DWI}}}{S_0} \right).$$

In clinical practice the ADC calculation is done automatically for each voxel of the diffused weighted image. An important feature is that in order to calculate the ADC map we have to have at least two b-values. Areas with slow molecular diffusion are presented with low signal intensity on ADC map conversely to diffusion weighted image.

3 Conclusion

Incredibly rapid development of magnetic resonance imaging and the increasing use require a deep understanding of basic physical principles of

spin-echo sequence. This technique allows to obtain excellent images in T1 and T2 relaxation time by an appropriate choice of technical parameters. The possibility of further implementation of DGs to spin-echo method favours to obtain images weighted in diffusion. The patterns of diffusion in combination with standard sequences are sufficient for the visualization of pathological tissues and processes in clinical practice. The quantitative evaluation (ADC map) is mandatory and useful addition to the diffusion weighted images.

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