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OPTIMAL SIGNAL SUPPRESSION OF BONE AND CARTILAGE IN MRI

The robust procedures for the separation of bone and cartilage tissues in magnetic resonance (MR) images are presented. Increased differentiation by contrast and signal-to-noise ratio (SNR) in proposed methods is based on pulse sequence dependence. First method is based on the new pulse sequence for MR bone and cartilage imaging, which allows simultaneous suppression of the signal from one tissue and visualisation of another one or vice versa. Second method is an optimization of balanced steady-state free precession (bSSFP) sequence. Mathematical modeling shows direct increasing of tissue differentiation under optimal values of these pulse sequences obtained for high contrast between bone and cartilage and a high SNR.

Key words: MR bone and cartilage imaging, SNR, contrast, pulse sequence optimization.

Introduction. Bone and cartilage are difficult to distinguish in MR images obtained by standard pulse sequences (such as fast spin echo, gradient recall echo, turbo spin echo, bSSFP) [3; 4; 6–9]. As a result these images cannot be segmented. But segmented images are more meaningful and easier to analyze and could be used for practical applications such as measurement of tissue volumes, treatment planning and study of anatomical structure. Each of the pixels in a region of segmented image is similar with respect to some characteristic or computed property, such as color, intensity, or texture. On MR tomograms these properties are functions of such physical characteristics as equilibrium magnetization, spin-lattice and spin-spin relaxation times and parameters of pulse sequence used for visualization. Having considered all arguments, it may be pointed out that study of anatomical structure and implants manufacturing need a special pulse sequence for bone and cartilage MR imaging. This problem could be solved in two ways. First way is calculation of parameters for a new pulse sequence for MR bone and cartilage imaging. Second way is optimization of standard pulse sequence for bone and cartilage separation.

In this work we perform calculation and optimization of parameters for a new pulse sequence and bSSFP sequence by contrast and SNR for MR bone and cartilage imaging. Optimization was provided by mathematical modeling according to the theoretical model.

Theoretical model. Model of bone and cartilage for MR could be described with three pairs of parameters: equilibrium magnetization per unit volume, spin-lattice and spin-spin relaxation times of bone and cartilage.

Spin-lattice and spin-spin relaxation times of bone and cartilage are listed in Table 1 [1–3; 5; 10].

Table 1

Spin-lattice relaxation times T_1, spin-spin relaxation times T_2 and densities of bone and cartilage

Tissue	T_1, ms	T_2, ms	Density, kg/m^3
Bone	554±27	140±12	1850
Cartilage	1060±160	42±7	1050

The following approximations were introduced for assessment of the equilibrium magnetization per unit volume M_0 of bone and cartilage:

1. Water is a major component of bone and cartilage tissues. That is why only the protons of water molecules were considered in the calculations.

2. Distribution of water in bone and cartilage is uniform: $M_0 \neq f(\vec{r})$.

The equilibrium magnetization as a function of magnetic field B_0 and temperature T is obtained according to the 26% mass fraction of water in bone and 70% mass fraction of water in cartilage.

The value of the equilibrium magnetization per unit volume in the high-temperature approximation is obtained using the following formula:

$$M_0 = \frac{N \cdot \hbar^2 \cdot \gamma^2 \cdot I \cdot (I+1)}{3 \cdot k \cdot T} \cdot B_0 . \qquad (1)$$

Total number of protons is calculated by the following formula:

$$N = \frac{2 \cdot \rho_{H_2 0} \cdot V_{H_2 0} \cdot N_A}{M_{H_2O}} , \qquad (2)$$

where $\rho_{H_2 0}$ – water density, $V_{H_2 0}$ – water volume in tissue, N_A – Avogadro constant, M_{H_2O} – molar mass of water.

Mass fraction of water in bone is calculated as follows:

$$\frac{m_{H_2O,b}}{m_b} = \frac{\rho_{H_2O} \cdot V_{H_2O,b}}{\rho_b \cdot V_b} = 0.26 , \qquad (3)$$

where m_b – bone mass, $m_{H_2O,b}$ – water mass in bone, ρ_b – bone density, $V_{H_2O,b}$ – water volume in bone, V_b – bone volume.

$$V_{H_2O,b} = \frac{0.26 \cdot \rho_b}{\rho_{H_2O}} \cdot V_b = \frac{0.26 \cdot 1850}{1000} \cdot V_b = 0.481 \cdot V_b , \qquad (4)$$

From (2) and (4) the total number of protons in the bone per unit volume is:

$$N_b = 0.481 \cdot \frac{2 \cdot \rho_{H_2 0} \cdot N_A}{M_{H_2O}} . \qquad (5)$$

From (1) and (5) the equilibrium magnetization per unit volume for bone is:

$$M_{0,b} = 0.481 \cdot \frac{\hbar^2 \cdot \gamma^2 \cdot I \cdot (I+1)}{3 \cdot k \cdot T} \cdot B_0 \cdot \frac{2 \cdot \rho_{H_2 0} \cdot N_A}{M_{H_2O}} . \qquad (6)$$

Mass fraction of water in cartilage is calculated as follows:

$$\frac{m_{H_2O,c}}{m_c} = \frac{\rho_{H_2O} \cdot V_{H_2O,c}}{\rho_c \cdot V_c} = 0.7 , \qquad (7)$$

where m_c – cartilage mass, $m_{H_2O,c}$ – water mass in cartilage, ρ_c – cartilage density, $V_{H_2O,c}$ – water volume in cartilage, V_c – cartilage volume.

$$V_{H_2O,c} = 0.7 \cdot \frac{\rho_c}{\rho_{H_2O}} \cdot V_c = 0.7 \cdot \frac{1050}{1000} \cdot V_c = 0.735 \cdot V_c , \qquad (8)$$

From (2) and (8) the total number of protons in the cartilage per unit volume is:

$$N_c = 0.735 \cdot \frac{2 \cdot \rho_{H_2 0} \cdot N_A}{M_{H_2O}} . \qquad (9)$$

From (1) and (9) the equilibrium magnetization per unit volume for cartilage is:

$$M_{0,c} = 0.735 \cdot \frac{\hbar^2 \cdot \gamma^2 \cdot I \cdot (I+1)}{3 \cdot k \cdot T} \cdot B_0 \cdot \frac{2 \cdot \rho_{H_2 0} \cdot N_A}{M_{H_2O}} . \qquad (10)$$

Methods. New pulse sequence for MR bone and cartilage imaging. Inversion-recovery method was chosen because of almost double spin-lattice relaxation times difference:

$$\frac{T_{1,c}}{T_{1,b}} = \frac{1060}{554} \approx 1.91, \quad (11)$$

where $T_{1,c}$, $T_{1,b}$ – spin-lattice relaxation times of cartilage and bone respectively.

Pulse sequence consists of two parts. In the first part of the sequence only cartilage was visualized (bone signal was suppressed). It can be realized in such a way. A 180° pulse is applied first. This pulse rotates the net magnetization down. A 90° pulse is applied after applying of 180° pulse with delay:

$$T_b = T_{1,b} \cdot \ln(2) \approx 384 \text{ ms}. \quad (12)$$

It should be noted that at this time net magnetization of bone is zero. In the second part of the sequence only bone was visualized (cartilage signal was suppressed). This part is similar to the first part of sequence, but the delay between the radio frequency (RF) pulses is:

$$T_c = T_{1,c} \cdot \ln(2) \approx 734.7 \text{ ms}. \quad (13)$$

Sequence repetition time TR is 3 seconds, because z-component of the magnetization vectors must return to equilibrium.

bSSFP uses rapid excitation radiofrequency pulses combined with fully balanced gradient pulses to acquire images. It is based on a low flip angle GRE sequence and also includes transverse magnetizations from overlapping echoes along with longitudinal magnetizations from GRE.

Simulation is selected for verification of results, because the simulator is available in comparison with the real MRI system. 3D simulations are very time consuming, that is why 2D sample "2D 2-spheres" (Fig. 1) was selected as the object in MRI simulator JEMRIS. This sample allows to explore two tissues simultaneously.

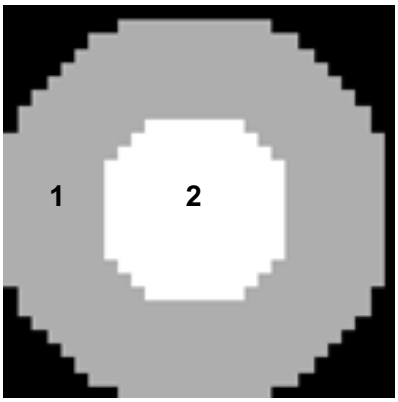

Fig. 1. Sample "2D 2-spheres":
1 – bone, 2 – cartilage

Results. Parameters used in the simulation correspond to real parameters of bone and cartilage.

The new pulse sequence. After signal suppression images shown in Fig. 2 were obtained. Contrasts are 90±10% and 86±9%, SNR are 5.00 and 6.59 respectively.

Optimization of delay between RF-pulses by a high contrast and maximum SNR of reconstructed tomogram was done. Examples of obtained images shown in Fig. 3.

Delays are 400 ms and 750 ms respectively. Results of mathematical modeling showed that the contrasts in these cases are 63±12% and 90±10% and SNRs are 6.51 and 7.87 respectively.

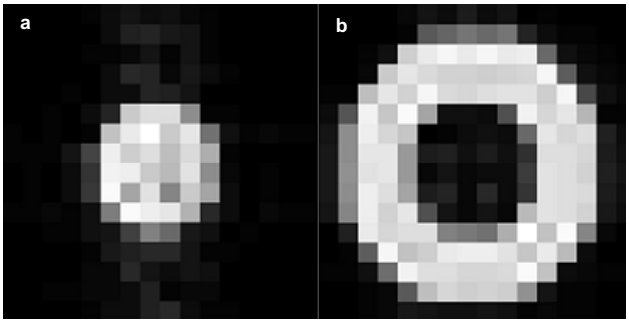

Fig. 2. Reconstructed tomograms:
a – the cartilage image
(the bone signal was suppressed);
b – the bone image
(the cartilage signal was suppressed)

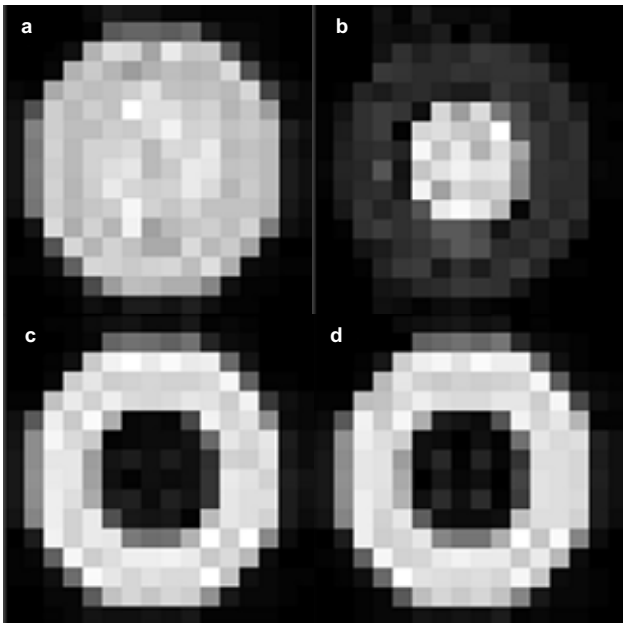

Fig. 3. Reconstructed tomograms were obtained
by the new pulse sequence with delay between RF pulses in:
a –300 ms; b –400 ms; c –700 ms; d –800 ms

bSSPF studies consist of SNR dependence upon repetition time TR, excitation time TE and flip angle. Fig. 4 shows SNR as a function of TR from both cartilage and bone with excitation time TE = 0.5TR and flip angles of 22° and 53°.

It can be seen that SNR decays with the increase of TR. Fig. 5 shows image degradation at long repetition times TR.

For studying SNR dependence upon excitation time TE, TR was set to 6 ms based on the aforementioned results. Flip angles remained the same. TE was changed from 1 to 5 ms with increment of 1 ms.

No visible changes of SNR were registered, but cartilage contrast is slightly better with TE = 3 ms (~ 0.5 TR).

The most important task was to find an optimal angle, so that both cartilage and bone SNR were of satisfactory values. TR =6 ms and TE = 3 ms were chosen.

Fig. 6 shows SNR dependence upon flip angle with clearly visible maximum cartilage SNR of ~17 around 20–23° with bone SNR of ~12 at the same angle. Choosing flip angle that maximizes bone SNR (53°) is not rational because of significant cartilage SNR drop at this angle.

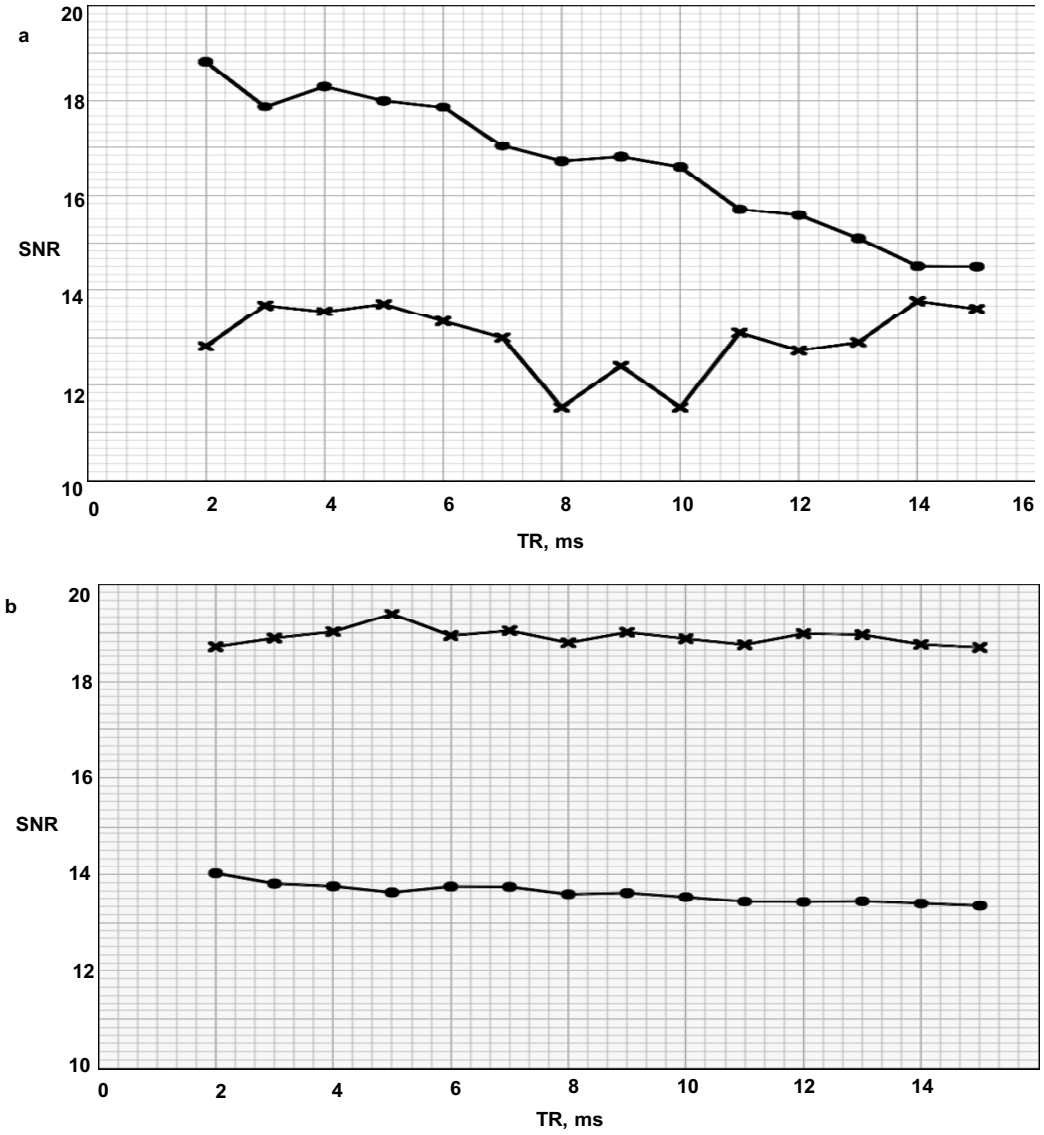

Fig. 4. SNR dependence upon TR, flip angle 22°(a) and 53°(b).
SNR of cartilage is marked with dots, SNR of bone is marked with crosses

Fig. 5. Image degradation with TR = 40 ms
and TE = 20 ms (right) compared to TR =4 ms and TE = 2 ms (left)

To compare SNR-efficiency the image with the same SNR of both tissues using classic GRE pulse sequence (TR/TE/angle = 500ms/10ms/51°) was acquired. Time to achieve SNR of 17 for cartilage and 12 for bone was 15 m 13 s using GRE and 10 m 17 s using bSSFP, making bSSPF 22% more SNR-efficient than classic GRE sequence. This difference in efficiency can be used for shortening scan time or for improving overall image resolution.

Conclusion. We have proposed two procedures for optimal signal suppression of bone and cartilage in MRI. Inversion-recovery is the most effective when delays between RF-pulses are 400 ms and 750 ms for bone and cartilage suppression. The contrasts are 63±12% and 90±10% and SNRs are 6.51 and 7.87 respectively. bSSFP achieves the best separation with TR/TE/flip angle = 6 ms/3 ms/22° acquiring cartilage and bone SNRs of 17 and 12 respectively.

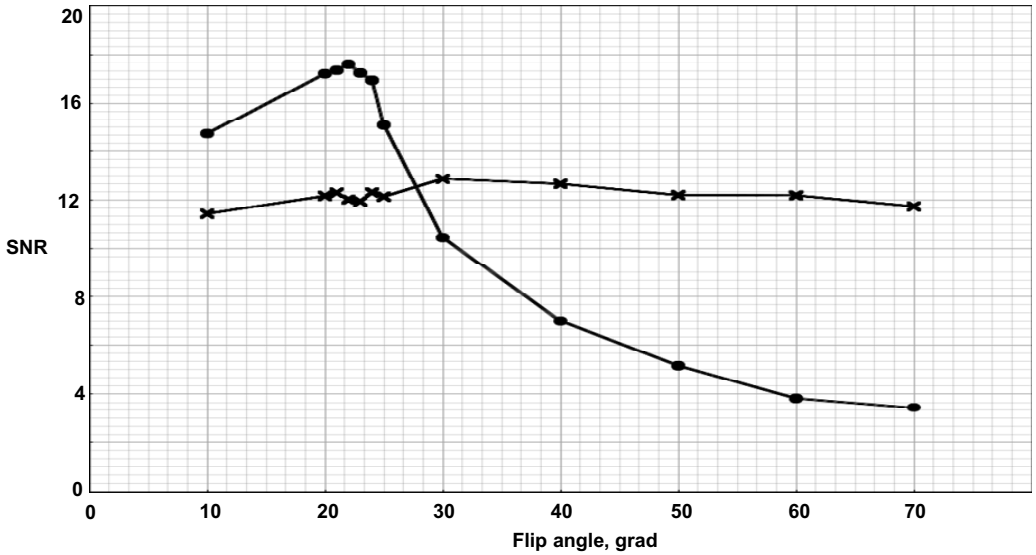

Fig. 6. SNR dependence upon flip angle with TR = 6 ms and TE = 3 ms. SNR of cartilage is marked with dots, SNR of bone is marked with crosses

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ОПТИМАЛЬНЕ ПРИДУШЕННЯ СИГНАЛІВ КІСТКИ І ХРЯЩА В МРТ

Представлено надійні процедури для відділення кісткових і хрящових тканин на магнітно-резонансних (МР) зображеннях. Збільшення диференціації за контрастом і співвідношенням сигнал-шум у запропонованих методах засновано на залежності від імпульсної послідовності. Перший метод базується на новій імпульсній послідовності для МР зображень кісток і хрящів, яка дозволяє придушити сигнал від однієї тканини і візуалізувати іншу і навпаки. Другий метод – це оптимізація послідовності balanced steady-state free precession. Математичне моделювання показує пряме збільшення диференціації тканин при оптимальних значеннях цих послідовностей, отриманих для високого контрасту між кісткою і хрящем і великого співвідношення сигнал-шум.

Ключові слова: магнітно-резонансні зображення кісток і хрящів, співвідношення сигнал-шум, контраст, оптимізація імпульсної послідовності.

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ОПТИМАЛЬНОЕ ПОДАВЛЕНИЕ СИГНАЛОВ КОСТИ И ХРЯЩА В МРТ

Представлено надежные процедуры для отделения костных и хрящевых тканей на магнитно-резонансных (МР) изображениях. Увеличение дифференциации по контрасту и соотношению сигнал-шум в предложенных методах основано на зависимости от импульсной последовательности. Первый метод базируется на новой импульсной последовательности для МР изображений костей и хрящей, которая позволяет подавить сигнал от одной ткани и визуализировать вторую и наоборот. Второй метод – это оптимизация последовательности balanced steady-state free precession. Математическое моделирование показывает прямое увеличение дифференциации тканей при оптимальных значениях этих последовательностей, полученных для высокого контраста между костью и хрящом и большого соотношения сигнал-шум.

Ключевые слова: магнитно-резонансные изображения костей и хрящей, соотношение сигнал-шум, контраст, оптимизация импульсной последовательности.