Hydrogen interactions in magnetic resonance imaging Histogram-based segmentation of brain tissues

S. Gułkowski*, J. M. Olchowik

Institute of Physics, Lublin University of Technology, Nadbystrzycka 38, 20-618 Lublin, Poland

Magnetic resonance imaging (MRI) is a very popular, non-invasive medical diagnostic technique providing images of the human body. The method, commonly used in clinics and hospitals all over the world, is based on the magnetic properties of hydrogen atom and its interactions with magnetic field. MR images show differences in water content and distribution in various body tissues. Even different tissues in the same organ, such as the gray and white matter in brain, can easily be distinguished. For a precise analysis of the images obtained in the result of MR examination, special computer applications are used. The paper presents the basis for the magnetic resonance imaging and description of the method of brain segmentation which could be a useful software tool in daily medical diagnosis practice.

Key words: *magnetic resonance imaging; histogram-based segmentation*

1. Introduction

Present diagnostic technologies provide us with various methods of imaging of human body tissues. One of these methods commonly used in clinics and hospitals all over the world is magnetic resonance imaging (MRI). MRI is based on the principles of nuclear magnetic resonance (NMR), a very important method for studying nuclear and atomic systems.

MRI is based on the magnetic properties of hydrogen atom and its interaction with a strong external magnetic fields and radiowaves to produce highly detailed images with excellent soft tissue contrast. Moreover, for more accurate analysis of a tissue or its pathology, special computer applications are created. These methods of image processing are very helpful in daily work of physicians.

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*Corresponding author, e-mail: s.gulkowski@pollub.pl

2. Basis for magnetic resonance imaging

Human body is composed primarily of fat and water, thus also of many hydrogen atoms. The nucleus of the hydrogen atom contains a single proton which is chracterized by spin, being a magnetic moment vector. In nuclear magnetic resonance, such unpaired magnetic moments are of great importance. In a static, strong magnetic field $\vec{B}_0$, each of them alings in one of the two possible orientations: parallel (low-energy state) and anti-parallel (high-energy state) to the direction of the field. In addition to aligning with $\vec{B}_0$, protons precess at a frequency given by the Larmor equation:

$$\omega_0 = \gamma B_0 \tag{1}$$

This equation tells us that the frequency depends on the strength of the external magnetic field and on the ratio specific for a given nucleus – gyromagnetic ratio γ, which for hydrogen is 42,58 MHz/T [1].

At room temperature, the number of spins in the low-energy state n^{up} slightly outnumbers the number of spins in the high-energy state n^{down}. The Boltzmann distribution tell us that [2]:

$$\frac{n^{\text{down}}}{n^{\text{up}}} = \mathrm{e}^{E/kT} \tag{2}$$

where: E is the energy difference between the spin states, k is Boltzmann constant, and T is the temperature in kelvins.

The difference between the number of spins (magnetic moments) in the low-energy- and high-energy states creates the net magnetization. In the equilibrium, the net magnetization vector is aligned with the direction of the applied magnetic field $\vec{B}_0$, and does not precess. The z-component of net magnetization is called longitudinal magnetization M_z. There is no transverse magnetization M_{xy} [2], being the xy-plane component of the net magnetization. The higher hydrogen concentration in a specific human tissue, the stronger is magnetization of the tissue. This is the main idea of spin density technique of MRI [3].

It is possible to change net magnetization. If we apply a specific resonance frequency pulse, called 90° pulse of $\vec{B}_1$ magnetic field with energy equal to the energy difference between the spin states, the net magnetization spirals down around the z-axis to the xy plane. The $\vec{B}_1$ field should be orthogonal to the main field $\vec{B}_0$.

After the $\vec{B}_1$ magnetic field is applied, some "lower energy" spins move to the "higher energy" level, hence the longitudinal magnetization disappears. Simultaneously, individual spins are precessing in phase, so transversal magnetization appears. This transversal magnetization rotates about the z-axis and induces in the receiver a signal called free induction decay (FID). After RF pulse, the magnetization $\vec{M}$ re-

turns to its equilibrium state during the relaxation process. The *z*-component of the net magnetization M_z gets longer in T_1 relaxation time and the *xy*-component gets shorter in T_2 relaxation time. This signal contains information about the density of the matter and the relaxation times in the examined object.

3. Method of imaging

The FID signal that we get for a given $\vec{B}_0$ field after the application of a 90° pulse is the signal from every hydrogen magnetic moment (spin) all over the body. The aim is to distinguish the signal from each (*x,y,z*) point and to obtain a 3D image.

3.1. Gradient of magnetic field – slice selection

Slice seletion in MRI is achived by applying a one-dimensional, linear magnetic field gradient during the period the RF pulse is applied. A 90° pulse applied in conjunction with a magnetic field gradient will rotate magnetization vectors which are located in a slice or plane through the object [2]. It is possible because magnetic moments from different planes (along *z* axis) experience different value of the static field ($\vec{B}_z$), so they have unique resonance frequency ω_z. Applying the RF pulse with resonance frequency, we select the spins from the one plane. Spins located above and below this plane are not affected by RF pulse.

3.2. Gradient of phase and frequency encoding

To distinguish information from a chosen slice of the scanning object, phase and frequency encoding gradients are applied. Every spin from the selected slice precesses at the same Larmor frequency given by the applied magnetic field. If we apply the phase encoding gradient $\vec{G}_x$ along the *x*-axis, the net magnetizations at different locations begin to precess with different frequencies. When the phase encoding gradient is turned off, all net magnetization vectors precess at the same frequency but at different phase along the *x*-axis. After the phase encoding gradient, the frequency encoding gradient is applied. As a result, the spin packets precess at rates dependent on their *y*-location [2]. This sequence results in each of the net magnetization vectors of a chosen slice which is characterized by unique phase and precessional frequencies. Performing the Fourier transform, we can obtain these information from the FID signal.

In order to visualise the information about hydrogen density or relaxation times obtained from the FID signal, special computer programs are employed which can convert these data into images. Many medical applications give the display of images in three planes of imaging.

For a more accurate analysis of the tissues of interest or their pathology, the medical software tools have implemented special image processing methods, such as level-window or pseudocolouring, which improve quality of images. Many applications use more advanced techniques to select interesting information from images. The next section shows an example of the histogram based on segmentation technique, which uses the Levenberg–Marquardt optimization function.

4. Computer analysis of anatomical structures in medical image

A very important task in computer analysis of an MR image is the segmentation of white and gray matter from the human brain image. Segmentation means classifying each voxel of the image (voxel is defined as a volume element of a slice corresponding to the appriopriate pixel of image) as belonging to a specific type of tissue [4].

As shown in the histogram of MR slice of healthy head (Fig. 1), it contains three distinct peaks. These peaks correspond to different kinds of tissues. First represent the cerebrospinal fluid (CSF), second – gray matter and third – white matter of brain. In order to segmentate these structures, we have to found the boundaries between the adjacent peaks.
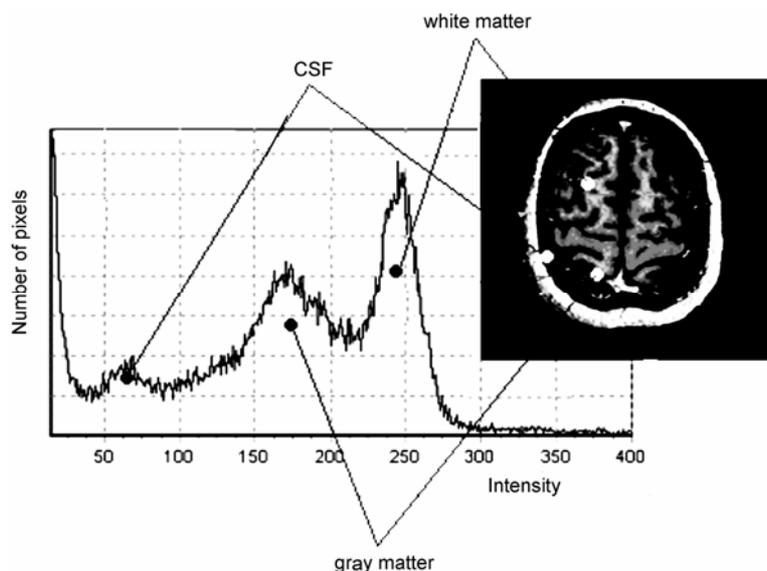

Fig. 1. Histogram of MR T1-weighted slice with peaks corresponding to various tissues.
Image of brain taken from [5]

Because of the overlapping, the choice of the intensity levels and the threshold values is very subjective and depends on the kind of the slice. It is necessary to use an algorithm that could follow the shape of the histogram. A good solution is the

Levenberg–Marquardt (L–M) algorithm which allows one to fit precisely a model function $f(a,x_i)$ into a histogram. The model function is a sum of Gaussian distributions:

$$f\left(a, x_i\right) = f\left(A_k, \mu_k, \sigma_k, x_i\right) = \sum_{k=1}^{Q} A_k \exp\left[-\frac{(x_i - \mu_k)^2}{\sigma_k^2}\right] \tag{3}$$

where Q is the number of Gaussians and A_k, μ_k, σ_k are characteristic parameters of each Gaussian.
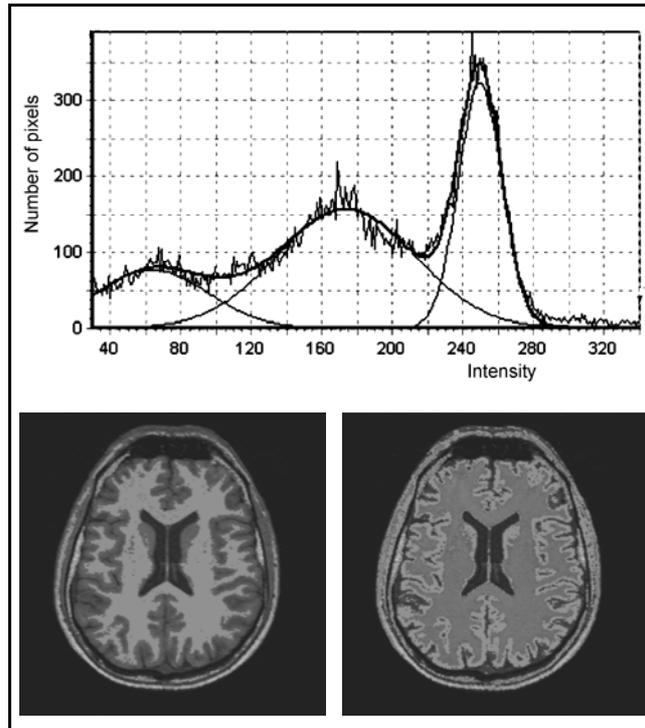

Fig. 2. A histogram of MR image after L-M optimization;
white and gray matter that have been segmented from the image

On the basis of these parameters, it is possible to find threshold values between peaks corresponding to a given tissue and the segmentation of these tissues from the image. The results of the white and gray matter segmentation are shown in Fig. 2.

5. Conclusions

Magnetic resonance imaging (MRI) is a powerful, non-invasive medical technique that providing highly-detailed images with excellent soft tissue contrast. With its ad-

vanced software applications, it allows accurate and high-precision medical diagnosis. The gray and white matter segmentation results obtained by L–M optimization of the image histogram brought us to the conclusion that the presented segmentation technique could be very useful for medical imaging. This solution allows one to analyse MR images quickly and precisely, hence it should be useful in a variety of medical applications, additionally to such simple and widely used methods as window/level or pseudocolours. Although MRI is very popular in clinics and hospitals all over the world, it is still quite a young and progressing technique in the diagnostics of human body.

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