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OPTIMIZATION TECHNIQUES IN THE MAGNETIC RESONANCE IMAGING

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Abstract The main purpose of this work is to analyze the Magnetic Resonance Imaging (MRI) methodologies used in non-invasive clinical investigation, with the aim of developing suitable denoising filtering techniques to improve the image quality and to allow the reduction of the acquisition time during an health examination. In this way the number of tests that can be performed in the same amount of time can be increased.

This work was carried out in collaboration with the staff of Martina Franca Hospital (Italy), which provided ourselves the material necessary for the development of the simulation algorithm and the sample images on which the algorithm has been tested.

Keywords: magnetic resonance imaging, brain image, anisotropic, diffusion filter, noise removal.

1. MRI BASICS

Nuclear magnetic resonance imaging (MRI) is a common clinical procedure used to scan whole body patient giving three dimensional images. Recent technological innovations have encouraged the use of this technology for noninvasive coronary, heart, and chest investigation or for research applications. The quality of images obtained from this technique depends on many factors.

MRI is based on magnetic properties of hydrogen that is the most common chemical element in the body; it is mainly in the water (H_20) content of the tissues. Hydrogen is the simplest element (one proton/electron pair) and it has the highest sensitivity to magnetic resonance with respect to any other chemical element.

The MRI scanner is a tube surrounded by a large circular magnet. The patient is placed on a moveable bed that is gradually inserted into the magnet. The magnet creates a strong magnetic field (ranging between 0.2 T and 1.5 T) that aligns the nuclear magnetic moments (also called spins) of hydrogen atoms in the human body which are normally randomly oriented. So the magnetic moments feel a torque proportional to the strength of the static magnetic field that causes the oscillation of these spins at a rotation frequency proportional to the field [1]:

$$\varpi_0 = -\gamma B_0 \qquad (1)$$

where ϖ_o is the precession frequency (called *Larmor frequency*), γ is the gyromagnetic ratio that is a constant specific of the nucleus and B_0 is the magnetic field strength.

Hence, in equilibrium, the net magnetization vector lies along the direction of the applied magnetic field B_o and is called equilibrium magnetization (M_o). The nuclear magnetization is very weak and so it is difficult to measure it when it is aligned with the strong static magnetic field. To overcome this problem, it is necessary to tip the moment away from the static field; this is accomplished by applying a time-varying magnetic field that lies in the plane transverse to B_o. The rotation angle of the magnetization is termed *flip angle* (φ).

Exposing the hydrogen atoms to suitable rotating radiofrequency field (RF pulses) with rotation frequency equal to ω_0, it is possible to perturb the equilibrium by tipping the magnetization in a plane perpendicular to the static magnetic field. In this way, the magnetization of the tissue under test can be measured. The particular excitation pulse that rotates the longitudinal magnetization into the xy-plane is termed *90° RF pulse*. The RF pulse is generated by a coil placed in *xy* plane used as a transmitter as well as a receiver.

If we place ourselves in a frame of reference that also rotates at precession frequency ω_0, this second field appears stationary. In this case the Larmor relation is still hold, and the precession frequency is proportional to the amplitude of RF pulse.

When the RF pulse turns off, the nuclear spins realign themselves with the static field, so the transverse magnetization (M_{xy}) decays and the longitudinal magnetization (M_z) recovers. The time constant describing how fast the longitudinal magnetization returns to its equilibrium value is often called *relaxation time* (T_1). In practice T_1 is the time required to realign a certain percentage of the tissue's nuclei and it is typically equal to about 1 s. It depends on the amplitude of magnetic field and on the characteristic properties of the tissue under examination.

The rate of recovery of the longitudinal component of the magnetization toward equilibrium after it has been perturbed by an RF pulse is given by [1]:

$$M_Z = M_0(1 - e^{-t/T_1}) \qquad (2)$$

It is possible to adjust some image acquisition parameters in order to emphasize the differences in

relaxation times associated to different tissues. In this case the image contrast will depends mainly on the T_1 time constant of the analyzed tissues and the MRI is referred as T_1 –weighted imaging.

Moreover, during relaxation, the nuclei loss energy emitting their own RF signal. This signal also known as *free-induction decay* (FID) can be detected by the same coil used to produce RF pulse. The FID is an oscillating signal with an exponentially decaying envelope. The time constant which describes the return to equilibrium of the transverse magnetization (M_{XY}), is called the *spin-spin relaxation time* (T_2) [1] and its trend is described by equation (3).

$$M_{XY} = M_0 e^{-t/T_2} \quad (3)$$

In this case it is possible to adjust some image acquisition parameters in order to emphasize the differences in T_2 times related to different tissues. The obtained MR images are termed as T_2 –weighted.

The transverse magnetization cannot be directly measured, because it decays quickly. So, it is necessary to generate an echo of the original signal. The delay between the creation of transverse magnetization and the detection of the echo is called *eco time* (T_E) [2].

In order to obtain all information to generate an MR image, it is necessary to excite several times each tissue with the RF pulse. The time between the application of an RF pulse and the successive, is called *repetition time* (T_R). This parameter affects the image contrast.

The signals received by the coils are recorded in a 3-D matrix representing the spatial frequencies of the image. By means of Fourier analysis of the signals [3],[4] we can obtain a map of spatial distribution of spins (called *k-space*).

2. DENOISING TECHNIQUES

Magnetic Resonance images are affected by random noise which limits the accuracy of any quantitative measurements on the data.

Many manufacturers of MR devices modify technical parameters to improve the image quality. The most commonly adopted technique is the increasing of the magnetic field. This solution decreases the noise but introduces some disadvantages such as: 1) a non uniform illumination of the images due to the difficulty to control the strength of the magnetic field; 2) the high power required to supply the device and 3) the high maintenance costs [5].

A lower cost solution consists in the application of some noise filtering technique before visual inspection or in the application of noise sensitive post-processing methods; in both cases it is possible to decrease significantly the image noise and simultaneously to preserve some fine details in the acquired images. The trade-off between noise reduction and the preservation of actual image features is a particularly delicate and difficult task [6].

The noise modelling is a crucial step in the filter developing. Conventional noise filtering schemes applied to MRI assume that in a first approximation the noise is Gaussian distributed.

However, several studies about statistical analysis of MRI have proved that the magnitude of MR data are Rice distributed [7], [8], [9]. Therefore starting from this assumption, suitable denoising filters were developed and applied to several MR images.

Many filtering methods are based on the signal averaging principle which uses the spatial redundancy in the image. In this sense, Gaussian filters have been largely used in MRI applications but they have the disadvantage to introduce of blurring edges. In fact, a low-pass filter reduces the amplitude of the noise fluctuations, but also degrades sharp details such as lines or edges. This type of filtering does not respect region boundaries or small structures, and the resulting images appear blurry and diffused.

To overcome this limit, in our study, a non-linear anisotropic diffusion denoising technique has been tested and its performances have been compared with common Gaussian filtering techniques. Anisotropic diffusion [9] is a selective and non-linear filtering technique which improves the quality of the image removing the noise while preserving and even enhancing details. The anisotropic diffusion employs the diffusion coefficients to determine the amount of smoothing that should be applied to each pixel of the image [5]. These coefficients depend on the gradient magnitude.

The diffusion process is described by means of the diffusion equation [10]:

$$I_{i,j}^{t+1} = I_{i,j}^{t} + \lambda\left(c_N \cdot \nabla_N I + c_S \cdot \nabla_S I + c_E \cdot \nabla_E I + c_W \cdot \nabla_W I\right) \quad (4)$$

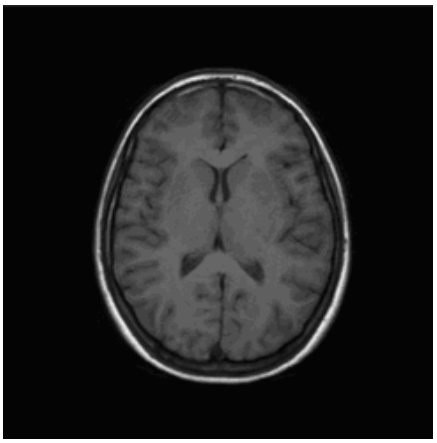
where $I_{i,j}^{t}$ is the intensity of the pixel at position *i,j* and time *t*, c_N, c_S, c_E, and c_W are the diffusion coefficients in the four directions (*north, south, east* and *west*), $\nabla_N, \nabla_S, \nabla_E$ and $\cdot\nabla_W$ are the nearest-neighbour differences in the four directions and λ is a coefficient which assures the stability of the model; it ranges in the interval [0-0.25]. The initial condition (t=0) of diffusion equation is the intensity pixels of original image.

The value of diffusion coefficients is determined taking into account the local properties of the image through the scales. The expected outcome of the proposed diffusion process is that the strong as well as the weak edges are well depicted and noise is effectively removed. For this reason, the filter action must be stronger on that image zones where there are no details (*homogeneous* zones) and weaker on image zones where edges are present (*edge* zones).

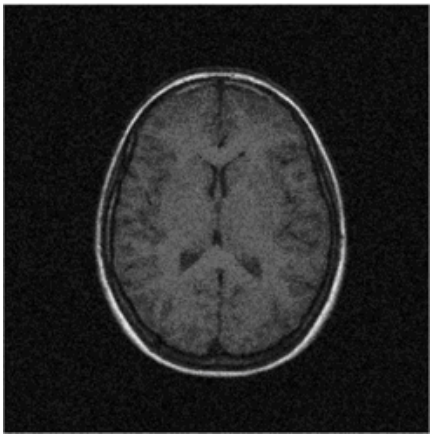
3. EXPERIMENTAL RESULTS

In our study we have analyzed some MR images of human brain which represent “the bottleneck” for clinical diagnostic, due to very long acquisition time. The pulse sequence used in the simulation algorithm is the Spin Eco [11] that is the most adopted sequence in analysis of brain images.

The MR images analysed (256x256 pixels of size) are acquired by means of a Philips Gyroscan 1.5 Tesla scanner set up in the Martina Franca Hospital. These images are in DICOM (Digital Imaging and Communications in Medicine) format, which is a standard for handling, storing,



(a)



(b)

Figure 1 - (a) Original image, (b) image corrupted with Rice noise

printing, and transmitting information in medical imaging. It includes a file format definition and a network communication protocol allowing easy information transfer among several sanitary structures [12]. The DICOM standard groups information into data sets. It consists of a number of attributes, including important information such as image size and format, acquisition parameters, equipment description, and patient information [13].

In our study, we have used a MATLAB based DICOM viewer to import the brain images (axial, coronal and sagittal views) supplied by the Martina Franca Hospital;

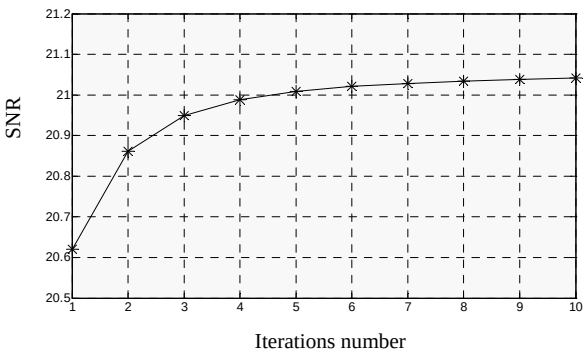


Figure 2 - SNR of filtered image versus iterations number

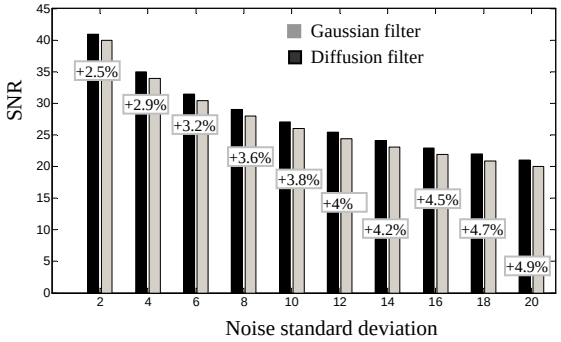


Figure 3 - SNR obtained by applying Gaussian and diffusion filter versus noise standard deviation

moreover we have derived the information about the values of characteristic parameters used in this structure to set the scanner. The protocol used by the staff of Martina Franca Hospital is divided in two main acquisition forms including the presence or the absence of the contrast agent that sometimes is used to improve the image brightness. We have analysed MR images acquired without the administration of the contrast agent because it is difficult to evaluate the variation of the characteristic parameters of the brain tissues due to this agent.

In a first step, all MR images were corrupted with Rice-distributed noise to simulate low quality images (Figure. 1). In particular the percentage of noise was varied from 1% to 20%. (typical range of MR image noise).

The estimation of the noise level in corrupted the image is based on calculation of the standard deviation of the pixels in the homogeneous zone [14]. For this reason the pixel indexes of the original image background identifying the zones where there is no signal ($I(i,j)$=0) are firstly calculated. Then these indexes are used to calculate the standard deviation in the corrupted image.

Several simulations were performed by applying the diffusion filter. To evaluate the effect of the filtering process the *Signal Noise Ratio* (SNR) was calculate by means of the following relationship.

$$SNR = 20\log\frac{\sqrt{\frac{1}{M\cdot N}\sum_{i=1}^{N}\sum_{j=1}^{M}\left[I(i,j)\right]^2}}{\sigma} \qquad (4)$$

where M and N are the number of the image rows and columns respectively, $I(i,j)$ is the pixel intensity of the original image and σ is the standard deviation of the noise.

In Figure 2 the trend of the *SNR* (obtained with the maximum noise level) versus the number of iterations is reported; analysing this trend is possible to estimate the minimum number of iterations for a good diffusion process,. The simulations show that using more than five iterations the reduction in *SNR* is negligible.

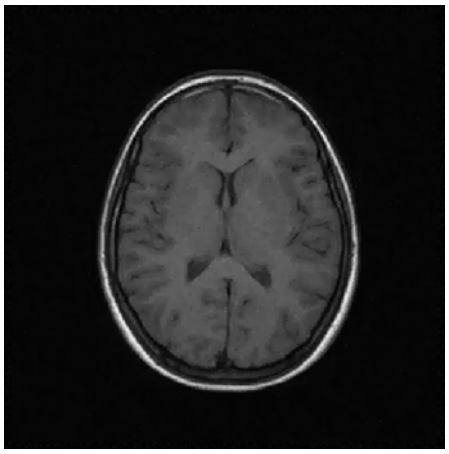


Figure 4 - Image filtered using diffusion filter and σ=20

Finally, Gaussian filter was tested and its performance was compared applying the diffusion filter. Figure 3 shows the results obtained applying the two filters for different noise standard deviation. It is possible to observe that diffusion filter offers always the greater SNR especially for high noise level. In this case the SNR rise about 5%.

Figure 4 shows an image obtained using diffusion filter for greater noise standard deviation.

4. CONCLUSIONS

MRI is a technique, widely used in medical settings to produce high quality images of the human body structure. For this feature, the MRI is the most powerful and flexible tool to diagnose several pathologies or physiological alterations of living tissues.

In this work we have tested different filtering techniques and compared their performances. The analysis of simulation on real DICOM images treated with Gaussian and diffusion filters permit us to estimate the value of acquisition parameters in order to obtain good images reducing the acquisition time.

The improvement of image quality by means of the application of denoising filter is an important issue in neuroimaging applications such as the computational analysis of brain structure and the assessment of potential pathologies.

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