FROM MAGNETIC RESONANCE IMAGING TO DIELECTRIC PROPERTIES OF TISSUES

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Abstract: A process to extrapolate the dielectric properties of human tissues from magnetic resonance imaging will be presented. The process is divided into a few steps which comprise the evaluation of the T_1 relaxation time and the water content of the tissue. Magnetic resonance image signals and parameters (e.g. image intensity values, relaxation times) will be addressed from a mathematical point of view. Afterwards, the water content evaluation (dependent on T_1 mapping) will be treated. Different ways to the same goal will be outlined and discussed in order to draw a roadmap for subsequent studies and refinements.

1 INTRODUCTION

Models of the human body are widely used in several kinds of studies. In particular, models which give information both on human anatomy and on tissues' dielectric properties (dosimetric human body models) are used in the evaluation of the electromagnetic field absorbed by the human body under several exposure conditions (Bernardi et al., 2004). As a consequence, during the last years, several dosimetric models have been generated, mainly starting from magnetic resonance tomography data (Christ et al., 2009; Dimbylow, 2002; Dimbylow, 1997).

However, the technique used to develop such kind of models is slow (due to iterations for organ identification and mesh building) and needs the involvement of people expert in human anatomy. For these reasons the developed models are commonly used as "average" human body models for evaluating electromagnetic power absorption in representative cases. On the contrary, it would be very useful to develop the body model of a specific person, by using fast and automatic techniques. One of the possible fields of application of customized body models is the treatment planning in thermal-ablation therapies (Goldberg et al., 2000). These therapies use the electromagnetic field radiated by a microwave antenna to remove unhealthy tissue (e.g. a tumor), by heating the target area up to temperatures of about 60 °C. If a customized model of the patient, including the tumor area, could be developed, a computer simulation could be run before the treatment, in order to optimize power and time of application to the specific tumor, according to its position within the body, shape, dimension, and dielectric properties.

In the following, an automatic procedure will be illustrated to derive information on the dielectric properties of body tissues from a Magnetic Resonance Imaging (MRI) scanning. The procedure has been first proposed in 2003 (Mazzurana et al., 2003) and it aims at linking the voxel (a 3D pixel) intensities, recorded in the Nuclear Magnetic Resonance (NMR) scanner, to the relaxation time of the tissue filling the corresponding body area. Then, from the relaxation time, the tissue water content is derived; finally, from the water content, the tissue dielectric properties are obtained.

Relationships between relaxation times, related to signal intensity, and water content of tissues have been reported in the literature (Fatouros and Marmarou, 1999; Whittall et al., 1997). The scope of this work is to present a possible way to implement the above mentioned tasks. Moreover, an optimization of the procedure previously proposed and alternatives to reach the same goal will be suggested. In fact, main limits of Mazzurana's approach can be found in the MRI sequence chosen and in the procedure used to evaluate dielectric properties from water content.

2 METHODS

The overall process (see figure 1) could be summarized as follows:

- From an MRI scanner a set of slices is obtained. Each slice is composed by a voxel matrix, which represents the intensity of the received signal.
- From the intensity of each voxel, relaxation times, which are linked to water content, can be evaluated.
- Some mathematical models are available (Schepps and Foster, 1980) to calculate the dielectric properties of a tissue starting from its water content.

The aim is to develop a computer code able to automatically perform all the necessary tasks to complete the job. To this end, a general choice was done: to use open source software. In fact, in order to have a product and a process suitable for medical operators, simplicity and performance should be joined. *Ad hoc* applications developed on open source platforms can realize both items. In the following, the implementation of the above cited steps will be presented, starting from the NMR images acquisition, up to the evaluation of the dielectric properties of tissues.

It is important to remark that some steps could be performed using different algorithms (e.g. relaxation time evaluation) or models (e.g. the equation to evaluate dielectric properties) depending on MRI acquisition subject and conditions (e.g. value of static electromagnetic field). In the following, the different steps will be presented both with reference to Mazzurana's work and discussing alternative ways to obtain the same information with lower computational loads.

Figure 1: Process Overview.

2.1 MRI Scanning

The MRI scanning supplies the acquired images in a specific standard, named "Digital Imaging and Communications in Medicine (DICOM)" (ACR-NEMA, 1985), which gives general information on the data scanning (DICOM header) and then yields the corresponding slice. An MRI scanner is able to use different sequences which can be modeled by the following equations and similar others.

$$S = \rho \frac{\left(1 - e^{-\frac{TR}{T_1}}\right) \sin\alpha}{1 - e^{-\frac{TR}{T_1}} \cos\alpha} e^{-\frac{TE}{T_2}} \tag{1}$$

$$S = k\rho e^{-\frac{TE}{T_2}} \left(1 - 2e^{-\frac{TR - \frac{TE}{2}}{T_1}} + e^{-\frac{TR}{T_1}}\right) \tag{2}$$

Equation (1) models a "Fast Low Angle SHot (FLASH)" MRI sequence where:

- α is the angle of which the net magnetization is rotated or tipped relative to the main magnetic field direction via the application of an RF excitation;
- ρ is the proton density;
- TR is the repetition time of the RF pulse, it is the amount of time that exists between successive pulse sequences applied to the same slice;
- TE is the echo time of the signal;
- T_1 is the longitudinal relaxation time constant, it indicates the time required to regain longitudinal magnetization following an RF pulse;
- T_2 is the transversal relaxation time constant, it is a measure of how long the resonating protons rotate "in phase" following an RF pulse;
- S is the intensity of the MRI signal.

Equation (2) models a Spin-Echo sequence where, besides parameters already cited, there is a k-factor which depends on other parameters and tissue properties. The Spin-Echo sequence has been used in (Mazzurana et al., 2003) while the FLASH sequence is reported here because of its characteristics: it increases T_1 weighting (when α is large).

2.2 Image Acquisition

To automatically load the NMR images, the "Insight Tool Kit (ITK)" software library has been used (Kitware, 1999). The software program acts as an orchestrator: it asks to ITK module to read the DICOM header, then it performs an auto-setting and, finally, it reads the intensity values building a matrix with the corresponding data (see figure 2).

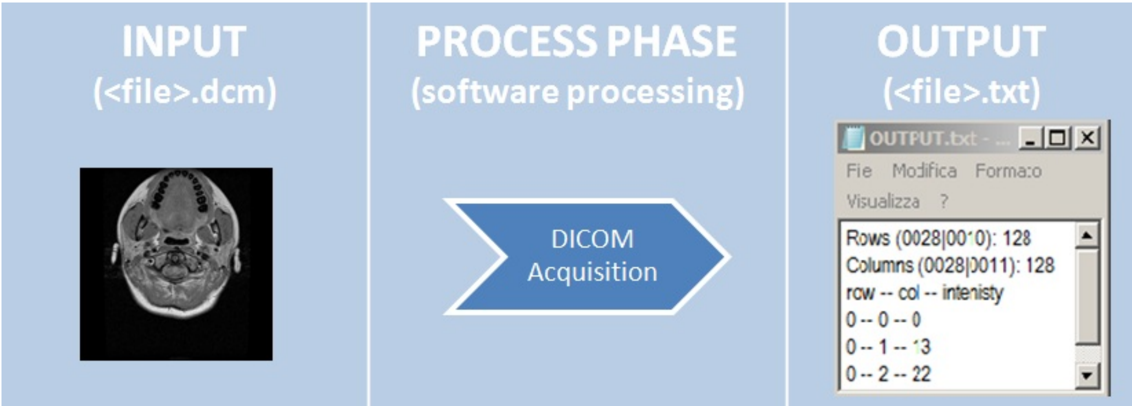

Figure 2: "DICOM Acquisition" details.

2.3 T_1 Evaluation

This step depends on the kind of sequence used in MRI scanning. According to the procedure introduced by Mazzurana and colleagues (Mazzurana et al., 2003), each slice must be acquired twice, with different sequence parameters. A Spin-Echo sequence using a short (450 ms) and a long (2100 ms) TR during the first and second slice acquisition was proposed. In fact, to derive the T_1 value from the signal S, some unknowns (e.g. relaxation time T_2, proton density ρ) must be cut out from eq. (2). It is possible to eliminate these ones considering the ratio between the signal S obtained, performing a double acquisition, where only one parameter is changed, e.g. the repetition time.

In this article two sequences are taken into account: the Spin-Echo (in order to follow Mazzurana's approach), and the FLASH sequence, which has been more recently proposed, and where the T_1 evaluation is easier.

Applying this procedure to the above reported FLASH and Spin-Echo sequences (eqs. (1) and (2)), the following equations can be derived (with a small and a large flip angle and with a short and a long TR, respectively):

$$\frac{S_{\alpha_1}}{S_{\alpha_2}} = \frac{1 - e^{-\frac{TR}{T_1}} \cos\alpha_2}{1 - e^{-\frac{TR}{T_1}} \cos\alpha_1} \frac{\sin\alpha_1}{\sin\alpha_2} \tag{3}$$

$$\frac{S_{TR_{short}}}{S_{TR_{long}}} = k_{ratio} \left(\frac{1 - 2e^{-\frac{TR_{short} - \frac{TE}{2}}{T_1}} + e^{-\frac{TR_{short}}{T_1}}}{1 - 2e^{-\frac{TR_{long} - \frac{TE}{2}}{T_1}} + e^{-\frac{TR_{long}}{T_1}}} \right) \tag{4}$$

According to Mazzurana and colleagues' procedure, T_1 should be derived from equation (4), i.e. from the inversion of a transcendental equation. To avoid this cumbersome operation, a look-up table can be built where the T_1 values are pre-evaluated for a selected range of values (see figure 3). This has been implemented first, and results will be shown in the following. However, as reported in eq. (3), a possible alternative way could be to use the FLASH sequence where the T1 value can be obtained more readily.

Whatever the way used, at the end of this step, a matrix filled with T_1 values is ready to be passed to the "water content evaluation" step.

2.4 Water Content Evaluation

For this evaluation, the equation proposed by Fatouros and Marmarou (1999) will be used, where A and B are parameters experimentally found out, and f_w is the water fraction (the unknown).

$$\frac{1}{f_w} = A + \frac{B}{T_1} \tag{5}$$

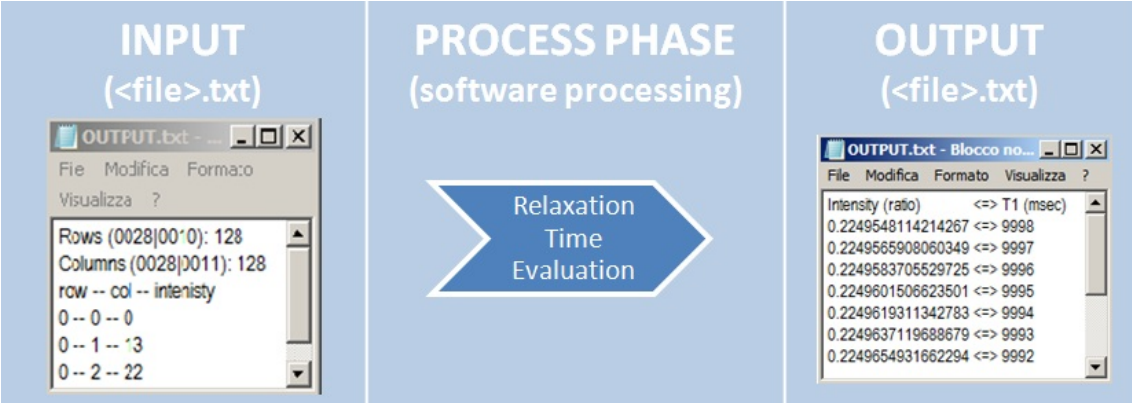

Figure 3: “Relaxation Time Evaluation” details.

A and B are factors that depend on magnetic field strength. Experimental values are available only for 1 Tesla (Fatouros and Marmarou, 1999). Eq. (5) comes out from a study which measures the brain water content by considering the influence of total water content, hydration fraction, and magnetic field strength. Fatouros and Marmarou verified the linearity between $1/T_1$ and $1/f_w$ both in gelatin solutions of varying water content and in an experimental animal model of brain edema.

As said before, this equation is not the only one available in the literature. Other and more recent studies (Shah et al., 2011) have recently been published. In particular, the relative amount of water in an image voxel is evaluated comparing the number of resonant protons in a region of interest (ROI) of 100% H_2O with the number of protons in a voxel.

2.5 Dielectric Property Evaluation

Schepps and Foster (1980) reported the dependence of tissue dielectric properties, as a function of frequency on water content. This equation, evaluated for each voxel, delivers the 3D dielectric human body model.

The assumption behind the method of Schepps and Foster for the calculation of dielectric constants from the water content of a tissue is that, in the microwave range, cell membranes have very low impedance and tissues can be compared to suspensions of proteins in water. Eq. (6) shows these relationships:

$$\varepsilon = \varepsilon_w \left(\frac{1-P}{1+(K-1)P} \right) \left(1 + \frac{KP\varepsilon_P}{\varepsilon_w(1-P)} \right) \tag{6}$$

where

- ε_w is the permittivity of water, it is given at a specific microwave frequency;
- P is the volume fraction of suspended solid,
- ε_P is the permittivity of the protein molecule;
- K is a factor which depends on geometry and ε_P.

Mazzurana and colleagues decided to proceed with the formulation of an empirical transfer function which relates directly the image signal with the relative permittivity and conductivity at a known frequency. Anyway, some verifications about this step took place.

3 RESULTS

Following the process described in the previous section, the first steps, up to the evaluation of the T_1 value, have been executed. Spin-Echo MRI scanning has been performed. The MRI scanner setup is reported in tables 1 and

2. The objects used for the test are gel tubes (provided by Diagnostic Sonar Ltd.) and a tube filled with water. All the tubes are put together in a phantom called Test Object 5 (TO5). In figure 4, the positions of gel tubes are shown. For each gel the relaxation times T_1 and T_2 are available.

Table 1: Scanner Setup for Spin-Echo sequence. Environment and general settings.

VARIABLE - PARAMETER	VALUE
Temperature	23 °C
Static EM Field	3 T
Scanning Sequence	Spin-Echo
Protocol Name	Turbo Spin-Echo
Matrix Size	512x512
Bits Allocated and Stored	16
Pixel Spacing	0.39

Table 2: Scanner Setup for Spin-Echo sequence. The same slice has been scanned with a different TR.

SERIES NUMBER	TR (ms)	TE (ms)
1	200	19
2	400	19
3	2000	19

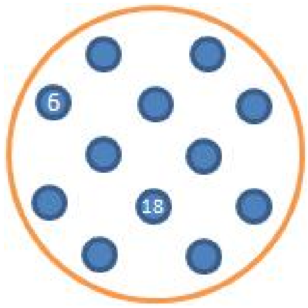

Figure 4: Gel 6 and 18 positions.

The MRI scanning phase produced a set of three DICOM files (see figure 5) which are fetched up by a software program whose task is to identify and extrapolate voxel information.

In table 3, the T_1 evaluations and the corresponding expected values are reported. For the uniformity of tube geometry and composition, it is enough to consider only a pixel of a slice for each tube. Providing three DICOM files to our software, two T_1 values will be obtained. Indeed, the program will perform an average of these two values.

The information about gel composition is not yet available: it is known only the gadolinium presence but not the amount. Moreover, as reported before, having used a 3T scanner, the A and B values are not defined. Thus, the third step (water content evaluation) could only be outlined, at this moment.

4 CONCLUSIONS

This article describes a semi automatic process to evaluate dielectric properties of a body part, starting from MRI images. The MRI images are manipulated in order to extrapolate information on the relaxation time T_1 shown

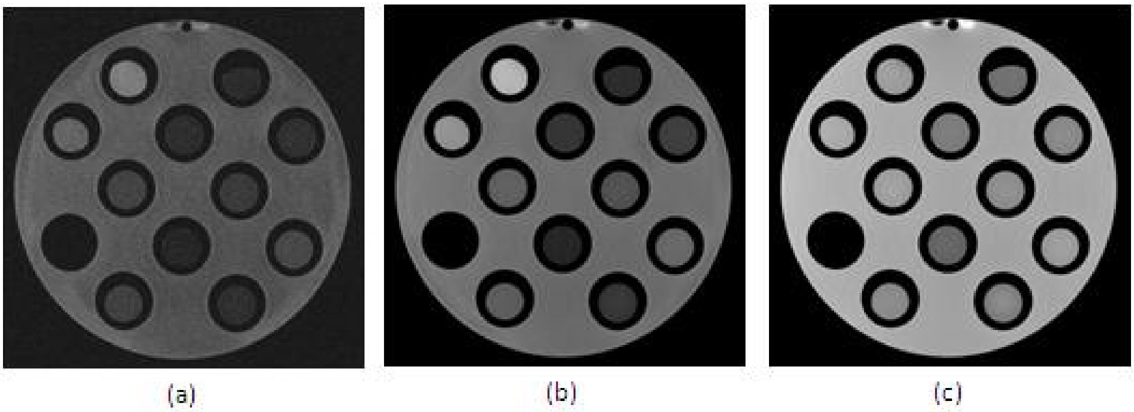

Figure 5: Spin-Echo DICOM series with repetition times of 200 ms (a), 400 ms (b), 2 s (c).

Table 3: T_1 values.

GEL ID	EXPECTED T_1 (ms)	T_1 FROM SPIN-ECHO SEQUENCE (ms)
6	444	396
18	1576	1584

by each voxel in the image. Then, the T_1 value can be linked to the water content of the tissue (voxel by voxel). Examples have been given on the T_1 evaluation of a gel material. New fast and reliable strategies to quantitatively measure the absolute water content (Shah et al., 2011) can be considered for the future. The successive step will be to link the tissue water content to the corresponding dielectric properties. Afterwards, further validations will be performed in order to show the applicability of the proposed procedure to automatically create electromagnetic human body models to be used in therapeutic procedure planning.

The proposed method allows to obtain electromagnetic models of part of the body of specific patients in order to optimize protocols in minimally invasive therapeutic techniques as microwave ablation. To the Author's knowledge, up to now, neither authomatic nor fast procedures are available to derive such models.

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