RADIOFREQUENCY POWER DEPOSITION DURING MAGNETIC RESONANCE DIAGNOSTIC EXAMINATIONS

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INTRODUCTION

Magnetic Resonance Imaging and Spectroscopy (MRI, MRS) require that subjects be exposed to a radiofrequency field, and the corresponding energy absorption leads to tissue heating. The main question, thus, to be considered in connection to safety and health aspects is related to the specific absorption rate (SAR) in the imaged subject and the exposure durations which might put a practical limit on the pulse sequence which can be used (1).

This explains why current safety regulations applying to MRI and MRS examinations give information on levels of exposure to radiofrequency electromagnetic fields associated with these diagnostic procedures (2). In particular, IRPA/INIRC is issuing guidelines giving some guidance on the magnitude of temperature increases at which no adverse health effects are expected and on the corresponding acceptable cumulative exposure (SAR times Exposure duration).

In this paper some models and experimental results for radiofrequency power deposition in MRI and MRS machines are reviewed. Models show that energy dissipation is a function of the frequency, RF incident power density, exposure duration, coupling between the RF coil and the subject, and several properties of the exposed tissue, including conductivity, dielectric constant, specific gravity, size, and orientation relative to the field polarization. The ability of the body's normal thermoregulatory responses to cope with high levels of RF energy deposition must be also taken into account (3).

THEORY

Simple theoretical estimates of the average, maximum, and spatial variation of the radiofrequency power deposition in terms of SAR during proton nuclear magnetic resonance imaging have been recently deduced for homogeneous spheres and for cylinders of biological tissues (4-6). Results have been obtained with uniformly penetrating linear RF fields directed axially and transverse to the cylindrical axis. In the RF frequency range below 100 MHz and under the near-field geometric conditions presently used in MRI devices, approximately 90% or more of the absorbed energy results from tissue currents induced by the magnetic component of the field (7). The specific absorption rate is related to the average induced electric field, $E/\sqrt{2}$, the tissue conductivity, σ, the tissue den-

(o) Member, IRPA International Non Ionizing Radiation Committee sity, ρ , and the duty cycle, D, of the applied RF field by the equation:

$$SAR = \sigma E^2 D/2\rho \qquad (1)$$

The duty cycle D is equal to t/T, where t is the pulse duration and T is the pulse repetition interval. Being the magnetic component of the field dominant in the low range of RF frequencies used for MRI, the sinusoidal induced electric field for a circular loop of tissue of radius R is $E = \pi\nu RB$, in which B is now understood to be the amplitude of magnetic component of the RF field perpendicular to the plane of the cross-section. In addition, the rotating component of the field, B_1, must satisfy the Larmor resonance condition. For an arbitrary flip angle, θ , SAR is given by the equation:

$$SAR = \frac{4kR^2\sigma\theta^2\nu^2}{\pi^2\rho\, tT} \qquad (2)$$

where k is equal to 6.81×10^{-19} in SI units.

The SARs predicted by Eq.(2) are plotted for a human torso model and for a human head model in Fig.1. The SAR at frequencies from 1 to 100 MHz is presented as a double logarithmic plot with $tT = t^2/D$ as the abscissa. It is evident that the SAR increases as the pulse repetition interval (T) decreases and/or the duty cycle (D) increases. At 30 MHz the SAR in the torso model equals the average body basal metabolic rate in the resting condition when t^2/D is approximately 10^{-5} s^2. The quadratic dependence of the SAR on the loop radius as shown in Eq.(2) has been confirmed experimentally (5). These diagrams provide useful interface between MRI and MRS spectrometer operating conditions and safety guidelines for RF power deposition.

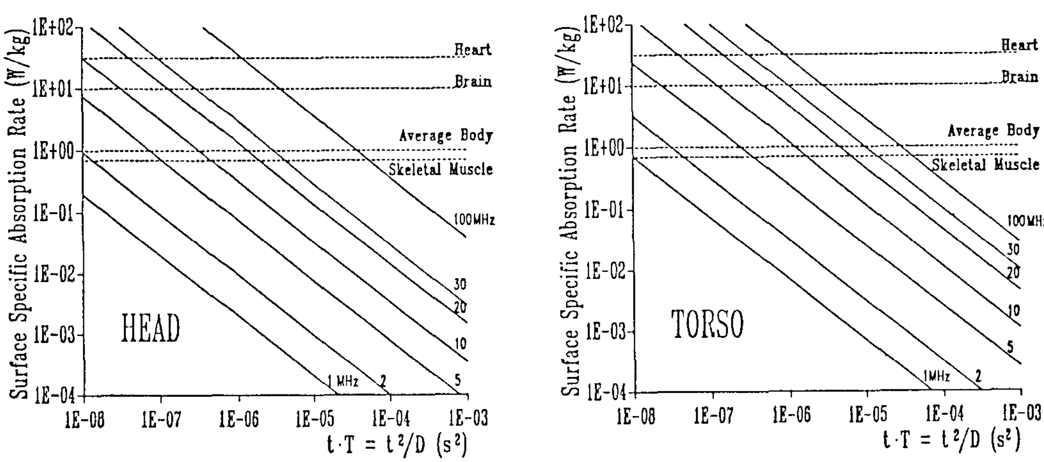

Fig.1 - Surface specific absorption rates predicted by Bottomley et al (5) for human head and torso models

A number of numerical techniques have also been developed to evaluate SAR in the human body, for which an analytical solution is impossible due to the complicate geometry. This techniques are generally based on the subdivision of the body, or parts of it, in cells small enough to assume that both the field intensity and the dielectric properties of the tissue are constant troughout each cell.

This approach in general requires large computer memories and long processing time. Recently a method has been developed which considerably reduces these requirements. This is based on the modeling of portions of the body using an impedance network. The application of circuit theory allows to determine internal currents, and therefore SAR, based on the knowledge of the dielectric constant and conductivity of each cell (8-10). The impedance method has been successfully applied to the evaluation of power deposition in magnetically induced hyperthermia. At present it is being exploited also for the evaluation of SAR in MRI, both in the ideal case of uniform magnetic field and in that of field generated by a real applicator.

EXPERIMENTAL DOSIMETRY STUDIES

Experimental measurements of SAR have been performed using a number of techniques, including infrared thermography and direct temperature measurements both in living specimens and in phantoms filled with saline or tissue-equivalent materials.

The simplest method to determine the SAR associated with the RF fields used in MRI and MRS is to measure the change in the quality factor Q of the coil upon introduction of the specimen. Bottomley et al (6) adopted this procedure to compare previous theoretical estimates with the experimental total power deposited in the bodies of nine adult male volunteers. The results for the average power deposition agree within about 20% for the exact model of the cylinder with axial field, when applied to the exposed torso volume enclosed by the RF coil. The average values predicted by the simple spherical and cylindrical models with axial fields, the exact cylindrical model with transverse field, and the simple truncated cylinder model with transverse field were about two to three times that measured, while the simple model consisting of an infinitely long cylinder with transverse field gave results about six times that measured.

Recently, Guy and Shuman (11) performed MRI dosimetry studies based on infrared thermography using a full scale phantom human body and spherical head models. The technique involved the fabrication of a phantom model with a synthetic liquid tissue for quantifying average SAR and a gel tissue for determining SAR distribution within the exposed body. The resulting change in temperature is related to the SAR by the known thermal properties of the synthetic tissue, and a useful map of the energy deposition pattern can be obtained.

Superficial and deep tissue heating during high specific absorption rate RF irradiation has been measured in dogs to see if significant temperature changes could be produced by a clinical MRI device operating at 1.5 T (12). Temperature probes were placed in the subcutaneous tissues of the dog axilla and groin, in the muscle of the neck and thigh, in the liver and mid-peritoneum, and in the urinary bladder. Temperatures were recorded before, during, and after a 26 minute spin-echo sequence. During RF exposure, in each dog all tissues experienced a linear temperature change of several degrees; maximal average change was 4.6° C in the urinary bladder. Deep tissue heating was slightly greater than that of superficial tissues.

CONCLUSIONS

Many authors have determined, both theoretically and experimentally , the RF power deposition within patients and animal models. These studies have shown that SAR levels resulting from human exposure to maximum possible MRI and MRS RF magnetic fields can exceed limits currently recommended.

It appears possible that multi-echo, multi-slice techniques may well deposit dangerous amounts of power. According to Gandhi et al (10), hot spots can also be expected due to constriction of current paths associated with some anatomically thin regions not insulated from large current paths.

The main conclusion is that when imaging infants, cardiac compromised patients, and patients with altered thermoregulation, particular attention should be paid to SAR. In addition, relatively low room temperature and humidity should be maintained along with good airflow through the magnet bore (8).

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