RF Magnetic Field Penetration, Phase Shift and Power Dissipation in Biological Tissue: Implications for NMR Imaging

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Received 22 December 1977, in final form 23 February 1978.

ABSTRACT. The magnetic field penetration, phase shift and power deposition in planar and cylindrical models of biological tissue exposed to a sinusoidal time-dependent magnetic field have been investigated theoretically over the frequency range 1 to 100 MHz. The results are based on measurements of the relative permittivity and resistivity dispersions of a variety of freshly excised rat tissue at 37 and 25 °C, and are analysed in terms of their implications for human body nuclear magnetic resonance (NMR) imaging.

The results indicate that at NMR operating frequencies much greater than about 30 MHz, magnetic field amplitude and phase variations experienced by the nuclei may cause serious distortions in an image of a human torso. The maximum power deposition envisaged during an NMR imaging experiment on a human torso is likely to be comparable to existing long-term safe exposure levels, and will depend ultimately on the imaging technique and NMR frequency employed.

1. Introduction

A recent advance in the nuclear magnetic resonance (NMR) technique has been to produce cross-sectional images which represent the spatial distribution of mobile protons in heterogeneous systems (Lauterbur 1974, Kumar, Welti and Ernst 1975, Hinshaw 1976, Andrew, Bottomley, Hinshaw, Holland and Simaroj 1977, Mansfield and Maudsley 1977b). This new development, combined with the observation that excised biological tissues exhibit different NMR properties depending on their health and origin (Hazelwood, Cleveland and Medina 1974, Hollis, Saryan, Eggleston and Morris 1975, Kiricuta and Simplaceanu 1975), has led to the suggestion that the method may prove a useful tool in medical diagnosis (Grannell and Mansfield 1975, Damadian, Goldsmith and Minkoff 1977, Hinshaw, Bottomley and Holland, 1977, Holland, Bottomley and Hinshaw 1977, Mansfield and Maudsley 1977a) and several laboratories are at present engaged in the development of NMR imaging systems capable of scanning the whole human body (Andrew 1977).

Since objects of such size are of comparable dimension to the electromagnetic skin depth in biological tissue at radiofrequencies (RF) (Johnson and Guy 1972), skin effects arising from the resonant RF magnetic field cannot be neglected and any consequent amplitude or phase variation of this field towards the centre of the object will have a two-fold effect on the medical viability of NMR imaging. Firstly, it will tend to degrade the image quality by altering the nature of the NMR signal derived from this region, and, secondly, the
power loss associated with the induced fields produced by the RF pulse will result in heating of the specimen. In medical applications this heating should be kept below recognised danger levels.

Although the effects of RF electromagnetic (EM) radiation on biological tissue are fairly well documented (Schwan 1965, Johnson and Guy 1972, Michaelson 1972, Lin, Guy and Johnson 1973), the NMR imaging situation differs in that the resonant applied alternating magnetic field has only a very small associated electric field so that the electrically induced component of the absorbed power is negligible in comparison with the magnetically induced component. In addition, variations in phase, as well as amplitude, are important where phase-sensitive detection is employed (Hinshaw 1976, Mansfield, Maudsley and Baines 1976).

The present paper represents a first attempt at quantifying the extent of RF magnetic field penetration, phase shift and power deposition in biological tissue samples during an NMR imaging experiment. Calculations showing the behaviour of the magnetic field within samples assume semi-infinite planar (Bottomley 1977) and infinitely long cylindrical models of homogeneous biological tissue. The latter model is particularly relevant to thin section NMR imaging (Hinshaw et al. 1977, Mansfield and Maudsley 1977, Hinshaw, Andrew, Bottomley, Holland, Moore and Worthington 1978) where end effects can be ignored in all but the extremities of a specimen. This is because only the NMR signal derived from a well defined cross-section is of interest. The power deposition estimates also assume a cylindrical homogeneous tissue model. All of these calculations assume a uniform applied RF field, and refer to the distortions and power deposition of the RF field prior to NMR absorption. The calculations are based on measurements of the resistivity and relative permittivity of freshly excised rat lung, brain, liver, hepatoma D23, kidney, heart and abdominal wall muscle tissues at 37 and 25 °C over the frequency range 1 to 100 MHz using a technique developed in this laboratory (Bottomley 1978).

2. Theory

2.1. Semi infinite planar model

The magnetic induction \( \mathbf{B} \), in a linear, homogeneous and isotropic medium must satisfy the wave equation

\[
\nabla^2 \mathbf{B} - \epsilon \mu \frac{\partial^2 \mathbf{B}}{\partial t^2} - \frac{\mu}{\rho} \frac{\partial \mathbf{B}}{\partial t} = 0
\]

(1)

where \( \epsilon \) is the permittivity, \( \mu \) the permeability and \( \rho \) the resistivity of the medium. The solution of eqn (1) which describes the behaviour of a sinusoidal time-dependent magnetic induction field in a semi-infinite plane of conducting material is the same as that which describes the propagation of plane electromagnetic waves in conducting media (Lorrain and Corson 1970). Thus if the conducting material which occupies the region \( x > 0 \) (fig. 1(a)) is subjected to
the magnetic induction field
\[ \mathbf{B}(x \geq 0) = B_0 \exp(j\omega t) \mathbf{\hat{z}} \] (2)
then within the medium
\[ \mathbf{B}(x \geq 0) = B_0 \exp[-K_I x + j(\omega t - K_R x)] \mathbf{\hat{z}} \] (3)
where
\[ K_I = \omega \left( \frac{1}{2} \mu \left( \left[ 1 + \left( \frac{1}{\rho^2} \epsilon^2 \omega^2 \right) \right] + 1 \right) \right) \] (4)
\[ K_R = \omega \left( \frac{1}{2} \mu \left( \left[ 1 + \left( \frac{1}{\rho^2} \epsilon^2 \omega^2 \right) \right] + 1 \right) \right) \] (5)
and \( \omega \) is the angular frequency, and where it is understood that we are considering only the real parts of eqns (2) and (3).

2.2. Cylindrical model
Consider an infinitely long cylinder of radius \( r_0 \) of the conducting material coaxial with the z-axis and immersed in the alternating magnetic field described by eqn (2). Cylindrical polar coordinates \( r, \theta \) and \( z \) are employed (fig. 1(b)). By symmetry, the field inside the cylinder will be a function of the radial position coordinate only, and if we assume a solution of (1) of the form
\[ \mathbf{B} = B_0 R(r) \exp(j\omega t) \mathbf{\hat{z}} \] (6)
then the function \( R(r) \) must satisfy the modified Bessel equation
\[ \frac{\partial^2 R}{\partial r^2} + \frac{1}{r} \frac{\partial R}{\partial r} - K^2 R = 0 \] (7)
of order zero and independent variable \( Kr \), where \( K = K_I + j K_R \) is the complex
wave number. The solution of (7) that remains finite for all \( r \leq r_0 \) is

\[
R(r) = \frac{I_0(Kr)}{I_0(Kr_0)}
\]

where \( I_0 \) is a modified Bessel function of the first kind of order zero, and the integration constant, \( [I_0(Kr_0)]^{-1} \) has been chosen so that \( R(r) = 1 \) at \( r = r_0 \). This result is similar to that which describes uniform cylindrical electromagnetic waves with the \( E \) or \( B \) vectors parallel to the axis (Schelkunoff 1943).

Eqn (8) can be expressed in polar form by expanding the modified Bessel function into a power series (Spiegel 1968) and invoking De Moivre’s formula. Thus the solution (6) can be written

\[
\mathbf{B} = B_0 \frac{|I_0(Kr)|}{|I_0(Kr_0)|} \exp[i(\omega t - \xi(r))] \hat{z}
\]

where

\[
|I_0(Kr)| = \sqrt{\text{Re} I_0(Kr)^2 + \text{Im} I_0(Kr)^2}
\]

\[
\xi(r) = \arctan \left( \frac{\text{Im} I_0(Kr)}{\text{Re} I_0(Kr)} \right) - \arctan \left( \frac{\text{Im} I_0(Kr_0)}{\text{Re} I_0(Kr_0)} \right)
\]

\[
\text{Re} I_0(Kr) = 1 + \frac{s^2 r^2}{2^2} \cos 2\phi + \frac{s^4 r^4}{2^2 4^2} \cos 4\phi + \frac{s^6 r^6}{2^2 4^2 6^2} \cos 6\phi + \ldots
\]

\[
\text{Im} I_0(Kr) = \frac{s^2 r^2}{2^2} \sin 2\phi + \frac{s^4 r^4}{2^2 4^2} \sin 4\phi + \frac{s^6 r^6}{2^2 4^2 6^2} \sin 6\phi + \ldots
\]

\[
s = (K_r^2 + K_i^2)^{\frac{1}{2}} \quad \text{and} \quad \phi = \arctan \left( \frac{K_r}{K_i} \right)
\]

Here the attenuation in amplitude of the magnetic field is described by the ratio \( |I_0(Kr)|/|I_0(Kr_0)| \) and the phase variation is described by \( \xi(r) \). The infinite series \( \text{Re} I_0(Kr) \) and \( \text{Im} I_0(Kr) \) converge rapidly in practice.

2.3. Power deposition

Power losses in the form of joule heating within the specimen will derive from eddy currents induced by the alternating magnetic field. In the above configuration, the induced electric field \( E \), as deduced from Faraday’s Law, is

\[
E = -\frac{j\omega B_0 I_1(Kr) \exp(j\omega t)}{K_1(Kr_0)} \hat{\phi}
\]

where \( I_1 \) is a modified Bessel function of the first kind of order one (Schelkunoff 1943). The absorbed power density resulting from both conduction and displacement current losses is given by (Johnson and Guy 1972)

\[
P = \frac{1}{2} \sigma |E|^2
\]
where \( \sigma = 1/\rho \) is the effective (measured) conductivity of the tissue, and

\[
|E| = \left| \frac{\omega B_0 I_0(Kr)}{K I_0(Kr_0)} \right|
\]

is the amplitude of the real part of \((10)\).

If the field is pulsed, as in an NMR imaging experiment, the power density is averaged over a complete pulse cycle (Schwan 1965). Thus, if \( T \) and \( \tau \) are respectively the NMR pulse length and the pulse repetition period, the time averaged absorbed power density is

\[
P = \frac{\sigma T}{\tau} |E|^2.
\]

It may be noted that the amplitude of the RF field, \( B_0 \), is also related to the pulse length, \( T \), via the Larmor equation, and for 90° pulses at the proton resonance, \( B_0 = 5.87 \times 10^{-9}/T \) tesla, \( T \) in seconds.

3. Experiment

The values of the permittivity and resistivity of biological tissue used in the present investigation were obtained from measurements of the tissue impedance using a type 4815A Hewlett-Packard RF vector impedance meter (Bottomley 1978). The measurements were performed at frequencies of 1, 2, 3, 4, 5, 10, 15, 20, 25, 30, 40, 50, 60, 70, 80, 90 and 100 MHz at 37 °C and 25 °C. The temperature was controlled to within \( \pm 1 \) °C during measurements by blowing dry heated air over the meter probe and specimen cell assembly.

The resistivity and relative permittivity dispersions of seven different tissue types at 37 °C are shown in figs 2(a) and 2(b). All of the tissue samples were taken from freshly killed white Wistar rats and all measurements were completed within six hours of death. The curves shown for each tissue type represent an average from measurements taken from six tissue samples. Individual samples exhibited curves which deviated from these mean curves by less than 5%. This can be attributed to a combination of experimental error and the natural heterogeneity of biological specimens. Temperature dependence over the range 25 °C to 37 °C is approximately \(-1\% \) per °C for the resistivity and approximately \(+1.2\% \) per °C at 1 MHz decreasing to approximately \(-0.3\% \) per °C at 100 MHz for the relative permittivity of the tissues.

These experimental results are in good agreement with results measured at 25, 50 and 100 MHz by Osswald (1937) and with data collected by Schwan (1965). A possible cause of the difference between the results for healthy and tumorous liver tissues is the increase in water content and change in the ratio of free to bound water of tumorous tissue relative to its host (Kiricuta and Simplaceanu 1975).
RF Magnetic Field Behaviour in Tissue

Fig. 2. Resistivity (a) and relative permittivity (b) of rat lung (A), brain (B), liver (C), kidney (D), heart (E), liver hepatoma D23 tumour (F) and abdominal wall muscle (G) tissues at 37 °C as a function of frequency.

4. Results

Curves showing the variation of the skin-depth of penetration of the RF magnetic field from 1 to 100 MHz in the seven tissue types at 37 °C are shown in fig. 3(a). The corresponding phase shifts at one skin depth and 37 °C are shown in fig. 3(b) for the planar tissue model. The curves are calculated from eqns (3), (4) and (5) with the data shown in fig. 2 and assuming a tissue magnetic permeability equivalent to that of free space. The skin depths of abdominal
Fig. 3. Skin depths (a) and corresponding phase shifts at one skin depth (b) of the tissues as a function of frequency at 37 °C.

wall muscle, tumour, heart, liver and kidney at 1, 10, 27, 41 and 100 MHz are in tolerable agreement with values characteristic of tissue of high water content published by Johnson and Guy (1972). However, the large variations in skin depth with tissue type observed here emphasise the difficulty in assigning a single meaningful representative value to such a broad range of tissues.

The behaviour of the RF magnetic field within a cylinder of tissue at 37 °C, as described by eqn (9), is illustrated in figs 4 and 5. In figs 4(a), 4(b) and 4(c) the field amplitude, phase shift and combined amplitude–phase variation at the axis of tissue cylinders of radius 20 cm, as a function of frequency, are shown. Cylinders of this size give a fair approximation to an average human torso. Fig. 5(a) shows the amplitude–phase variation at the axis of a muscle cylinder
Fig. 4. Magnetic field amplitude (a), phase (b) and amplitude-phase (c) variation at the axis of tissue cylinders of radius 20 cm as a function of frequency. These are represented by the functions $|I_0(Kr_0)|^{-1}$, $\xi(0)$ and $|I_0(Kr_0)|^{-1} \cos(\xi)$ respectively.
Fig. 5. Part (a) shows the amplitude–phase variation \( \left( \frac{I_0}{K_r} \right) \cos (\xi) \) at the axis of a muscle cylinder as a function of the cylinder radius for different frequencies. Part (b) shows the amplitude–phase variation \( \left( \frac{I_0}{K_r} \right) \cos (\xi) \) along the radius of a muscle cylinder of fixed radius 20 cm for different frequencies.

as a function of the cylinder radius for different frequencies. Fig. 5(b) shows the amplitude–phase variation along the radius of a muscle cylinder of fixed radius 20 cm for different frequencies.

The results for the cylindrical tissue model differ significantly from those for the planar tissue model. Within the cylinder, the variation in field amplitude with position is far from the exponential decay exhibited in the planar model (eqn 3) and may even increase towards the centre of the cylinder. This increase is due to a combination of the reduced wavelength in the medium and the radius of curvature which can produce a focussing of the RF field that more than compensates for the conduction and displacement current losses. The zero’s in the curves of figs 4(c), 5(a) and 5(b) occur at phase shifts of \( n\pi/2 \), where \( n \) is an odd positive integer.

The absorbed power density associated with the induced eddy current losses in a 20 cm radius tissue cylinder is shown in fig. 6. The calculated curves (eqn (11)) assume that the cylinder is subjected to a string of 10 μs 90° RF pulses at the proton resonance, spaced at intervals of 10 ms. This is comparable to our existing NMR image operating conditions (Hinshaw et al. 1978). Power absorption estimates for different \( T \) or \( \tau \) values can be obtained from fig. (6) by appropriate scaling as indicated by eqn (11).

Greatest power absorption occurs at the surface of the cylinder where the eddy current loops are linked by the greatest flux. Within the cylinder, the power absorption density decreases monotonically to zero at the axis where the flux linkage is zero. For muscle tissue this is shown in fig 6(a) for different frequencies. The surface absorbed power density for all of the tissues as a function of frequency is shown in fig. 6(b). At the higher frequencies the surface power absorption is less than would be expected from a low frequency extrapolation because the magnetic field penetration is reduced, thus giving rise to a decreased flux linkage.
Fig. 6. The absorbed power density along the radius of a muscle cylinder of fixed radius 20 cm for different frequencies is shown in part (a). Part (b) shows the absorbed power density at the surface of the 20 cm radius cylinder for most of the tissues as a function of frequency. The curve for heart falls midway between those of muscle and kidney and the curve for tumour lies midway between that of muscle and heart. The calculated curves assume $T$ and $\tau$ are $10 \mu s$ and $10$ ms respectively.
5. Discussion

In the planar tissue model, fig. 3 indicates that a skin depth of 20 cm or less is approached by most tissues at frequencies of approximately 20 MHz, with corresponding phase shifts of around 80°. Remembering that the skin depth represents an attenuation in amplitude to 37% of that at the surface, it would appear that we might expect serious RF field attenuation and phase problems to occur at frequencies of 20 MHz and higher in human body sized NMR imaging experiments. The results from the cylindrical tissue model lend qualitative support to this conclusion although the RF field attenuation is less severe owing to the surface curvature.

From the NMR imaging viewpoint, an attenuation in amplitude to 37% at the centre of the specimen will mean that an NMR pulse which is 90° at the surface will be reduced to a 33° pulse at the centre resulting in a 45% reduction in the NMR signal. Similarly, an attenuation of 50% results in a 30% reduction in the NMR signal at the centre. For a 20 cm radius muscle cylinder, the latter condition is reached at a frequency of nearly 60 MHz (fig. 4(a)). The effect on an NMR image would be to produce ‘fading’ or loss of sensitivity in this region.

Phase variations in the RF field throughout a specimen will produce distortions in the NMR signal and hence the resultant NMR image. For example, if the NMR signal is pure absorption mode at the cylinder surface, it will be pure dispersion mode at regions where the phase shift is 90°, and inverted at regions where the phase shift is 180°, thus giving rise to a large negative signal in these regions. The extent of the image deterioration caused by these phase variations will depend upon the imaging technique employed. A possible solution to this problem would be to use the power spectrum of the NMR signal. For the cylindrical model, phase shifts of $\pi/4$, $\pi/2$ and $\pi$ are reached at frequencies of 14, 30 and 74 MHz respectively, within the 20 cm muscle cylinder (fig. 4(b)).

Present power density safety standards are all expressed in terms of flux levels. The maximum recommended safe power density flux for long term human exposure in the United Kingdom is 10 mW cm$^{-2}$ (Ministry of Technology 1968). This corresponds to an absorbed power density of order 1 mW cm$^{-3}$ or a current density of approximately 3 mA cm$^{-2}$ (Schwan 1972). The surface power absorption in an NMR imaging experiment operating under the specified conditions exceeds this figure at frequencies greater than about 5 MHz (fig. 6(a)) for the 20 cm radius tissue cylinders. The averaged power absorption throughout the cylinders exceeds this figure at frequencies greater than about 10 MHz. Different imaging techniques may increase or reduce power absorption levels. For example, with $T$ and $\tau$ values of 60 and 800 $\mu$s respectively as employed in the technique of Damadian et al. (1977), the above absorbed power level estimates would be doubled, and the surface power absorption would exceed 1 mW cm$^{-3}$ at operating frequencies greater than about 3-5 MHz. It should be noted however that power levels much greater than the recommended value are used in clinical diathermy treatment of many areas of the human body (Schwan 1965, Johnson and Guy 1972).
Finally, there are several additional considerations which must be taken into account when applying the above results to a real human body sized NMR imaging situation. Firstly, there are the problems which arise from the assumptions made in the derivation of the simplified theory presented. These are: (i) that the applied RF field is perfectly uniform and unaffected by the specimen and (ii) that the specimen is composed of a single homogeneous tissue infinite in extent and either of planar or cylindrical geometry. In heterogeneously composed specimens, reflections and refractions at tissue interfaces should also be taken into account (e.g. Kritikos and Schwan 1976). Secondly, we have considered only the behaviour of the RF field at the resonating nuclei. The NMR signal itself may undergo additional distortions prior to detection. There will also be an additional power absorption component resulting from nuclear magnetic relaxation processes, but this is likely to be several orders of magnitude below that due to eddy–current losses (Grannell and Mansfield 1975).

6. Conclusion

Evidence based on simple planar and cylindrical tissue models indicates that the distortion in both amplitude and phase of the resonant magnetic field is sufficient to cause serious problems in NMR imaging of the human body at frequencies greater than 30 MHz. Should it be desired to operate at such frequencies in order to improve the signal-to-noise ratio for example (Hinshaw 1976), it may be possible to correct for these distortions using the theory and results presented.

Power deposition during an NMR imaging experiment on a human torso will depend on the imaging technique employed. Although the power deposition estimates presented relate to our existing NMR image operating conditions, with a simple modification they can be made to apply to other NMR imaging systems. With the operating conditions specified, the maximum recommended safe power density for long term human exposure is exceeded at NMR imaging frequencies greater than 5 MHz. However, the power levels achieved below 100 MHz are no greater than those used in clinical diathermy treatment, and with the relatively short exposure times envisaged for a completed image (Andrew et al. 1977, Hinshaw et al. 1977), it is improbable that RF heating will occur at dangerous levels.

We would like to thank Dr. M. R. Price of the Nottingham University Cancer Research Laboratory for providing the tissue samples. The work described forms part of an MRC funded project on NMR imaging undertaken by a group consisting of E. R. Andrew, P. A. Bottomley, W. S. Hinshaw, G. N. Holland and W. S. Moore.

RÉSUMÉ

Pénétration à haute fréquence de champ magnétique, déphasage et dissipation de puissance dans les tissus biologiques: implications pour la représentation de la RMN

La pénétration d’un champ magnétique, le déphasage et le dépôt de puissance dans des modèles planaires et cylindriques de tissus biologiques exposés à un champ magnétique sinusoidal à dépen-
dance temporelle ont été étudiées sur le plan théorique sur une gamme de fréquences de 1 à 100 MHz. Les résultats sont fondés sur des mesures des dispersions relatives de la constante diélectrique et de la résistivité sur divers tissus de râie récemment excisés, à 37 et 25°C, et leurs implications pour la représentation de la résonance magnétique nucléaire (RMN) du corps humain sont analysées.

Les résultats indiquent qu'à des fréquences de RMN beaucoup plus élevées que celles se situant aux alentours de 30 MHz, l'amplitude du champ magnétique et les variations de phase subies par les noyaux sont susceptibles de provoquer de graves distorsions sur une image du tarse humain. Il est probable que le dépôt maximum de puissance envisagé pendant une expérience de représentation de RMN sur un tarse humain est comparable aux niveaux d'exposition existants sans danger à long terme, et qu'il dépendra en définitive de la technique de représentation et de la fréquence de RMN utilisée.

ZUSAMMENFASSUNG

Durchdringung des Magnetfeldes im Hochfrequenzbereich, Phasenverschiebung und Energiezerstreuung in biologischen Geweben: Bedeutung für die kernmagnetische Resonanzabbildung

Theoretisch untersucht wurde die Durchdringung des magnetischen Feldes, die Phasenverschiebung und Energieablagung in ebenen und zylindrischen Modellen biologischer Gewebe, die einem sinusförmigen, zeitabhängigen Magnetfeld ausgesetzt wurden. Der bei dieser Untersuchung zugrundegelegte Frequenzbereich war 1 bis 100 mHz. Die Ergebnisse stützen sich auf Messungen der relativen Permissivität und der Resistivitäten bzw. deren Verteilung an einer Vielzahl von Geweben, die unmittelbar vor der Messung bei 37 und 25°C an Ratten ausgeschnitten wurden. Der Analyse erfolgte mit Hinsicht auf die Bedeutung, die für die kernmagnetische Abbildung am menschlichen Körper besteht.

Die Ergebnisse deuten an, dass bei kernmagnetischen Resonanzfrequenzen, die wesentlich über 30 mHz liegen, die magnetischen Feldamplituden und die Phasenunterschiede, die die Krone auf weisen, zu schwierig werdenden Verzerrungen bei der Abbildung des menschlichen Körpers führen können. Die maximal zu verursachende Energieabladung bei der kernmagnetischen Resonanzabbildung am menschlichen Körper dürfte denjenigen Werten nahekommern, die für die auf langer Dauer vom Menschen ohne Schaden zu ertragenden Werte der Energieführung gelten. Letztlich hängen diese Werte wohl von der verwandten Abbildungsmethode und der kernmagnetischen Resonanzfrequenz ab.

REFERENCES


RF Magnetic Field Behaviour in Tissue


