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Phil. Trans. R. Soc. Lond. B 1980 **289**, 511-518
doi: 10.1098/rstb.1980.0070

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Phil. Trans. R. Soc. Lond. B **289**, 511–518 (1980) [511]
Printed in Great Britain

Experimental considerations in implementing a whole body multiple sensitive point nuclear magnetic resonance imaging system

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The multiple sensitive point technique of n.m.r. imaging is discussed in terms of the mode of n.m.r. excitation and detection, provision of field gradients and the methods of data processing. The design criteria of an instrument to implement this and related imaging methods at whole body size are described and particular attention is paid to techniques employed to increase the versatility and ease of use of the apparatus.

Introduction

The scaling up of a nuclear magnetic resonance (n.m.r.) system to whole body sizes presents special problems due to the low frequency, large size and inherently low sensitivity of the experiment; this paper discusses the design criteria of such an apparatus.

We begin by giving a brief explanation of the mode of operation of the existing n.m.r. imaging apparatus, developed in our laboratory over the last three years, which can produce n.m.r. images of objects of up to 8 cm diameter. The image production technique is a development of the sensitive point method of Hinshaw & Moore (1974) and Hinshaw (1974, 1976) and the steady state free precession (s.f.p.) technique is used to acquire the n.m.r. signals (Hinshaw 1979). The images, of which numerous examples have been published elsewhere (Hinshaw *et al.* 1977, 1978, 1979), made on this apparatus show the theoretical expected resolution of 128×128 picture points. On the basis of our experience with this apparatus, we go on to discuss the factors that have governed our design of the scaled-up apparatus that we have constructed to image the human head and torso. This latter apparatus is now producing n.m.r. signals and will, in the near future, produce images on a 128×128 matrix, of comparable quality to those produced on the smaller machine.

The Imaging Method

(a) *N.m.r. technique*

Since the n.m.r. signal : noise ratio is the limiting factor in picture acquisition times, it is essential to employ an efficient method of signal detection. In a conventional c.w. n.m.r. experiment it is easily shown (Abragam 1961) that the maximum possible n.m.r. voltage that can be induced in an r.f. coil surrounding a specimen is that induced by a precessing nuclear magnetization of magnitude $\frac{1}{2}M_0(T_2/T_1)^{\frac{1}{2}}$, where M_0 is the equilibrium nuclear magnetization, T_2 is the transverse relaxation time, or dephasing time of the individual nuclear precessions, and T_1 is the longitudinal relaxation time, or thermal equilibration time of the nuclei with the bulk material. In this instance, the maximum n.m.r. signal : noise power is proportional to T_2/T_1. This can easily be seen by realizing that $1/T_2$ is the spectral width of the resonance,

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and that the maximum power that can continuously be put *into* the nuclear system is proportional to $1/T_1$. This is the optimum method, and all other methods, particularly pulse methods, at best equal this sensitivity when $T_1 = T_2$ (in liquids) or tend to be less sensitive in the ratio T_2/T_1 when $T_2 \ll T_1$ (solids). W. S. Hinshaw (personal communication) has shown that, when the method of multiple side band (m.s.b.) n.m.r. is employed, where the excitation is by means of a string of very short, coherent, equally spaced r.f. pulses, the response is a periodic function of frequency, with periodicity $2\pi/\tau$, where τ is the pulse spacing. Consideration of his earlier paper (Hinshaw 1976) shows that the sensitivity reaches the theoretical maximum if the pulse length and power are chosen appropriately to match the values of T_1 and T_2. This is the method that we have used in the smaller of our two systems, which operates at 30 MHz in a magnetic field of 0.7 T. Typically we use 20 μs pulses spaced by 2 ms (Hinshaw *et al.* 1979). A further subtlety of the m.s.b. detection method is the fact that, if $\tau < T_1, T_2$, the driven equilibrium n.m.r. response signal is truly periodic in time and is almost symmetric about the mid-point of the interpulse interval, τ. Thus, if the n.m.r. signal is sampled by phase detecting the pick up coil voltages induced by the x and y components of the rotating nuclear magnetization (quadrature detection) and the discrete Fourier transform is used to analyse the signal, then the components of the discrete frequencies that are obtained exactly correspond to the maxima of the response function of the n.m.r. system to the string of pulses. If, further, the number of point pairs (x- and y-magnetization) $N = 2^n$, the fast Fourier transform (f.F.t.) algorithm of Cooley & Tukey (1965) may be used to advantage to perform the analysis quickly in real time on a computer.

(*b*) *Gradients*

Lauterbur (1973) has clearly stated the fundamental idea behind all methods of n.m.r. imaging and demonstrated its viability. Briefly stated, the n.m.r. frequency response of an extended sample in a linear magnetic field gradient is the one-dimensional projection at right angles to the gradient direction of the n.m.r. response of the object. The method works well because the n.m.r. response of nuclei in fairly mobile molecules is sharp (100 Hz $> 1/T_2 >$ 1 Hz), so that a weak gradient can map in frequency with good resolution the spatial distribution of the nuclear signal along the gradient direction. Since 1973, there have been many variants of this basic idea (Mansfield *et al.* 1973; Hinshaw & Moore 1974; Hutchison *et al.* 1974; Kumar *et al.* 1975; Hoult 1979). In the sensitive point method (Hinshaw 1976) the three gradients, $\partial B_z/\partial x$, $\partial B_z/\partial y$, $\partial B_z/\partial z$, are driven asynchronously to define in an extended space a small region where, due to the intersection of the null field planes of the gradients, there is effectively no resultant field. The position of this region in space can easily be moved electronically in each particular coordinate direction by adjusting the zero field plane of the pair of coils producing the particular gradient. The n.m.r. signals from everywhere but this point are time-dependent and will average to zero, and so the n.m.r. response from a definable point within a large object can be singled out, essentially by destroying the signal from all the rest of the object. Images can then be built up by scanning this point in a raster. Clearly, this is an inefficient method, since information is being collected from a small part of the object over a limited band width of $1/T_2$. Accordingly, in our small apparatus we have used the rotating gradient provided by driving the gradients $\partial B_z/\partial z$ and $\partial B_z/\partial x$ in phase quadrature with low frequency a.c. derived from the r.f. frequency. This procedure defines a strip, in the y-direction, of zero resultant field, with a time-dependent field everywhere else. The position of this strip can be moved in the z-direction by alteration of the position of the zero-field plane of the z-gradient

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($\partial B_z/\partial z$) coil pair. The signals from points along this strip are differentiated by application of the static gradient, $\partial B_z/\partial y$. This is, then, a line scanning method and the time to collect the n.m.r. signal from the strip is the same as that for a point in the previous method, because of the frequency multiplexing involved. Signals are now being acquired from a strip of the object and over a much larger band width. In our apparatus there were effectively 128 discrete volume elements along the strip of dimensions 60 mm × 0.5 mm × 2 mm (see § 2 (*c*)) and the total band width used at the r.f. of 30 MHz was *ca.* 64 kHz (500 Hz per point, allowing $T_2 \geqslant 2$ ms to be resolved). An optimum method would collect the n.m.r. signals from all of the specimen all of the time, although the minimum band width necessary for this rapidly becomes impractical as the effective resolution array size increases beyond about 32 × 32, and as the r.f. frequency is decreased.

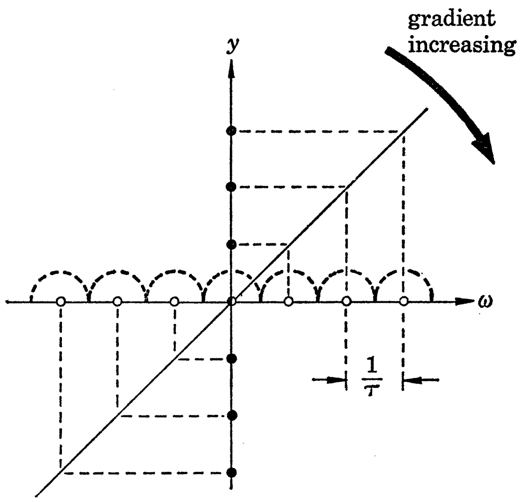

FIGURE 1. The relation between the spatial response of an isolated strip of the specimen in the *y*-direction when subjected to a magnetic field gradient $\partial B_z/\partial y$, if the n.m.r. frequency response is analysed by f.F.t. ●, 'sensitive volumes' along the strip; ○, f.F.t. frequency components, illustrated for $N = 8$. Also indicated on the horizontal axis is the multiple side band method response function (Hinshaw 1979).

(*c*) *Combined response to r.f. pulses and gradients*

When the strip of sample defined by the gradients in § 2 (*b*) is excited by the pulse method of § 2 (*a*) the response is as shown in figure 1. The periodic frequency response is mapped via the static gradient $\partial B_z/\partial y$ to a periodic spatial response along the strip isolated by the rotating gradient. Thus the computer-generated f.F.t. of the average time signal between pulses is one line of the n.m.r. image and the n.m.r. signal amplitude can be displayed as intensity modulation on an oscilloscope display. In our apparatus, the n.m.r. signal between the pulses is integrated over 128 equal intervals, digitized rapidly compared to the integration time and accumulated in a purpose-built 2 × 128 channel averager for about 1000 such intervals, then rapidly transferred to a computer where it is Fourier transformed and displayed on a storage oscilloscope while the selected strip is moved on automatically by a distance equal to its width, and 1000 new interpulse signals are accumulated. Efficiency is ensured by averaging over all of the time, except that while the strip is being moved and the gradients are changing. The time for the f.F.t. (0.1 s) would only limit the total picture formation time, which is the time to process 128 lines, if total times of significantly less than 30 s were employed. The picture production times we have used lie between 50 s and 10 min. The whole arrangement, as modified for the

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whole body apparatus, is shown schematically in figure 2. To make sure that the rotating gradient removes, by the averaging process, the signals from the material outside the selected strip, the number of interpulse interval signals that are averaged is the nearest whole number to an integral number of gradient revolutions.

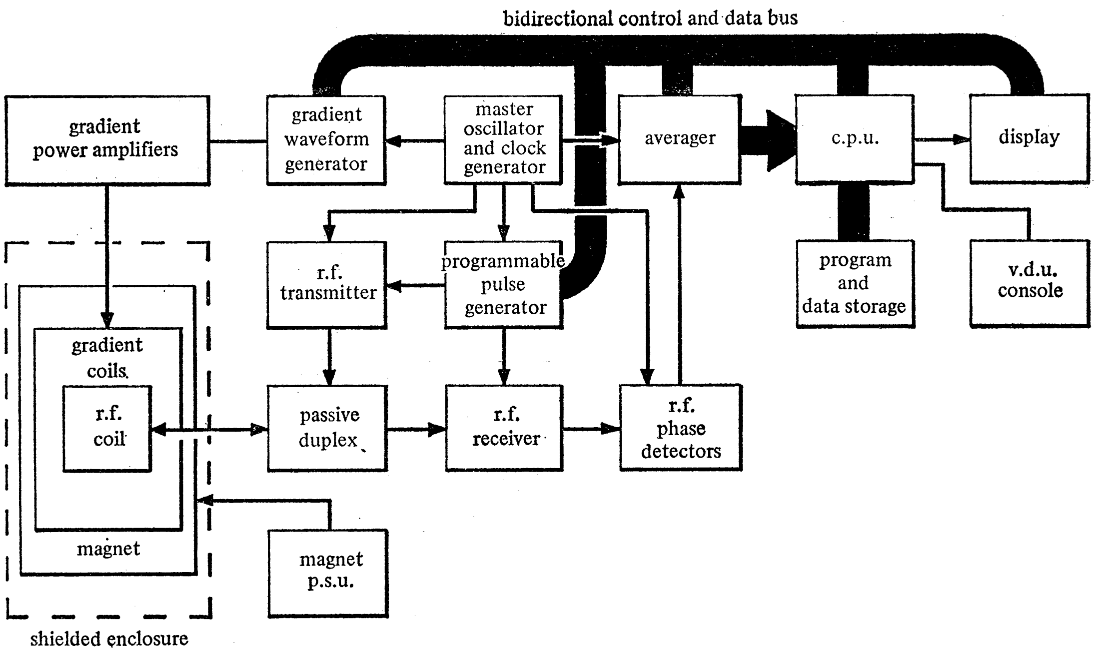

FIGURE 2. Functional diagram of the complete whole body n.m.r. apparatus.

3. THE WHOLE BODY APPARATUS

(*a*) *Magnetic field*

N.m.r. signal strength increases with increasing magnetic field. Hence the maximum possible field B_0 is desirable. However, r.f. field penetration measurements (Bottomley & Andrew 1978) have shown that the maximum usable r.f. frequency before the attenuation and phase shift caused by the conductivity of biological tissue becomes severe is size-dependent and ranges from 10 MHz to 15 MHz for the human torso and head respectively. Magnetic field and r.f. resonant frequency, ω_0, are related by $\omega_0 = \gamma B_0$, where γ is specific to a given nucleus, and for proton (hydrogen nucleus) n.m.r. these figures correspond to B_0 between *ca.* 0.2 and 0.3 T, whereas, for phosphorus n.m.r., B_0 ranges from *ca.* 0.5–0.75 T. Large volume homogeneous fields of this magnitude are impractical to produce with closed flux path iron magnets and so the choice lies between air-cored coils and superconducting systems. Air-cored systems are very wasteful of power above about 0.2 T and superconducting systems have their own problems. Thus, in common with most other groups to date, we have chosen to work at 0.1 T and have used an air-cored, four-coil system of 60 cm bore, designed by Walker Scientific. The homogeneity over a 30 cm diameter spherical volume in the centre of this and similar magnets is easily adjustable in practice to 1 part in 10^4, even with nearby iron objects and in a laboratory constructed of reinforced concrete. Homogeneities of 1 part in 10^5 would be much more difficult to attain over this volume in similar magnetic environments.

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(*b*) *R.f. system*

The static gradient strength G in conjunction with the r.f. pulse spacing τ and the number of points N in the f.F.t. determine the system band width to be $\Delta\omega = 2\pi N/\tau$. The centre frequency, ω_0, is determined by the strength of the static field, B_0, and is $\omega_0 = \gamma B_0$, where γ is the gyromagnetic ratio, in this case for protons, and is 4.26 MHz for $B_0 = 0.1$ T. The band width per point given by the f.F.t. of the interpulse time data (§ 2 (*c*)) from the selected strip is $2\pi/\tau$ and in order to resolve regions of the sample with short T_2, and hence response band width,

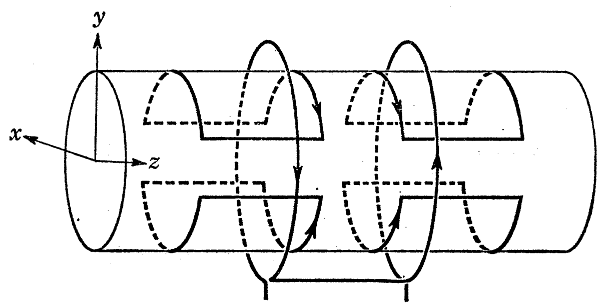

FIGURE 3. The sample tube with *y*- and *z*-gradient coils shown, which fits into the bore of the four-coil magnet. The *x*-gradient coil is identical to the *y*-gradient coil, but rotated through 90° about the tube axis.

$1/T_2$, τ must be made as short as possible. However, the use of a large tuned circuit around the human sample to pick up the n.m.r. signal introduces a natural band width, ω_0/Q, from its quality factor, Q. In addition, in order to excite uniformly all the side bands in the m.s.b. method spanning $\Delta\omega = 2\pi N/\tau$ from the selected strip of sample, the pulse length, δ, must be less than $\tau/2N$. These factors, together with the coil volume, V, to accommodate the sample, and the requirement that the r.f. pulse must nutate the nuclear magnetization through an angle of about $\frac{1}{2}\pi$ for the m.s.b. method, combine to define the r.f. power, P, required via

$$P \approx \frac{2\pi^3}{\mu_0\gamma^2}\frac{VN^3}{\tau^3},$$

by use of the defining equation for the field B_1 from a coil

$$B_1 \approx (\mu_0 PQ/\omega_0 V)^{\frac{1}{2}}$$

and elimination of B_1 via the nutation angle requirement $\frac{1}{2}\pi = \gamma B_1 \delta$. There are thus severe practical limits as to how short τ can be made, and how large is N before P becomes excessive, since V is fixed. There are, however, compelling reasons for making τ as short as possible. First, when τ is short, the gradient must be increased (§ 2 (*c*)) to span the sample, thus circumventing the need for a highly homogeneous field, B_0. Secondly, a short τ enables the widest possible range of T_2 in the sample to be imaged at maximum resolution. In our system we have designed and built a solid state r.f. power amplifier that can deliver up to 3.5 kW of pulse power (Holland & Heysmond 1979), and this fed to an r.f. coil of Q *ca.* 80 imposes a minimum τ of *ca.* 1.3 ms, hence minimum resolvable T_2 of *ca.* 1.3 ms, all for $N = 128$.

The r.f. transmitter coil is of a novel design inspired by a suggestion by Hoult & Richards (1975). It is in the form of two plane ellipses crossing at right angles wound on the inside of the 50 cm diameter cylindrical fibreglass tube which fits into the bore of our magnet and will hold the patient (figure 3). This coil couples to the magnetization perpendicular to the tube axis as it

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has to, since the static field, B_0, is along the axis. The receiver coil for head imaging is of the same design but 25 cm in diameter and with its magnetic axis crossed with that of the transmitter coil to minimize cross-coupling.

(c) *Gradient coils*

The signal band width $\Delta\omega = 2\pi N/\tau$ determines the gradient strength, G, necessary to collect the signals from the isolated strip of length L, via $G = 2\pi N/\gamma\tau L$, where $\Delta\omega/\gamma$ is the spread in field corresponding to n.m.r. signals of band width, $\Delta\omega$. The figures in § 3(*b*) lead to a maximum gradient strength (for imaging the head) of 10^{-4} T cm^{-1}. The gradients themselves must be uniform over at least a 30 cm diameter in the centre of our sample tube in order to image the human torso without geometric distortion. We have used Golay (1958) coils wound on the outside of the fibreglass tube to generate the gradients $\partial B_z/\partial x$ and $\partial B_z/\partial y$, where the cylinder axis is z and x and y are orthogonal. Computer calculations and subsequent measurements indicate that this geometry gives adequate homogeneity. These coils require approximately 0.5 kW of power to drive them in the worst case (the head at high resolution). The gradient $\partial B_z/\partial z$ is simply generated by a Helmholtz pair of coils driven in opposition, mounted between the fibreglass cylinder and the inner coil pair of the four-coil magnet.

(d) *Computer control system and display*

Although the quality of an n.m.r. imaging system is initially determined by the method used to encode and acquire the n.m.r. signal, its versatility is enhanced by its computer processing, control and display facilities.

In our whole body n.m.r. imaging apparatus an on-line dedicated minicomputer system based on the Data General Nova 2–10, equipped with 32k of core is used for the above functions. The basic computer is interfaced to a wide variety of proprietary and custom peripherals. Two 2.5 Mbyte hard disks are available for mass storage of programme and raw data and a dual cassette for long term storage of processed image data. Operator input is principally via a v.d.u., with a fast line printer for hard copy. Image display is on either a monochrome storage oscilloscope or a colour television. Image hard copy can be obtained by photographing either video screen or by using a custom designed photoscanner (Bottomley *et al.* 1978).

Interactive experimental control is achieved by having all machine variables under direct computer control. A 16 bit parallel bidirectional control bus is interfaced to each area of apparatus, such as the pulse programmer, gradient controller, etc. Currently an additional control capability in the form of a 16 bit microprocessor system, based on the TMS 9900, is being implemented for use whilst the Nova is busy.

It is important for such a complex experiment as n.m.r. imaging that all processing be performed in real time, in order to obtain operator feedback. We have made a high priority of reducing processing time to a minimum. All software has been written in assembly level language, which is significantly faster than a high level language such as Fortran. A hardware multiplier has been designed and constructed to reduce the time taken to perform the f.F.t. Finally, a method has been devised for real time alteration of processing parameters. A set of analogue potentiometers can be accessed and the voltage on each can be sampled, digitized to 8 bit accuracy and the resultant 8 bit word inserted into programme loops. Thus we have real time control of such variables as linear and first order phase corrections for the f.F.t., and display window levels. The same method can also be used to vary pulse lengths/intervals, averaging time, gradient frequency and amplitude, and scanning plane thickness. The effect

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of changing each variable can be seen by observing a single line spectrum live on the video display screen, whence it is possible to optimize image parameters prior to image formation.

For example, having the interpulse interval τ, pulse length δ and r.f. power gain under direct computer control allows the computer to calculate automatically δ for a given τ value based on the relation $\delta = \tau/2N$ and to produce a pulse of sufficient power so that α, the nutation angle, is $\frac{1}{2}\pi$ by accessing the predetermined look-up table for the r.f. power gain.

Other techniques, such as the linearization of scanning gradients by software, are also possible with the apparatus in its present configuration.

In summary, therefore, the rationale behind the construction of a 'programmable' imaging system is twofold. Firstly, it allows the effect of the variation of machine parameters on image formation to be monitored and evaluated simply, which we believe to be essential in an experimental apparatus. Secondly, it allows changes in the imaging technique to be performed without recourse to hardware redesign. This is also essential at this stage since it is by no means clear whether detailed n.m.r. data from a point, high resolution in the image, or speed of image production will be the most useful attribute of n.m.r. imaging. We are thus able, in our system, to use the single sensitive point method, the line-scanning method described in detail, and, also, the faster reconstruction from projection method.

4. Conclusion

We have discussed the mode of operation of our whole body n.m.r. imaging apparatus in terms of both the particular n.m.r. technique employed and the design principles on which the machine is based. The results produced in the next few months will indicate whether the approach adopted is justified.

The current work is being funded by a grant from the Wolfson Foundation to one of the authors (W.S.M.) and to Professor E. R. Andrew. The 30 MHz system was constructed with funding provided by the Medical Research Council.

We would particularly like to thank Dr R. C. Hawkes who, since joining the project, has contributed greatly to the implementation of many of the ideas discussed in this paper. The contribution of Dr W. S. Hinshaw, who has since left the project, is also gratefully acknowledged. Thanks are also due to Mr R. Erickson, Mr E. Heysmond and Mr B. Hill for their great help in electrical and mechanical construction.

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