

LETTER TO THE EDITOR

Multi-planar image formation using NMR spin echoes

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Abstract. A new method of two- or three-dimensional spin density imaging by nuclear magnetic resonance (NMR) is proposed, which exploits the properties of spin echoes in time-dependent magnetic field gradients. An analysis shows that simultaneous observation and differentiation of signals, arising from all spins distributed in a plane or set of planes within the specimen, is possible. The method is thus capable of producing visual pictures faster than previously described planar imaging methods.

Quite recently we described a method of planar NMR imaging which allows the formation of spin density images simultaneously from a distribution of points in a plane or set of planes throughout the object under study (Mansfield and Maudsley 1976). This method of imaging relies for its operation on selective excitation of the specimen in switched magnetic field gradients and is an extension of the line-scan method of imaging (Mansfield *et al* 1976), which in turn is based on general imaging methods by selective irradiation described by Garroway *et al* (1974). A number of other different imaging methods have also been reported (Damadian *et al* 1976, Lauterbur 1974, Hinshaw 1976, Kumar *et al* 1975).

In this Letter, a new planar imaging method is proposed, which achieves spatial selectivity either in part or fully by exploiting the properties of spin echoes. The method employs time-dependent magnetic field gradients and in many respects is the Fourier transform equivalent of some earlier experiments (Mansfield and Grannell 1975). In part of this earlier work we were concerned with the determination of periodic structures using Fourier transformation in a continuous time domain. In the present experiment we consider what might be regarded as the complementary situation, namely, the determination of a continuous spin density distribution when periodicity in the time domain is imposed.

As with our previous planar imaging method, the object of this experiment is to produce, at high speed, cross-sectional NMR images corresponding to the mobile spin density variations throughout a living biological specimen. This speed is imperative if NMR is to be successfully used in medical imaging.

For simplicity we shall restrict the preliminary discussion of the new method to two-dimensional imaging in one plane. Consider an extended specimen, figure 1, placed in a large uniform static magnetic field B_0 which defines the equilibrium spin polarization

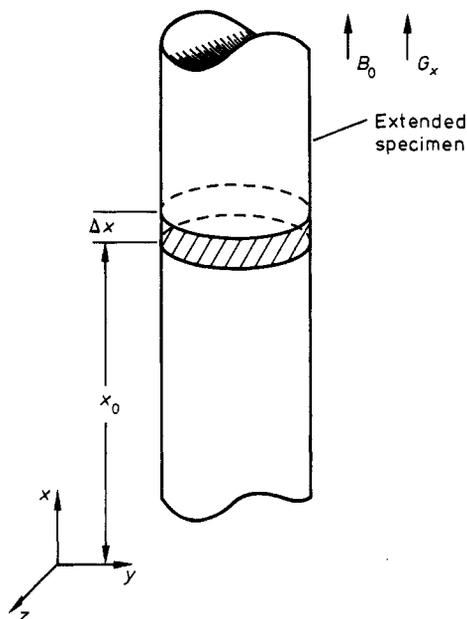


Figure 1. Sketch showing selectively excited layer of spins of thickness Δx in the plane x_0 within an extended specimen.

axis. A linear magnetic field gradient $G_x = \partial B_x / \partial x$ is also applied and at the same time a tailored 90° RF pulse excites the spins within a single slice of thickness Δx at x_0 , giving a free induction decay (FID). Immediately following this excitation pulse, the gradient G_x is switched off and the FID is observed in a switched gradient $G_y = \partial B_x / \partial y$ and a steady gradient $G_z = \partial B_x / \partial z$.

The effect of gradient switching is best understood if we study what happens when $G_z = 0$. Let us suppose that, in a time τ_b , a steady gradient G_y has caused the FID amplitude to decay completely to zero. If this decay alone were sampled and Fourier transformed it would of course yield the projection profile of the spin distribution along the y axis in the x_0 plane. However, by reversing the gradient direction (or by applying a 180° RF pulse) the decayed FID signal can be made to grow into a spin echo in a further time τ_b which then decays again. Further gradient reversals can thus recall the signal P times provided that $t_b = 2P\tau_b < T_2$, where T_2 is the spin-spin relaxation time of the specimen. Recalling the signal in this way and sampling the full spin echo train imposes a discreteness on the Fourier transformed projection profile. The discrete frequency spacing is given by $\Delta\omega_y = \pi/\tau_b$. The finite sampling time t_b will introduce some slight broadening on the discrete lines giving them all an angular frequency width of approximately $2\pi/t_b$. External shaping of the spin echo train gives the possibility of further broadening the discrete lines into rectangular or other desired profiles.

In the full two-dimensional experiment, signal sampling is performed with the additional steady gradient G_z which broadens the individual discrete lines to yield, for a single echo train plus Fourier transformation, a complete set of resolved cross-sectional profiles of the spin distribution across the thin layer in the specimen. The profiles can then be appropriately formed into a rectangular array of data points within a computer memory and then output in a television display to form a visual image (see for example Baines and Mansfield 1976).

In a generalization of this experiment to three dimensions, a multi-planar selection process can be incorporated by modulating both G_x and G_y , while maintaining G_z steady. In this case, the initial selective excitation pulse can be replaced by a conventional non-selective 90° (or θ°) RF pulse. As we shall show in the following analysis, the effect of gradient switching together with digital sampling of the signal is to impose on an otherwise continuous spin density distribution, a discrete lattice point distribution with spatial periodicities a , b and c .

The FID signal in the rotating reference frame at time t following the pulse is given by

$$S(t) = \text{Re} \iiint \rho(x, y, z) \exp [i\gamma \int_0^t (xG_x(t') + yG_y(t') + zG_z) dt'] dx dy dz \quad (1)$$

where $\rho(x, y, z)$ is the continuous spin density distribution of the sample and γ is the magnetogyric ratio.

We first consider the effect of the time-dependent x gradient alone in equation (1) and write the integral over x as a new function $f(y, z, t)$. If $G_x(t)$ is periodic, modulo $2\tau_a$, then we have

$$\begin{aligned} f(y, z, t) &= \int dx \rho(x, y, z) \exp [i\gamma \int_0^t xG_x(t') dt'] \\ &= \sum_{p=0}^P F[y, z, (t - 2p\tau_a)]. \end{aligned} \quad (2)$$

If the modulation is square wave and τ_a is chosen to be long enough for the FID to decay to zero amplitude, then with $G_y = G_z = 0$, equation (2) integrated over y and z yields a spin echo train. When all three gradients are applied with appropriate square-wave modulation, we may write the density $\rho(x, y, z)$ as a function of angular frequency $\rho(\omega_x, \omega_y, \omega_z)$. For large P , the Fourier transform of $f(y, z, t)$ becomes therefore

$$g(\omega_x, \omega_y, \omega_z) = \rho(\omega_x, \omega_y, \omega_z) 2\pi \delta(\omega_x - l\Delta\omega_x) \quad (3)$$

where $\delta(\omega_x - l\Delta\omega_x)$ is the Dirac delta function.

A similar transformation with respect to the integral over y introduces a second delta function $\delta(\omega_y - m\Delta\omega_y)$. In both delta functions l and m are integers and the angular frequency intervals between points are given by

$$\Delta\omega_x = \pi/\tau_a = \gamma a G_x \quad (4)$$

etc. The inverse Fourier transform of equation (3) when substituted back into equation (1) together with the corresponding substitution for the y integrand, yields finally

$$S(t) = (ab/\gamma G_z) \int d\Omega \rho(l\Delta\omega_x, m\Delta\omega_y, \omega_z) \cos \Omega t \quad (5)$$

where the angular frequency Ω is given by

$$\Omega = l\Delta\omega_x + m\Delta\omega_y + \omega_z. \quad (6)$$

Digital sampling of $S(t)$ for a time τ_c introduces a discreteness along the z axis. The points are spaced at $z = z_0 + nc$ (n integer) which corresponds to an angular frequency interval $\Delta\omega_z = 2\pi/\tau_c = \gamma c G_z$. Taking this into account we may rewrite equation (5) as the discrete sum

$$S(t) = \sum \rho_{lmn} \cos(l\Delta\omega_x + m\Delta\omega_y + n\Delta\omega_z) \Delta v_{lmn} \quad (7)$$

where $\Delta v_{lmn} = abc$ is the unit cell volume, the spins of which contribute to the signal at

each lattice point. If the modulation periods (and hence the gradients) are chosen so that

$$\Delta\omega_x/M = \Delta\omega_y = N\Delta\omega_z \quad (8)$$

where M and N are the largest values of m and n respectively in the imaging field, then we see from equation (7) that all points in the distribution ρ_{lmn} are uniquely defined in the frequency domain. Fourier transformation of $S(t)$ in equation (7) will thus yield in one calculation the complete three-dimensional spin density distribution function ρ_{lmn} . This Fourier transform nesting which in effect converts a three- (or two-) dimensional transformation to one-dimensional form, is the basis of our previous planar imaging method (Mansfield and Maudsley 1976).

Provided the conditions on gradient amplitude and periodicity are maintained, it is straightforward to see from equation (1) that the square waveforms of the field gradient modulations may be replaced by cosine waveforms to good approximation. For optimum working of this experiment, the gradient modulations must be phase coherent.

If the time saving gains are to be effective in planar imaging, the complete signal sampling cycle must be repeated frequently so that data acquisition approaches a continuous process for signal/noise ratio enhancement purposes. The coherent nature of the method described lends itself to the incorporation of complementary store cycles along the lines of the driven equilibrium Fourier transform (DEFT) technique for signal averaging (Becker *et al* 1969) or possibly the steady state free precession (SSFP) technique (Carr 1958, Hinshaw 1976) both methods making signal observation essentially independent of spin relaxation effects if required.

A full account of this work including the various cyclic variants for practical realization of high-speed imaging will be presented elsewhere.

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