Chapter 3

NMR Imaging in theory

There are many methods to construct NMR Images from the NMR signal received from RF coil. The effect of applying a strong static or time - dependent magnetic field gradient to a NMR sample is to introduce spatial dependence into the corresponding NMR signal (Bottomley 1981: 3). This is obvious from consideration of the Larmor equation. NMR imaging methods exploit this property : carefully controlled, well-defined gradients modulate the NMR signal in a known manner such that the spatial information can later be decoded and plotted as an image. Typically, the gradients are adiabatic and chosen with linear spatial dependence so the NMR frequency spectrum directly corresponds to position, or even one or more spatial coordinate axes. The imaging methods differ mainly in the nature of the gradient time-dependence (static, continuously time-dependent, or pulsed), and in the type of NMR pulse sequence employed.

It is important to note that the gradients referred to here are in the z-component of the magnetic field where, by convention, the z-axis is chosen as the direction of the resonance-producing main magnetic field B_0. Contributions from the gradients in the transverse plane affect the Larmor frequency to second order only and are usually disregarded. In Cartesian coordinates, the imaging gradients can thus be represented by the derivatives $\partial B_0/\partial x$, $\partial B_0/\partial y$, and $\partial B_0/\partial z$, hereinafter termed the x-, y-, and z- gradients respectively. The amplitudes of the gradients are denoted by G_x, G_y and G_z throughout and, where appropriate, the periods for which they applied by t_x, t_y, and t_z.

To discuss the different imaging methods, by using the classification scheme in Fig.3-1. The sample volume is divided into n_x, n_y and n_z volume elements along the three respectively Cartesian axes so $n_x n_y n_z$ independently measured values of the NMR signal are required to

completely reconstruct the image. These values may be obtained from N experiments, where $N \leq n_x n_y n_z$. In the simplest type of imaging experiment, each volume element is sampled by only one of the N experiments. Hence, $N = n_x n_y n_z$ and the imaging method is termed a sequential point technique. In sequential line methods, volume elements along an entire selected line, for example in the x-direction, are simultaneously observed, reducing for the complete image to $N = n_y n_z$. Yet more efficient methods permit simultaneous observation of an entire imaging plane. These are termed sequential plane methods for which $N = n_z$. The simultaneous observation and resolution of an entire plane of points in this manner requires a dynamic range capable of some $n_x n_y$ pixels. This places stringent demands on instrumentation and bandwidth. To circumvent this problem, some alternative schemes have been devised that resolve, for example, n_x lines in each experiment while simultaneously sampling data from the entire plane of $n_x n_y$ elements. Although $N = n_y n_z$ for these techniques, they are classed in the sequential plane class because they exhibit similar sensitivity per image time. For the most experiment, N = 1 and all volume elements are sampled simultaneously leading to a simultaneous or three-dimensional NMR imaging method. In analogy to the sequential plane class of methods, we include in this class techniques that require $N = n_y n_z$ experiments per image in which each experiment involves data collection from all $n_x n_y n_z$ points simultaneously.

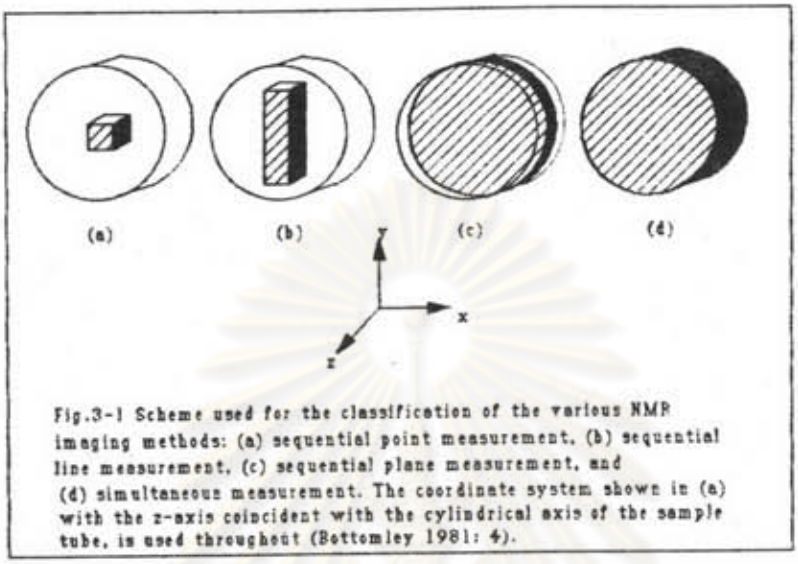

Fig.3-1 Scheme used for the classification of the various NMR imaging methods: (a) sequential point measurement, (b) sequential line measurement, (c) sequential plane measurement, and (d) simultaneous measurement. The coordinate system shown in (a) with the z-axis coincident with the cylindrical axis of the sample tube, is used throughout (Bottomley 1981: 4).

Sequential Point Methods

Sequential point methods that represent an advance on surface coil NMR (Bottomley 1981: 5). High-order (quadratic and quartic) magnetic field gradients are applied to carefully profile the main static magnetic field to generate field that is homogeneous only over a small fixed volume centered on the region of interest. Outside the sensitive volume, the field homogeneity deteriorates rapidly. Spin in the sensitive volume are excited by a conventional NMR coil, a surface coil, or for improved selectivity, a shaped RF field that is maximum at the sensitive region and decreases rapidly outside the region. The complete NMR scan is performed by moving the object through the volume. Localized T_1 and chemical shift information have been generated by this means.

Hinshaw's sensitive point technique is the most sophisticated of the sequential point methods, requiring no moving coils or gantries for moving the object. Spatial localization in three dimensions is achieved by application of three orthogonal time-dependent linear gradient magnetic fields in the presence of a continuous string of closely spaced phase-alternated RF pulses. A time-dependent gradient magnetic field of the form $(x-x_o)G_x\cos(w_xt)$, where x_o, G_x, and w_x are constants denoting,

respectively, position, gradient amplitude and frequency, introduce corresponding time-dependence on the NMR signal derived from all regions of the sample, except the plane at x=xo where the time-dependent gradient vanishes (Fig.3-2). The RF pulse scheme, known as *steady-state free precession* (SFP), provides a continuous and large component of the transverse magnetization which contains both the time-dependent component is removed by a low pass filter leaving signal originating only from the sensitive plane at x=xo.

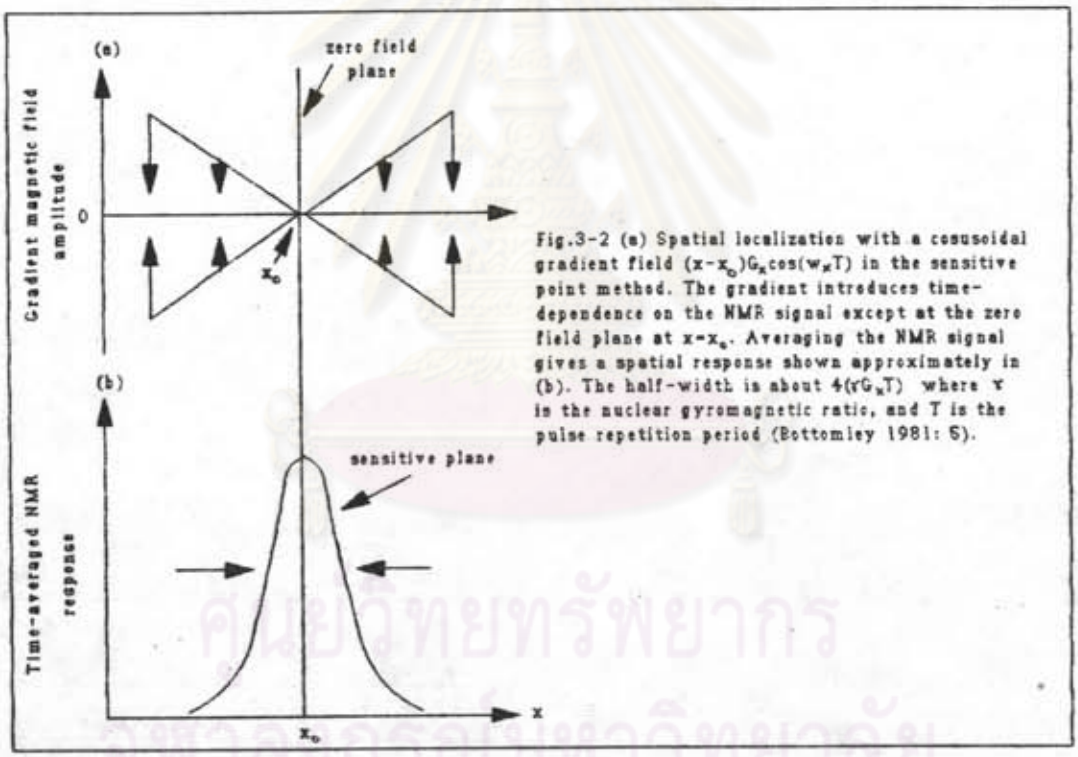

Fig.3-2 (a) Spatial localization with a consusoidal gradient field $(x-x_o)G_x\cos(w_xT)$ in the sensitive point method. The gradient introduces time-dependence on the NMR signal except at the zero field plane at $x=x_o$. Averaging the NMR signal gives a spatial response shown approximately in (b). The half-width is about $4(\gamma G_xT)$ where γ is the nuclear gyromagnetic ratio, and T is the pulse repetition period (Bottomley 1981: 6).

If two additional time-dependent fields, $(y-y_o)G_y\cos(w_yt)$ and $(z-z_o)G_z\cos(w_zt)$, are applied only the spins at or near the point (x_o,y_o,z_o) produce a time-independent NMR signal, and localization to a sensitive-point results. Here, the three gradient frequencies are all different, but alternatively two frequencies may be identical, providing that their phases are in quadrature. The position of the sensitive point is scanned across a sample by changing the ratio of currents in each half of the proper gradient coil set, and its size varied by altering

the RF pulse repetition period or gradient strength. The NMR signal intensity is a complex function of spectrometer operating conditions and $T_1(x_o,y_o,z_o)$, the transverse relaxation time $T_2(x_o,y_o,z_o)$, and the nuclear spin density $\rho(x_o,y_o,z_o)$. It is most for T_2's approaching T_1 as in liquid-like samples, and is thus often referred to as the mobile nuclear spin density at (x_o,y_o,z_o). In addition to mobile spin density images, images reflecting different T_1's have been produced by introduction of a 180 degree RF pulse into the SFP sequence at a different repetition rate. The concept of using sinusoidal gradients and the SFP sequence has also been applied to achieve spatial localization line and sequential plane type, respectively.

Sequential line methods

The sensitive line or multiple sensitive point technique use two orthogonal time-dependent magnetic field gradients, a SFP pulse sequence, and signal averaging to spatially localize the NMR spectrometer sensitivity to a line, as in the sensitive irradiation cycle (Fig.3-3) contains three intervals, $2t_z$, t_y, and t_x, during which the z-, and y-, and x- gradient fields are applied. In the first interval, t_z, a thin plane within the sample is defined by saturating the magnetization of spins lying everywhere outside the plane. This is achieved by applying a RF pulse modulated so its frequency spectrum selectively excited by a tailored 90 degree RF pulse applied in the presence of the y-gradient, G_y. The RF pulse has a narrow frequency spectrum which is sure that only a narrow strip of spins are perturbed. Finally, the FID resulting from excitation of the selected line is observed in the interval t_x during which the x-gradient, G_x, is applied. The Fourier transform of the FID yields the NMR signal distribution along the line in the x-direction. Successive strips are scanned electronically by changing the excitation frequency of the 90 degree RF pulses by computer control. Because only one strip is excited at a time,

the pulse repetition rate is not limited by the T_1 of the sample as different lines can be consecutively irradiated and observed in less than T_1.

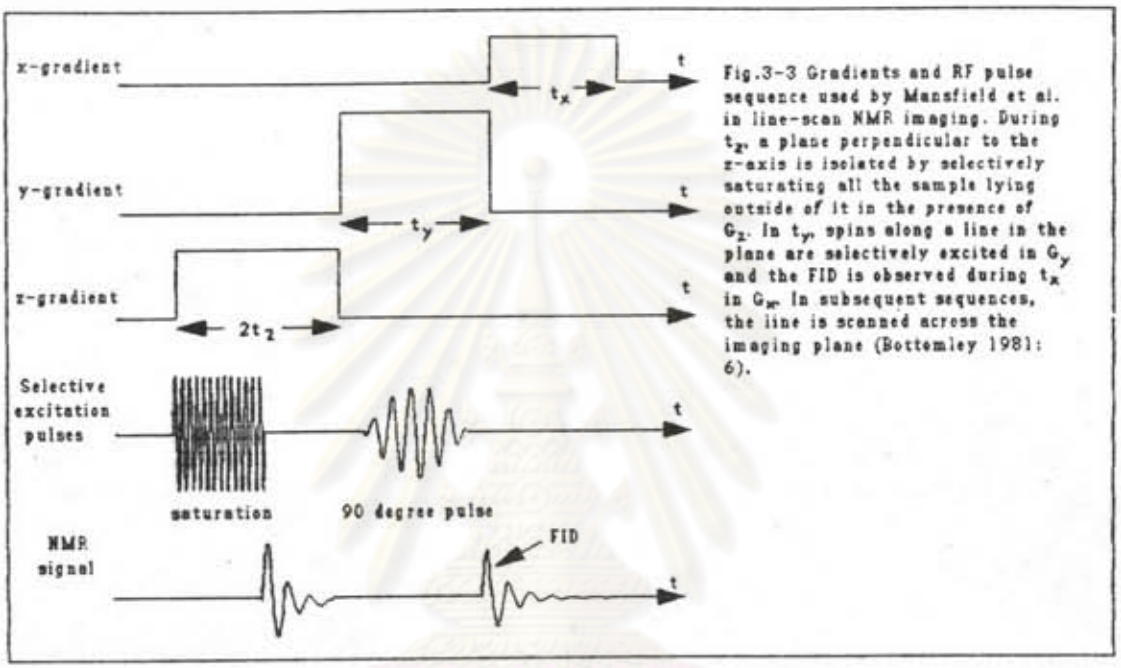

Fig.3-3 Gradients and RF pulse sequence used by Mansfield et al. in line-scan NMR imaging. During t_z, a plane perpendicular to the z-axis is isolated by selectively saturating all the sample lying outside of it in the presence of G_z. In t_y, spins along a line in the plane are selectively excited in G_y and the FID is observed during t_x in G_x. In subsequent sequences, the line is scanned across the imaging plane (Bottomley 1981: 6).

A subtle but import modification to the selective excitation procedure that has led to some controversy is including a time reversal step immediately following the excitation pulse. When applied to the above line-scan configuration, the time reversal can be introduced by reversing the sign of G_y for a short period following the tailored pulse. The gradient reversal rephases the spins forming an echo, at the center of which G_y is removed. At this point, G_x is applied and the FID observed and Fourier transformed as before. Another variation is to add an inverted x-gradient component synchronous with the reversed y-gradient component to the above sequence. This moves the position of the echo maximum along the time axis.

In this method, the selective saturation pulse in the interval t_z is replaced by a tailored 180 degree pulse which inverts only those spins in the chosen plane (Fig.3-4). This is followed by a selective 90 degree pulse and switch y- and x-gradients as above. The recorded echo

comprises the magnetization from the entire plane that was selected by the 90 degree pulse in the y-direction, minus all those spins that were inverted by the selective 180 degree pulse. The inverted spins lie along the line formed by the intersection of the selected z- and y-planes. To determine the signal arising from the selected line alone, the sequence is repeated without the inverting pulse to get a signal from the y-plane, which contains no inverted spin magnetization. The two signals recorded with and without the 180 degree preparation pulse are subtracted and Fourier transformed to field the spin distribution along the selected line. In the latter case, the order of the RF pulses is reversed to give a selective 90°-180° sequence that results in a spin echo signal derived from only the selected line. The images are thus formed without the above subtraction procedure and are sensitive to T_2 rather then T_1. In fact, T_2 images can be generated by repeating the 180 degree pulse as in a Carr-Purcell sequence.

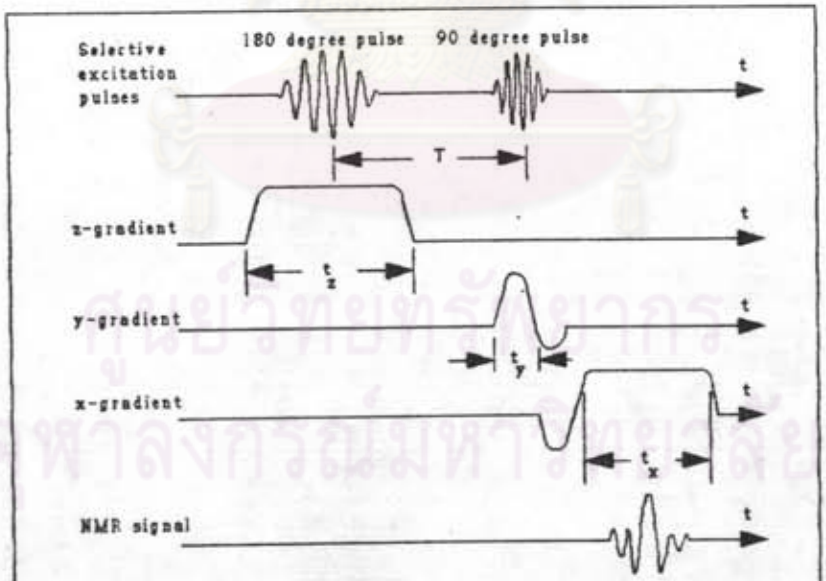

Fig.3-4 Imaging sequence in line-scan imaging scheme of Hutchison et al. In t_z, spins in a z-plane are inverted by a tailored 180 degree pulse. A selective 90 degree pulse during t_y tips inverted and noninverted spins in a chosen y-plane into the transverse plane, and an FID is observed during t_x. The FID contains signal from the entire y-plane minus the signal from the line at the intersection with the z-plane. Repetition of the sequence without the 180 degree pulse enables the contribution from the selected line alone to be obtained (Bottomley 1981: 7).

Sequential plane methods

The two-dimensional spatial variation of image of a physical property of an object can be reconstructed from a series of one-dimensional projections of the parameter that are records at different orientation relative to the sample (Bottomley 1981: 7). This principle forms the basis of the highly successful x-ray computed tomography imaging method used in diagnostic medicine and also the projection reconstruction zeugmatography NMR imaging technique. In this book, this earliest technique will be absented. The another method is the two-dimensional Fourier transformation method in which there are many variations. The general form of the 2DFT method was first proposed by Kumar et al (1975). (D G Taylor 1988: 644-654). However, it is the variation on this theme as developed by Edelstein et al (1980) that is in common use now. The latter is depicted in Fig.3-5 which show a typical pulse and gradient timing sequence.

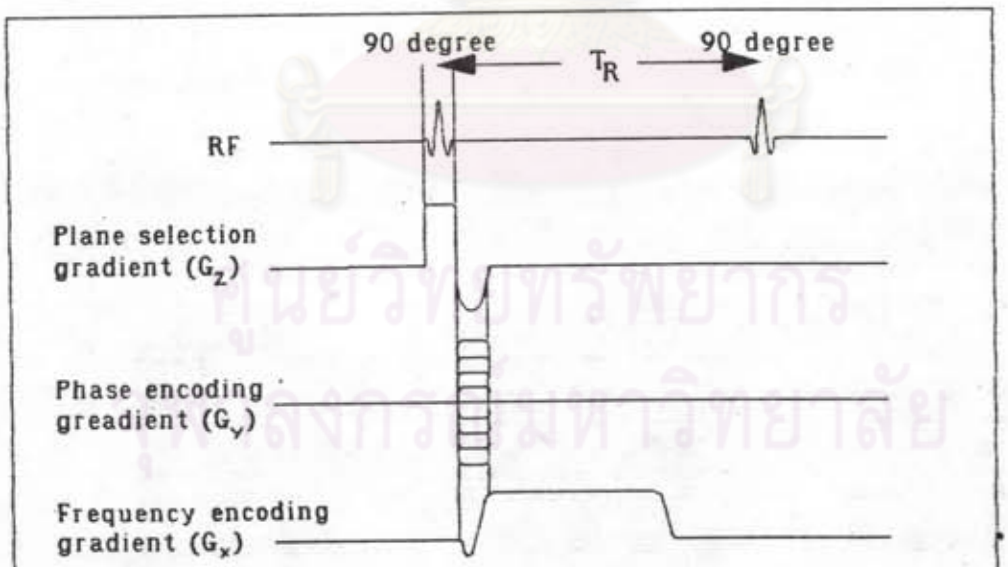

Fig.3-5 A saturation recovery sequence in the 2DFT imaging technique. Following each application of a selective 90 degree pulse only the amplitude of phase encoding gradient (G_y) changes (D G Taylor et al 1988: 644).

The slice selection procedure is exactly the same as for the projection reconstruction method. The acquisition period is also very similar since the FID is acquired in the presence of a gradient G_x. However, here this particular gradient changes neither in magnitude nor in direction, i.e. the projection is always onto the same axis. To see how spatial encoding is achieved in the y direction we have to look at the effect of the gradient G_y before the acquisition period.

While G_y is on, each spin-voxel along the y direction precesses with a frequency $w(y) = \gamma(B_0 + G_y \cdot y)$. When G_y is switch off, all spin-voxels along the y direction even in the presence of G_x. The phase that each voxel accumulates during the time G_y is on, t_y, is

$$\theta(y) = w(y) \cdot t_y = \gamma(B_0 + G_y \cdot y) \cdot t_y \qquad (3.1)$$

assuming a rectangular G_y gradient pulse.

Following the G_y gradient pulse the phase of the precessing magnetization varies linearly along the y axis. A spatial frequency has thus been imposed in the y direction. This process is repeated for several different y gradient pulse amplitudes corresponding to a set of spatial frequencies.

In the presence of a static gradient in the x direction G_x, each column of spin parallel to the y axis is uniquely characterised by a frequency of precession. The complete imaging sequence consists of collecting N data points for each of N amplitudes of G_y. This $N \times N$ data set is then Fourier transformed in two dimensions to yield the final image.

The equation of this reconstruction method is illustrated in Eq.(3.2). It give the NMR signal S obtained from a two-dimensional distribution of proton density $\rho(x,y)$ subjected to gradients G_x and G_y for detection following the application of a selective 90 degree pulse

$$S = KM_0 \int_x \int_y \rho(x,y) \exp[iy(xG_x t_x + yG_y t_y)] \exp[-(t_x + t_y)/T_2^*] dx dy \quad (3.2)$$

The steady state equilibrium value of the magnetization (M_0) present in a sample in a field B_0 is given by equation

$$M_0 = \frac{\rho \gamma^2 \hbar^2 I(I+1) B_0}{3kT_s}, \quad (3.3)$$

where ρ is the concentration of spins within the sample and T_s is the temperature.

It is clear from the form of this equation that $\rho(x,y)$ may be recovered by a two-dimensional Fourier transformation of a set of signals obtained by variation in either the gradient amplitude or its duration.

T1 weighted image

A second pulse sequence that may be used to weight an image with T_1 is the inversion recovery (IR) sequence. Fig.3-6 shows the pulse and gradient timing diagram using the 2DFT method. The first pulse is a non-selective pulse. After its application M_0 has been completely inverted, i.e. its instantaneous value is $M_z = -M_0$. The pulse sequence may be written as $(180\text{-TI-}90\text{-TR})_n$ where TI, the inversion time, is the interval between the 180 degree and the 90 degree pulse.

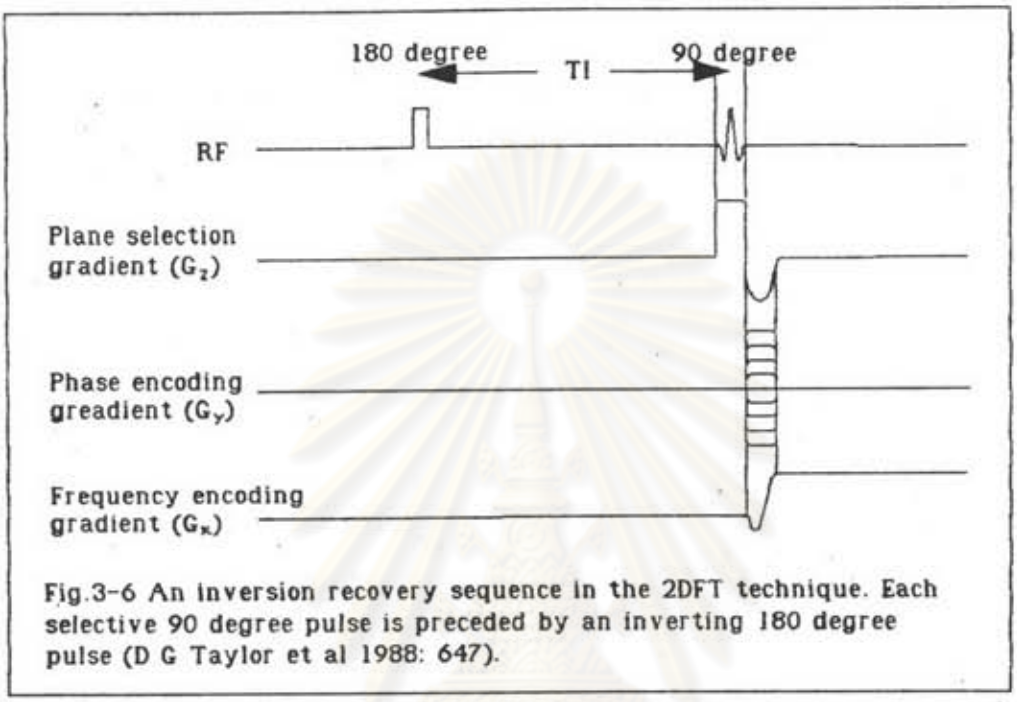

Fig.3-6 An inversion recovery sequence in the 2DFT technique. Each selective 90 degree pulse is preceded by an inverting 180 degree pulse (D G Taylor et al 1988: 647).

T2 weighted image

A spin-echo pulse sequence that is used to weight an image with T_2. The density image weighted with T_2 produce images which are quite different from T_1 weighted images. T_2 values in biological tissue are about 10 times shorter than T_1.

Since T_2 characterises the irreversible fall of M_{xy} to zero, to weight the detected signal with spin-spin relaxation (T_2), we have to chart the free induction decay (FID). However, as pointed out earlier, other external factors also affect the FID and pure T_2 effects can be studies only by the use of the spin-echo (SE) sequence. The pulse-train may be written as $(90^\circ - T_{E/2} - 180^\circ - TR)_n$. A typical pulse and gradient timing diagram is shown in Fig.3-7. An echo signal is formed at time T_E (see spin-spin relaxation in chapter 2).

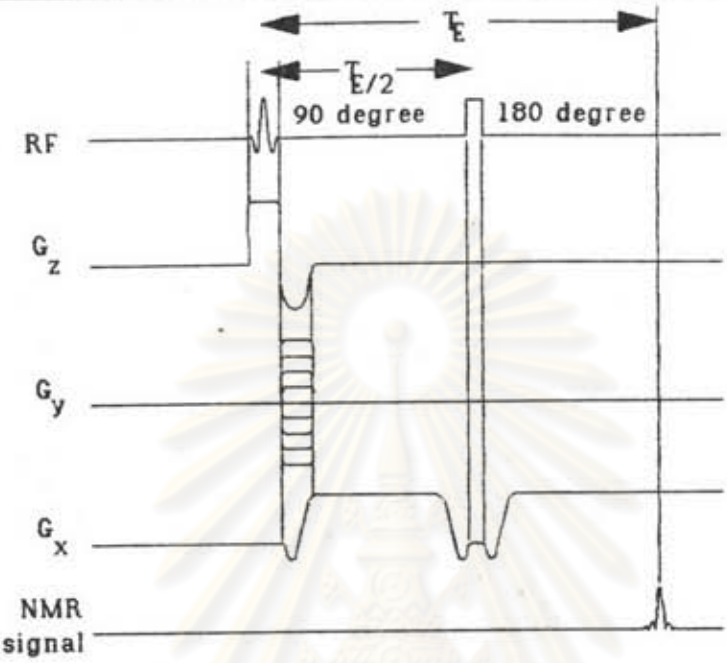

Fig.3-7 A spin-echo pulse sequence in the 2DFT technique. Each selective 90 degree pulse is followed at a time $T_{E/2}$ by a non-selective 180 degree pulse (D G Taylor 1988: 649).

A variation on the spin-echo sequence is the Carr Purcell sequence with a Meiboom Gill modification (CPMG) (Farrar and Becker 1971). This is essentially a $90°-T_{E/2}-180°$ pulse sequence followed by a train of 180 degree pulse every T_E second later. The RF phase of the 180 degree pulses in 90 degree relative to the initial 90 degree pulse. A train of spin echoes is produced at time T_E, $2T_E$, $3T_E$.... An image may be generated for each of these echoes or alternatively T_2 may be evaluated for each pixel and plotted as a T_2 map. This multi-echo method is, therefore, very efficient in minimising the time required to generate a series of T_2 weighted images with varying contrast.

Diffusion image

The random thermal motion will influence the signal as spins move from one part of the sample to another and experience a different magnetic field strength due to static field inhomogeneities. Variation in the frequency and phase of these mobile spins introduces a phase

incoherence which causes a reduction in the signal-amplitude. The amplitude of the presence of a gradient G is given by Eq.(3.4), where the random motion is characterised by the diffusion constant D (~10^{-5} cm^2s^{-1} in bilogical samples)

$$M' = M_0 \exp\left(-\frac{T_E}{T_2}\right) \cdot \exp\left[-\frac{2D\gamma^2G^2}{3}\left(\frac{T_E}{2}\right)^3\right] \qquad (3.4)$$

By applying field gradient pulses of known strengths (that is a known inhomgeneity) before and after the 180 degree pulse in the spin-echo sequence, it is possible to evaluate D for spins diffusing in a direction parallel to the field gradient direction. Such a sequence may be easily incorporated into a standard imaging experiment as shown in Fig.3-8, from which diffusion data may be mapped.

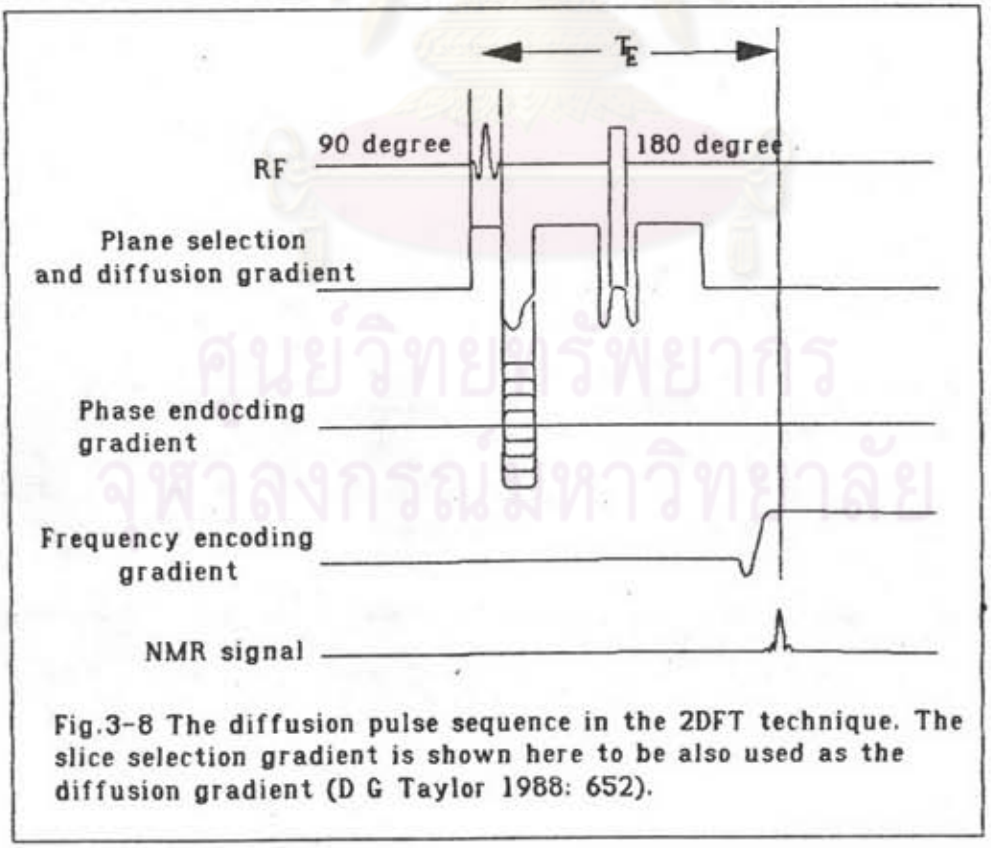

Fig.3-8 The diffusion pulse sequence in the 2DFT technique. The slice selection gradient is shown here to be also used as the diffusion gradient (D G Taylor 1988: 652).

Consider the motion of nuclei in a direction perpendicular to the imaging plane. If a saturation recovery sequence is performed with a short repetition time T_R, the intensity of signal recorded will depend on the number of nuclei which enter the imaging plane within the time T_R. As T_R is increased so does the intensity until it reaches a maximum at a certain time $T_R=t_m$, when all the nuclei within a region of interest are replaced by fresh nuclei.

The spin-echo technique as described for diffusion measurements can be extended to the study of flow since latter may be regarded as anisotropic diffusion. Consider the application of a pulsed gradient G_z before 180 degree pulse in the spin-spin sequence. The phase acquired by a spin at location z_1, under the influence of G_z for a time (t_1-t_2) is

$$\phi = \gamma \int_{t_1}^{t_2} G_z \cdot z_1 \, dt = \gamma G_z z_1 (t_2 - t_1) \qquad (3.5)$$

If the 180 degree pulse is followed by an exact repetition of the gradient pulse G_z, the phase get by a stationary spin during this period exactly cancels that acquired before the 180 degree pulse. However, if the spin moves with a constant velocity v in the direction of G_z, the total phase acquired by a spin at the time of the echo is

$$\phi = \frac{1}{2} \gamma G_z \cdot v (t_2^2 - t_1^2) \qquad (3.6)$$

From this it is possible to see that the phase acquired is directly proportional to the velocity. The extraction of phase information from the NMR signal is easy with phase sensitive detection and the grey level mapping of phase information produces images that are sensitive to the magnitude and direction of flow.

Chemical shift image

The chemical shift characterises the chemical environment around the spin (D G Taylor 1988: 653-654). The interaction between the applied magnetic field and the electrons produces a small reduction in the net magnetic field at the site of the nuclei. The effect is to lower slightly the resonant of frequency as a function Bo. The interest in the parameter, aside from spectroscopy is two-fold; the first is the possible production of artefacts at the difference between water and fat protons. Water protons and lipid protons have differing chemical shifts due to their differing chemical environment.

The basic aim is to produce an image that solely reflects either water proton or fat proton distribution. Several methods of achieving this have been proposal, one of which relies on the acquisition of two spin-echo images in which the water and fat produce signals with reversed relative phases. Algebraic manipulation then allows one to separate the water and fat components.

Technique which do not require the acquisition of two images have also been proposed. In one of these, a 180 degree echo pulse is selectively tailored such that only protons of a specific chemical shift coalesce to give a spin-echo. An alternative method is where a selective 90 degree pulse is used to suppose the signal from the unwanted protons before a normal spin-echo sequence. Fig.3-9 shows such a sequence.

Note that the suppressing 90 degree pulse is immediately followed by a 'spoiler' gradient pulse to induce a rapid transverse dephasing of the magnetization. The following 90 degree pulse then produces a signal at the coil which is solely from the shift of interest.

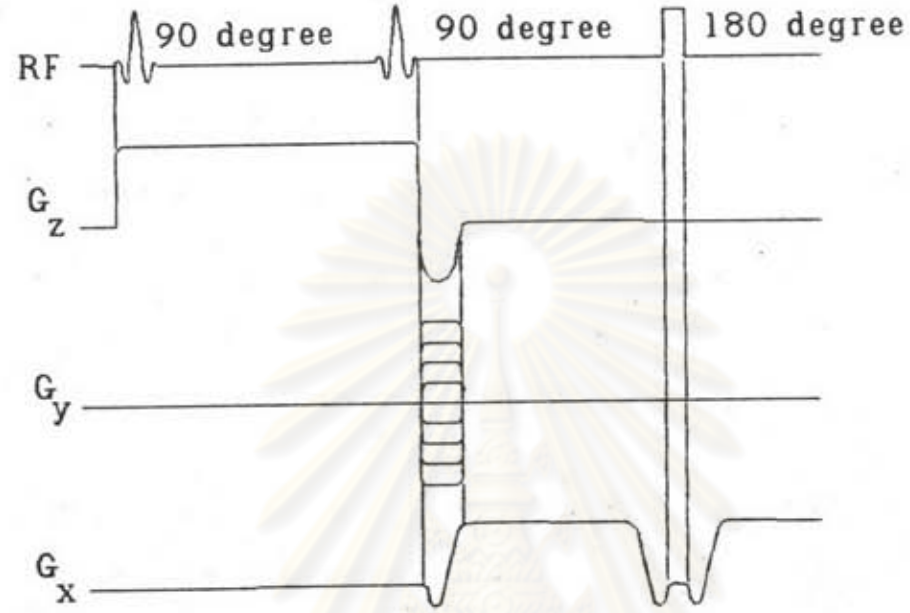

Fig.3-9 A chemical shift imaging pulse sequence in the 2DFT technique (D G Taylor 1988: 653).

Echo-planar image

In this method, images are obtained by imposing periodicity in the time domain rather than the spatial (frequency) domain (Bottomley 1981: 10), The excitation sequence is shown in Fig.3-10. Application of a tailored 90 degree RF pulse concurrent with the z-gradient excites the spins in a selected plane giving a FID. G_z is switched off and the FID observed in a periodically switched y-gradient plus a static x-gradient. The effect of the switched y-gradient alone is to induce a train of spin echoes which imposes a discreteness on the y-projection obtained when the signal is Fourier transformed. Addition of the x-gradient broadens the discrete lines of the projection to yield, from the Fourier transformed echo train, a complete set of profiles representing the spin

distribution across the selected plane. Since all spins in the plane are excited, the sensitivity loss meeting in planar imaging from unirradiated spins is avoided.

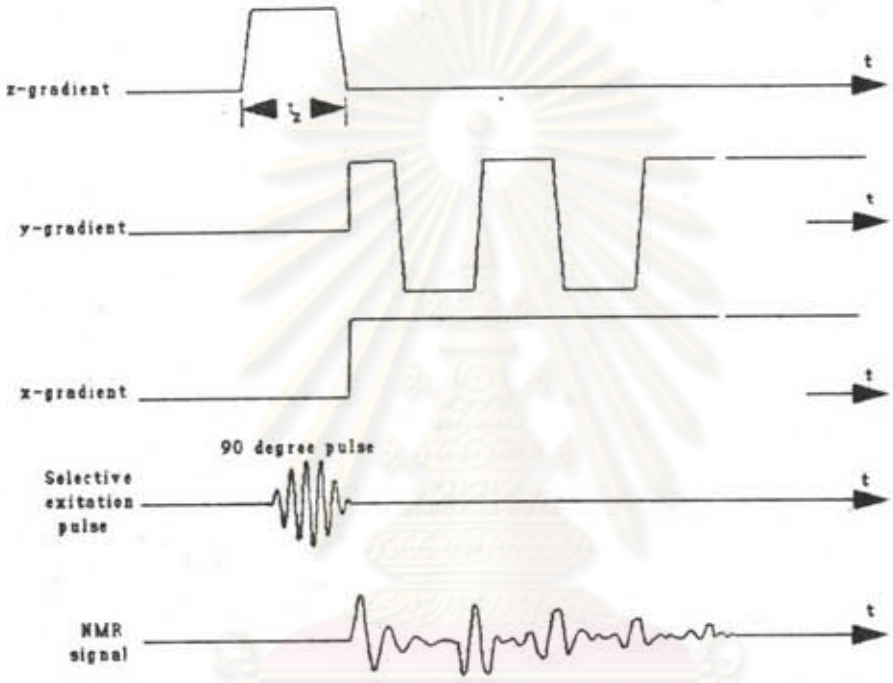

Fig.3-10 Echo-planar imaging sequence. A tailored 90 degree pulse selects the z-plane and the FID is observed during a switched y-gradient and a static x-gradient. The resultant echo train is Fourier transformed to obtain an image of the plane (Bottomley 1981 : 10).

Three-dimensional or Simultaneous methods

It might expect to find the excitation sequences employed in three-dimensional imaging methods in general simpler, notwithstanding the difficulties met in spatially encoding the NMR signal in the third dimension (Bottomley 1981: 11).

To do three-dimensional rotating frame zeugmatography is reconstructed by a second RF gradient applied in the x-gradient for the variable period t_x with the static gradient now applied in the

z-direction (Fig.3-11). Or alternatively, instead of using a second RF gradient, the RF field can be modulated at a second frequency during t_x. This method is employed, the three-dimensional image is reconstructed by three-dimensional Fourier transformation of the FID about t_x, t_y, and t_z.

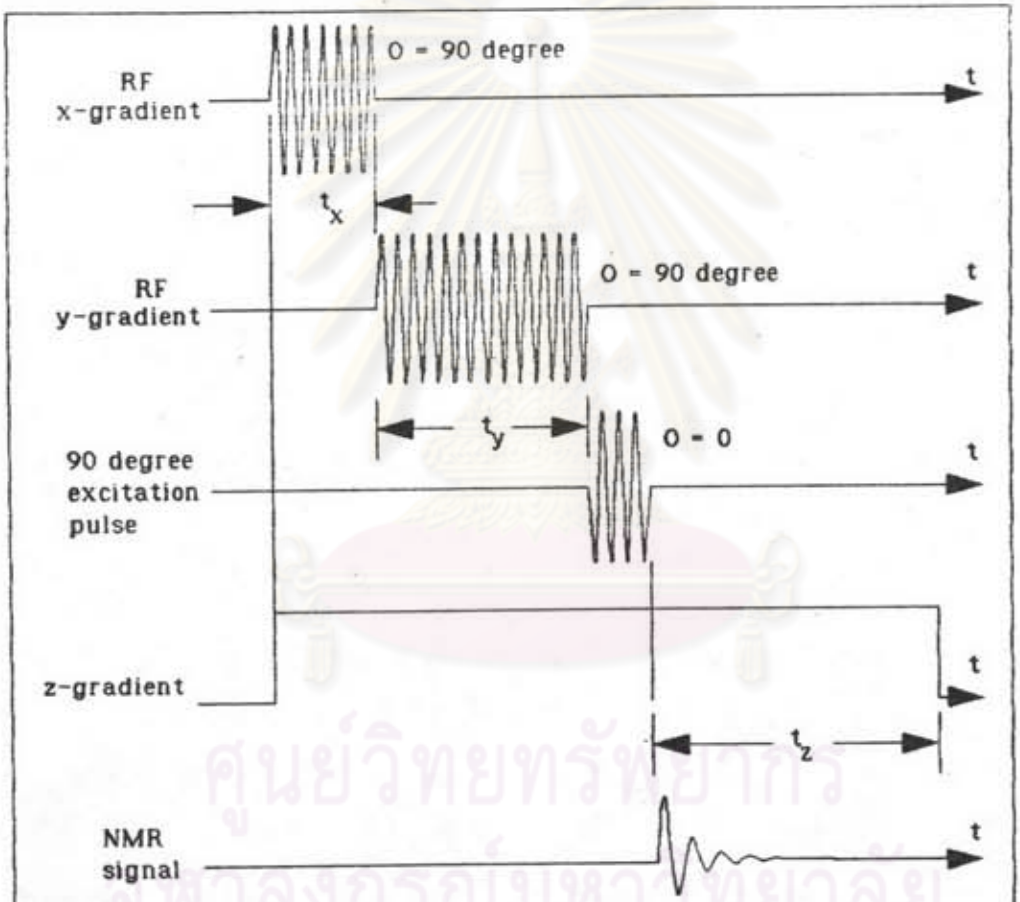

Fig3-11 Timing diagram for three-dimensional rotating frame zeugmatography. RF field gradients are applied in the x- and y-gradients for variable intervals t_x and t_y. The FID is observed during t_z in a static z-gradient and a three-dimensional image reconstructed by three-dimensional Fourier transformation with respect to t_x, t_y and t_z (Bottomley 1981: 13).