Study of R.F., Gradient Pulse and Magnet Instability Effect in NMR Tomography

C.H. Oh, H.S. Kim, H.W. Park, W.S. Kim, S.W. Lee and Z.H. Cho*
Korea Advanced Institute of Science, Seoul, Korea

ABSTRACT

Newly emerged NMR imaging requires careful studies on r.f. pulse shapes and sequencing for the selection of the region, gradient pulsing for the 2- or 3-D spatial coding, and suitable signal handling technique for the compensation of the inherent instability of the system, especially fluctuation of the static magnetic field.

Above subjects are discussed in detail and a new method which would be useful for the line integral projection reconstruction is proposed. The method could equally be applied to other 2-D NMR imaging techniques such as KWE (Kumar-Welti-Ernst) direct Fourier reconstruction or planar integral projection type reconstruction.

I. INTRODUCTION

NMR tomographic imaging is relatively new (1, 2, 3, 4) and still in its infancy, especially in the field of NMR electronics and instrumentation. It is also unfortunate at this time that the NMR CT is in heavy commercial competition and consequently many valuable new techniques and methods developed during the last several years become unavailable for public use in sufficient detail.

In this paper, we will discuss a few key problems arising in NMR imaging due to the pulse shape, finite pulse rise time, and stability factor of the instrument, namely, selective excitation by variety of r.f. pulses and effects of their pulse shapes, gradient pulse and its finite pulse rise time, and finally the random phase fluctuation effect due to the magnetic field instability. Each topic is analysed in detail and problems and solutions are indicated. A new scheme is proposed with which most of the problems can be overcome and both experimental as well as theoretical studies of this new proposed method were carried out.

Finally, experimentally obtained human images with KAIS-NMR Tomograph using the proposed technique are shown.

Although the method is applied to line-integral projection reconstruction, its basic concept can be applied to other NMR imaging systems and methods.

II. ANALYSIS OF R.F., GRADIENT PULSES AND MAGNETIC FIELD INSTABILITY EFFECT

1. Selection of slices by 90° r.f. pulses

Ideal selective 90° r.f. pulse which is supplied along x'-direction flips the spins only in a given frequency band into y'-direction and does not affect the spins which are located outside of the given frequency band (5, 6). After the r.f. pulse, the magnetization M starts precessing around z-axis. Then the FID signal becomes

\[ s(t) = \int_{\omega - \frac{\Delta \omega}{2}}^{\omega + \frac{\Delta \omega}{2}} M(\omega) e^{j \omega t} d\omega \]

where \( \omega \) is a Larmor frequency, \( \Delta \omega \) is a frequency band, and \( M(\omega) \) is the spatially dependent magnetization which is sometimes noted as spin density function \( \rho(\omega) \). By Fourier transform of \( s(t) \), \( M(\omega) \) is obtained.

In actual system, however, an ideal selective pulse which provides rectangular frequency band selection does not exist. In other words, the spins in the selected frequency band are not all flipped same amount into y'-axis so that they possess a unique set of \( M_{y'}(\omega) \) and \( M_{x'}(\omega) \) (7, 8, 9). In addition, the spins outside the rectangular frequency band are also excited giving spurious unwanted signals.

If we put complex magnetization \( \hat{M}(\omega) \) as

\[ \hat{M}(\omega) = M_{y'}(\omega) + iM_{x'}(\omega) \]

then, FID signal becomes

\[ s(t) = \int_{\omega - \frac{\Delta \omega}{2}}^{\omega + \frac{\Delta \omega}{2}} \hat{M}(\omega) e^{j \omega t} d\omega \]

Therefore, by Fourier transform of eq. (3), complex magnetization \( \hat{M}(\omega) \) is obtained.

For the search of an optimum r.f. pulse shape, spin responses to the different types of r.f. pulse have been measured and corresponding computer simulations were performed. In the experiment, a small amount of water contained in a vessel was excited by r.f. pulses of different shapes at near the Larmor frequency. The inhomogeneity within the volume of water was less than 100 Hz and was relatively small compared to the bandwidth of the r.f. pulse. By Fourier transform of FID, the complex magnetization \( \hat{M} \) was obtained and its magnitude and phase were calculated as

\[ |\hat{M}| = (M_{y'}^2 + M_{x'}^2)^{1/2} \]

This paper was presented at the 1982 Nuclear Science Symposium, Oct. 20-22, 1982 Washington D.C.

* Also at Dept. of Radiology, Columbia Univ., New York, N.Y.
and
\[ \phi = \tan^{-1} \frac{M_y}{M_x}, \]  

(4)

The experiment was repeated changing the frequency for each pulse shape and the distributions of \(|\tilde{M}|\) and \(\phi\) versus frequency were obtained. The experiment was carried out for several pulse shapes and the corresponding results are shown in Fig. 1. At each figure, the r.f. pulse shape, the magnitude of magnetization v. s. frequency, and the phase v. s. frequency are shown at the left, at the middle, and at the right, respectively. Fig. 1(a) is for the r.f. pulse of Hanning type which is given by

\[ R(t) = \frac{(1-\cos(2\pi t_p t))}{2}, \quad -t_p \leq t \leq 0. \]  

(5)

with corresponding Fourier transform of

\[ F(\omega) = \frac{e^{i\omega t_p^2}}{2\omega} \left[ \sin\omega t_p + i(1-\cos\omega t_p) \right]. \]  

(6)

Fig. 1(b) and Fig. 1(c) are the pulse shapes and their responses of the Hanning functions of the first and second half. In Fig. 1(d), a truncated sinc shape r.f. pulse and its responses are shown. Used sinc function is given by

\[ R(t) = \frac{\sin\omega_t (t+\frac{t_p}{2})}{\pi(t+\frac{t_p}{2})} \quad \text{for} \quad -t_p \leq t \leq 0, \]  

(7)

with its Fourier transform as

\[ F(\omega) = \frac{1}{2\pi} \left[ \left( u(\omega+\omega_1) - u(\omega-\omega_1) \right) \frac{\sin\omega/2}{\omega/2} \right] e^{-t_p^2/2}, \]  

(8)

where \(t_p\), \(2\omega_1\), and \(u(\omega)\) are pulse width, frequency bandwidth, and unit step function, respectively. Fig. 1(e) and Fig. 1(f) are the pulse shapes and their responses of the first and second half of Fig. 1(d), respectively. And finally Fig. 1(g) is for a square pulse.

Observing the distributions of the magnitudes and phases of the magnetization v. s. frequency, two problems can be noticed, namely, selectivity definition and phase distribution. In terms of selectivity, full truncated sinc and full Hanning appear most optimum. The phase distribution, however, is very widely spread.

The experimentally observed results were confirmed by computer simulations, i.e., by solving Bloch equations numerically, generating FID signals, and then Fourier transformation. These simulation results are also shown in Fig. 1. As is seen, the close agreement between the experimental results and simulations was found. It has been suggested that the dephasing effect which results in wide distribution of phase can be corrected by application of gradient reversal (10).

The effects of rephasing by gradient reversal obtained by computer simulation are shown in Fig. 2 for the full Hanning and full truncated sinc pulse. Gradient reversal with full truncated sinc or full Hanning could provide accurate spin rephasing in ideal condition, i.e., uniform excitation of the spins in a selected region. As is shown, gradient reversal requires negative gradient as shown in Fig. 2 with the same area shown with A and B. By examination of the responses of different r.f. pulses, it appears that the full truncated sinc pulse is an optimum choice.

2. Rise time effect of gradient pulse on image quality

Right after the r.f. excitation, if a gradient pulse is applied, e.g., along x-direction, an FID signal is obtained during the gradient pulse application which is given by

\[ s(t) = \oint \hat{G}(t) \ dx \]  

(9)

where \(\hat{G}(t)\) is a time varying gradient, \(\rho(x, y)\) is spin density and \(P_g(x)\) is a line integral projection data, respectively. If gradient pulse is constant during the integration time \(t\), eq. (9) can be written as

\[ s(t) = \int P_g(x) e^{i\gamma G_x t} \ dx \]  

Fig. 1 Frequency domain magnitude and phase distributions as a function of frequency are shown for the several different r.f. pulse shapes. (a) Hanning (b) Hanning', (c) Hanning", (d) Sinc, (e) Sinc', (f) Sinc", (g) Rectangular.
Fig. 2 Frequency domain magnitude and phase distributions as a function of frequency for the full Hanning and full truncated sinc shapes with spin rephase. Spin rephase is obtained with the gradient reversal.

\[
\text{or} = \frac{1}{\gamma G_x} \int P_x(x) e^{i\omega t} \, dx
\]

where \( G_x \) is a constant gradient and \( \omega = \gamma x G_x \). Projection data is then obtained via Fourier transform of FID signal. In actual system, however, gradient pulse has a finite rise time and its effect on image is detrimental unless proper measures are taken.

Then eq. (9) can be written as

\[
s(t) = \int P_x(w) e^{i\omega t} e^{2\pi i \omega t} \, dw
\]

where

\[
t_{\text{eff}} = \frac{1}{\gamma G_x} \int_0^T C_x(t') \, dt' \quad \text{and} \quad \omega = \gamma x G_{\text{MAX}}.
\]

In general, the obtained FID experiences time varying \( G_x(t) \).

The effect, however, could be eliminated if we take into account the actual pulse shape in the implementation of the Fourier transform operation (to obtain the projection data), i.e.,

\[
P_x(w) = \int s(t) e^{i\omega t} e^{2\pi i \omega t} \, dt_{\text{eff}}.
\]

An example of the rise time effect was tested by both computer simulation and experiment using an exponentially rising gradient pulse of

\[
G(t) = A(1-e^{-t/\tau}),
\]

where \( \tau \) is rise time and \( A \) is amplitude constant which was assumed as unity.

In Fig. 3, the results obtained by computer simulation are shown. Fig. 3(a) is the image obtained by simple Fourier transform of FID signal while (b) is the one obtained by using eq. (12). In Fig. 4, experimental results obtained with similar condition are shown for demonstration of the usefulness of the method in practice.

3. Correction of phase instability by using magnitude method.

In the practical system, magnetic field fluctuates both in short as well as long term period. Among others, main magnetic field fluctuation is the most serious problem in NMR imaging since it adds random phase fluctuation in the FID signal thereby degrading the resolution as well as the uniformity of the reconstructed image. In this section, we will analyse the problem and suggest a solution.

Let \( s(t) \) be an inverse Fourier transform of projection data \( P(w) \), where \( P(w) \) is a real value. It can be written as

\[
s(t) = \left\{ \begin{array}{ll}
  s_R(t) + i s_I(t) & \text{for } t \geq 0 \\
  s_R(t) - i s_I(t) & \text{for } t < 0,
\end{array} \right.
\]

where \( s_R(t) \) and \( s_I(t) \) are real and imaginary components of \( s(t) \), respectively.

In the conventional NMR system, however, data is collected only for \( t \geq 0 \) (see Fig. 5) and actual FID signal obtained is then

\[
s(t) = \left\{ \begin{array}{ll}
  s_R(t) + i s_I(t) & \text{for } t \geq 0 \\
  0 & \text{for } t < 0,
\end{array} \right.
\]

where \( s_R(t) \) and \( s_I(t) \) are usually obtained by quadrature phase sensitive detection. Then, \( s(t) \) given in eq. (15) can also be written in a divided form as

\[
s(t) = s_R(t) + s_I(t),
\]
where
\[ s_1(t) = \begin{cases} \frac{1}{2}(s_R(t) + is_I(t)) & \text{for } t \geq 0 \\ \frac{1}{2}(s_R(t) - is_I(t)) & \text{for } t < 0, \end{cases} \]
and
\[ s_2(t) = \begin{cases} \frac{1}{2}(s_R(t) + is_I(t)) & \text{for } t \geq 0 \\ \frac{1}{2}(-s_R(t) + is_I(t)) & \text{for } t < 0. \end{cases} \]

In this case, Fourier transform of \( s(t) \) would result
\[
F(s(t)) = F(s_1(t) + s_2(t)) = \frac{1}{2} (P(\omega) + iQ(\omega)) e^{i\phi},
\]
where \( Q(\omega) \) is a real value. Theoretically, therefore, from eq. (17) and (18), projection data can be obtained by Fourier transform of FID signal and by taking the real part only.

In most of NMR systems, due to finite power supply stability (10 parts per million), center frequency fluctuates at the rate of 10 - 100 Hz.

Because of this center frequency instability, phase delay between FID signal and reference signal exists. Therefore, eq. (19) appears as
\[
F(s(t)) = \frac{1}{2} (P(\omega) + iQ(\omega)) e^{i\phi},
\]
where \( \phi \) is the phase fluctuation resulting from magnetic field instability, by taking magnitude we have
\[
|F(s(t))| = P(\omega).
\]
This full data set can be obtained by 180° spin-echo technique that will be described in the following.

III. AN ALTERNATIVE NEW SCHEME PROPOSED FOR LINE INTEGRAL PROJECTION RECONSTRUCTION

In the above sections, we have examined: (a) necessity of rephasing the spins using gradient reversal for 90° r.f. selection pulse, (b) effect of gradient rise time and its correction, and (c) use of possibility of correcting field instability by taking FID signals of both \( t \geq 0 \) and \( t < 0 \). A new method which could solve the above three problems simultaneously is proposed and its pulse scheme is shown in Fig. 5. In this scheme, the whole object is excited by 90° r.f. pulse without application of any gradient pulse. Immediately after the end of 90° r.f. pulse, x-y gradient pulse follows. FID signals obtained during this period is distorted by the rise time of the gradient pulse, hence it is discarded. After \( t_{\text{r1}} \) time delay a 180° pulse is applied together with z-selection gradient pulse and subsequent reading pulse allows us to collect spin echo FID signal. In this way, spins in the frequency band of r.f. pulse selectively flip 180° and spins are re-focused while all the spins outside the frequency band decay away. The first half of the spin echo pulse corresponds to FID signal of \( t < 0 \) while the second half is for \( t > 0 \) allowing us to obtain real value only by taking magnitude of the Fourier transformed value. In addition, the signals are obtained at the region where gradient pulse is constant.

IV. EXPERIMENTAL RESULTS

Computer simulations and experiments were performed with the new proposed pulse scheme for the line integral projection reconstruction. In Fig. 6, simulated results of phase instability with FID signals obtained for \( t \geq 0 \) and for both \( t < 0 \) are shown. In the latter case, by taking magnitude, the phase instability effect was eliminated. Fig. 7(a) and (b) are the experimentally obtained results with the pulse scheme shown in Fig. 5. As shown, substantial image quality improvement has been obtained with the new scheme. Human head images of sagittal and coronal views have also been obtained and are shown in Fig. 8(a) and (b), respectively. In Fig. 9, actual pulse scheme for line integral projection reconstruction using spin echo technique is shown. In Fig. 9, upper row shows the first 90° r.f. pulse which excites all the spins of the entire volume and the following 180° selective r.f. pulse which selects the slice (for example, region normal to z-direction) together with gradient pulse (z-direction) shown in the 2nd row. Combination of x- and y-gradients are added during the \( t_{\text{r1}} \) period for the FID signal read out. Actually \( t_{\text{r1}} \) determines the data acquisition time in each view.
Fig. 6 Simulation of magnitude method using spin echo technique. (a) Using spin echo signal of $t \geq 0$ only. (b) Using spin echo signal of both $t \geq 0$ and $t < 0$ to eliminate the phase fluctuation effect (see discussion in section II-3).

Fig. 7 Experimental results (images) obtained by magnitude method. (a) Using spin echo signal of $t \geq 0$ only. (b) Using spin echo signal of both $t \geq 0$ and $t < 0$.

Fig. 8 Human head images obtained by the pulse scheme shown in Fig. 5. (a) Sagital image of head. (b) Coronal image of head.

Fig. 9 Experimentally observed r.f., gradient pulse sequences and FID signal from the pulse scheme shown in Fig. 5. Pulses at 1st row show the $90^\circ$ and $180^\circ$ r.f. pulses, respectively. FID signal is shown at the right side of the bottom row. $180^\circ$ r.f. pulse and z-gradient select the desired image plane.

V. CONCLUSIONS

Various electronics and instrumental effects on NMR imaging were studied, especially, r.f. pulse shapes for selection of slices, finite gradient pulse rise time for the image degradation, and instability of the main magnetic field which causes random phase fluctuation and eventually results image blurring. Finally we have proposed an alternative method which could overcome those problems. This new proposed scheme with $180^\circ$ spin echo has been successfully applied in our KAIS 1 K Gauss NMR tomographic system. This $180^\circ$ spin echo slice selection technique would be useful to other NMR imaging methods, e.g., KWE direct Fourier imaging as well as plane integral total volume imaging.

REFERENCES


<table>
<thead>
<tr>
<th>Pulse shape</th>
<th>Pulse function $R(t)$ for $-\tau_p \leq t \leq 0$</th>
<th>Fourier transform of $R(t)$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Han</td>
<td>$\frac{1}{2}(1-\cos 2\pi t/\tau_p)$</td>
<td>$\frac{1}{2}[(\frac{\tau_p}{\pi})^2 - u] \sin ut + \frac{1}{2}[(\frac{\tau_p}{\pi})^2 + u] \cos ut$</td>
</tr>
<tr>
<td>Han'</td>
<td>$\frac{1}{2}(1+\cos 2\pi t/\tau_p)$</td>
<td>$\frac{1}{2}[(\frac{\tau_p}{\pi})^2 - u] \sin ut - \frac{1}{2}[(\frac{\tau_p}{\pi})^2 + u] \cos ut$</td>
</tr>
<tr>
<td>Han'</td>
<td>$\frac{1}{2}(1-\cos 2\pi t/\tau_p)$</td>
<td>$\frac{1}{2}[(\frac{\tau_p}{\pi})^2 - u] \sin ut - \frac{1}{2}[(\frac{\tau_p}{\pi})^2 + u] \cos ut$</td>
</tr>
<tr>
<td>Sinc</td>
<td>$\sin u(t/t_p/2) / (\tau/t_p/2)$</td>
<td>$\frac{1}{\tau_p}[(u(\tau_p/2) - u)^2] \sin ut / u$</td>
</tr>
<tr>
<td>Sinc'</td>
<td>$\sin(u t / 2) / u t$</td>
<td>$\frac{1}{\tau_p}[(u(\tau_p/2) - u)^2] \sin ut / u$</td>
</tr>
<tr>
<td>Sinc'</td>
<td>$\sin(w t / 2) / (2u)$</td>
<td>$\frac{1}{\tau_p}[(u(\tau_p/2) - u)^2] \sin ut / u$</td>
</tr>
<tr>
<td>Rect</td>
<td>1</td>
<td>$\frac{1}{u} \sin ut + \frac{1}{u}[(1-\cos ut)]$</td>
</tr>
</tbody>
</table>