Original Research

Phase-Offset Multiplanar (POMP) Volume Imaging: A New Technique[1]

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Phase-offset multiplanar (POMP) imaging is a technique that excites several sections simultaneously for improved imaging efficiency. The centers of the reconstructed images from each of the POMP sections are offset from each other in the phase-encoding direction by means of view-dependent phase modulation of the radio-frequency (RF) excitation pulses and are placed adjacent to each other in the reconstruction. With a suitable reconstruction matrix size, the images can be made nonoverlapping and stored separately. At constant imaging time, signal-to-noise ratio (S/N), and resolution, POMP imaging produces a factor N_p more sections than a conventional sequence but with a reduced field of view. Alternatively, imaging time may be increased by the factor N_p to retain the same field of view but with the expected S/N advantage. The average RF power deposited by the 90° composite RF pulse is greater by the factor N_p, but the power for the 180° pulse is unchanged. The POMP method is discussed and compared with three-dimensional and Hadamard techniques.

Index terms: Image processing • Physics • Pulse sequences • Three-dimensional imaging

JMRI 1991; 1:457–461

Abbreviations: FOV = field of view, NEX = number of excitations, NSL = number of sections, POMP = phase-offset multiplanar, RF = radio frequency, S/N = signal-to-noise ratio, 3D = three-dimensional, 2D = two-dimensional, VERSE = variable-rate selective excitation.

BECAUSE THE LONGITUDINAL RELAXATION times of tissues are relatively long (on the order of 1 second), imaging times are long for T1-weighted sequences in which significant magnetization recovery is desired. Thus, multiplanar sequences, in which the time between planar excitations is used to excite other planes, are universally used to improve efficiency. For example, with a TE of 20 msec and a TR of 500 msec, about 10 sections can typically be excited in one pass. Sometimes this number of sections (NS) is adequate; however, with a section thickness of 3 mm, more than one pass (which increases imaging time accordingly) may be required to image the desired volume. Often the number of excitations (NEX) per phase encoding is more than one in such thin-section imaging in order to achieve sufficient signal-to-noise ratio (S/N).

One technique popular for thin-section imaging is three-dimensional (3D) (volume) excitation, since the S/N advantage of a 3D set with NS sections is identical to that of a two-dimensional (2D) image with NEX = NS. However, Fourier encoding produces severe intersection contamination because of well-known truncation effects when NS is small (typically eight or less) (1). An alternative approach is Hadamard encoding (2), in which NS phase-cycled combinations of selective radio-frequency (RF) pulses are used to simultaneously excite NS planes, where NS is a power of two. From the NS data acquisitions, all planes are uniquely determined by linear combinations of the complex data according to a Hadamard sequence. The advantage of this technique is that a small number of sections (eg, two or four) may be obtained without section profile degradation as in Fourier methods, since the profiles are determined by the RF pulse characteristics rather than by phase encoding.

The Hadamard technique has several disadvantages. First, the RF pulses require NS^2 times as much peak RF power and NS times as much average power. Second, the data must be preprocessed during reconstruction (inverse Hadamard transformation) in order to extract the individual section data. Third, imaging time must be increased by a factor of NS (although the S/N is increased by $\sqrt{NS}$, as expected).

The first disadvantage (ie, high RF power) has been

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mitigated recently by use of variable-rate selective excitation (VERSE) pulse methods (3). Thus, binary encoding techniques in which multiple sections are simultaneously excited can be readily used to improve imaging efficiency. The second and third disadvantages of Hadamard techniques persist, however.

It is desirable, therefore, to consider a multiplanar technique that has the advantages of Hadamard 3D imaging while not requiring any data preprocessing for reconstruction, and that can increase the number of sections acquired without necessarily increasing imaging time. In the present study, a method of satisfying this goal with phase-offset encoding of simultaneous multiple-section excitations is presented. By way of introduction, the technique of offsetting an image in the phase-encoding direction is first discussed. Preliminary work was reported previously (4,5).

• THEORY AND METHODS

Image Offset in the Phase-encoding Direction

Let $F(k_y)$ be the Fourier transform of a function $f(y)$. Then

$$f(y) = \int F(k_y)e^{-ik_y y}\, dk_y. \tag{1}$$

From a well-known theorem in Fourier analysis,

$$f(y - y_0) = \int F'(k_y)e^{-ik_y y}\, dk_y, \tag{2}$$

where

$$F'(k_y) = F(k_y)e^{-ik_y y_0}. \tag{3}$$

Thus, by simply multiplying F by a k_y-dependent phase factor, the spatial location of $f(y)$ can be offset by a desired amount y_0. This phase factor can be applied either during reception (before Fourier transform reconstruction) or during excitation by incorporation of the phase shift in the RF pulse.

For application to 2D Fourier transform MR imaging, let y represent the phase-encoding direction. Then k_y is proportional to the area under the G_y phase-encoding gradient; k_y may be represented by phase-encoding view number. Thus, the image can be offset in the phase-encoding direction by use of a view-dependent phase factor, according to Equation (3). This can be achieved during excitation either with a view-dependent $B_0(t)$ pulse or by phase modulating the RF excitation pulse with a phase factor that depends on view. With the latter technique, if $R(t)$ represents the RF envelope for a given linear-phase excitation pulse, then excitation with $R'(t,k_y)$, where

$$R'(t,k_y) = R(t)e^{ik_y y_0}, \tag{4}$$

will produce a reconstruction with the image offset by y_0. Thus, an image whose center is offset from the origin by y_0 may be produced by replacing the RF pulse envelope with one whose phase varies with view according to Equation (4). Let us now examine the technique of simultaneously exciting two or more sections with arbitrary offsets.

Phase-Offset Multiplanar (POMP) Technique

The POMP technique uses a composite RF pulse that combines two or more phase-modulated excita-

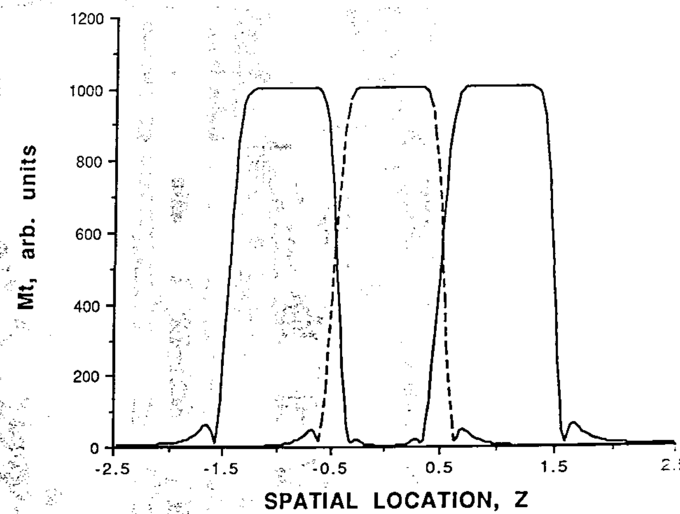


Figure 1. Calculated section profiles for single-section (dashed line) and POMP (solid line) 90° pulses. The sinc pulse had a phase length of 8π and was Hamming windowed; the POMP pulse was made by adding two spatially offset sinc pulses of the same type.

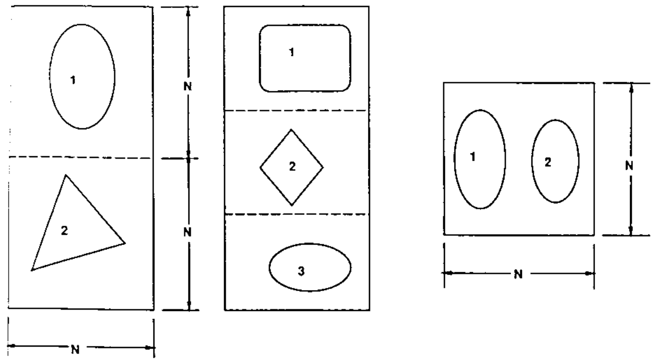
tions that have y_0 offsets chosen so that the resulting images do not overlap. Thus, if $R_i(t)$ represents the RF envelope for the ith desired section, a multisection POMP excitation is generated by the composite RF pulse

$$R'(t,k_y) = \sum_{i=1}^{N_p} R_i(t)e^{ik_y y_i}, \tag{5}$$

where y_i is chosen to offset each of the N_p sections to a different region of the output image space in the phase-encoding direction. If the field of view (FOV) is chosen large enough, then all N_p images can be made nonoverlapping by a suitable choice of y_i. After reconstruction, the N_p images can be separated and stored individually.

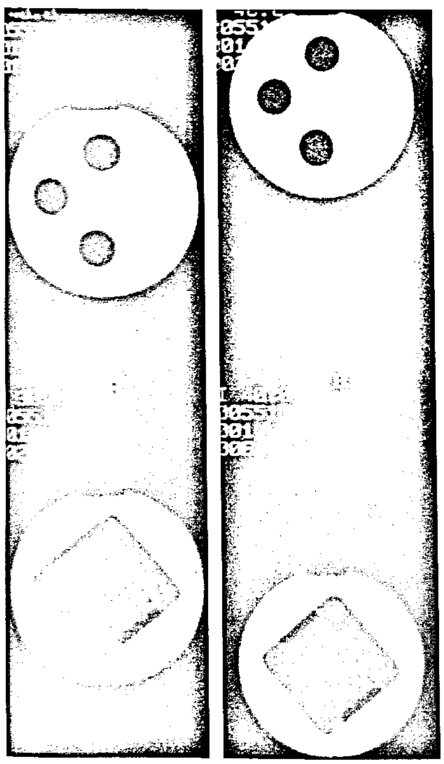
A desired property of the POMP pulse is the simultaneous excitation of all component sections with no loss of section profile fidelity. Equation (5) tacitly assumes linearity of the excitation process, an assumption that breaks down at flip angles greater than 90° but that is substantially valid for flip angles of 90° or less (2). This is shown in Figure 1, in which section profiles calculated by solving the Bloch equations are illustrated for a single-section 90° excitation with a sinc pulse and for a POMP pulse made by adding two offset sinc pulses together. The simulation shows that the two-section POMP excitation is nearly identical to that expected from simple addition of translated single-excitation profiles. Thus, the individual section profiles obtained with POMP excitation are nearly identical to those obtained from the single-section components, and therefore image contrast, section cross-talk, and so forth, should be nominally the same.

To create nonoverlapping POMP images, the reconstructed FOV (F_r) in the phase-encoding direction must be at least N_p times the FOV of the individual sections (F_0). That is, if the full FOV must be maintained,

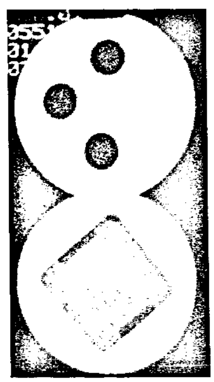


a. b. c.

Figure 2. Typical POMP reconstruction options: (**a**) N_p = 2, two NEX; (**b**) N_p = 3, two NEX; and (**c**) N_p = 2, one NEX.



a. b.



c.

Figure 4. Imaging of phantoms with POMP technique. (**a**) Conventional images of two sections. (**b**) With phase-offset excitation, images are displaced. (**c**) Simultaneous excitation of both sections with POMP pulse made by phase-cycled addition of pulses used in **b**.

$$F_r = N_p F_0. \qquad (6)$$

One way of creating a larger FOV is simply to decrease the k_y phase-encoding-step amplitude. This increases the FOV accordingly, but at the expense of resolution. Alternatively, the number of views can be multiplied by a factor N_p, and the k_y encoding step decreased by the same factor, as shown in Figure 2a. This will maintain resolution while increasing F_r, as desired. The S/N will be multiplied by $\sqrt{N_p}$, as expected, since the S/N behavior is equivalent to that of conventional N_p-NEX imaging in which averaging is used.

Equation (6) assumes that the object being imaged fills the entire FOV in the y direction. Note, however, that human cross sections are often rather elliptically shaped, so that it may be possible to relax the requirement in Equation (6), as illustrated in Figure 2b, 2c.

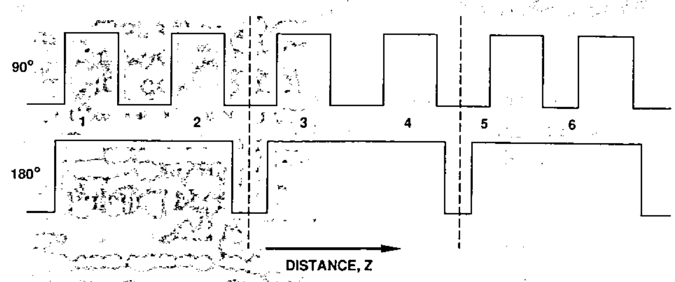


Figure 3. Section excitations for POMP imaging with N_p = 2. The 90° pulse is made by summing N_p individual section pulses, and it simultaneously excites two adjacent sections; the 180° pulse excites all N_p sections.

Here the phase-encoding direction is chosen to be along the short axis of the elliptical section, so that in Figure 2b, three images can fit into a field of view F_r = 2 F_0, instead of 3 F_0. Therefore, the POMP factor N_p is not intrinsically constrained by NEX but rather is limited by the size of the object to be imaged relative to the reconstructed image size in the phase-encoding direction. It is noteworthy, therefore, that the factor N_p need not be a power of two. This is a distinct advantage over Hadamard methods, for which binary encoding is a necessity and for which the total number of acquisitions increases with the number of sections.

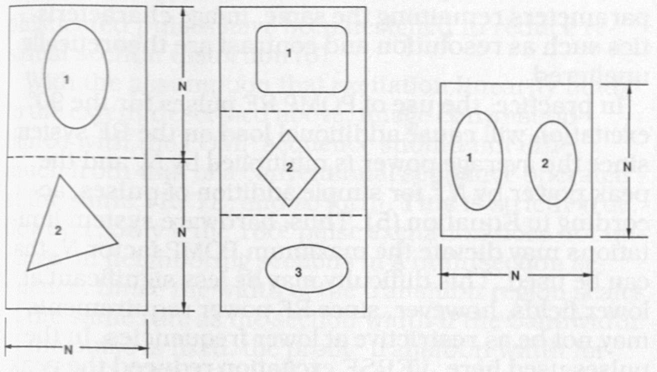
Implementation

The technique was implemented on a conventional 1.5-T clinical imager (Signa, Rev 3.3; GE Medical Systems, Milwaukee). Starting with a standard multisection pulse sequence, the excitation profile for the 180° pulse was modified as shown in Figure 3 so that its selection width encompassed all N_p adjacent POMP sections. Therefore, no additional RF power was required for this pulse. The conventional 90° sinc pulse was replaced by a POMP pulse that was view dependent and made from combinations of offset sinc waveforms, according to Equation (5). Examination of Equation (5) shows that N_p different pulses are needed to cover the cyclic POMP phase encoding. These 90° pulses were VERSE pulses to reduce average RF power to about the same level as that of the conventional pulse. POMP factors (N_p) of two and three were examined.

A POMP reconstruction program was written that first performs a conventional image reconstruction, with a matrix size of 256 in the readout direction and either 256 or 512 in the phase-encoding direction, depending on the NEX and FOV chosen. This was followed by extraction of the N_p sections from the resulting POMP image(s) and storage in individual image files.

• RESULTS

Figure 4 illustrates the POMP method. Conventional single-section images of phantoms are shown in Figure 4a; Figure 4b shows the result of phase shifting of the separate excitation pulses, according to Equation (4). A two-section (N_p = 2) POMP acquisition is shown in Figure 4c, which uses an RF pulse made by adding together (Eq [5]) the two RF pulses



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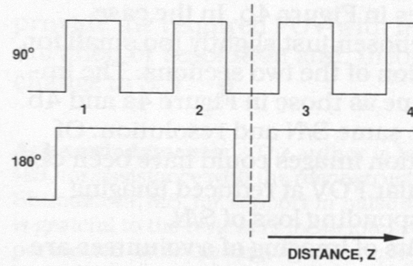

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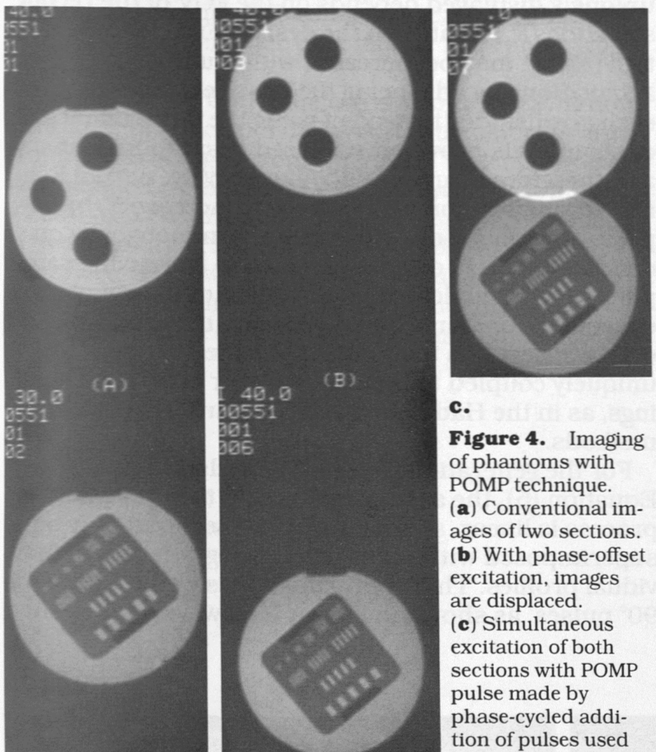

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Typical in vivo results of imaging of a volunteer are shown in Figure 5. Twenty-four-section imaging was performed with an N_p of three. The protocol chosen would correspond to eight-section imaging if performed with a conventional sequence. Figure 5a shows one of the eight POMP triplet reconstructions obtained (256 × 512 image) before separation. Figure 5b shows one of the resulting images after separation and storage in image files. In this case, the acquisition with an N_p of 3 was accomplished in a two-NEX imaging time (and resultant image size of 512 pixels in the phase-encoding direction), without overlap, because of the asymmetric nature of the human anatomy chosen for this example. Thus, the sequence has provided three times as many images in the same imaging time as conventional imaging, with nominally the same contrast, S/N, and resolution.

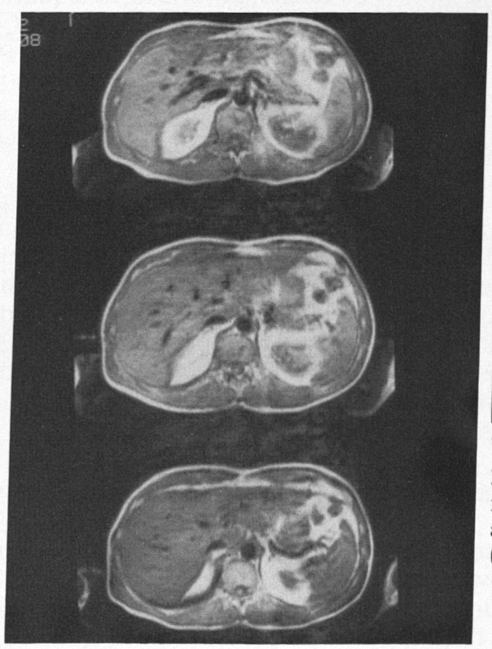
• DISCUSSION

POMP is a volume-excitation sequence in which several sections are simultaneously excited, with the individual sections appearing in different regions of the output image space in the phase-encoding direction. No preprocessing of the data (Fourier or Hadamard decoding) is needed. Thus, the technique can be applied to any pulse sequence by simply replacing the 90° pulse with POMP pulses and by increasing the selection width of the 180° pulse to encompass the several consecutive POMP sections. The resulting sequence produces N_p times as many sections as the conventional sequence, with repetition rate and other parameters remaining the same. Image characteristics such as resolution and contrast are theoretically unaltered.

In practice, the use of POMP RF pulses for the 90° excitation will cause additional load on the RF system, since the average power is multiplied by N_p and the peak power by N_p^2 for simple addition of pulses, according to Equation (5). Thus, hardware system limitations may dictate the maximum POMP factor N_p that can be used. This difficulty may be less significant at lower fields, however, since RF power requirements may not be as restrictive at lower frequencies. In the pulses used here, VERSE excitation reduced the peak and average power requirements and allowed pulses with an N_p of 3 to be used in body-coil imaging (Fig 5).

Apart from RF power considerations, the number of sections that may be simultaneously encoded and uniquely displayed depends on the FOV of the reconstruction (F_r) relative to the desired FOV of each section (F_0). F_r may be increased with multiple-excitation acquisitions by advancing the phase encoding for each acquisition (instead of using the more usual averaging). This provides increased k_y sampling density and therefore a larger FOV. Alternatively, partial Fourier reconstruction may be used to increase F_r by disposing the phase-encoding views asymmetrically on one side of the k_y origin, thereby also increasing sampling density while leaving the total number of phase encodings constant. Note, therefore, that the number of POMP sections that may be obtained is not uniquely coupled to the total number of phase encodings, as in the Hadamard and 3D Fourier transform methods.

For the generation of the POMP pulses described in Equation (5), the assumption is that the excitation process is linear, so that individual sections may be superimposed without significant degradation of individual profiles. This approximation is well realized for 90° pulses, as shown in Figure 1. However, more so-



a.

b.

Figure 5. In images of volunteer, three abdominal sections fit in 512-pixel space when NEX = 2 and N_p = 3; (**a**) 256 × 512 image before separation; (**b**) one of the images after separation.

that created the images in Figure 4b. In the case shown, the FOV was chosen just slightly too small for unambiguous resolution of the two sections. The imaging time was the same as those in Figure 4a and 4b, and the image had the same S/N and resolution. Of course, the single-section images could have been obtained with a rectangular FOV at reduced imaging time but with a corresponding loss of S/N.

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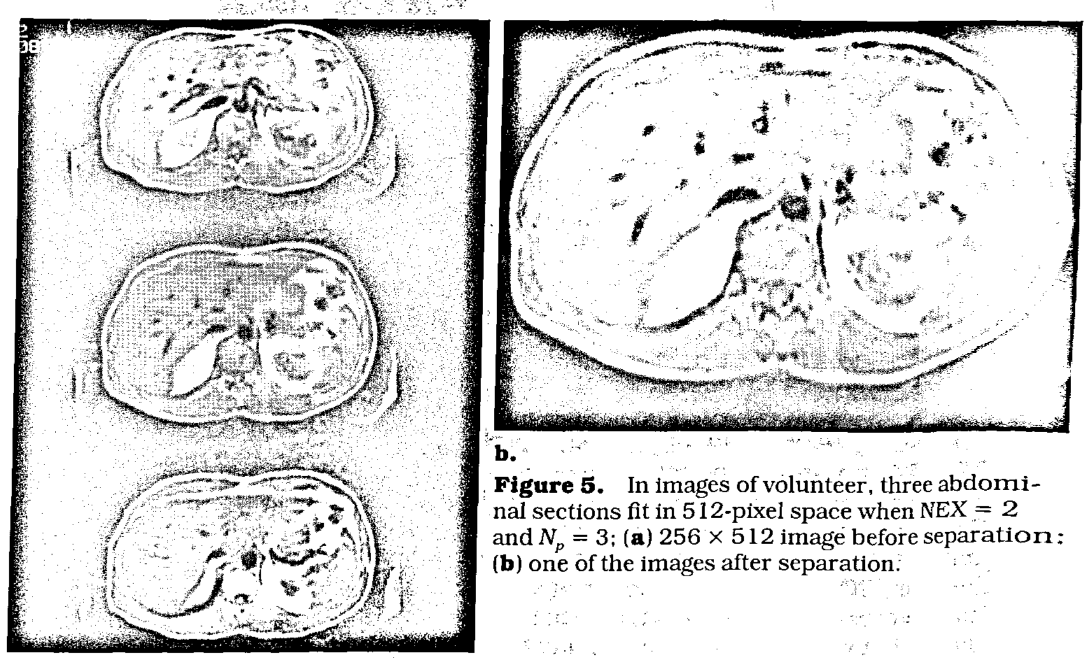

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phisticated pulses have been designed to reduce residual section distortion (6).

With the assumption that excitation linearity holds to the extent described above, image contrast obtained with the POMP sequence should not differ much from that of a conventional sequence operated at the same TR, TE, and so forth. This is strictly true, however, only if the 180° pulse excitation does not impinge on neighboring sections in a multisection sequence. Since the width of the transition region scales at the same rate as the section width if the bandwidth of the pulse is fixed, the profile transition width for the POMP 180° pulse will be wider by a factor of N_p than for the conventional pulse. This means that contiguous-section imaging could be degraded more with the POMP method, unless the 180° pulse is tailored to have the same transition width as in the conventional sequence. Although this can be done, the resulting RF power requirements for the tailored 180° pulse will be greater.

In summary, POMP imaging is a volume-excitation technique in which only 2D Fourier transforms are required for image reconstruction. The technique may be used when the FOV in the phase-encoding direction exceeds the dimension of the object by a factor of two or more. This can occur, for example, when signal averaging would normally be used because of S/N considerations. In such cases, the increased number of phase encodings can be used in POMP to provide the required FOV with no change in imaging duration or resolution and, theoretically, with no change in S/N. ●

Acknowledgments: The author is indebted to Kirk Udovich, MS, for assistance with the reconstruction program and Ann Shimakawa, MS, for support in pulse sequence development. He is grateful to the reviewers for helpful comments. This work was performed while the author was employed by GE Medical Systems, whose support is gratefully acknowledged.

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