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Halving MR Imaging Time by Conjugation: Demonstration at 3.5 kG[1]

Conjugation can be used to synthesize half of the data acquired during a conventional two-dimensional Fourier transform imaging procedure, thus reducing imaging time by nearly half. The images acquired by this process have the same object contrast and spatial resolution as conventional images do, but with a 40% reduction in the signal-to-noise ratio (S/N). Conjugation can be used to advantage in magnetic resonance imaging units in which S/N levels are higher than needed to permit imaging with a single acquisition of each projection.

Index terms: Magnetic resonance (MR), technology

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THE cost-effectiveness and patient acceptance of magnetic resonance (MR) imaging depend to a certain extent on imaging time. Shorter imaging times can sometimes result in improved diagnostic efficacy, since they reduce the likelihood of voluntary motions of the subject. Furthermore, as specific time thresholds are crossed, the effects of biologic motion such as breathing, peristalsis, and heart beating are avoided. We present here an imaging strategy that reduces imaging time to approximately one-half of the conventional two-dimensional Fourier transform (2DFT) mode, and can be used in conjunction with other speed up strategies, which are also discussed.

THEORY

Consider the time domain data acquired in a conventional 2DFT mode (Fig. 1). Each line of data is produced from a spin echo at an echo time TE. The interval between consecutive lines of data is the repetition time TR. Each data line encodes position information along the x-axis and contributes position information to a column along the y-axis. If after the x-axis Fourier transform we extract all the points along a constant value of x (a column) and plot them, they have the shape of a spin echo. The acquired spin-echo signal is the result of signal readout along the x-axis and is called the "time domain signal." Because of the mathematically equivalent form of the signal shape when plotted along the y-axis, it is sometimes called the "pseudo time domain signal."

The spatial resolution along x is obtained by applying an x-gradient during spin-echo readout. The spatial resolution along y is obtained by applying a y-gradient some time before readout. This gradient dephases the nuclei on the basis of their position in the object and is called the phase-encoding gradient. A Fourier transform applied to the data of Figure 1 first along one axis and then along the other produces an MR image, hence the 2DFT name.

The number of resolved vertical lines (lines of constant x) for a given field of view is given by the number of discrete points obtained in the spin-echo acquisition. Similarly, the number of resolved horizontal lines (lines of constant y) will depend on the number of echoes acquired with incremental strengths of y-gradient, since each contributes one discrete point to the pseudo echo. The imaging time (t) is

$$t = \mathrm{TR} \times n \times N, \quad (1)$$

where TR is the interval between acquisition of successive echoes, n is the number of times that echoes with the same phase encodings are acquired, and N is the number of horizontal lines in the field of view. The use of values of n larger than one serves two purposes: The first is for improving signal-to-noise (S/N) levels, which increase in proportion to $\sqrt{n}$. The second is more subtle: $n = 4$ acquisitions can be used to cancel certain artifacts and $n = 2$ will cancel a subset of these artifacts. The origin of the artifacts and the way they are canceled are beyond the scope of this paper and are discussed elsewhere (1); however, it is important to note that S/N alone is not the sole determinant of whether $n = 1$ procedures can be implemented.

Given a desired spatial resolution, to decrease imaging time we may reduce N with a consequent reduction in the field of view and in S/N, which varies as $\sqrt{N}$ if all other factors are kept constant. Shortening TR also reduces imaging time, but it has a well-understood detrimental impact on sensitivity to subtle lesions, which are typically more confidently detected in long TR imaging spin-echo sequences (2-5). To shorten time further, various strategies can be used. The most dramatic reductions are those obtained by Mansfield et al. using echo-planar techniques (6). The basic concept here is that successive spin echoes can be used to encode position information. Rather than phase encoding a line of data every TR interval, several encodings can be acquired from a train of successive spin echoes, that is, every TE. The basic limitation of this technique (which is called echo planar) is the time limit imposed by T2 decay. Most soft tissues have a T2 value on the order of 50 msec. If an image is to encode 128 lines, and if these lines are to be acquired during a time not to exceed

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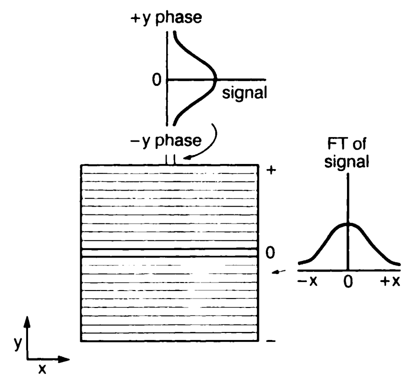
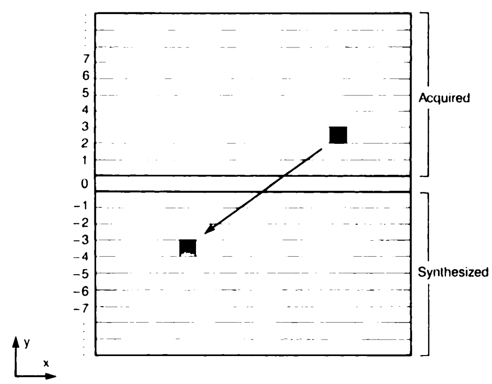
T2, then the interval between echoes has to be 0.5 msec, that is, TE = 0.5 msec. Because this is a technically demanding task, particularly for gradient switching, hybrid techniques have been used in which only two or four echoes are obtained after each excitation, with consequent savings in time (7, 8). Because each echo is reduced by T2, depending on the way this technique is implemented either the field of view or spatial resolution (or both) become T2 dependent. Another time-saving technique is to shorten TR but at the same time reduce the flip angle to less than 90°, thus recovering some of the object contrast characteristics of long TR techniques. Mills et al. have shown that time savings on the order of a factor of three are thus possible (9).

The three imaging speed-up strategies described above all change the output image characteristics. Another strategy is possible, one which maintains object contrast as in spin-echo imaging and can be used simultaneously with any of the previously described techniques. As is well known from MR spectroscopy, the time domain echo signal is symmetric around its peak. This symmetry is exact only if T2 is much longer than the echo width, which is a reasonable approximation in imaging. By analogy, the pseudo echo is also expected to be symmetric. Therefore, if only half of the data are acquired (top or bottom half in Fig. 1), the second half is, in principle, recoverable by a process called conjugation, which is described mathematically in the Appendix. With the peak signal as a reference point, each line of data is reflected about the peak, that is, translated to the opposite side and reversed left to right (Fig. 2). The data set thus obtained is then reconstructed by a standard 2DFT algorithm. Because the image has all the characteristics of the 2DFT spin-echo image except for lower S/N as described below, we chose to characterize those images by the parameter n being set to $1/2$ in equation (1). Had we set N to be $1/2$ the value, this would imply a smaller field of view in the output image. While only half the data plus the peak are needed, it is useful in practice to accumulate a few extra projections to permit better definition of the peak. For a 256-line image, accumulating 138 projections allows a savings in time of 46% rather than the 50% it would be for 128 projections (Fig. 3). Nevertheless, the reliability increase provided by the additional projections is well worth the slight increase in imaging time.

There are some points worth making regarding this process. First, S/N decreases to about 70% because of the halving in the number of data sets (Fig. 3). Second, the added computational complexities associated with conjugation are not worth the effort unless the system allows for $n = 1$ operation, since conjugation of an $n = 2$ procedure would have the same resolution and S/N of a conventional $n = 1$ procedure. Conjugation also needs to be differentiated from other techniques that use one-half of a data set. For instance, in Figure 1 only the inner half of the data set could be accumulated, and the outer two quarters zero-filled to a 256-line set. This in effect yields a low-resolution image on a higher-resolution matrix, containing no added information to that that would follow from an interpolation in image space (Fig. 4). A second acquisition mode would skip every other line in Figure 1 and fill it with an interpolated value. This is equivalent to a reduction in the field of view, resulting in fold-over artifacts. Both techniques decrease S/N to 70% because of the smaller data set used. Reconstruction of the half data set without prior conjugation pro-

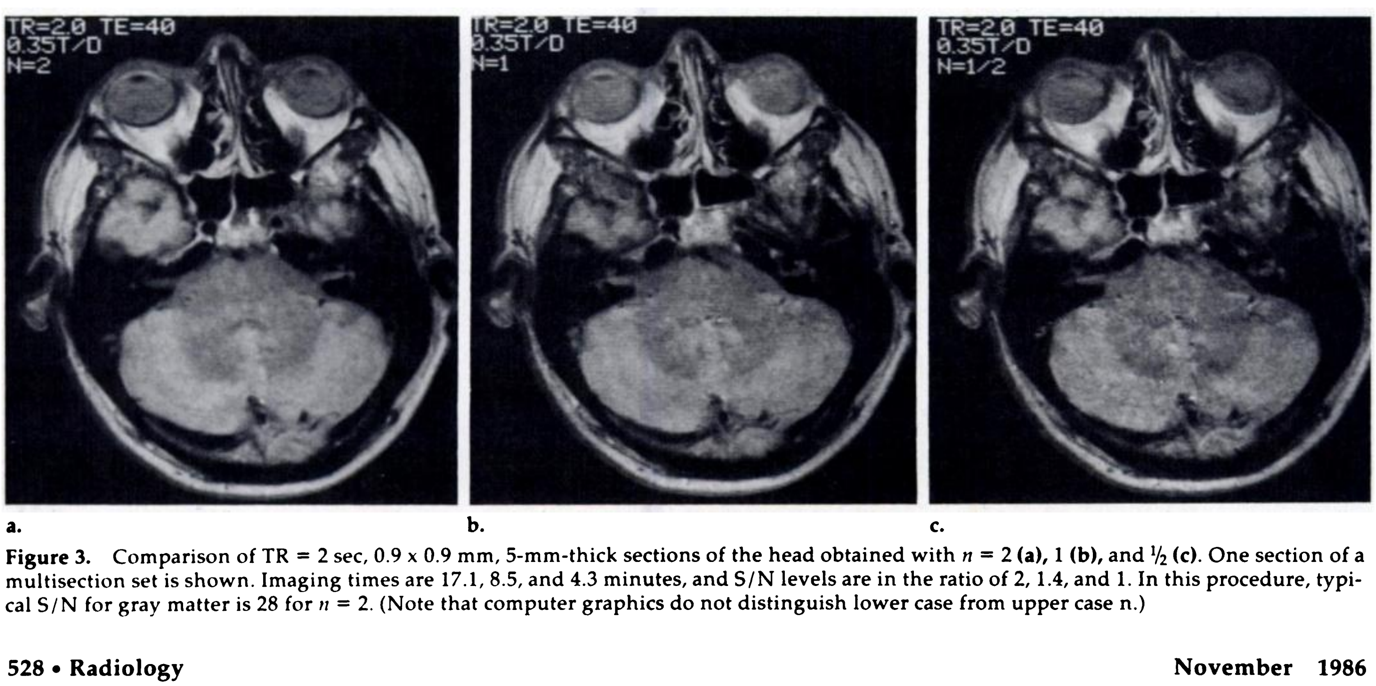


1.



2.

Figures 1, 2. **(1)** The square represents a 2DFT time domain set. Each line of the set has position encoded by a gradient along the x-axis. The lines differ by the amount of phase encoding applied along the y-axis. The Fourier transform of the spin-echo signal of a line provides one projection of the object. If a column is assembled by using one data point from each projection, each with the same value of x, a pseudo echo is obtained. Although it is produced by phase encoding, this echo is mathematically equivalent to a spin echo. **(2)** Because of symmetry, if only half of a 2DFT time domain set is acquired, as shown, the other half can be synthesized by reflecting the data about the center. For instance, the signal intensity value at $x = 61$, $y = 37$ would be transposed to $x = -61$, $y = -37$.



a. b. c.

Figure 3. Comparison of TR = 2 sec, 0.9 x 0.9 mm, 5-mm-thick sections of the head obtained with $n = 2$ **(a)**, 1 **(b)**, and $1/2$ **(c)**. One section of a multisection set is shown. Imaging times are 17.1, 8.5, and 4.3 minutes, and S/N levels are in the ratio of 2, 1.4, and 1. In this procedure, typical S/N for gray matter is 28 for $n = 2$. (Note that computer graphics do not distinguish lower case from upper case n.)

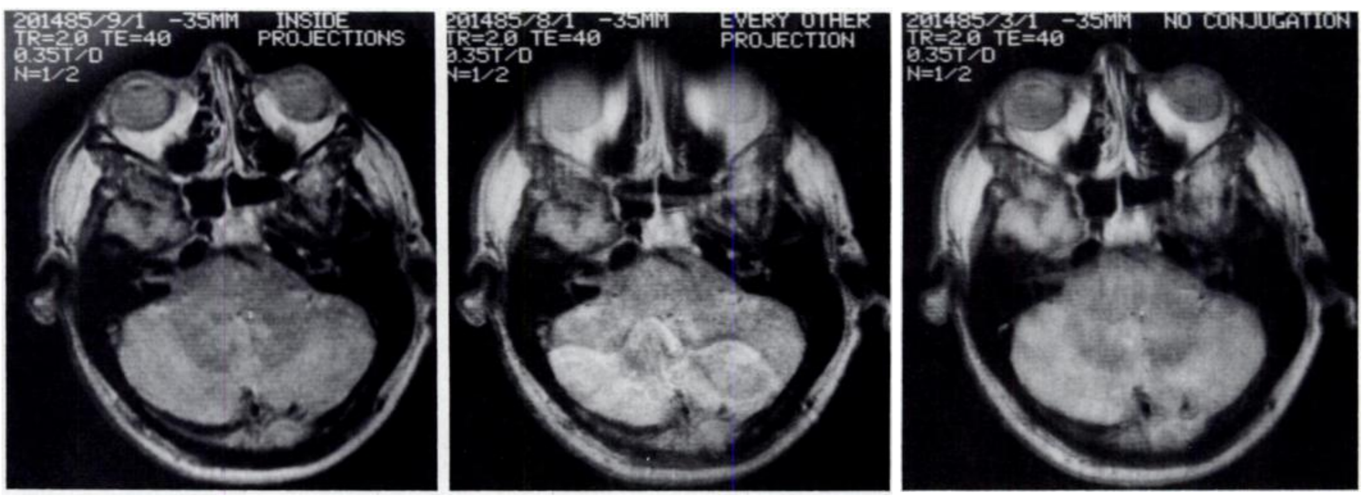
vides a recognizable image but with significant artifacts (Fig. 4). Conjugation recovers all of the spatial information present in a conventional 2DFT image, and only S/N is reduced as a direct result of the shortened imaging time.

EXAMPLES

Conjugate synthesis imaging was implemented in a prototype of an MT/S imager (Diasonics, South San Francisco, Calif.) operating at 3.5 kG. Head and body imaging was performed with quadrature detection coils. In Figures 5 and 6 we compare 10- and 5-mm-thickness images obtained with n = 2, 1, and ½, a resolution of 0.9 mm in a 256 x 256 matrix, and a TR of 2 sec. The acquisition times are 17.1, 8.5, and 4.3 minutes, respectively, for 20 two-echo sections in the 10-mm-thickness mode, and 16 or 32 sections for double- or single-echo 5-mm-thick sections. The ability to detect the large and small abnormalities in these patients demonstrates the rapid screening potential of the technique for head imaging (Fig. 7). This is of particular importance with anxious patients, patients with trauma, and children. Figure 8 shows body images obtained at a rate of 11 sec for three sections. Thickness is 10 mm, and resolution is 1.7 mm. The subject was asked to hold his breath, and the imaging procedure started just after the breath was held.

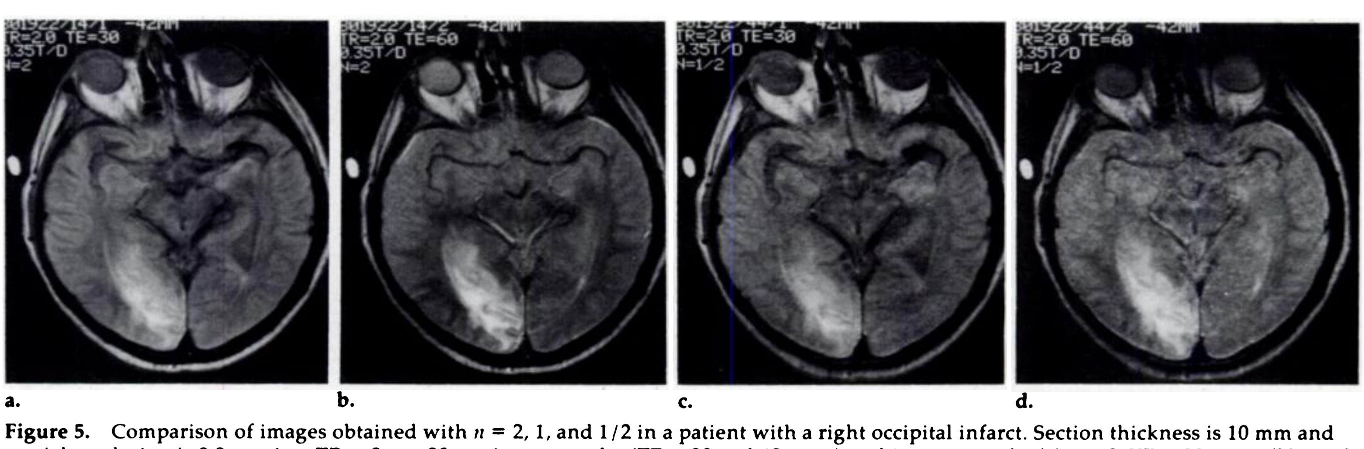
DISCUSSION

At 3.5 kG, S/N has increased to the point that for many procedures (e.g., head images with section thicknesses of 10 and 5 mm and resolution of 1.7 and 0.9 mm, and 10-mm-thick body images with resolution of 1.7 mm) it is possible to use conjugation to reduce imaging time by a factor of two while still obtaining diagnostically adequate images. In the head, the whole brain can be imaged in 2.1 minutes and 4.3 minutes for low- and high-resolution images, respectively, using a highly sensitive technique with TR = 2 sec. It is worth emphasizing that conjugate synthesis of data is only advantageous in MR imaging systems that have an intrinsic S/N level high enough to permit imaging with n = 1. If adequate diagnostic studies require for instance n = 2, then conjugation would reduce S/N to prohibitively low levels. While the conjugation procedure adds no new information to that available from longer imaging times, it improves patient acceptance; in systems in which photography does



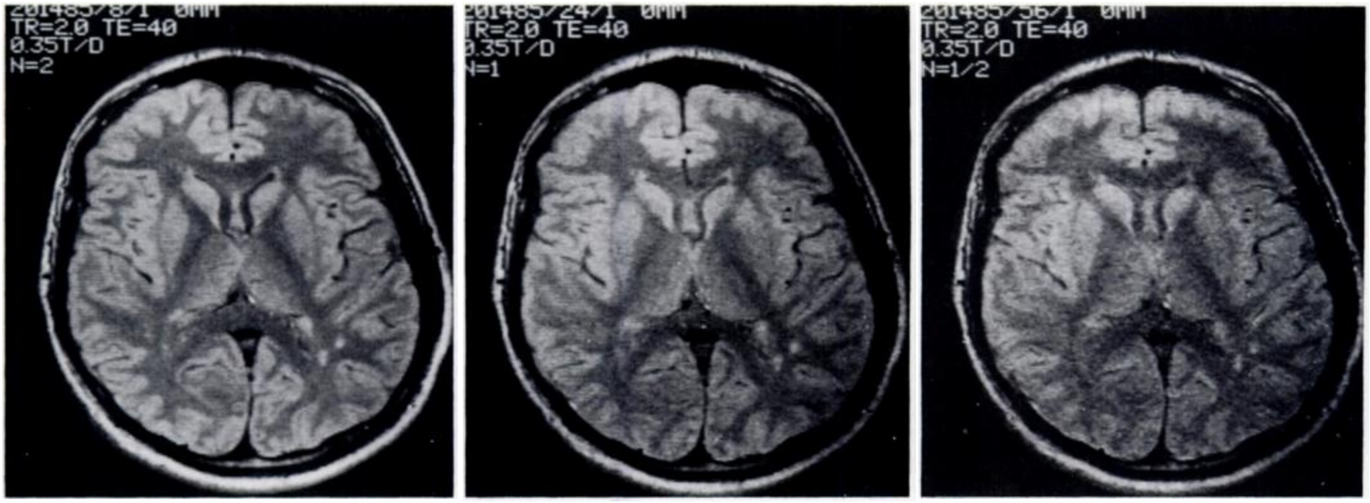
a. b. c.

Figure 4. This figure demonstrates the consequences of using strategies other than conjugation to reconstruct a 128-projection data set onto a 256-line matrix. **(a)** Only the inner 128 projections are acquired, and the data are zero-filled to 256 projections. This corresponds to interpolation in the spatial domain. Effective resolution is 1.8 mm, even though matrix pixel size is 0.9 mm. **(b)** Result of obtaining every other projection and interpolating. If no interpolation had been done, the field of view would be one-half that of a full data set. The interpolation does not fully recover the correct data, and overlap artifacts become apparent. **(c)** If the half data set is reconstructed without conjugation, blurring along the y-axis becomes evident.



a. b. c. d.

Figure 5. Comparison of images obtained with n = 2, 1, and 1/2 in a patient with a right occipital infarct. Section thickness is 10 mm and spatial resolution is 0.9 mm in a TR = 2 sec, 20-section, two-echo (TE = 30 and 60 msec) multisection mode. **(a)** n = 2, TE = 30 msec; **(b)** n = 2, TE = 60 msec; **(c)** n = 1/2, TE = 30 msec; **(d)** n = 1/2, TE = 60 msec.



a. b. c.

Figure 6. Transverse sections in a subject with left occipito-parietal white matter asymptomatic lesions. Section thickness is 5 mm, resolution is 0.9 mm, TR = 2 sec. Sixteen two-echo or 32 one-echo sections can be acquired. **(a)** $n = 2$, time = 17.1 minutes; **(b)** $n = 1$, time = 8.5 minutes; and **(c)** $n = 1/2$, time = 4.3 minutes.

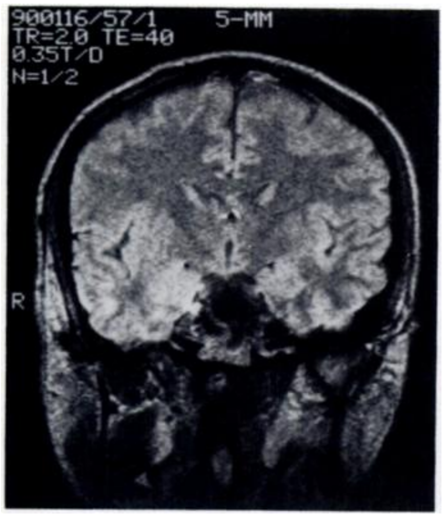
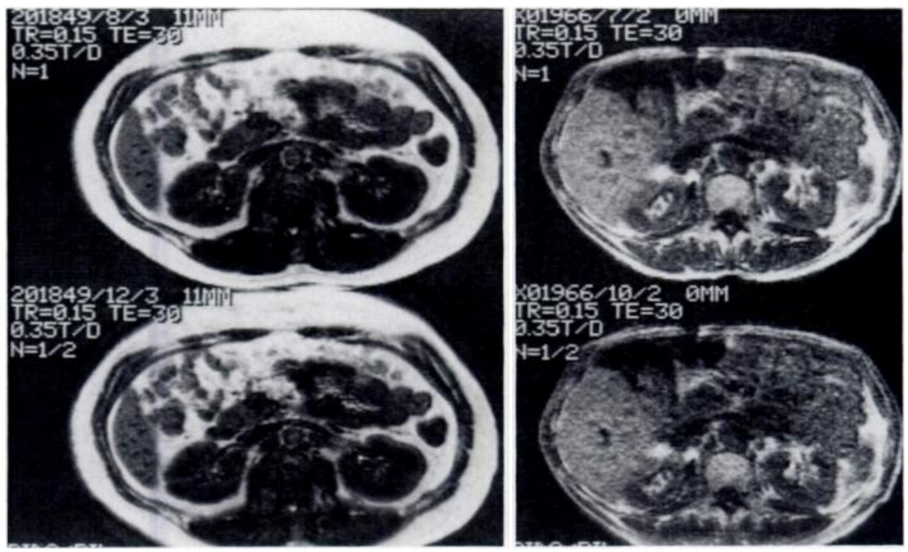


Figure 7. One of a multisection set of coronal $n = 1/2$ images of 5-mm-thickness and 0.9-mm resolution. For TR = 2 sec, 32 one-echo sections can be acquired in 4.3 minutes.



a. b.

Figure 8. Abdominal sections obtained with $n = 1/2$ at a rate of 11 sec for three sections. The subjects were asked to hold their breath prior to start of the procedure. **(a)** Healthy volunteer. Top: $n = 1$ (19 sec), bottom: $n = 1/2$ (11 sec). **(b)** Sixty-six-year-old alcoholic with impaired cognitive functions and occluded descending aorta. Specifications are same as in **a.**

not affect acquisitions, the conjugation procedure increases throughput significantly. In the body, fast MR imaging procedures permit single-breath-hold imaging. Conjugation can be used with other time-reducing procedures, such as partial flip-angle imaging with radio frequency refocusing pulses, echo-planar imaging, and "hybrid" imaging.

APPENDIX

The presence of conjugate symmetry in a spin echo has been known by MR spectroscopists for many years. This symmetry about the echo's central peak is a consequence of the memory of phase in the net magnetization vector during the echo formation process. In spectroscopy, conjugate symmetry offers the possibility of averaging the two halves of the echo for the purpose of increasing S/N, but complications may arise from T2 decay during the echo period when T2 is much shorter than TE. In MR imaging, T2 decay is constant throughout the multiple spin echoes acquired with different phase-encoding gradient strengths, G_y. The signals are conjugate symmetric, $M(\phi) = M^*(-\phi)$, about $\phi = 0$, the signal acquired with no net G_y. As mentioned by other investigators (7), the above relationship permits in theory a 2DFT spatial image to be acquired with $(N/2) + 1$, where N is the total number of image lines. Here it is convenient to define ϕ as the spatial frequency K_y in the Fourier transform's reciprocal space (time domain data),

$$K_y \cdot y = \gamma y \int G_y(t) dt,$$

where G_y is incremented in each cycle of the pulse sequence, and

$$K_x \cdot x = \gamma x \int G_x(t) dt,$$

where G_x is a constant gradient applied during the signal readout period. The time domain data is acquired with both $+k$ and $-k$ spatial frequencies for 2DFT,

$$\int_{-N/2}^{+N/2} \int_{-N/2}^{+N/2} M(k_x, k_y) \cdot \exp[-\gamma i(xk_x - yk_y)] \cdot dk_x \cdot dk_y = m(x,y),$$

where $m(x,y)$ is the two-dimensional image data.

In the experiments reported here, we

have computer synthesized nearly half the signals by applying the conjugate symmetry rule to the phase-encoded spatial frequencies $M(+k_x,+k_y) = M^*(-k_x,-k_y)$, noting that $(R + iI)^* = (R - iI)$ where R and I are the real and imaginary components, respectively, of a complex number. It is significant to flow imaging that the Fourier transform of such conjugate symmetric data produces only real components in the complex image data $m(x,y)$ except in spatial locations where phase shifts result from movement of spins in the direction of a magnetic gradient.

Conjugation is not limited to 2DFT imaging but can be extended to 3DFT MR imaging (1). Conjugation synthesis may also be used in Fourier transform velocity imaging (10) or in Fourier transform spectroscopic imaging to reduce imaging time. For example, $M(\pm k_x, -kv_z) = M^*(\pm k_x, +kv_z)$, where the velocity vector component on the z-axis (V_z) has been phase encoded with pairs of incremented gradient pulses. ■

References

1. Feinberg DA, Hoenninger JC, Crooks LE, Kaufman L, Watts JC, Arakawa M. Inner volume MR imaging: technical concepts and their application. Radiology 1985; 156:743–747.
2. Brant-Zawadzki M, Norman D, Newton TH, et al. Magnetic resonance of the brain: the optimal screening technique. Radiology 1984; 152:71–77.
3. Crooks LE, Ortendahl DA, Kaufman L, et al. Clinical efficiency of nuclear magnetic resonance imaging. Radiology 1983; 146:123–128.
4. Ortendahl DA, Hylton N, Kaufman L, et al. Analytical tools for magnetic resonance imaging. Radiology 1984; 153:479–488.
5. Droege RT, Wiener SN, Rzeszotarski MS. A strategy for magnetic resonance imaging of the head: results of semi-empirical model. II. Radiology 1984; 153:425–433.
6. Mansfield P, Maudsley AA, Baines T. Fast scan proton density imaging by NMR. J Phys E 1976; 9:271–278.
7. van Uijen CM, den Boef JH, Verschuren FJ. Fast Fourier imaging. Magn Reson Med 1985; 2:203–217.
8. Haacke EM, Bearden FH, Clayton JR, Linga NR. Reduction of MR imaging time by the hybrid fast-scan technique. Radiology 1986; 158:521–529.
9. Mills TM, Ortendahl DA, Hylton NM. Investigation of partial flip angle magnetic resonance imaging. IEEE Trans Nucl Sci 1986; NS-33:496–500.
10. Feinberg DA, Crooks LE, Sheldon PE, Hoenninger JC, Watts JC, Arakawa M. Magnetic resonance imaging the velocity vector components of fluid flow. Magn Reson Med 1985; 2:555–566.