Original Research

Rapid Three-dimensional T1-weighted MR Imaging with the MP-RAGE Sequence[1]

John P. Mugler III, PhD • James R. Brookeman, PhD

The authors investigated the application of three-dimensional (3D) magnetization-prepared rapid gradient-echo (MP-RAGE) imaging to the acquisition of small (32 × 128 × 256) T1-weighted 3D data sets with imaging times of approximately 1 minute. A theoretical model was used to study the contrast behavior of brain tissue. On the basis of these theoretical results, 3D MP-RAGE sequences were implemented on a 1.5-T whole-body imager. Thirty-two-section 3D data sets demonstrating good signal-to-noise ratios and resolution and strong T1-weighted contrast were obtained in 1 minute. Compared with standard short TR/TE spin-echo sequences with the same imaging times and comparable sequence parameters, the 3D MP-RAGE sequence delivered increases of more than 50% in the white matter/gray matter signal difference–to-noise and white matter signal-to-noise ratios, and provided almost twice as many sections. These sequences may find a clinical role in 3D scout imaging and screening and in patients with claustrophobia or trauma.

Index terms: Artifact • Brain, MR studies, 10.1214 • Image processing • Magnetic resonance (MR), experimental • Model, mathematical • Pulse sequences • Rapid imaging • Three-dimensional imaging

JMRI 1991; 1:561–567

Abbreviations: FLASH = fast low-angle shot, MP-RAGE = magnetization-prepared rapid gradient echo, RF = radio frequency, SE = spin echo, S/N = signal-to-noise ratio, SSI = steady-state incoherent, 3D = three-dimensional, WGSD = white matter/gray matter signal difference.

WE RECENTLY INTRODUCED three-dimensional (3D) magnetization-prepared rapid gradient-echo (MP-RAGE) imaging (1), which can acquire high-resolution, high-contrast 3D data sets in imaging times reasonable for clinical use. The 3D MP-RAGE technique is an extension and generalization of the prepare-acquire imaging philosophy introduced by Haase et al with snapshot FLASH (fast low-angle shot) imaging (2,3). A typical 3D MP-RAGE pulse sequence uses a three-step cycle: (*a*) magnetization preparation for contrast control, (*b*) data acquisition with a short TR gradient-echo sequence, and (*c*) magnetization recovery for additional contrast control. In the first implementations, the sequences were designed to acquire large T1-weighted 3D data sets (eg, 128 × 128 × 256) from regions such as the whole head or abdomen, so that the data could be reformatted to obtain high-quality images in any plane. The resulting acquisition times were approximately 4–6 minutes for image sets covering the whole head.

In the present study, we investigated the application of the 3D MP-RAGE sequence to the acquisition of smaller 3D data sets (32 × 128 × 256) obtained with much shorter imaging times (approximately 1 minute), within the constraints of maintaining reasonable signal-to-noise ratios (S/Ns) and in-plane resolution and strong T1-weighted contrast. Specifically, we developed a theoretical model to predict the contrast between brain white matter and gray matter as a function of the relevant pulse sequence parameters, including the preparation and excitation pulse flip angles and the phase-encoding order. This model was used to determine suitable pulse sequence parameters for experimental measurements in healthy volunteers. The resulting 3D MP-RAGE images were compared with standard short TR/TE spin-echo (SE) images with the same acquisition times and comparable sequence parameters. Applications of this implementation of the 3D MP-RAGE sequence include (*a*) rapid screening of large volumes of interest (eg, the whole head) and (*b*) rapid high-resolution 3D imaging of smaller volumes of interest (eg, the pituitary).

During this investigation, we found that potentially severe image artifacts, secondary to gradient eddy

[1] From the Departments of Radiology and Biomedical Engineering, MRI Facility, Box 190, University of Virginia Health Sciences Center, 1105 W Main St, Charlottesville, VA 22908. From the 1991 SMRI annual meeting. Received April 4, 1991; revision requested June 4; revision received and accepted July 12. Supported by the Whitaker Foundation and Siemens Medical Systems. **Address reprint requests to** J.P.M.

current effects, may be produced when centrally reordered phase encoding is used. We implemented a gradient preparation scheme during the magnetization preparation period of the 3D MP-RAGE sequence that proved effective in suppressing these image artifacts.

• METHODS

A theoretical model of the 3D MP-RAGE sequence based on the Bloch equations was formulated and implemented in FORTRAN on a VAX 11/750 computer (Digital Equipment, Maynard, Mass). The magnetization preparation period was modeled as a nonselective radio-frequency (RF) pulse, with a flip angle between 0° and 180°, followed by a variable delay. It was assumed that any transverse magnetization generated by the preparation pulse was adequately spoiled with gradients. The gradient-echo acquisition was modeled as an ideal spoiled FLASH-type (4) sequence (ie, the contribution of the transverse magnetization excited by a given pulse to succeeding acquisitions was neglected) that used a nonselective excitation for the 3D volume acquisition. The magnetization recovery period was a variable delay. T1 relaxation was incorporated throughout the pulse sequence. Because the transverse magnetization generated by any given RF pulse was assumed to be ideally spoiled before the succeeding RF pulse, T2 relaxation was incorporated only between each excitation pulse for the data acquisition sequence and the succeeding TE. The pulse sequence timing diagram is shown in Figure 1.

The signal intensity for a given tissue was taken as that which occurred when the zero spatial frequency component was sampled. For signal intensity and contrast calculations, the complete sequence cycle was repeated until a steady state was achieved. When the preparation and excitation angles were both relatively small (preparation, $<50°$; excitation, $<10°$), steady state was reached in 10–30 cycles. For other preparation and excitation angles, including those used in the experimental measurements, steady state was reached in less than 10 cycles. The signal intensity and contrast calculations were based on the assumption that magnitude reconstruction was used in the images.

The theoretical model was used to study the contrast behavior for representative tissue parameters as the total acquisition time for 32 sections was reduced to approximately 1 minute and as the sequence parameters were varied. In particular, we calculated the signal intensity difference between white matter (T1/T2/proton density = 550 msec/90 msec/0.65 [normalized]) and gray matter (940/100/0.75) as a measure of T1-dependent contrast in the brain, using relaxation values from the literature (5,6). Both standard sequential and centrally reordered phase-encoding schemes were considered for the 32-step phase encoding (third dimension) performed within each sequence cycle. Within each cycle the magnetization is sampled during a transient that depends on the tissue relaxation times and the acquisition sequence parameters. Because tissue contrast is determined primarily by the low-spatial-frequency components, the temporal location of these spatial frequency components within the acquisition period can significantly affect

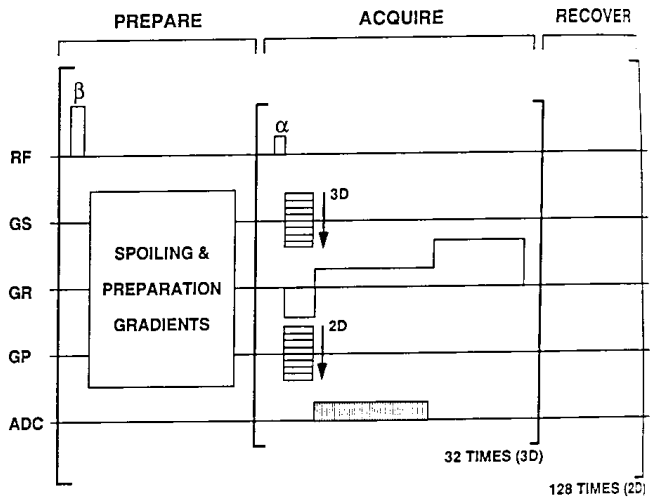


Figure 1. Pulse sequence timing diagram for the rapid T1-weighted 3D MP-RAGE sequence, demonstrating the basic prepare-acquire-recover sequence structure. The magnetization preparation period consists of a nonselective RF pulse (β) followed by a variable delay. During this delay, gradients are applied for spoiling and/or stabilization of eddy currents. Data acquisition is performed with a FLASH gradient-echo sequence, with 32 phase-encoding steps in each sequence cycle. *ADC* = analog to digital converter.

contrast. With use of centrally reordered phase encoding, image contrast reflects the magnetization state at the end of the preparation period, before the magnetization is significantly modified by relaxation and acquisition (7–9). For centrally reordered phase encoding, the zero spatial frequency component is sampled first and sampling then proceeds from low to high spatial frequencies, alternating between positive and negative.

Standard sequential phase encoding was used for the in-plane direction. Considered separately, the acquisition order for in-plane phase encoding should not significantly affect contrast unless low spatial frequencies are acquired before a steady state of the sequence cycle is reached. The effects of independently reordering phase encoding for the in-plane direction, or of jointly reordering phase encoding for both the in-plane and 3D directions, were not investigated.

The performance of the sequence was compared theoretically to a standard steady-state incoherent (SSI) gradient-echo acquisition with the same total acquisition time. The MP-RAGE and SSI sequences were assumed to use data sampling periods of equal duration. The theoretical results were used as a guide to selecting the pulse sequence parameters for experimental tests. The 3D MP-RAGE sequences were implemented on a standard 1.5-T whole-body imager (Magnetom 63SP; Siemens Medical Systems, Iselin, NJ), and tested in healthy volunteers after informed consent was obtained.

As a control, the 3D MP-RAGE sequences were compared experimentally with standard SE sequences with the same imaging times. The SE sequences were chosen from the standard sequences supplied with our imager. For these sequences, the shortest possible TEs were chosen within the con-

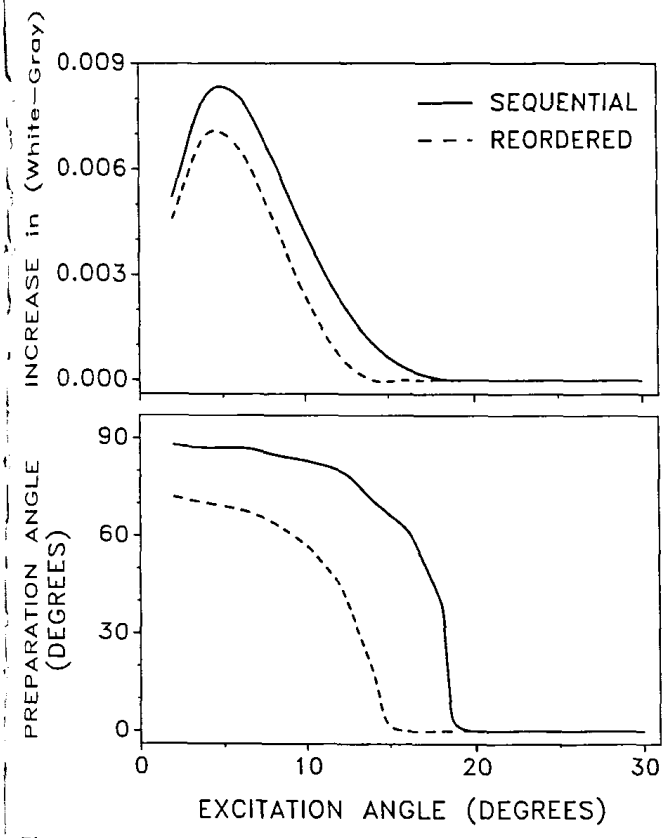


Figure 2. *In the top graph, the WGSD maximum over the* range of preparation pulse flip angles minus the WGSD for no preparation pulse (*INCREASE in [White-Gray]*) is plotted as a function of the excitation angle used in the gradient-echo acquisition. The ordinate is in units of normalized transverse magnetization. In the bottom graph, the preparation pulse flip angle yielding the maximum WGSD is plotted as a function of the excitation angle. In both graphs, curves are shown for standard sequential and centrally reordered phase encoding. Pulse sequence parameters: preparation—RF pulse followed by a 140-msec delay; acquisition—FLASH sequence with TR/TE of 10/4 and 32 phase-encoding steps per cycle; recovery—none.

straints of section thickness requirements, thus allowing the maximum number of sections. The TRs for the SE sequences were set equal to the duration of one sequence cycle of the corresponding 3D MP-RAGE sequences. For comparison with the 3D acquisition, the SE sequences acquired contiguous sections with an interleaved acquisition order. Other sequence parameters such as section thickness, in-plane matrix size, and field of view were the same for both sequence types. White matter/gray matter signal difference (WGSD)–to-noise ratios and white matter S/Ns were calculated for each sequence. In all cases, the data sampling bandwidth was 33% narrower for the SE sequences, providing a relative decrease in the noise level. No corrections were applied for bandwidth differences.

Another informative experimental comparison would be between the 3D MP-RAGE and an RF-spoiled steady-state gradient-echo sequence. However, because of current hardware limitations, we could not perform this comparison.

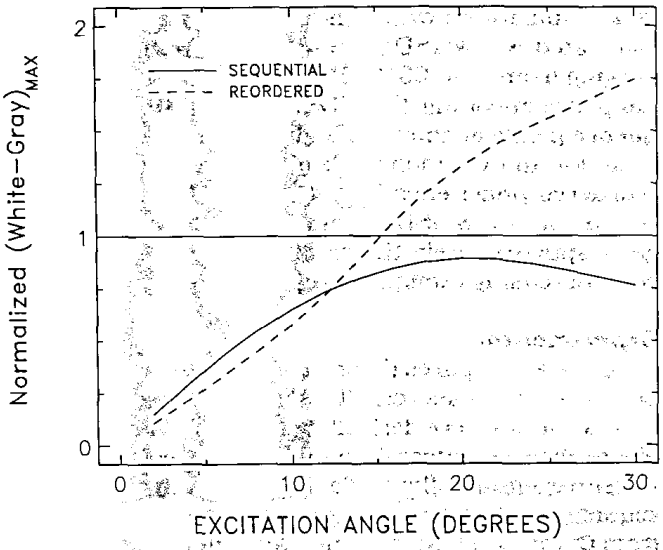


Figure 3. WGSD maximum over the range of preparation pulse flip angles, normalized by the WGSD maximum for an SSI sequence with equal imaging time, is plotted as a function of the excitation angle used in the gradient-echo acquisition. Curves are shown for standard sequential and centrally reordered phase encoding. Pulse sequence parameters: preparation—RF pulse followed by a 140-msec delay; acquisition—FLASH sequence with TR/TE of 10/4 and 32 phase-encoding steps per cycle; recovery—none.

• RESULTS

Theoretical

Assuming an acquisition of 32 phase-encoding steps per cycle and a TR of approximately 10 msec for the gradient-echo acquisition, an acquisition time of approximately 1 minute for a 128 × 256 in-plane matrix allows a maximum time of 100–200 msec in each cycle for the magnetization preparation and recovery periods. Because the T1 values of interest are longer than 100–200 msec, all this time was allocated to the magnetization preparation period to allow contrast to develop after the preparation pulse (ie, the magnetization recovery period was set to zero).

Results for a 3D MP-RAGE sequence with a preparation period of 140 msec and a TR msec/TE msec of 10/4 for the gradient-echo acquisition, yielding an imaging time of 1.0 minutes for a 32 × 128 × 256 matrix, are shown in Figures 2 and 3. The plots demonstrate the changes that occur in the WGSD as the preparation and excitation pulse flip angles are varied.

Figure 2 demonstrates the effectiveness of the preparation pulse for introducing T1-dependent contrast. The preparation pulse provides significant increases in WGSD for excitation angles below approximately 15° but no increase in WGSD for excitation angles above 20°. The results for sequential and reordered phase encoding are similar; in general, the preparation pulse is somewhat less effective for reordered phase encoding.

In Figure 3, the WGSD maximum over the range of possible preparation pulse angles (see Fig 2, bottom graph) is plotted, normalized by the WGSD maximum of the SSI sequence (14.4/4, 23° flip angle). The SSI

WGSD maximum was found by a global search of the calculated SSI WGSD values for excitation flip angles ranging from 0° to 90° in 1° increments. For sequential phase encoding, the WGSD for the MP-RAGE sequence peaks at about 90% of the SSI WGSD maximum for an excitation angle of approximately 20°. For reordered phase encoding, the MP-RAGE WGSD equals the SSI WGSD maximum for an excitation angle of approximately 15°, and continues to increase with increasing excitation angle.

Experimental

Figure 4 compares theoretically predicted and experimentally measured WGSD values. The experimental values were derived from coronal head images of a healthy volunteer. Ten image sets were acquired, with an excitation flip angle of 5° for the gradient-echo sequence and preparation pulse flip angles ranging from 0° to 90°. The experimental values follow the same trend as the theoretical curve and agree reasonably well with the theoretical values.

Sets of 3D MP-RAGE images covering the whole head (32 sagittal sections, 5 mm thick) were acquired in 1.0 minutes with no preparation pulse, a 20° excitation pulse for the gradient-echo acquisition, and centrally reordered phase encoding. These image sets demonstrated that the whole head could be imaged with contiguous 5-mm sections displaying good contrast and S/N, in only 1 minute. Compared with the SE reference sequence (460/10, 18 sections), the 3D MP-RAGE sequence yielded a 64% increase in the WGSD-to-noise ratio and a 52% increase in the S/N for white matter, and provided 14 additional sections.

When centrally reordered phase encoding was used, as in the aforementioned example, we found that image artifacts were generated. These artifacts, consisting of a pronounced blurring in the 3D direction and other intensity distortions, were apparently caused by time-dependent eddy currents that produced shifts in the position and phase of the echo for the first several phase-encoding steps of the gradient-echo acquisition in each sequence cycle. Gradient preparation (ie, executing the non-phase-encoding gradient elements of the gradient-echo sequence several times before the start of data acquisition) was used to stabilize the eddy currents before the start of data acquisition and in turn to suppress the artifacts.

Figure 5 shows sagittal head images that demonstrate typical artifacts seen with centrally reordered phase encoding and suppression of these artifacts with gradient preparation. The images in Figure 5a and 5b were acquired without gradient preparation. Note the artifactual intensity outside the head (Fig 5a) and the intensity distortions in the region of the upper cervical spine (Fig 5b). The images in Figure 5c and 5d were acquired with the same sequence parameters as in Figure 5a and 5b, except that the data acquisition in each sequence cycle was preceded by 14 repetitions of the non-phase-encoding gradient elements of the gradient-echo sequence. As seen in Figure 5c and 5d, the artifacts were suppressed by the gradient preparation.

Figure 6 shows images of four 2-mm transverse sections from a set of 32 sections acquired in 1.1 minutes with a 90° preparation pulse, a 15° excitation

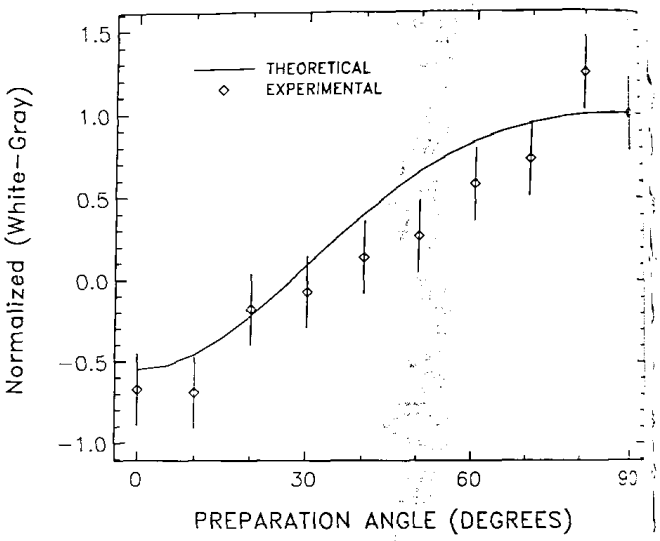


Figure 4. Comparison of theoretically predicted and experimentally measured WGSDs for an excitation pulse angle of 5° and preparation pulse angles ranging from 0° to 90°. The vertical bars centered on each experimental value represent plus and minus one standard deviation based on the thermal noise level. The WGSDs are normalized with respect to the value for a 90° preparation pulse. Pulse sequence parameters: preparation—RF pulse followed by a 140-msec delay; acquisition—FLASH sequence with TR/TE of 10/4, 32 × 128 × 256 matrix, 250-mm field of view, 5-mm section thickness, sequential phase encoding; recovery—none; total acquisition time, 1.0 minutes.

pulse for the gradient-echo acquisition, and sequential phase encoding. The images in Figure 6 demonstrate that even relatively thin sections can be obtained in approximately 1 minute with reasonable image quality. Compared with the SE reference sequence (525/15, 17 sections), the 3D MP-RAGE sequence yielded a 300% increase in the WGSD-to-noise ratio and a 65% increase in the S/N for white matter, and provided 15 additional sections.

Coronal sections reconstructed from transverse acquisitions are shown in Figure 7. The image in Figure 7a was reconstructed from the data set represented in Figure 6. In Figure 7b, the sequence parameters were the same except that the preparation pulse was removed. The two reconstructed images demonstrate similar contrast behavior, but the image obtained with no preparation pulse shows a pronounced signal intensity banding along the section-select direction, due to incomplete spoiling of transverse magnetization.

• DISCUSSION

At low excitation angles, contrast properties are dominated by the effects of the preparation pulse (3), and as such the preparation pulse is effective in providing large increases in WGSD (Fig 2) by converting the contrast from proton-density weighted (no preparation pulse) to T1 weighted (relatively high preparation pulse angle). As the excitation angle increases above 15°–20°, the net effect of the 32 excitation pulses outweighs that of the preparation pulse. As signal behavior becomes dominated by the excitation pulses, the maximum WGSD for a given excitation

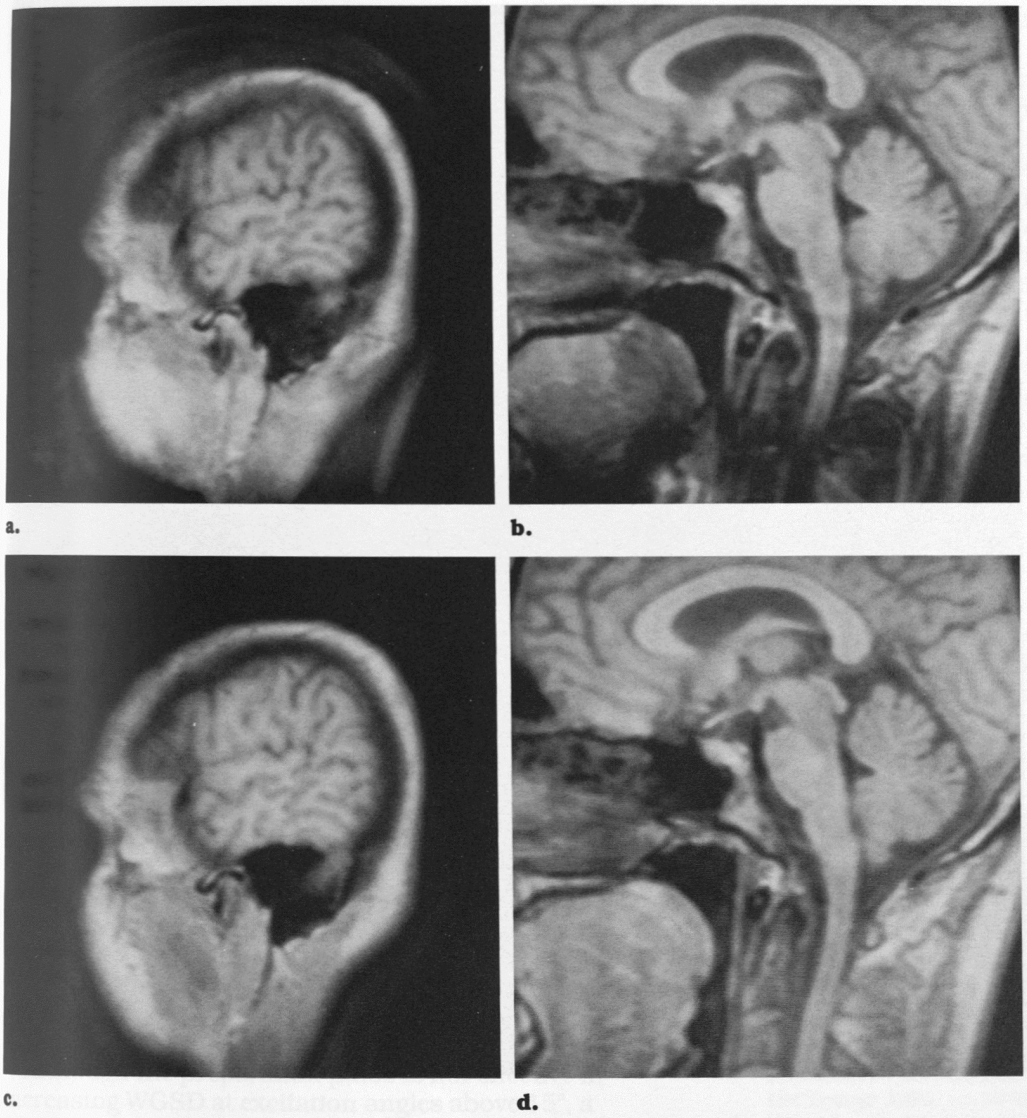


Figure 5. Sagittal head images of healthy volunteer acquired without **(a, b)** and with **(c, d)** gradient preparation before the start of gradient-echo acquisition in each sequence cycle, demonstrating the effects of eddy currents on the acquisition when centrally reodered phase encoding is used. Pulse sequence parameters: preparation—140-msec delay **(a, b)** or 140-msec gradient preparation **(c, d)**; acquisition—FLASH sequence with TR/TE of 10/4, 15° flip angle, 32 × 128 × 256 matrix, 250-mm field of view, 5-mm section thickness, centrally reordered phase encoding; recovery—none; total acquisition time, 1.0 minutes.

angle becomes much larger for centrally reordered phase encoding (Fig 3).

The contrast properties that can be achieved with the 3D MP-RAGE sequence are very dependent on the time made available during each sequence cycle for magnetization preparation and recovery, relative to the relaxation times of interest. For the rapid 3D acquisitions being investigated, we have severely restricted the preparation and recovery periods by limiting the total acquisition time to 1 minute. Despite this limitation, the theoretical calculations indicate that WGSD values significantly larger than the maximum value possible with an SSI sequence can still be achieved with use of relatively large excitation angles and centrally reordered phase encoding (Fig 3). Indeed, extrapolation of the theoretical values suggests that a 90° excitation angle would provide even greater WGSD values. However, for relatively large preparation pulse angles, the characteristics of the corresponding point spread functions become unacceptable. In addition, even at moderate flip angles (eg, 30°) at which the point spread functions are still well behaved, there are other problems in trying to realize these large WGSD values. In our experience with head imaging, section-to-section intensity changes due to incompletely spoiled transverse magnetization (10) become noticeable when the excitation angle reaches 15°–20°. The exact values depend on several factors, such as the section thickness and the details of any gradient-spoiling scheme applied. As a result, we have been limited for the FLASH-type acquisition to the range of excitation angles that produce WGSD values approximately equal to those obtainable with an ideal SSI sequence. As a possible solution to this problem, we are currently investigating the use of gradient-rephased sequences (eg, FISP [fast imaging with steady-state precession], GRASS [gradient-recalled acquisition in the steady state]) for the data acquisition. Preliminary results indicate that WGSD values significantly larger than the maximum for an SSI sequence can be achieved with these sequences. The flip angles and phase-encoding schemes chosen for the experimental examples reflect parameter combinations that do not result in any significant section-to-section intensity changes.

Compared with the SE sequences used as a control, the 3D MP-RAGE sequences provided significantly increased WGSD-to-noise ratios, S/Ns for white mat-

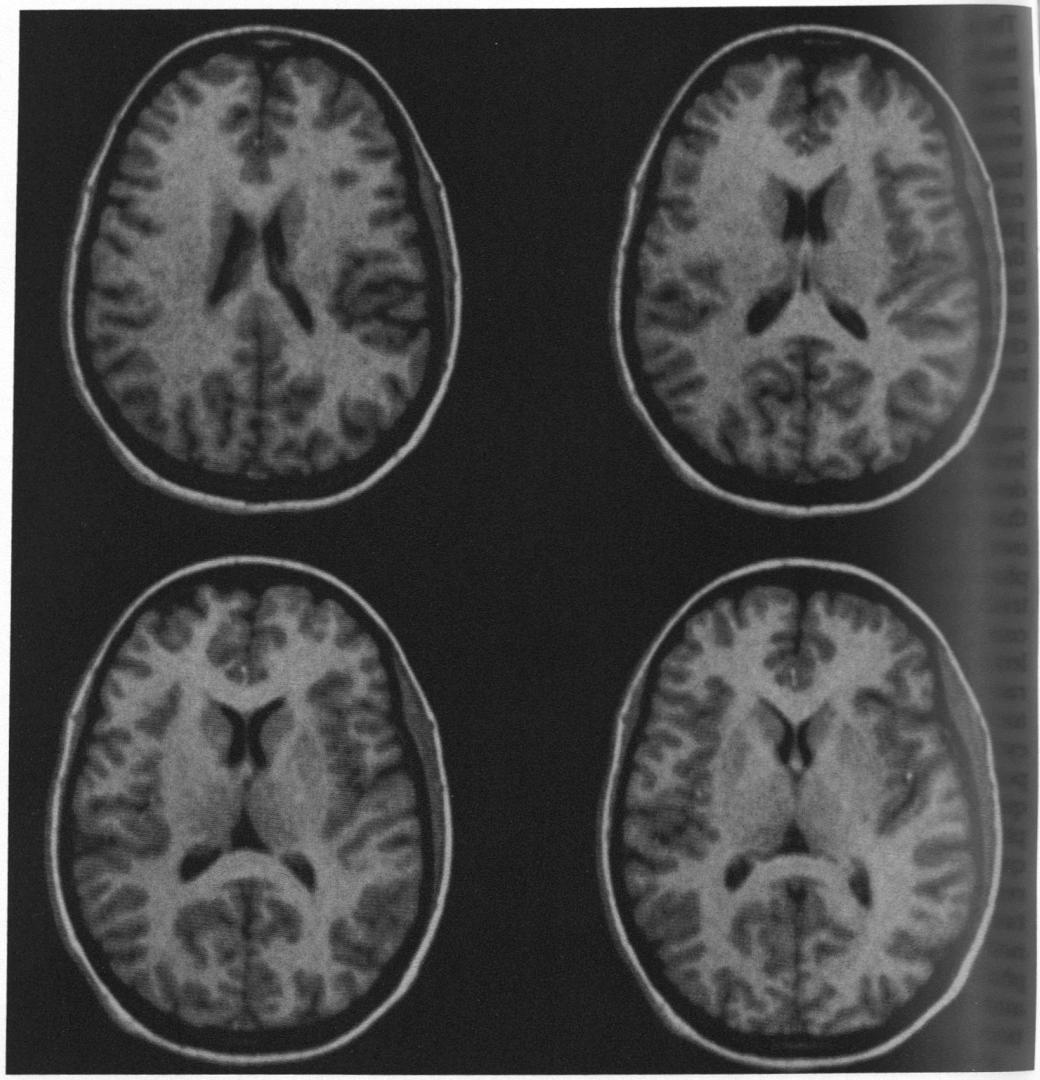
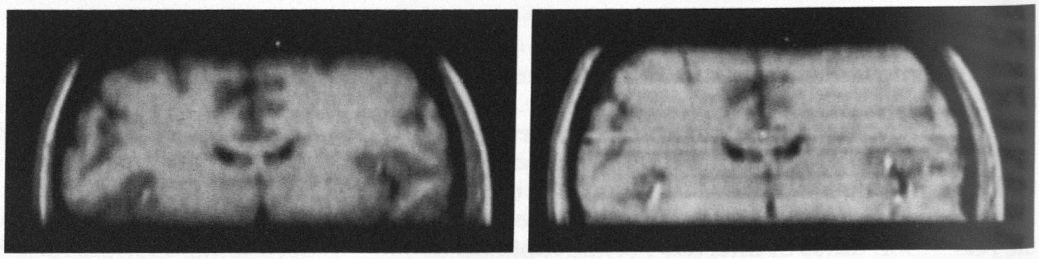


Figure 6. Transverse head images (2-mm-thick sections) of a healthy volunteer from a 3D set of 32 sections. Total acquisition time was 1.1 minutes. Pulse sequence parameters: preparation—90° RF pulse followed by a 140-msec delay; acquisition—FLASH sequence with TR/TE of 12/5, 15° flip angle, 32 × 128 × 256 matrix, 250-mm field of view, sequential phase encoding; recovery—1 msec.



a. **b.**

Figure 7. Coronal sections reconstructed from transverse acquisitions. **(a)** Reconstruction from the data set represented in Figure 6. **(b)** Reconstruction from an acquisition with the same parameters as in **a** but no preparation RF pulse.

ter, and coverage (sections per unit time). Given that the SE sequences used contiguous sections for comparison with the 3D acquisition, longer RF pulses (for better section definition and reduced section-to-section crosstalk) would probably provide improved S/Ns and WGSD-to-noise ratios for SE sequences, but with the penalty of further reducing coverage. One advantage retained by the SE sequences is their relative immunity to susceptibility-based artifacts. The combination of short TEs, small voxel dimensions, and 3D acquisition results in minimal susceptibility artifacts on gradient-echo images. For example, the 3D MP-RAGE sequence used in Figure 6 (1 × 2 × 2-mm voxels; TE, 5 msec) produces images nearly equivalent to SE images with respect to intensity artifacts at susceptibility interfaces such as between air and tissue. However, these susceptibility artifacts become apparent with the increased section thickness in the 3D MP-RAGE sequence used to cover the entire head in a 1-minute acquisition (1 × 2 × 5-mm voxels; TE, 4 msec). With this latter sequence, susceptibility artifacts were more pronounced in the regions of the sinuses and pituitary gland, relative to the SE sequence.

When 3D data sets are acquired with relatively thick sections, truncation artifacts in the section-select direction may present a problem (11). The truncation artifacts can be easily suppressed in one of two ways.

The data can be windowed in the section-select direction before Fourier transformation, to reduce the magnitude of the side lobes of the corresponding point spread function. Alternatively, the same signal versus phase-encoding step relationship can be obtained with use of variable flip angles in the gradient-echo acquisition (12). In either case, the side-lobe amplitudes are reduced at the expense of an increase in the width of the central lobe of the point spread function. However, for a rapid screening sequence, such as the 1-minute 3D MP-RAGE sequence, a slight increase in the effective section thickness is probably not objectionable.

The use of gradient preparation proved effective in suppressing image artifacts that result from the effects of gradient eddy currents on the centrally reordered gradient-echo acquisitions. However, we find that with the present sequence implementations, overall image quality is still superior with sequential phase encoding. Given the potential benefits of centrally reordered phase encoding with respect to image contrast behavior, further investigation of the problems unique to this phase-encoding scheme is warranted. Presumably, eddy currents cause similar distortions in the first several echoes of each sequence cycle when sequential phase encoding is used. We postulate that similar artifacts are not seen with sequential phase encoding because the first several phase-encoding steps correspond to high spatial frequencies, and changes in these components would result in intensity artifacts much smaller in magnitude and with an appearance similar to background noise. In addition, we believe that centrally reordered phase encoding may be more sensitive to echo phase and position errors; this is a topic under investigation.

Although the preparation pulse is not effective in increasing WGSD at excitation angles above 15°, a nonzero preparation pulse angle can still provide significant improvements in image quality when a FLASH-type gradient-echo sequence is used for data acquisition (Fig 7). For the sequence parameters of Figure 7, theoretical calculations indicate that WGSD values vary over a range of only about 15% as the preparation pulse angle varies from 0° to 90°. However, as shown in Figure 7, section-to-section intensity variations are pronounced for no pulse as compared with the 90° pulse. We postulate that intensity banding is significantly reduced by the 90° preparation pulse because of the overall reduction in average signal level with this pulse and its effect on the development of coherent transverse magnetization. With a reduction in overall signal strength, the amount of transverse magnetization available to produce coherence artifacts is in turn reduced. The presence of the preparation pulse perturbs the development of transverse coherences across sequence cycles.

In summary, we have shown that 3D data sets ($32 \times 128 \times 256$) demonstrating strong T1-weighted contrast and relatively high S/N can be obtained in imaging times of only 1 minute with the 3D MP-RAGE technique and standard commercial whole-body imaging hardware. These sequences may be clinically useful for quick 3D scout imaging, for screening, and for patients with claustrophobia or trauma; the potential clinical roles of these applications are currently being investigated.

With a FLASH-type sequence for data acquisition, a WGSD approximately equal to that obtained with an ideal SSI sequence can be achieved. However, the 3D MP-RAGE sequence offers at least two advantages over the standard SSI acquisition. First, thin sections can be obtained without section-to-section intensity variations, although the same result can be achieved with standard gradient-echo sequences in conjunction with RF spoiling. Second, there is time available during the preparation period to perform secondary preparations such as spatial or chemical presaturation. The anatomic coverage and S/Ns and WGSD-to-noise ratios for the 3D MP-RAGE sequence were significantly greater than those for standard SE sequences with equivalent imaging times and comparable sequence parameters. Currently, we are investigating the use of gradient-rephased sequences (eg, FISP, GRASS) with larger excitation angles ($> 20°$), possibly in conjunction with RF spoiling, for the 3D MP-RAGE data acquisition sequence. This sequence configuration may potentially yield rapid acquisition with contrast values substantially greater than those that can be achieved with an RF-spoiled steady-state gradient-echo sequence. ●

Acknowledgments: The authors thank Wayne S. Cail, MD, Director of Neuroradiology, and Eduard E. de Lange, MD, Director of Body MR, for their many helpful discussions concerning the clinical goals of MR imaging.

References

1. Mugler JP III, Brookeman JR. Three-dimensional magnetization-prepared rapid gradient-echo imaging (3D MP RAGE). Magn Reson Med 1990; 15:152–157.
2. Haase A, Matthaei D, Bartkowski R, Duhmke E, Leibfritz D. Inversion recovery snapshot FLASH MR imaging. J Comput Assist Tomogr 1989; 13:1036–1040.
3. Haase A. Snapshot FLASH MRI: applications to T1, T2, and chemical shift imaging. Magn Reson Med 1990; 13:77–89.
4. Haase A, Frahm J, Matthaei D, Hanicke W, Merboldt K-D. FLASH imaging: rapid NMR imaging using low flip-angle pulses. J Magn Reson 1986; 67:258–266.
5. Bottomley PA, Hardy CJ, Argersinger RE, Allen-Moore G. A review of 1H nuclear magnetic resonance relaxation in pathology: are T1 and T2 diagnostic? Med Phys 1987; 14:1–37.
6. Brix G, Schad LR, Deimling M, Lorenz WJ. Fast and precise T1 imaging using a TOMROP sequence. Magn Reson Imaging 1990; 8:351–356.
7. Norris DG, Bottcher U, Leibfritz D. A simple method of generating variable T1 contrast images using temporally reordered phase encoding. Magn Reson Med 1990; 15:483–490.
8. Holsinger AE, Riederer SJ. The importance of phase-encoding order in ultra-short TR snapshot MR imaging. Magn Reson Med 1990; 16:481–488.
9. Mugler JP III, Spraggins TA. Improving image quality in snapshot FLASH and 3D MP RAGE sequences by employing reordered phase encoding (abstr). In: Book of abstracts: Society of Magnetic Resonance in Medicine, 1990. Berkeley, Calif: Society of Magnetic Resonance in Medicine, 1990; 1310.
10. Wood ML, Runge VM. Artifacts due to residual magnetization in three-dimensional magnetic resonance imaging. Med Phys 1988; 15:825–831.
11. Carlson J, Crooks L, Ortendahl D, Kramer DM, Kaufman L. Signal-to-noise ratio and section thickness in two-dimensional versus three-dimensional Fourier transform MR imaging. Radiology 1988; 166:266–270.
12. Mugler JP III. Shaping the signal response of short TR gradient-echo sequences during the approach to steady state: a technique for improving image quality (abstr). In: Book of abstracts: Society of Magnetic Resonance in Medicine, 1991. Berkeley, Calif: Society of Magnetic Resonance in Medicine, 1991; 847.