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Correction of CSF Motion Artifact on MR Images of the Brain and Spine by Pulse Sequence Modification: Clinical Evaluation

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A modification of the standard spin-echo pulse sequence designed to suppress motion artifacts was clinically evaluated on T2-weighted MR images of the cervicocranial region. A retrospective study involving 40 patients, half of whom were examined with a standard T2-weighted multislice spin-echo sequence and half of whom were examined with a gradient waveform modification of the same sequence, uniformly demonstrated restoration of CSF signal intensity on images obtained with the gradient modified sequence. The cervical subarachnoid spaces, cisterna magna, medullary cistern, pontine cistern, fourth ventricle, and aqueduct were more consistently and brightly represented. However, the phase-encoding artifacts arising from CSF motion were not significantly reduced by using the gradient waveform modified pulse sequence.

Digital subtraction of an image obtained with the standard sequence from an image of the same slice with the gradient modified sequence provides a direct image representation of CSF flow.

Vascular pulsations transmitted to the CSF can cause loss of signal and may also result in phase-encoding artifact on T2-weighted spin-echo MR images of the brain and spine [1–7]. Techniques proposed to correct these flow-induced phenomena include cardiac gating [2], even-echo rephasing [4, 5], judicious choice of phase-encoding direction [8], and averaging of additional excitations [9]. Of these methods, the most extensively employed to date has been cardiac gating, but this involves additional patient preparation and requires that the pulse repetition time (TR) be a multiple of the cardiac cycle time [10].

A pulse sequence modification entitled "motion artifact suppression technique" and abbreviated MAST* [11, 12] has recently been introduced. This technique reduces motion artifact on T2-weighted spin-echo images by employing tailored read and slice-select gradient waveforms. This report describes initial clinical assessment of this pulse sequence modification on T2-weighted spin-echo sagittal MR images of the head and cervical spine. Images incorporating the MAST modification were compared with similar studies of the head and cervical spine obtained with a standard T2-weighted spin-echo sequence.

Background

The most extensively used pulse sequence in clinical MR imaging currently is the multislice spin-echo technique (Fig. 1). Radiofrequency pulses and pulses of magnetic field gradients along three orthogonal spatial axes (x,y,z) are employed to generate electrical (radiofrequency) signals that can be processed into images depicting the location of tissue-containing hydrogen. To generate a diagnostically useful image, this pulse sequence (which lasts less than a few seconds) must be repeated a number of times with variation in the amplitude of the magnetic gradient field in the phase-encoding direction. Each such cycle, referred to as a phase-encoding step, contributes to the quality of the entire image. Corruption or loss of signal in one or more such cycles during the course of a scan, whether related to instrument malfunction or to motion within the patient, results in artifacts on the resultant image. When

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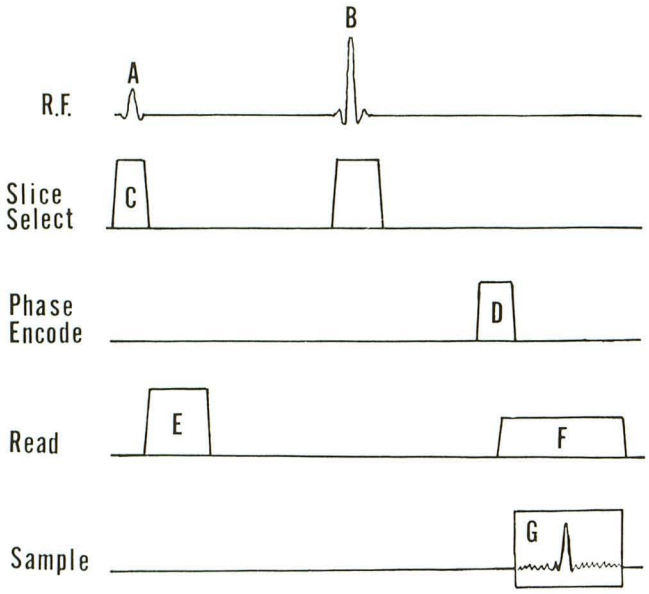

Fig. 1.—Pulse sequence diagram of standard T2-weighted (TE = 100) multislice spin-echo imaging sequence. The five channels depict the RF pulses, the amplitudes of the magnetic field gradients, and the data acquisition as each occurs in time. A selective 90° RF pulse (A) in the presence of the slice-select gradient (C) generates transverse magnetization for only a thin slab of tissue. This transverse magnetization is dephased by the action of the first lobe of the read gradient (E). A selective 180° RF pulse at time TE/2 (B) prepares the transverse magnetization so that an echo will form during detection (G) after another interval of TE/2 in the presence of the second lobe of the read gradient (F). The phase-encoding gradient (D) is stepped in amplitude for each phase-encoding step and is necessary to locate tissue in the phase-encoding direction. A long TE causes the time interval between the lobes of the read gradient, (E) and (F), to also be long, which sensitizes this pulse sequence to motion in the read direction.

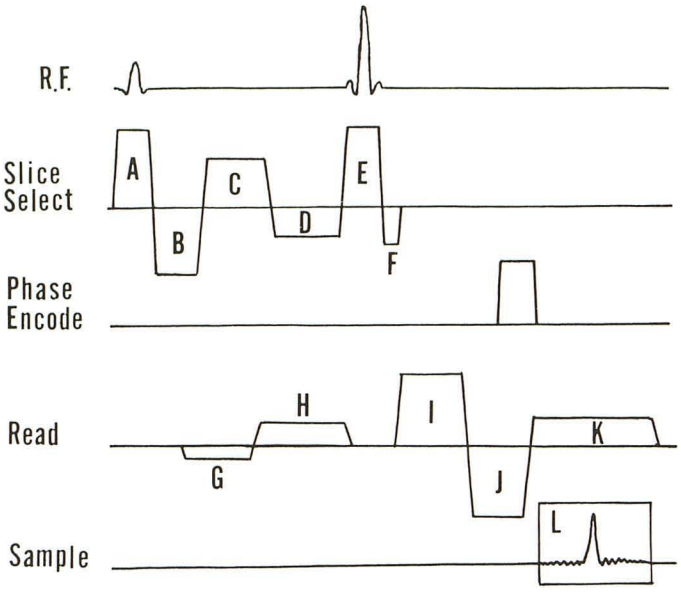

Fig. 2.—Pulse sequence diagram for the MAST gradient waveform modification of the standard T2-weighted (TE = 100) multislice spin-echo imaging sequence. The five channels plot the same parameters as the standard spin-echo sequence (Fig. 1). A mathematically derived waveform with many calculated lobes, (A–F) for the slice-select gradient and (G–K) for the read gradient, results in a pulse sequence with much less motion sensitivity than the standard spin-echo sequence. Transverse magnetization is forced to give an echo exactly in the middle of the signal detection period (L) regardless of whether tissue is stationary or in motion.

periodic motion is responsible for the corruption of signal from successive phase-encoding steps, an artifact resembling faint reproductions of the moving structure displaced in the phase-encoding direction ("ghosting") on the image results [1–4]. Signal intensity is also lost from structures that have motion, causing them to appear hypointense on the image (flow-void sign) [10]. If the phase-encoding steps can be synchronized with the motion (possible only if the motion is periodic), step to step differences can be minimized with resultant improvement in image quality [5, 6, 8].

Another approach to the problems caused by motion is to desensitize the pulse sequence to motion. This can be accomplished by employing a more complex waveform (MAST) for the read and slice-select gradients in an otherwise ordinary multislice spin-echo sequence (Fig. 2) [11, 12]. The duration, position in time, and amplitude of each lobe of these more complex gradient waveforms are calculated so that transverse magnetization forms an echo at time TE (in the center of the sampling window), regardless of whether tissue is stationary or undergoing physiologic motion of small amplitude. This complex modification of gradient waveform minimizes variations in signal intensity of moving tissue in successive phase-encoding steps. Moving tissues are depicted with a more uniform and brighter signal intensity when the echo is generated by this technique. Even though the pulse sequence is not synchronized to periodic displacements of the tissue of interest, phase-encoding artifacts can be reduced.

Movement of CSF may be depicted in a model as a combination of the fundamental orders of motion: flow, acceleration, and pulsatility. Flow is defined as the uniform change in position of a material with time; acceleration is the uniform change in flow with time; and pulsatility is the uniform change of acceleration with time. The more of these orders of motion that are included in the design of the model, the more accurately it represents the complex motion of CSF. The MAST gradient modification corrects for physiologic motions by removing the effects of flow, acceleration, and pulsatility in both the read and slice-select directions [11, 12].

Materials and Methods

An experiment involving a phantom was performed to illustrate the nature of the artifacts that arise in MR from pulsatile fluid flow and to assess the effectiveness of the gradient waveform modification in correcting these artifacts. A single loop of 6-mm internal diameter (3-mm wall thickness) flexible Tygon[†] plastic tubing was taped to a cardboard support and held rigidly in the sagittal plane. The tubing was filled with water doped with 0.25 mM manganese chloride to simulate the relaxation behavior of CSF. An eyedropper squeeze-bulb was attached to one end of the tubing and a reservoir to the other end. Displacements of 10 mm were periodically induced in the water column within the tubing by squeezing the rubber bulb located outside the bore of the magnet during the course of the imaging experiment. The water pulsations were made at roughly 1-sec intervals but were not synchronized to the data acquisition. The motion produced by this technique was that of the water column; the tubing itself did not move a significant amount since it was held firmly to the cardboard support. T2-weighted, 2000/100 (TR/TE), multislice spin-

[†] VWR Scientific Company, San Francisco, CA.

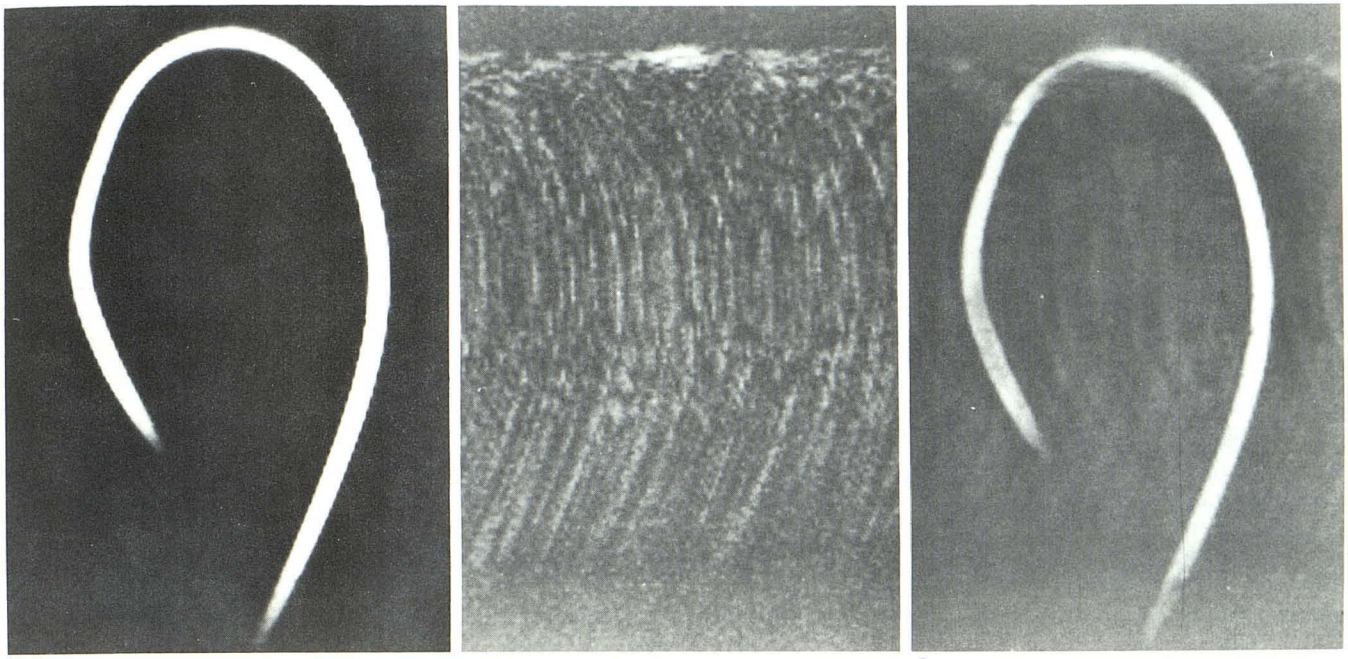

Fig. 3.—Central images of a sagittal multislice spin-echo sequence of a flow phantom (2500/100, horizontal phase encoding, 5-mm slice thickness). The phantom is a single loop of flexible 6-mm internal diameter plastic tubing filled with water.

***A*, Standard spin-echo sequence; water stationary in tubing. Water column in tubing appears bright, displaying expected uniform intensity.**

***B*, Standard spin-echo sequence; 10-mm periodic displacements of water column in tubing during the course of the imaging sequence, not synchronized with data collection. Water within tubing is not visualized. Extensive phase-encoding artifacts, appearing as faint reproductions of water and displaced in the phase-encoding direction, are present.**

***C*, MAST gradient waveform modified sequence; 10-mm periodic displacements as in *B*. Water within tubing is visualized with nearly uniform intensity and displays significantly less phase-encoding artifact.**

echo images were obtained during this experiment without and with the gradient waveform modification (see Fig. 3).

To evaluate the effect of the gradient waveform modification of the spin-echo pulse sequence on clinical images of the posterior cranial fossa and cervical spine, sagittal images of 40 patients were reviewed retrospectively. Twenty of these patients had undergone MR imaging of the head and/or cervical spine with the standard multislice spin-echo pulse sequence, while the other 20 had been examined with the gradient waveform modified pulse sequence.

All images were obtained on a 0.5-T instrument[‡] with a maximum gradient strength of 3 mT/m and a 20-msec sampling window. Imaging parameters were: single spin echo, eight contiguous slices, 5-mm slice thickness, 2000–2500/100/2 (TR range/TE/number of excitations), 30-cm field of view, 512 complex sampling points, and 256 phase-encoding increments (horizontal phase encoding). A butterfly-shaped cervical surface coil that partially wrapped around the patient's neck (30 cm long, 18 cm high) was used for detection, and the 55-cm body coil was used for excitation.

The only criterion for selection was that the patient had T2-weighted sagittal spin-echo images of the cervicocranial region obtained with a cervical surface coil with a 5-mm slice thickness. Consecutive cases were obtained from a time period immediately before (standard spin-echo sequence) and immediately after the MAST software upgrade (gradient modified sequence) was installed. No other modifications were made to the instrument during this time interval and there was no indication of a change in instrument performance as might be detected in the quality of the T1 images obtained on these same patients. Images displaying severe artifact from the patients' head or neck motion were excluded from this study. The group of patients examined with the standard spin-echo pulse sequence included 13 women ages 23 to 60 years (mean, 43 years) and seven men ages 37 to 73 years (mean, 56 years). The patient group examined with the gradient modified pulse sequence included 12 females ages 11 to 73 years (mean, 43 years) and eight men ages 32 to 55 years (mean, 40 years). All images were evaluated for the presence of phase-encoding artifacts as well as for variation in CSF signal intensity. The phase-encoding artifacts arising from CSF motion were classified as "absent," "mild" (small amount of artifact visible in adjacent tissue), or "moderate" (obliteration of details in adjacent tissue). The number of patients demonstrating loss of signal intensity for CSF was tabulated for each specific CSF space: upper cervical subarachnoid space, medullary cistern, pontine cistern, fourth ventricle, and cisterna magna.

Images of a volunteer (25-year-old woman) were obtained with both the standard and the gradient waveform modified pulse sequences sequentially without repositioning the subject between scans. This allowed direct comparison of images with the only variable being the modification to the gradient waveforms. The gain of the receiver was kept constant for the two scans, and the resulting magnitude images were digitally subtracted to yield images that depicted flow (see Fig. 6).

Results

The stationary water column within the plastic tube phantom was of uniform bright intensity on a standard T2-weighted sagittal image (Fig. 3A). When the identical imaging sequence

[‡] Vista 2055, Picker International, Inc., Highland Heights, OH.

was repeated on the phantom with periodic displacements of the water column during the entire sequence, severe phase-encoding artifacts resulted, and the water inside the tubing could not be clearly visualized (Fig. 3B). However, images of the periodically pulsating water-filled phantom obtained with the gradient waveform modification of the same pulse sequence clearly displayed the contents of the tubing with only minor inhomogeneity of signal and faint phase-encoding artifacts (Fig. 3C).

Sagittal T2-weighted spin-echo images of the head and cervical spine of the 40 patients in this study demonstrated similar findings. In the 20 individuals examined with the standard multislice T2-weighted spin-echo pulse sequence, the CSF-containing spaces in the posterior cranial fossa and in the cervical and upper thoracic spinal canal often appeared hypointense. The normal myelogramlike homogeneously bright appearance of CSF on T2-weighted MR images was not consistently observed (Fig. 4). This apparent flow-void sign [1] was seen less frequently in the lower thoracic spine. On many of these studies, it was not possible to accurately differentiate the spinal cord from the surrounding CSF due to the variations in the CSF intensity. Signal loss was identified in the upper cervical subarachnoid space in 17 (85%) of 20 individuals, in the medullary cistern in 18 (90%) of 20, in the pontine cistern in all 20 (100%), in the fourth ventricle in 15 (79%) of 19, and in the cisterna magna in 11 (55%) of 20. Phase-encoding ("ghosting") artifacts were often observed originating from the CSF spaces and were categorized as moderate in four (20%) of the 20 patients, mild in 13 (65%) of 20, and absent in the remaining three patients (15%).

Demonstrable improvement in the uniformity of signal intensity of the CSF was noted on the images of the 20 individuals examined with the gradient waveform modified pulse sequence (Fig. 5). On these images, CSF spaces appeared uniformly bright, and the flow-void sign was not observed in any patient in any of the visualized CSF spaces. However, there was no significant reduction in the frequency or severity of phase-encoding artifacts: four (20%) of 20 patients displayed moderate artifact, 12 (60%) of 20 mild artifact, and four (20%) of 20 no artifact.

Images obtained in the volunteer (Figs. 6A and 6B) provided a clear comparison of the standard and the gradient waveform

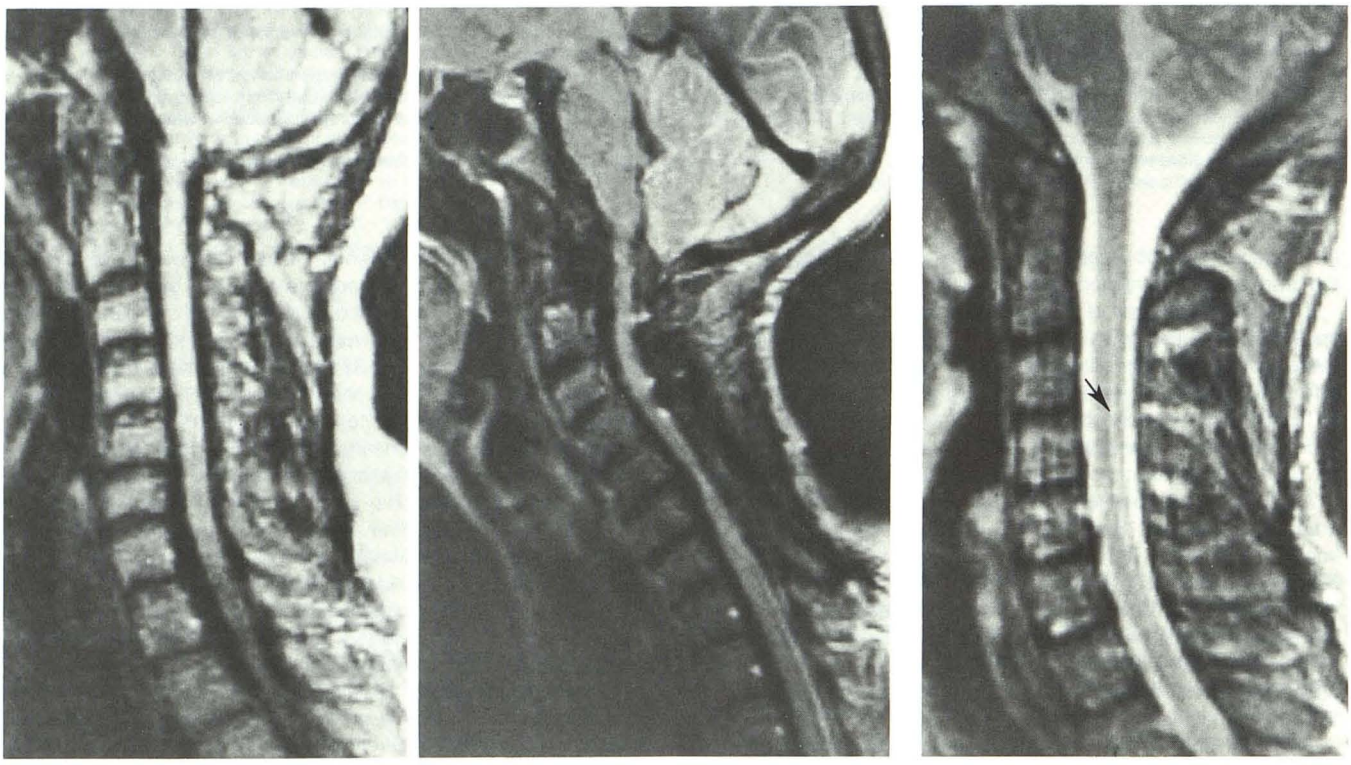

Fig. 4.—*A* and *B*, Approximately midsagittal MR images of cervicocranial region obtained with standard multislice spin-echo sequence (2000–2500/100, horizontal phase encoding, 5-mm slice thickness) in two patients. CSF appears nonuniform and variable. In both patients, the subarachnoid spaces of the cervical and upper thoracic spinal canal, the anteroinferior portion of the cisterna magna, the pontine cistern, and the anterior aspect of the fourth ventricle are hypointense, while the posterosuperior portion of the cisterna magna displays the expected homogeneous hyperintensity.

Fig. 5.—Image obtained with MAST gradient waveform modified multislice spin-echo sequence (2400/100, horizontal phase encoding, 5-mm slice thickness). CSF appears uniformly bright. Note bright linear band projecting within spinal cord (*arrow*) and parallel hypointense bands overlying the bony and soft tissues anterior and posterior to spinal canal; these represent phase-encoding ("ghosting") artifacts that are not eliminated by the MAST gradient waveform pulse sequence modification.

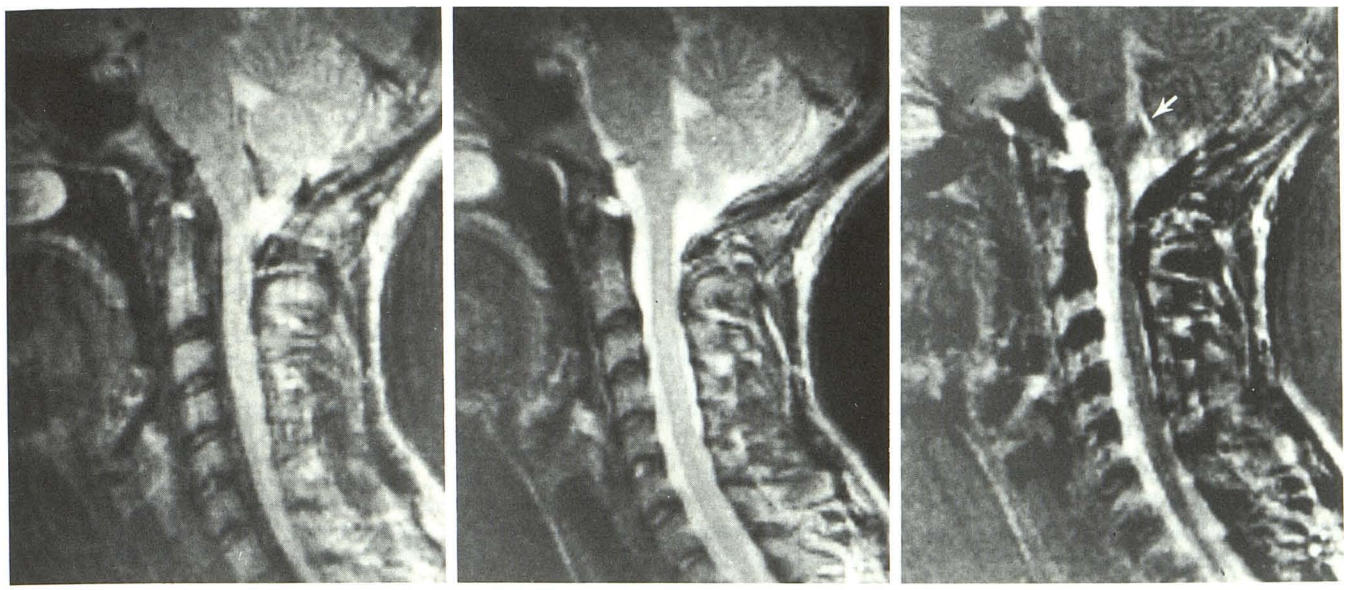

Fig. 6.—Image depicts motion of CSF generated by combining the data from standard and gradient waveform modified sequences obtained in same anatomic slice in a volunteer.

***A*, Flow-uncompensated image.**

***B*, Gradient waveform modified image of the identical slice, obtained sequentially without repositioning the subject.**

***C*, Flow image obtained by digitally subtracting image *A* from image *B*. Areas of high signal intensity depict relatively high flow in the anterior cervical subarachnoid space, medullary and pontine cisterns, and cisterna magna. Lesser signal intensity in fourth ventricle indicates relatively slower flow. A linear band of high intensity (*arrow*) depicts increased flow through foramen of Magendie.**

modified sequences in the same anatomic section. The image obtained with the gradient waveform modification exhibited uniformly bright CSF that was clearly demarcated from the adjacent brainstem, cerebellum, and cervical spinal cord (Fig. 6B). This was in striking contrast to the appearance of these structures on the standard T2-weighted image (Fig. 6A). Upon digitally subtracting the standard spin-echo image from the gradient waveform corrected image, all tissue except that undergoing motion was eliminated, leaving an image that depicted CSF flow (Fig. 6C).

Discussion

In this retrospective study, the MAST gradient waveform modification for multislice spin-echo pulse sequences resulted in consistent representation of CSF with uniformly bright signal intensity on T2-weighted sagittal images of the cervicocranial region. Areas of signal loss (hypointense on T2-weighted images) in the aqueduct, fourth ventricle, cisterns of the posterior fossa, and cervical subarachnoid spaces—a nearly universal occurrence on standard T2-weighted spin-echo images and a common cause of difficulty in the interpretation of such images—were eliminated. No gating or other modification in procedure was required beyond what was normally involved for a spin-echo sequence and the time required for a patient examination was identical to that of the normal T2-weighted spin-echo sequence.

On the basis of the appearance of the air background adjacent to tissue on patient images, the occurrence of phase-encoding ("ghosting") artifacts associated with swallowing and breathing appeared to be reduced with the modified pulse sequence. Although it was more difficult to evaluate the phase-encoding artifacts caused by CSF motion because of the presence of tissue adjacent to the CSF-containing spaces, there appeared to be no significant reduction in frequency or severity of such artifacts despite the evident reduction of such artifacts on the phantom study. Conversely, a pronounced hyperintense linear band extending down the central axis of the cervical spinal cord was frequently noted on the gradient waveform modified T2-weighted images (Fig. 5). This band does not represent the central canal of the spinal cord but is most likely a truncation artifact (Gibb phenomenon), a consequence of the Fourier transformation of data obtained within a limited sampling period. This artifact appears on MR images as faint periodic ripples of alternating intensity emanating from edges separating regions with large signal differences and is most pronounced in the phase-encoding direction. What makes the area around the spinal cord and canal unique in regard to this phenomenon is that it includes several parallel interfaces between hyperintense CSF and hypointense tissue on T2-weighted sagittal spin-echo images. The ripples generated by these parallel interfaces may add up (constructive interference) to generate a very bright band within the image of the spinal cord [13, 14].

Digital subtraction of normal spin-echo images from gradient waveform corrected images results in images that reveal the sites and to some extent the amplitudes of CSF motion (Fig. 6). The areas that are depicted as having the greatest flow in this volunteer are the upper cervical subarachnoid

space, medullary and pontine cisterns, and cisterna magna. Lesser flow is evident in the fourth ventricle, but Figure 6 suggests greater flow in the foramen of Magendie. This procedure could be improved if the two sequences were interlaced and executed simultaneously rather than obtained sequentially as was done in this study. Misregistration of pixels would be reduced, and a more complete cancellation of stationary tissue would result. The sensitivity to motion revealed in these flow images arises from displacements in the read and slice-select directions. This technique may find diagnostic utility in the quest for a more complete understanding of the dynamics of CSF motion.

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