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SIMA: Simultaneous Multislice Acquisition of MR Images by Hadamard-Encoded Excitation

S. P. Souza, J. Szumowski, C. L. Dumoulin, D. P. Plewes, and G. Glover

Abstract: We present a method of multislice magnetic resonance imaging that utilizes simultaneous binary-encoded excitation. Signals are acquired from all slices at once, and the images are separated in the reconstruction process. This simultaneous multislice acquisition method has been implemented for multislice spin-echo imaging, and the results are compared with those for a standard interleaved multislice method. Advantages include improved signal-to-noise ratios and flexible slice placement. Phantom and volunteer studies are presented and evaluated in comparison with competing methods. **Index Terms:** Magnetic resonance imaging, techniques—Magnetic resonance imaging, physics and instrumentation—Magnetic resonance imaging, acquisition.

Effective clinical use of magnetic resonance (MR) imaging frequently requires acquisition of data from multiple parallel image planes, since the exact location and extent of features of interest are seldom known ahead of time. To simplify interpretation of the images, this should be done with resolution, signal-to-noise, and spin parameter weighting identical to that of an equivalent single-slice image, and with minimal restrictions on user-selected imaging parameters.

To remove the inefficiency of sequentially acquiring many single slices while providing a long repetition time (TR) between excitations for return to equilibrium, conventional multislice techniques interleave adjacent slice excitations. This gives spins in one slice time to relax while other slices are being excited and acquired. In such a technique the number of slices that may be acquired is limited by TR. If more slices are needed, only partial interleaving is possible, negating the efficiency advantage. The alternative is to lengthen TR to accommodate the required number of slices, but this may detrimentally alter the weighting of the image (1). In addition, sequences with short TR and large NEX are sometimes used to reduce the effects of motion (2). Thus, in common clinical situations, conventional multislice imaging requires compromise of either throughput (efficiency), tissue discrimination, or spatial coverage.

An alternative multislice method that avoids the short-TR problem is the volume-excited three-dimensional Fourier transform (3DFT) technique (3). In this method all slices are excited uniformly and at once, and spatial encoding is performed on one readout axis and two phase-encoding axes. Repetition time is not constrained by the number of slices. As in any volume excitation method, acquisition of signal from a large fraction of the noise-generating volume yields an improved signal-to-noise ratio, provided that the receiver has sufficient dynamic range. Slices are contiguous, and a large number of slices may be obtained in a reasonable scan time. These properties, which may make 3DFT the method of choice in some situations, can be limitations in others. First, contiguous slices are not always desirable: slice-to-slice variations in anatomy may not be significant, and no possibility exists for unequal interslice gaps such as might be useful in studies of symmetrical structures (e.g., temporomandibular joint). Second and more fundamental, it is well known that spectral estimation via the discrete Fourier transform is subject to truncation (Gibbs) artifacts (4). These artifacts are due to sidelobes in the point-spread function produced by the intrinsic windowing effect of a finite number of

From the General Electric Consulting Services Corporation, Albany (S. P. Souza), University of Rochester Medical School, Rochester (J. Szumouski and D. P. Plewes), GE Corporate Research and Development Center, Schenectady (S. P. Souza and C. L. Dumoulin), NY; and GE Medical Systems, Milwaukee, WI (G. Glover). Address correspondence and reprint requests to Dr. S. P. Souza at GE Corporate Research and Development Center, Post Office Box 8, Schenectady, NY 12301, U.S.A.

samples and may be somewhat ameliorated by low-pass filtering (apodization) although at some cost in spatial resolution. In 3DFT imaging this effect manifests itself as a noticeable leakage of spectral content between slices in areas of large slice-to-slice signal variation. It is most bothersome in the slice direction, in which the number of samples is generally small (≤64) and there is no intrinsic filtering by relaxation effects.

We elaborate here on a technique for simultaneous multislice acquisition (SIMA) as introduced in references (5,6) and subsequently discussed by Muller (7) and present the results of phantom and volunteer studies. Similar results were obtained in earlier unpublished work by one of the authors. Like 3DFT, SIMA is a volume excitation method yielding improved signal-to-noise ratios for a given number of slices and scan time. Encoding in the slice dimension is accomplished by modulating the phase of each slice excitation frequency in a binary pattern such as that given by the Hadamard matrix of dimension equal to the number of slices, in a manner similar to that suggested by Maudsley (8). All acquired data are used to reconstruct each slice. This technique is particularly well suited to moderate numbers of slices (4–16, depending on the flip angles and anatomical regions involved) and may have an optimum range of applicability intermediate between conventional multislice and 3DFT methods. In addition, SIMA may be useful in adding multiple slice capability to methods such as chemical shift (9,10) and inner volume (11) imaging.

THEORY

Customarily in MR imaging the excited slice is defined by a radiofrequency (RF) pulse applied to the sample simultaneously with a linear magnetic field gradient that is constant during the RF pulse. The band of frequencies transmitted is just the Fourier conjugate of the RF pulse envelope. Thus, for a square excitation profile the RF envelope should ideally be a sinc function. In practice, of course, it is not possible to follow the sinc to its infinite extremes, so the sinc is truncated at some small number of zero crossings and convolved with a low-pass (e.g., Hanning) filter to minimize ringing of the slice profile. Furthermore, the resonance process is itself nonlinear, so even a perfect sinc waveform extended to infinity will result in some excitation outside the desired slice. If it is desired to offset the slice from the gradient isocenter, the transmitted frequency must be offset by

$$\omega_0 = 2\pi\gamma G_{ss} z$$

where γ is the gyromagnetic ratio for the imaged species, G_{ss} is the applied slice selection gradient strength, and z is the desired physical offset. This is equivalent to multiplying the RF envelope by a complex sinusoid, so that the overall modulation of the on-resonance carrier is given by

$$\mathrm{RF}(t) = \mathrm{sinc}(at)e^{i\omega_0 t}$$

where a is a constant.

This complex excitation contains real (cosine) and imaginary (sine) components, which are typically realized as I and Q channels in the transmitter. Transmitting the cosine term alone (double sideband modulation) will excite spins on both sides of the isocenter with the same phase (Fig. 1a), and transmitting the sine term alone will excite both sides with opposite phase (Fig. 1b). In traditional offset slice selection both terms are transmitted simultaneously (single sideband modulation), yielding excitation in only one offset slice.

Suppose, however, that we excite spins and acquire data first with the cosine term (I channel) only, and then with the sine term (Q channel) only. Each data set contains information from slices on both sides of the isocenter, modulated in phase by the symmetric and antisymmetric excitation patterns shown above. In this simple case, adding the data from these two excitations gives a slice centered at $+\omega_0$, and subtracting gives a slice at $-\omega_0$. This excitation pattern can be described by the matrix

$$\begin{array}{cc} \omega_0 & -\omega_0 \end{array}$$
$$\begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} \begin{array}{l} \text{cos modulation} \\ \text{sin modulation} \end{array}$$

which is just the Hadamard matrix (transform) of order 2 (H^2). The Hadamard transform [see, for ex-

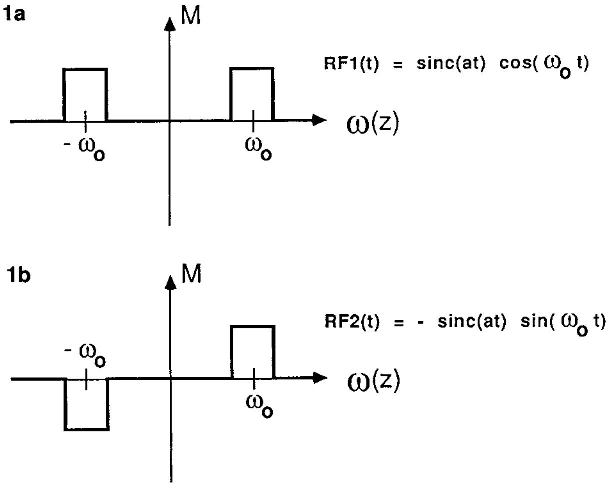

FIG. 1. Excitation patterns for two-slice Hadamard-encoded imaging. Transmitting **(a)** just the cosine term excites slices in phase, and transmitting the sine term **(b)** produces out-of-phase excitation.

ample, ref. (12)] is similar to the discrete Fourier transform but expands in terms of square waves instead of sines and cosines and thus is characterized by a square point-spread function. Furthermore, the Hadamard matrix is its own inverse. The Hadamard transform is defined for any positive integral order N, but as with the Fourier transform is computationally most efficient when N is a power of 2. In this case

$$H^N = \begin{bmatrix} H^{\frac{N}{2}} & H^{\frac{N}{2}} \\ H^{\frac{N}{2}} & -H^{\frac{N}{2}} \end{bmatrix}$$

If we represent the separate signals from slice locations S1 and S2 in a column vector

$$S = \begin{bmatrix} \mathrm{S1} \\ \mathrm{S2} \end{bmatrix} \begin{matrix} \text{corresponding to } \omega_0 \\ \text{corresponding to } -\omega_0 \end{matrix}$$

then the acquired data D can be written as

$$D = H^2\, S = \begin{bmatrix} \mathrm{S1} + \mathrm{S2} \\ \mathrm{S1} - \mathrm{S2} \end{bmatrix}$$

To extract the slices we take

$$S = (H^2)^{-1}\, D = H^2\, D$$

Thus, excitation and decoding are described by the same matrix.

To extend these principles to a larger number of slices, we must excite using multiple offset frequencies. Consider the case of $N = 4$ equally spaced slices, for which the excitation function is

$$\mathrm{RF}(t) = \mathrm{sinc}(at)e^{i\omega_0 t}e^{i2\omega_0 t}$$

This results in four excitation terms and the corresponding excitation matrix

$$\begin{matrix} 3\omega_0 & \omega_0 & -\omega_0 & -3\omega_0 & \\ \end{matrix}$$
$$\begin{bmatrix} 1 & 1 & 1 & 1 \\ 1 & -1 & -1 & 1 \\ -1 & -1 & 1 & 1 \\ -1 & 1 & -1 & 1 \end{bmatrix} \begin{matrix} \cos(\omega_0 t)\cos(2\omega_0 t) \\ -\sin(\omega_0 t)\sin(2\omega_o t) \\ i\cos(\omega_0 t)\sin(2\omega_0 t) \\ i\sin(\omega_0 t)\cos(2\omega_0 t) \end{matrix}$$

which can be rearranged via elementary row and column operations to give the equivalent matrix H^4.

Although this process is closely related to the Fourier transform, a clear distinction must be made between this technique and existing 3DFT methods. First, in SIMA, spatial encoding in the third dimension is accomplished by phase modulating the excitation envelope in discrete steps of π, rather than by adding a second phase-encoding magnetic field gradient pulse. This has the advantage of being exact for any value of N, rather than approximate as in any discrete Fourier transform. Second, it is not necessary that the slices be equidistant, that there be no slice at zero offset, or even that the distribution of slices be symmetric with respect to the gradient center. If, for example, we used modulation frequencies of ω_0 and $3\omega_0$ in the $N = 4$ case, we would have spacing between the central two slices twice that between either of the end pairs. Essentially arbitrary slice placement is possible in principle by appropriate choice of modulation patterns.

RESULTS

The theoretical development above was verified in phantom and volunteer studies on a 1.5 T imaging system (GE Medical Systems, Milwaukee, WI, U.S.A.). Radiofrequency waveform generation and Hadamard reconstruction software for 2, 4, 8, or 16 equally spaced slices were created, along with a SIMA-modified version of a spin-echo pulse sequence utilizing a nonselective time-reversal 180° pulse. For N slices the appropriate subset of N excitations is performed, first $N/2$ on the I transmitter channel only, then $N/2$ on the Q channel only. Reconstruction then consists of applying appropriate sign changes to the raw data from each excitation, summing all the excitations, and performing a conventional two-dimensional Fourier transform reconstruction with K-space filtering.

Figure 2 presents a set of 16 contiguous slices of a water-filled spherical phantom, obtained with a SIMA-modified spin-echo pulse sequence. A body coil was used for both transmitting and receiving. The slice thickness was 1.0 cm, the TR 400 ms, and the echo time 25 ms. There is no evidence of interslice interference arising from either the excitation/reconstruction method or nonideal RF excitation.

Figure 3 contains two sets of four sagittal slices of the head of a healthy volunteer, obtained with SIMA and 3DFT pulse sequences, respectively. Contrast and brightness have been set as necessary for clinical viewing. The 3D Fourier images show considerable interslice interference in the air-filled cavities due to signals arising from adjacent areas. This artifact results from the limited number of slices sampled in the Fourier method. In contrast, the SIMA images are free of any interslice sampling artifacts.

DISCUSSION

Volume excitation of any kind has both benefits and drawbacks. The primary benefit is improved signal-to-noise ratio for a given scan time, under certain conditions. Three-dimensional Fourier methods give many contiguous slices without re-

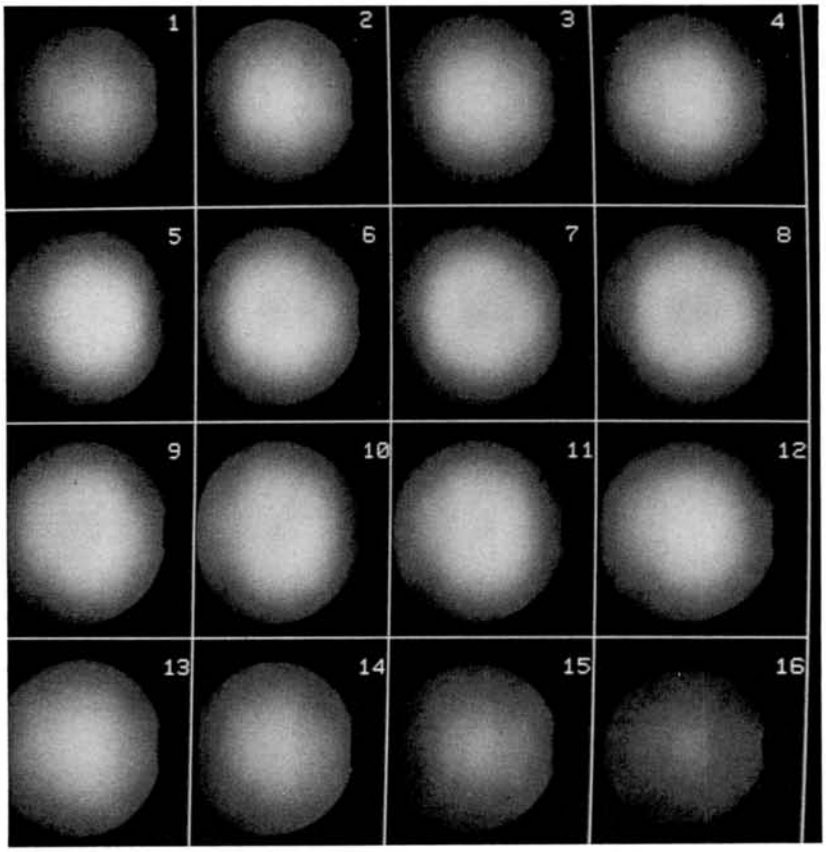

FIG. 2. Sixteen contiguous 1 cm slices of a water-filled spherical phantom, obtained with simultaneous multislice acquisition. A body coil was used for transmitting and receiving. No interslice interference is seen.

sorting to extremely long TR: this may or may not be an advantage. An important drawback of volume excitation is that, since data from all slices are acquired for each excitation, phase errors arising from processes in one slice will result in artifacts appearing in all slices. In particular this applies to motion artifacts including those due to flow. The sensitivity of 3DFT to such phase errors is also present in SIMA. Additionally, volume excitation often places stringent requirements on the performance of the imaging hardware. For example, since signals from a large volume are detected simultaneously, more dynamic range is required in the receiver and in the digitized data.

The SIMA technique presented here has its own unique strengths and limitations. Primarily, it is a general method to give multislice capability to almost any single slice MR procedure. Slice thickness and placement are flexible relative to 3D Fourier methods, and reconstruction of a finite number of slices is in principle exact. A signal-to-noise advantage similar to that for 3DFT methods is realized for a moderate number of slices. To achieve this, SIMA places some demands on system performance, particularly for the transmitter.

To perform properly, SIMA requires that the true excitation (magnetization) induced in the sample be close to that presented in the theory section. This in turn requires that the RF transmitter have high linearity. Harmonic distortion in the modulation envelope will appear as spurious excitation frequencies, introducing signals from unwanted slices. This is exacerbated by the fact that the peak RF power required is proportional to the number of slices excited. Similarly, the 90° phase difference and gain balance between I and Q transmitter channels must be closely maintained. Pulse-to-pulse stability is similarly important. If all these requirements are not met, the resulting images will display slice-to-slice leakage and may be contaminated by signals from material outside any of the desired slices.

The results described above indicate that SIMA can be accommodated on standard imaging systems without special adjustment. The continuous and discrete phantom results indicate that RF linearity, I/Q balance, and receiver dynamic range are sufficient to keep ghosts and artifacts to an acceptable level. Up to 16 slices have been acquired on phantoms, and initial experiments on volunteers indicate that at least four slices can be obtained for head and body imaging. Specific RF absorption and the high peak power required may be a problem when N is large. Both may be alleviated by pulse-shape power reduction techniques (13) and by the application of SIMA to limited flip angle, fast scan pulse sequences.

3a,b

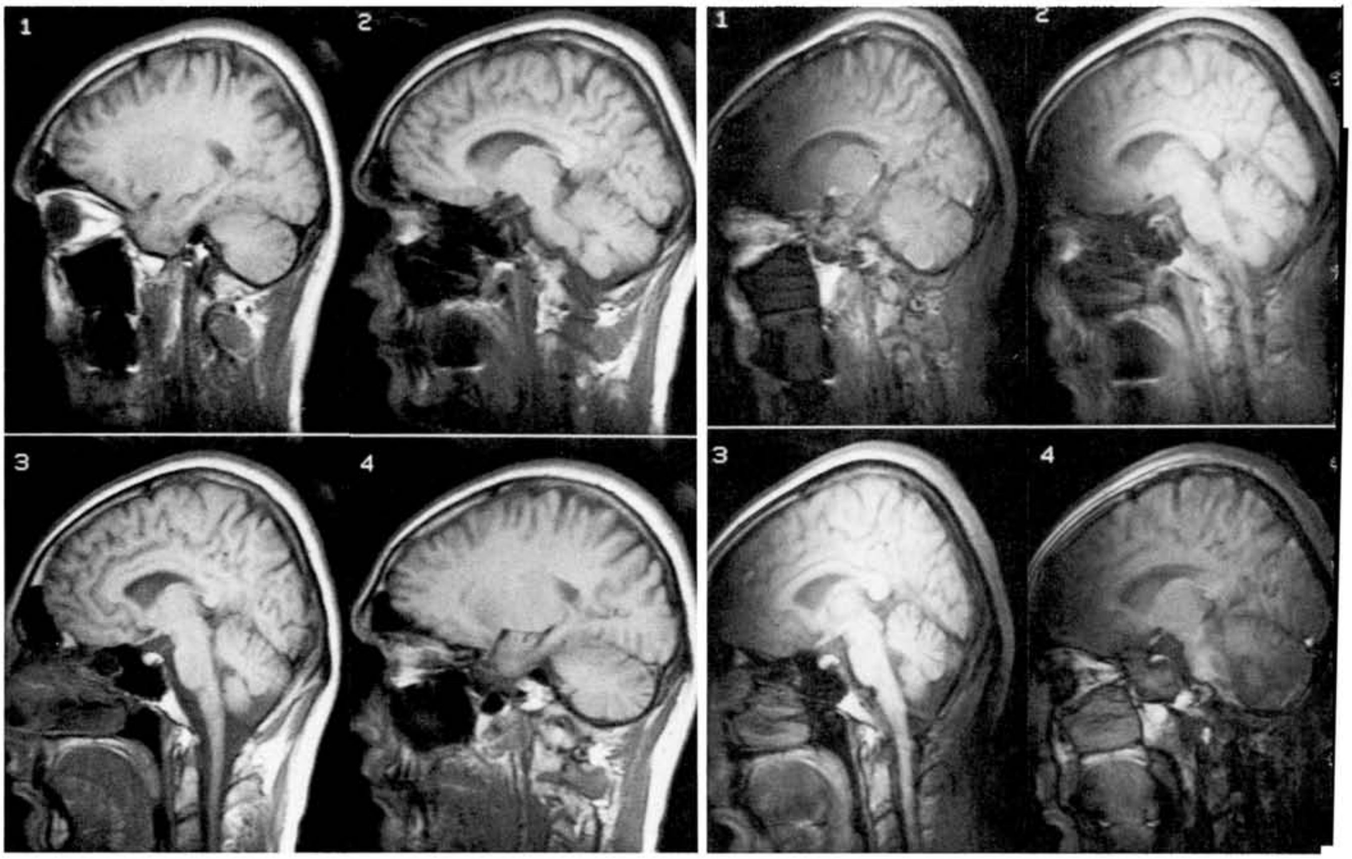

FIG. 3. Four sagittal slices of the head of a healthy volunteer obtained with **(a)** simultaneous multislice acquisition (SIMA) and **(b)** three-dimensional Fourier transform (3DFT) pulse sequences. The 3DFT images exhibit interslice interference, particularly in the sinuses, which contain little or no material and should be dark. The SIMA images are free of such artifacts.

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