A Solution to the Dynamic Range Problem in MRI Using a Parallel Image Acquisition

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ABSTRACT: A parallel image acquisition system has been developed to solve the dynamic range problem in MRI. The parallel receiver system was tested using 3D large matrix spin echo and gradient echo images. The dynamic range of the receiver system was extended to about 80 dB, which was sufficient for a large-matrix (256 × 256 × 512) 3D gradient echo image acquisition at a voxel resolution of 60 μm³. Our method can be a simple and efficient technique for solving the dynamic range problem in MRI. © 2006 Wiley Periodicals, Inc. Concepts Magn Reson Part B (Magn Reson Engineering) 29B: 161–167, 2006

KEY WORDS: MRI; dynamic range; k-space; receiver; analog-to-digital converter

INTRODUCTION

The dynamic range of an MRI receiving system is often insufficient because a typical MRI signal is only large around the center of the k-space. This problem becomes serious when a 3D image of a large object needs to be acquired in a high magnetic field, and several solutions to this problem have been proposed (1–5).

One straightforward solution is to use an intermediate frequency and direct digital sampling at a rate of several to several tens of megahertz (5). However, this approach requires high-speed data sampling and signal processing hardware. Another practical solution is to use a dual (or multiple) scan technique with low-gain and high-gain data acquisitions for the low and high spatial frequency components, respectively (3, 4). However, this approach requires extra scan time or a complicated scan protocol. In this work, we have developed a possible solution to this problem using a parallel image acquisition.

MATERIALS AND METHODS

Experimental Setup

Our study was performed using a homebuilt MRI system developed in our laboratory. The MRI system consisted of a vertical wide-bore 9.4 T superconducting magnet (JASTEC, Kobe, Japan), a homebuilt gradient probe, and an MRI console controlled using a personal computer (6–9).

Figure 1 shows a schematic diagram of the parallel MR data receiving system developed in this study. A 400-MHz NMR signal detected by an RF coil was amplified using a preamplifier and divided into two signals using a power divider (PD). One signal was further amplified (typically by +37 dB) using a second preamplifier, and directed into a compact MRI receiver (TR1, DTRX4, MRTechnology, Tsukuba, Japan). Another signal was directed into another receiver (TR2). Both quadrature-detected MRI signals were simultaneously digitized using two 14-bit two-channel ADC (analog-to-digital converter) boards (DATEL, PC-414G3, USA) installed in the computer running under the Microsoft Windows 98 operating system.
The MRI receiver was originally designed for wideband use (3.5–100 MHz), and was modified for use in our system. Thus, the receiver was not optimized for 400 MHz operation.

Figure 1  The parallel MR image data receiving system.

Figure 2 shows a parallel image data-acquisition program window. The MR signals that are simultaneously digitized with the high- and low-gain channels are displayed in the upper and lower data

Figure 2  The parallel MR data acquisition program running under the Windows 98 operating system. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]
windows, respectively. As shown in the upper window, the MR signal exceeds the ADC maximum range around the central region of the data acquisition window.

**MR Signal Synthesis**

Figure 3 shows the protocol of the MR signal synthesis from the two-channel MR data sets.

A minimum-volume ellipsoid outside of which no signal overflow was present in the high-gain channel was used for the signal synthesis. The minimum volume ellipsoid was determined as follows: at first, magnitude of the NMR signal was calculated along the positive readout direction for the central phase encoding line passing the k-space origin, and one end of the axis of the ellipsoid was determined as the first point that exceeded a threshold. The other end of the axis was determined as the first point that exceeded the threshold calculated along the opposite (negative) readout direction for the central phase encoding line. The center of the ellipsoid was determined as the central point of the first axis. The second axis was determined as a line segment passing through the center of the ellipsoid and perpendicular to the first axis. The algorithm to determine the location of the second axis was the same as that used for the first axis. The third axis was determined in the same way as that for the second axis. Although the procedure to determine the minimum volume ellipsoid may not be perfect for all the objects to be imaged, this method practically caused no problem. The gain and phase differences between two channel data sets were calculated between two ellipsoids: the inner ellipsoid described above and an outer ellipsoid slightly expanded (each axis was elongated by 20 data points).

Figure 4 shows histograms of the gain (A) and the phase differences (B) between the two receiver channels calculated for the region described in the text.

![Figure 4](image)

**Figure 4** Histograms of the gain (A) and the phase differences (B) between the two receiver channels calculated for the region described in the text.
locations in the histograms, because the peak locations do not depend on the size of the outer ellipsoid.

The MR signal data set was synthesized by combining the high-gain data from outside the inner ellipsoid and the gain and phase adjusted low-gain data from inside the inner ellipsoid. A 3D MR image was then reconstructed using a 3D FFT of the synthesized MR signal.

**Samples**

A water phantom and chemically-fixed human embryos were used to demonstrate the effectiveness of our method. The water phantom was made of a glass NMR sample tube (outer diameter = 12 mm, inner diameter = 11 mm) with a nylon sphere (diameter = 6.4 mm) placed at the bottom of the test tube. The test tube was filled with aqueous CuSO₄ solution. The human embryos were selected from the Kyoto Collection of Human Embryos (10–12). These embryos were chemically fixed with Bouin’s fluid about 40 years ago and stored in a 10% formalin water solution.

**Imaging Experiments**

The water phantom and human embryos were imaged using 3D spin echo (TR = 100 ms, TE = 12 ms) and

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Figure 5  Vertical cross-sections of a water phantom selected from the 3D image datasets acquired with the gradient echo sequence (A–E) and spin echo sequence (F). The gain differences were 0, 10, 20, 30, 40, and 30 dB for Figs. 5(A–F), respectively.
3D gradient echo (TR = 100 ms, TE = 5 ms, flip angle = 90°) sequences. The image matrix and voxel size were fixed to 256 × 256 × 512 and 60 μm³, respectively. The gain difference between the two receiver channels was varied between 0 and 40 dB for the water phantom, and was fixed at about 30 dB for human embryo samples.

RESULTS

Figure 5 shows vertical cross-sections of the water phantom selected from the 3D image data sets. Figures 5(A–E) show gradient echo images acquired with a 0 to 40 dB gain difference. Figure 5(F) shows a spin echo image acquired with 30 dB gain difference. The signal-to-noise ratios (SNR) of the water in Figs. 5(A–F) were 2.9, 5.9, 13, 20, 22, and 13, respectively.

Figure 6 shows a plot of the log of the average signal magnitude versus the log of the k-space radius for the synthesized MR signal of the water phantom. Figure 6(A) shows the power distribution of the spin echo images. The curve for a gain difference of 0 dB shows that the dynamic range of the receiver system was about 54 dB. The data in Fig. 6(A) show that the dynamic range of the receiver system increased with increasing gain difference up about 90 dB for the spin echo data. However, Fig. 6(A) shows that the spin echo data require a much larger dynamic range.

Figure 6(B) shows the power distribution of the gradient echo images. The dynamic range increased with increasing gain difference for the gradient echo data, as shown in the spin echo data in Fig. 6(A). This shows that an 80 dB dynamic range is sufficient for a gradient echo image with a 60 μm³ voxel resolution.

Figure 7 shows midsagittal cross-sections of a Carnegie Stage 22 human embryo (about 50 days after conception) acquired with the 3D gradient echo and spin echo sequences. These cross-sections were cut from 512 × 512 × 1024 voxel images obtained from zero-filled Fourier interpolation of the original 256 × 256 × 512 voxel image data.

The left-hand column shows images reconstructed from the data set acquired using the low-gain channel of the gradient echo signal. The middle column shows images reconstructed from the synthesized gradient echo signal with a 30 dB gain difference. The right-hand column shows images reconstructed using the synthesized spin echo signal with a 37 dB gain difference. The upper and lower figures show views of the entire embryo view and an enlarged view (×2.8) around the tong. The enlarged view clearly shows that the images reconstructed from the synthesized MR signal exhibit fine anatomical structures.

DISCUSSION

As shown in Fig. 6, the dynamic range of the single receiver was about 54 dB. This dynamic range was not due to the resolution of the ADC (14 bits) but the receiver itself, which was originally designed for wideband use (3.5–100 MHz) and therefore not optimized for 400 MHz operation. However, by using parallel image acquisition, the dynamic range was extended to about 90 dB, as shown in Fig. 6(A). Although there is no published data on the dynamic range of 400 MHz NMR receivers, this value is...
considered to be close to that of the state-of-the-art NMR receivers because about 90 dB dynamic range is reported for a commercial (Varian) receiver at 300 MHz in an earlier article (4). In our system, the dynamic range can be further extended by increasing the gain difference. The dynamic range achieved in our system was sufficient for a large matrix (256 × 256 × 512) 3D gradient echo image acquisition at a spatial resolution of 60 μm³.

Our method is conceptually identical to a dual or multiple scan approach (3, 4). However, it requires no additional scan and thus has the advantage in scan time and MR signal reproducibility and continuity in k-space. Current MRI systems usually have multiple receiver channels for use in parallel imaging. Therefore, our method can be used in many MRI systems with minimum system modification.

CONCLUSIONS

We have developed a parallel image acquisition system that extends the dynamic range of an MR receiver system. We have demonstrated the potential of the receiver system by imaging a water phantom and chemically fixed human embryos. Our approach can be a simple and efficient method to solve the dynamic range problem in MRI.

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REFERENCES