회전 경사자계와 사상 재구성을 이용한 무소음 자기 공명 영상법

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Silent Magnetic Resonance Imaging Using Rotating and Projection Reconstruction

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ABSTRACT

A new approach to silent MR imaging using a rotating DC gradient has been explored and experimentally studied. As is known, acoustic or sound noise has been one of the major problems in handling patients, mainly due to the fast gradient pulsings in interaction with the main magnetic field. The sound noise is also proportionally louder as the magnetic field strength becomes larger. In this article, we have described a new imaging technique using a mechanically rotating DC gradient coil as an approach toward silent MR imaging, i.e., a mechanically rotated DC gradient effectively replaces both the phase encoding as well as the readout gradient pulsings and data obtained in this manner provides a set of projection data which later can be used for the projection reconstruction or with some interpolation techniques one can also perform conventional 2-D FFT (Fast Fourier Transform) image reconstruction. We found, with this new technique, that the sound noise intensity compared with the conventional imaging technique, such as spin echo sequence, is reduced down to -20.7 dB or about 117.5 times. The experimental pulse sequence and its principle are described and images obtained by the new silent MR imaging technique are reported.

INTRODUCTION

An important obstacle in using MRI techniques has been the acoustical or sound noise generated in MRI scanners. Most of this noise is caused by mechanical forces generated by current pulsings in the gradient coils in interaction with the magnetic field in MRI scanners. When current pulses are applied to the gradient coils which are located within the strong magnetic field, Lorentz forces are induced resulting in acoustical vibration in the coils. The resulting rapid mechanical movement generates high-frequency sound noise in conjunction with the materials surrounding or supporting the coils. The sound noise also becomes proportionately louder as the magnetic field strength of the MRI scanner increases (1,2). Gradient pulsing noise has prevented the use of MRI in the diagnosis and treatment of many classes of patients, including the elderly, severely sick, small children, persons with psychiatric disorders, and those suffering from tinnitus, especially where the scanning time is extended. Besides physical discomfort, sound noise has many negative effects, such as the disturbances in fMRI (3,4) due to the unwanted stimulation of auditory sensory portions of the brain in the temporal lobes.

In the past, the noise problems associated with MRI scanning have been simply overlooked because of the powerful imaging capabilities of MRI. Early efforts include the technique of reducing vibrations by tightly binding the coils onto a massive or heavy support and by use of anti-phase noise cancellation technique (5). However, these techniques resulted in largely disappointing results. More recently, Mansfield et al., (6) and Botwell and Mansfield (7) have discussed a compensated or balanced type of gradient coil in an attempt to reduce Lorentz forces generated by the coils. These techniques, though partly successful in reducing induced sound noise intensity, did *not sufficiently* eliminated the objectionable noise. Therefore, there is a need for a simple and efficient method of reducing the acoustical noise inherent with pulsed gradient techniques used in magnetic resonance imaging.

PRINCIPLES OF TECHNIQUES

Among the three gradient coils used in MRI scanners, namely the X, Y, and Z gradient coils, the most dominant and the largest sound-producing gradient coils are the X and Y coils, which are the readout and phase encoding gradient coil, respectively. Data obtained with this technique provide a set of projection data to be used for projection reconstruction (8).

Let us assume that a slice is selected at $z = z_O$ using conventional spin echo imaging. In this case, the echo signal obtained can be noted as

$$S(t) \approx \sum_{m=1}^{m=M} \int_{-\infty}^{\infty} M_0(x,y;z_0)\, e^{i\gamma y m \Delta G_y T_P}\, e^{i\gamma x G_x t}\, dx = S(k_x, k_y) \qquad (1)$$

where ΔG_y, T_P, G_x, m, M, M_0, t, g, k_X, k_y are the usual notations used in MRI.

An equivalent form applicable in the projection-data-acquisition mode can be expressed as,

$$S(t) \approx \sum_{m=1}^{m=M_\phi} m\Delta\phi \int_{-\infty}^{\infty} M_0(x,y;z_0)\, e^{i\gamma x' G_{x'} n\Delta t}\, dx' \approx S(k,\phi) \qquad (2)$$

where x' is a rotated coordinate, M_ϕ and $\Delta\phi$ are given and related as, $\Delta\phi = \pi / M_\phi$.

$G_{x'}(t)$ is a combined amplitude of X and Y gradients and is a function of angle ϕ or at time t.

To implement the new silent MR imaging, $k_{x'}$ is replaced by a constant k (or G) throught data collection period, i.e.,

$$S(k_r, \phi) = \sum_{m=1}^{m=M_\phi} m\phi \int_{-\infty}^{\infty} M_0(x,y;z_0)\, e^{ix'k}\, dx'. \qquad (3)$$

EPERIMENTAL RESULTS

The pulse sequence employed for the new silent-MRI-spin-echo imaging is shown in Fig. 1. Note that the sequence has no phase encoding and no readout gradient pulsing. This pulse sequence is similar to the well-known projection-reconstruction technique except that the readout gradient is now replaced by a DC, rather than pulsing.

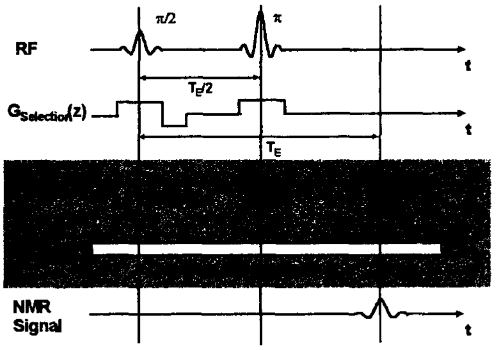

Fig. 1. Pulse sequence of the new silent spin-echo MRI. Note the absence of the phase encoding gradient as well as the newly added DC gradient applied in the readout gradient.

Experimental results obtained with a 2.0 Tesla whole-body MRI system indicates that virtually all the gradient-pulsing sound noise are effectively eliminated by using this method. A phantom image and its corresponding sinogram obtained from a 2.0 Tesla whole-body scanner with a hand-rotated gradient coil set are shown in Fig. 2(a) and (b).

Image
(a)

Sinogram
(b)

Fig. 2. A typical image and its corresponding sinogram obtained from the silent MRI technique shown in Fig. 1 using a 2.0 Tesla whole-body scanner with a hand-rotating gradient system.

In Fig.3 (a) and (b), a kiwi image and its corresponding sinogram obtained by the new silent-MR imaging technique are shown to demonstrate the high quality image obtainable with the new method.

(a) (b)

Fig. 3. A kiwi image and its corresponding sinogram obtained by experiment.

Specifically, measured sound noise signals with the present technique are approximately -20.7dB or 117.5 times less than noise obtained with the conventional spin-echo imaging sequence. As mentioned earlier, the accompanying sound noise measured from the new silent MRI technique is found to be more than two orders of magnitude smaller than sound noise of the conventional spin echo technique(See Fig. 4).

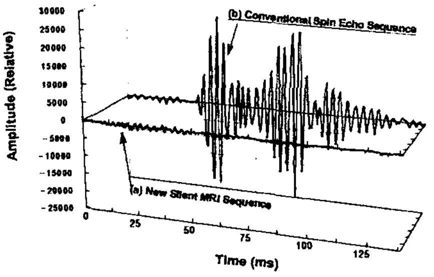

Fig. 4. Sound noise amplitudes measured and recorded from the new silent-MRI technique and the conventional spin-echo sequence, respectively. (a) Sound noise amplitude measured from the new silent MRI technique. (b) Sound noise amplitude measured from the conventional spin echo technique.

DISCUSSION AND CONCLUSIONS

Present method has two limitations, namely limitation in selecting the slice only in Z-direction due to the fact that of the X-gradient must be rotated around z axis and the other is the tilting effect of the selected slice at each angle while gradient is rotating thereby results in loss of selected slice volume. Because of a small DC gradient applied continuously during the application of selection gradient e.g. X-gradient in this case each selected slice is tilted an amount $\theta = tan^{-1}\frac{G_X}{G_Z}$ which amount to be a few degree from the Z-axis or the rotating axis. When this tilted slice is rotated 180°, the result will be a nutation of the selected slices or sum of the nutated slices which will result in a slice which is not fully overlapped thus results in poor resolution at the periphery. Although, in actual imaging situation, it is not a significant resolution degradation as long as the tilting angle is not that large a complete solution was desired, however. It can be solved by a synchronized motion of the patient bed so that the tilted angle q is compensated at each projection data collection.

In conclusion, we report some preliminary results obtained with the new silent MRI method, its apparatus, and its principles. We find that this method reduces sound noise emanating from MRI systems due to gradient pulsings. The proposed silent-MRI method (and apparatus) reduces the sound noise thereby fulfills the requirements of a sound-noise-free clinical-diagnostic environment, particularly with respect to pediatric patients, psychiatric patients, and functional-MRI studies. Another important application of the acoustic-noise-free MR technique is in the areas of NMR microscopy where the sample vibration caused by acoustic noise(in NMR microscopy acoustic noise is much larger due to the large gradient field employed in microscopic imaging) can result in resolution degradation.

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