Sliding Phase Encoding in Table-Motion Direction for Fast Continuously Moving Table Imaging

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Introduction

In MRI, FOV is restricted. In order to take an image with a large FOV, various techniques have been developed. Currently, the fastest imaging method for extended FOV is continuously-moving-table (CMT) imaging[1]. However, this method is limited in that the directions of phase-encoding and table-motion can not be parallel. Therefore, when homogeneous space is compressed in the table-motion direction, the read-out direction is forcibly set in the compressed direction. Then data acquired in one echo is reduced, and more echoes are needed for image reconstruction, resulting in longer imaging time. Furthermore, high resolution in the orthogonal direction of the table-motion results in larger numbers of phase encoding, that is, a longer imaging time. To prevent this increase in imaging time, we propose a new encoding method called sliding phase encoding (SPE) and reconstruction methods that replace conventional phase encoding and Fourier transform reconstruction. SPE is applicable in the table-motion direction, which enables us to set the read-out in the direction of a larger FOV or a higher resolution, and reduces imaging time.

Method

Figure 1 shows the direction of SPE and how the table moves. Imaging starts when one side of the total FOV is at the center of the signal-acquisition region and ends when the opposite side comes to the center of the signal-acquisition region. The direction of SPE is the same as the table-motion. SPE is implemented with a gradient pulse before the A/D period. The gradient pulse amplitude varies while the table continuously moves. Figure 2 shows the relationship between sliding phase encoding and the table position. The sliding phase encoding value changes from $-\pi$ to π each time the table traverses the length of the acquisition region. The SPE sequence diagram itself is similar to a conventional sequence using phase encoding, but it breaks the relationship between MR signals and the reconstructed images away from the Fourier Transform, because the signal-acquisition region differs in each echo. However, SPE still preserves the correspondence between reconstructed images and MR signals when the system parameters, such as the coil sensitivity, are given. The MR signals, s(n), acquired with SPE are written as

$$s(n) = \int_{r=0}^{FOV_{total}} m(r)\exp(-2\pi(r - r_{table}(n))k(n))w(r - r_{table}(n))dr \quad (1)$$

where n is the echo number, m(r) is the magnetization at position r, $r_{table}(n)$ is the table's displacement at echo n, k(n) is the SPE value shown in Fig. 2, and w(r) is the system parameter, such as the coil sensitivity. The image, m(r), is reconstructed by solving m(r) in Eq. 1. Unfortunately, this equation is in ill condition, but the optimization method works well to solve m(r). MR signals encoded with sliding phase encoding are simulated with phantom experiments, and the image is reconstructed from the correspondence between the reconstructed images and the MR signals.

Experiment

The proposed method was tested through phantom experiments. The SPE data was simulated from 128 images, which were took at 128 table positions, each slide to 5 mm. One echo from each image constructs the SPE data. Each image was taken in a stationary state. The receiver coil was fixed to a magnet. Only the coil sensitivity was considered as a scanner parameter for the reconstruction. Sensitivity, in cases where the table was centered on a static magnetic field, was applied for all table positions. The other imaging parameters were: matrix= 64 x 128, total FOV= 32 x 64 cm^2, sub FOV= 32 x 32 cm^2, slice= coronal, sequence= SE, TR= 200 ms, TE= 25 ms, FA= 90°, and thickness= 8 mm.

Results

Figure 3 shows a result of the SPE reconstruction. The result shows that the proposed methods successfully reconstructed an image when the MRI signal was affected by various scanner conditions, such as coil sensitivity, gradient non-linearity, magnetic field inhomogeneity, and RF inhomogeneity. A folding artifact appears at the right end of the total FOV.

Discussion

The folding artifact, which appeared at the right end of the total FOV, may depend on receiving coil sensitivity errors. This may be because only the coil sensitivity for tables centered on static magnetic fields was used in this reconstruction. At this stage, each signal is acquired in the stationary state and the imaging sequence is spin echo. For further research, signals should be acquired for moving states and various image sequences should be applied. Furthermore, information on the scanner conditions, such as gradient non-linearity, magnetic field inhomogeneity, and RF inhomogeneity should be easily built into our reconstruction process. How the accuracy of the reconstruction will improve when the scanner parameters are considered should be tested.

Conclusion

We propose a new MR imaging method, namely sliding phase encoding, for extended FOV. SPE is applicable in the table-motion direction and achieves fast imaging for cases when homogeneous space is compressed in the table-motion direction or high resolution is needed in the orthogonal direction of the table-motion. Phantom experiments show that the proposed methods are able to successfully reconstruct images.

Reference

[1] David G. Kruger, Stephen J. Riederer, Roger C. Grimm, and Phillip J. Rossman. Magn. Reson. Med. 2002; 47:224-231.

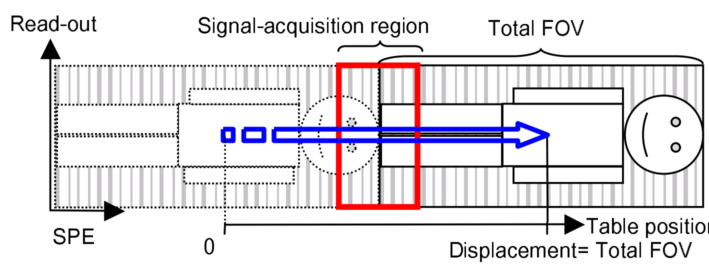

Fig. 1. Sliding phase encoding and table-motion. The axis used for the SPE runs in the same direction as the table motion.

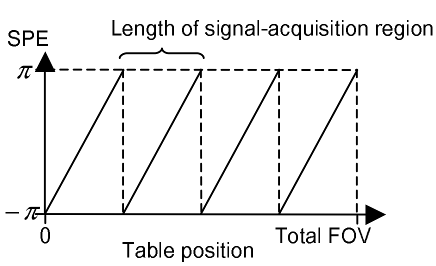

Fig. 2. Sliding phase encoding vs. table position: SPE value changes from $-\pi$ to π each time table traverses length of acquisition region.

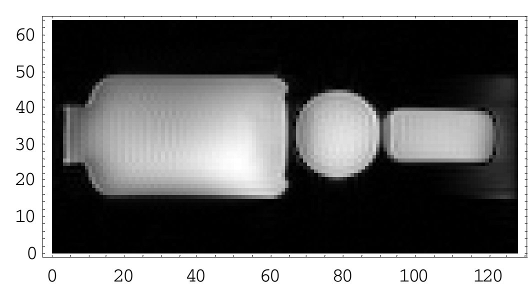

Fig. 3. Phantom experiment. The SPE data was simulated from 128 images, which were took at 128 table positions each slide to 5mm. One echo from each image constructs the SPE data.