Robust Estimation of Planar Rigid Body Motion in Magnetic Resonance Imaging

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ABSTRACT

In magnetic resonance imaging, some schemes require slow data acquisition in order to achieve acceptable spatial resolution and signal-to-noise ratio. However, long scanning periods allow patient motion to distort the reconstructed images. In cases of planar rigid body motion, rotational and translational transformations can be assumed to model the patient motion. In this study, we develop a novel technique for estimating the parameters of these transformations. It is based on fixing two markers to the patient and using a post-processing algorithm to estimate the location of these two markers during the scan time. The key factor in this algorithm is to incorporate *a priori* information about the geometric shape of the markers to achieve accurate estimation. Computer simulation results confirmed the ability of the algorithm to estimate the unknown parameters.

Key words: motion estimation, affine transformation, MRI, markers.

1. INTRODUCTION

In magnetic resonance imaging (MRI), data is acquired in the k-space, which is the Fourier transform of the imaged anatomical cross-section. According to the MRI scanning techniques, Hydrogen nuclei inside the imaged slice are excited by a radio-frequency signal. Returning back to their relaxation-state is accompanied by an emission of an echo signal, which is exploited to carry information about the spatial details of the imaged slice. To this end, temporal coding of the emitted signal is used to specify which part of the k-space being acquired. In a typical MRI acquisition, a slice is excited then the emitted signal is coded to represent one row of the k-space, referred to as a phase-encoding line. A period TR (Time-to-Repeat), in the order of seconds, is waited to give a chance for the nuclei to completely relax so that when re-excited, an appreciable signal can be measured. The excitation-relaxation cycle is continued until all the rows of the k-space are acquired. The direction of acquiring the rows is called the phase-encoding direction. Due to the delayed acquisitions, patient motion causes inconsistencies in the acquired k-space, eventually producing a blur in the reconstructed images which is a main drawback in the MRI.

Recently, compensation for rigid body motion is performed by estimating the parameters of the patient motion throughout the scan time via navigator echoes. A navigator echo is a readout signal along any axis in the k-space. From the projection theory, this navigator echo represents the Fourier transform of the image projection onto the readout axis. Examples include a technique that corrects for the motion artifacts during the head imaging using spatial-frequency-tuned markers (small liquid-filled acrylic containers) [1]. It uses navigator echoes, along two orthogonal axes, to determine the location of the two markers, which are fixed to the patient head. From their location, modeling of the head motion, and hence correction for it is possible. A variation of the navigator echo technique without using markers was proposed by Fu et al [2]. This technique uses an orbital navigator echo, which is a readout echo signal a circular path rather than a linear axis. Navigator echoes techniques provide good estimation of the motion provided that changes in successive echoes are solely due to changes in the object location, e.g. assume homogeneity of static magnetic field. Nevertheless, it requires modification of the acquisition procedure and may cause saturation to anatomic areas of particular interest. Moreover, a lengthy navigator echo should be measured to attain accurate estimation, for example, an echo with 1024-point is required to attain resolution of 0.35°[3].

In this study, a new technique, which is based on markers, is proposed. The innovation in this technique is to estimate the marker location using *a priori* information about its shape instead of the navigator echoes thus maintaining the acquisition time at its normal value.

2. THEORY AND METHODS

A necessary step in the current algorithm is to obtain the hybrid space of the object. The hybrid space derives its name from the fact that it carries frequency

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information along one direction and spatial information along the other. It is obtained by taking the inverse 1-D Fourier transform of the individual phase-encoding lines. This is described in figure 1. In the spin-warp MRI, the hybrid space contains the history of the patient motion along its phase-encoding direction. This is readily deduced from the fact that each phase-encoding line is acquired in a few milliseconds, a time that is short enough to assume stationarity. Yet, moving to the next line takes a time TR as explained above. As a result, most of the motion information exists only in the phase-encoding direction. To a certain extent, the object can be considered stationary during the acquisition of any segment containing few k-space phase-encoding lines. The previous assumption is valid for many acquisition procedures such as GRE and spin-warp, and ideal in FSE where each segment is acquired in few milliseconds. In this case, estimating the motion parameters, at the different acquisition times, corresponds to identifying the location of the markers at the different segments.

Usually MRI images are surrounded from both sides by blank areas, which we will refer to as Blank Vertical Bands (BVB). The presented algorithm makes use of two rectangular markers placed in the BVB as indicated in figure 2. Each complete line within the BVB region of the hybrid space, is the Fourier transform of a gate function representing the marker cross section. As will be shown in the next section, the marker location can be estimated from only few samples of this line. Next, combining information from the different lines of the BVB, we get a complete registration of the markers orientation.

2.a. Estimation of the marker cross-sections

Since the marker cross section in the phase-encoding direction is rectangular in shape, it can be represented by a gate function, i.e.,

$$f(x) = R(a, \varepsilon_1, \varepsilon_2) \tag{1}$$

Where $f(x)$ is the 1-dimensional cross section of the marker and R is a gate function of amplitude a. The two corners of the gate are at ϵ_1 and ϵ_2, and therefore, its center is at $(\epsilon_1 + \epsilon_2)/2$, and the width is $(\epsilon_2 - \epsilon_1)$. If the Fourier transform of $f(x)$, is $F(k)$, then differentiating both sides of equation (1) with respect to spatial index x and obtaining Fourier transform for the result gives

$$\bar{F}(k) = 2\pi jk\ F(k) \tag{2.a}$$

$$\bar{F}(k) = a\ e^{-2\pi j k \varepsilon_1} - a\ e^{-2\pi j k \varepsilon_2} \tag{2.b}$$

The above equation represents samples of two exponentially damped signal, then estimation of the unknown parameters is possible, given at least four samples of the signal spectrum [3]. This is achieved by a direct application of the modified linear prediction technique proposed by Kumaresean, et. al.[3]. First, consider the following prediction-error filter polynomial

$$g(2)Z^2 + g(1)Z^1 + 1 = 0 \tag{3}$$

where $g(1)$ and $g(2)$ are obtained by solving the following linear equation

$$\begin{bmatrix} \bar{F}(2) & \bar{F}(3) \\ \bar{F}(3) & \bar{F}(4) \end{bmatrix} \begin{bmatrix} g(1) \\ g(2) \end{bmatrix} = -\begin{bmatrix} \bar{F}(1) \\ \bar{F}(2) \end{bmatrix} \tag{4}$$

Then the roots of the polynomial, are given by

$$Z_m = e^{-2\pi j \varepsilon_m} \quad m = 1,2 \tag{5}$$

From these roots, precise location of the marker cross section is fully determined. By taking the average of the two edges, it is possible to locate the points of the marker central axis, which will be used later to estimate the motion parameters. Using this algorithm to estimate the marker edges at the different cross-sections results in determining the marker location and orientation in one segment. Repeating this procedure for the other segments gives full tracking of the marker during the scan time. Weighting the k-space phase-encoding lines by a hamming, or a Bartlett window [5] before obtaining the hybrid space is a necessary step to avoid the Gibbs ringing [5] results from the imaged object. This step is used only in the motion estimation stage, hence resolution degradation in the direction of the weighted-rows will not remain in the final corrected image.

2.b. Marker Orientation

In the aforementioned method of estimating the edges locations, the wider the marker cross-section, the more accurate the results. Therefore, in the estimation process, it is advantageous to use only the inner cross-sections, which is of the maximum width. However, only the marker central axis can be deduced from this incomplete set of cross-sections. Accordingly, any movement along this axis will not be detected because the estimated central lines of the markers before and after the motion will coincide with each other thus the relative motion cannot be estimated. This is illustrated in figure 3.a. Estimating the entire 2-D movements requires information from both markers together as shown in figure 3.b. Therefore, when the object moves in either directions or even rotates, the intersection of the markers central lines can provide sufficient information about the entire motion. This is discussed in the next section.

2.c. Motion Estimation

For each segment in the hybrid space, assume that we have obtained few points on each marker central line by applying the algorithm discussed in section (II.a). Determining the marker central line equation is then a simple curve-fitting problem. Complete registration of the marker motion, and hence the patient movements, can be achieved by comparing the location of the points A, B, and C, indicated in figure 2, at the different segments. As indicated, point A is the intersection of the two central lines of the two markers. Points B and C are taken along the central lines of marker M1 and M2, respectively, such that they are at predetermined fixed distance from point A. Assume the first segment is taken as the reference of the motion, and points (A, B, C) for this segment is obtained. Next, if points (A^*, B^*, C^*) was obtained for another segment, then the motion parameters can be obtained by solving a linear system which represents a simple affine transformation that maps (A,B,C) to (A^*,B^*,C^*).

3. RESULTS

Verification of the proposed algorithm is performed using computer simulation. In this simulation, the imaged object and the markers are as shown in figure 2. Motion was introduced to individual k-space segments. Each segment contains eight PE-lines. In order to test the sensitivity of the technique to slight patient movements, the true motion parameters were in the range (1:2.5 pixels) for translation and (-1:1 degree) for rotation. In figure 4, the true motion angles along with the estimated motion are shown. The ability of the proposed technique to estimate wide range of motion is limited by the number of the marker cross sections that maintain their maximum width even after the motion (the bold lines in figure 3). Therefore, instead of plotting the error at different motion ranges, the error is plotted against the number of these lines for the same range of the motion. This is shown in figure 5 and it is apparent that reducing the number of the cross sections increases the estimation error. The results above were achieved in presence of zero-mean uniform noise. The signal-to-noise ratio (SNR) was greater than 60 dB. The error for translation motion direction was found to be within 0.2 pixel. It should be noted that motion in the phase encoding direction does not reduce the number of the cross sections used in the estimation process, hence the only limiting factor is that the marker should lie inside the imaged field of view.

4. DISCUSSION

Estimating the motion parameters is followed by a correction step to obtain a blur-free image. The translational motion can be corrected by adding the appropriate phase shift to the different k-space samples. Correction for the rotational motion is achieved by rotating the k-space samples with angles equal to the estimated angles but in the opposite direction, a process that requires a gridding algorithm [6] to re-sample the rotated k-space points onto the ordinary rectangular grid. It was shown in section (2.a,), that only four samples are required to estimate the location of the boxcar cross sections. However, more robust results can be obtained if the segment size increases. This happens because equation (4) becomes an over-determined system, hence solving it using SVD, where small singular values are truncated, reduces the noise effect. However, increasing the number of lines in each segment is against the assumption that the object is to be stationary in the individual segments. It was found that eight lines per segment are sufficient for good estimation, meanwhile the assumption of the object-stationarity during the acquisition of one segment remains valid.

5. CONCLUSIONS

In this study, we have shown that navigator echoes are not necessary for estimating the motion parameters when using markers. This has the advantage of unchanging the acquisition procedure and in the same time still gains more accuracy in the estimation process. Moreover, as there is no navigator echoes the proposed technique does not saturate the imaged slice. The idea is to utilize a *priori* information about the marker cross section. The proposed algorithm does not involve much computations that is, solution of a few linear equations is only required to estimate the motion parameters. Accordingly, the algorithm can be used as part of real time reconstruction algorithm. When motion is estimated, it can be used to appropriately modify the gradient waveforms of the MRI system thus the collected data would follow the imaged object as it moves.

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Figure 1. Obtaining the hybrid space.

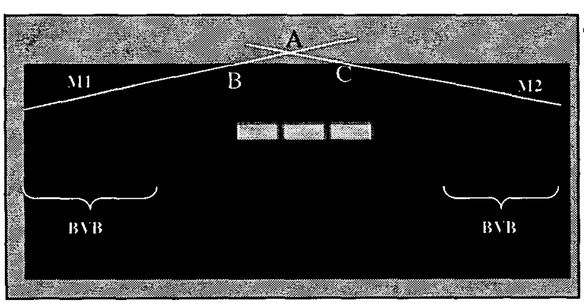
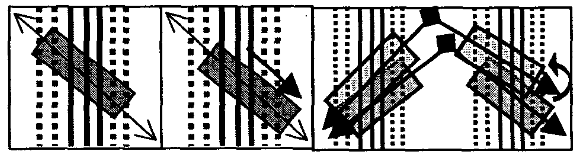


Figure 2. The imaged object and the markers (M1 an M2).



(a) (b)

Figure 3. Solid lines represent the marker cross-sections used in the estimation process. **(a)** The lines are not reflecting the motion in the direction of the marker central axis. **(b)** Object 2-D motion can be estimated easily from two markers.

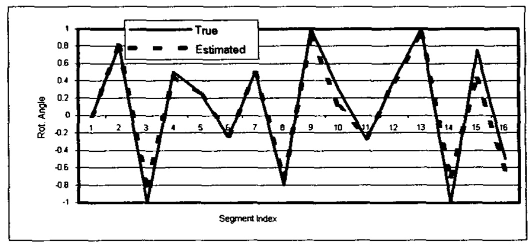


Figure 4. True and estimated rotation angle.

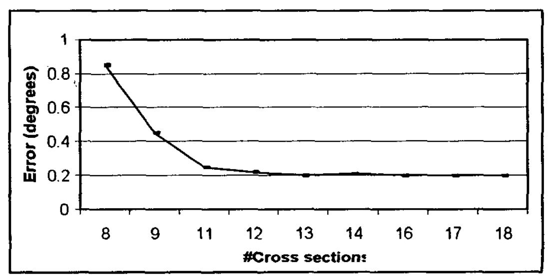


Figure 5. Rotation angle error versus number of marker cross sections used in the estimation.