Real-Time Organ Motion Tracking and Fast Image Registration System for MRI-Guided Surgery

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SUMMARY

The advantage of magnetic resonance imaging (MRI)-guided surgery, in which MR images taken during surgery are used to guide the surgery, has been recognized recently. However, there is a problem, due to long imaging time of MRI, that it is difficult to capture the respiratory organ motion. It is often a constraint that the patient must hold his or her breath, especially in the treatment of the abdominal region. Consequently, this paper proposes a method of eliminating this constraint in MRI-guided surgery by real-time monitoring of the movement of the organ involved in the surgery by means of the signal derived from MRI. Specifically, the navigator echo is extracted in imaging during surgery. The movement of the organ involved in surgery is detected on the basis of the change in the calculated projection profile, and the results are presented to the surgeon. In this study, the procedure is built into a 1.5-T clinical MRI system. The feasibility of the proposed system was examined under clinical conditions by evaluating its accuracy and delay in a phantom experiment, and also in an in vivo experiment on the liver of a volunteer. The results are reported in this paper. © 2005 Wiley Periodicals, Inc.

Key words: MRI-guided surgery; navigator echo; projection profile matching; image registration.

1. Introduction

In recent years, MRI-guided surgery, in which an MRI system is used to monitor the target lesion during therapy, has been attracting attention. MRI has various advantages, such as soft tissue discrimination, radiation safety, thus reducing the burden on both patient and physicians, and the possibility of arbitrarily setting the image slice direction. These features make MRI very useful as an image modality in guiding surgery.

Reports of clinical applications, such as biopsy [1], cryotherapy [2, 3], and local drug administration [4], have been presented. There are reports on the use of temperature maps derived from MR signals, such as laser hyperthermia [5, 6], coagulation by radio frequency (RF) waves [7] and microwaves [8], and ablation by focused ultrasound [9, 10].

On the other hand, there remain problems. In continuous scanning during surgery, an image is limited to 2D and its quality is much lower than conventional diagnostic MR images. It is difficult to continuously provide a clear tumor image during surgery. At least 1 or 2 seconds is required for a 2D image. In addition, there is a time delay of several seconds between image acquisition and the display of the
reconstructed image. Thus, the present MRI cannot be considered a suitable real-time tool for surgery in terms of time resolution and imaging delay. Artifacts due to movement (motion artifacts) also occur, and it is still difficult to perform surgery based on MR images when the organ continues to move.

As a method of compensating image quality during surgery, Gering and colleagues proposed a visualization system for MRI-guided surgery in which high-quality preoperative images and real-time intraoperative images are combined and the result is presented to the surgeon [11]. The system has the feature that the optically tracked hand piece allows surgeons to determine the imaging plane position and orientation interactively during operation. Another advantage is that a preoperative high-quality 3D image is resliced at the plane corresponding to the real-time imaging plane and displayed.

The preoperative 3D image is resliced and displayed with a time resolution of 10 Hz, which is the same as in tracking by the controller. Despite the constraint imposed by the rate of MR image updating, it is possible to make determinations such as the point of needle incision without difficulty. However, it should be noted that the intraoperative image is not updated at a rate of 10 Hz. Consequently, it cannot adapt to fast organ movement due to respiration and other factors.

Efforts to cancel the effect of object movement on the image in MRI-guided treatment have been made in studies by de Zwart [13] and Vigen [14] of body movement correction using the navigator echo [15] in temperature mapping images used in MRI-guided hyperthermia. The navigator echo is a special MR echo which is used to detect the movement of the imaging object. Compared to a conventional imaging, which requires generally 128 to 256 echo acquisitions, information on the position change of the imaged object can be acquired in a very short time (i.e., with high time resolution).

In the methods of de Zwart and Vigen, the navigator echo acquisition process is inserted in the imaging sequence, so that time-series data on the displacement of the imaging object are acquired while the image is being acquired. Using the acquired data, the phase deviation of the echo data in image reconstruction is corrected in order to reduce motion artifacts of the temperature image and degradation of accuracy in the temperature image.

Due to the constraints of the clinical MRI system, however, in past studies the navigator echo has been stored, together with the echo data for image reconstruction, and when all data for an image are ready, they are transferred as a block to the processing unit. Thus, although the navigator echo contains displacement information on the imaging object with higher time resolution than the image itself, the information is made available at the same timing as the images. Consequently, the displacement of the imaged object is not known in real time, even though the motion can be compensated in the image by postprocessing of the data.

This paper proposes a real-time organ tracking/fast image registration system intended to solve problems related to the time resolution, delay, and image quality of intraoperative MRI. Specifically, the following procedures are considered.

1. Delay is minimized by not storing all echo data, but instead transferring the data successively to the echo processing unit.
2. The displacement of the object is measured with a higher time resolution than intraoperative MRI, using navigator echo.
3. Based on the measured object displacement, motion artifacts in the intraoperative image are corrected, and the registration of the intraoperative image against the high-quality preoperative image is performed.

By this approach, the intraoperative image and the preoperative image can follow the movement of the treated area in real time. It also becomes possible to display the intraoperative MR image, which represents the latest state of the area, with motion artifacts eliminated. In this study, the prototype system is constructed on a 1.5-T clinical MRI system. The accuracy of displacement measurement and the delay between position change of the imaged object and determination of its displacement are evaluated. In addition, an in vivo experiment performed on the liver of a volunteer is reported. The feasibility of the proposed method in a clinical situation was investigated.

2. Method

2.1. Projection profile matching

Consider two-dimensional Fourier imaging. Let the frequency encoding direction on the 2D image plane be $x$ and the phase encoding direction be $y$. Magnetization is produced with a distributed $M(x, y)$ on the 2D plane. The gradient magnetic fields $G_x$ and $G_y$ are applied in the $x$ and $y$ directions, respectively.

Then, the following NMR signal is obtained:

$$ S(k_x, k_y) = \int x \int y M(x, y)e^{-i(k_xx+k_yy)} dxdy \quad (1) $$

Here, $k_x = \gamma G_x t_x$ and $k_y = \gamma G_y t_y$, $\gamma$ is the gyromagnetic ratio. $t_x$ and $t_y$ are the duration of application of $G_x$ and $G_y$, respectively.

The inverse Fourier transform is applied, only in the frequency encoding direction, to the echo signal obtained by setting $G_y$ to zero. Then, we obtain

*For the principle of MR imaging, see Ref. 12, for example. It is also outlined in the Appendix.*
This is the projection of the magnetization distribution onto the \( x \) axis.

Assuming that the motion of the imaging object consists of translation on the imaging plane, the motion is observed as shift of the projection. Consequently, by matching the projection profile at the start of measurement and the projection profile of the time under consideration, the change of position can be determined. This echo signal, which is obtained by setting \( G_y \) to zero, is used as the navigator echo. Similarly, by exchanging the direction of phase encoding and frequency encoding, the projection onto the \( y \) axis can be determined.

Below, only the \( x \) direction is considered. The projection on the \( x \) axis at time \( t = t_i \) is defined as follows:

\[
p_i(x) = \int_y M(x, y) \, dy
\]

(3)

Let the displacement of the imaging object from \( t = t_0 \) to \( t = t_i \) be \((\Delta x_i, \Delta y_i)\). Then we have

\[
p_i(x) = \int_y M(x - \Delta x_i, y - \Delta y_i) \, dy
\]

\[
= p_0(x - \Delta x_i)
\]

(4)

The contribution of parts that extend outside the range of integration (i.e., the range of imaging) is ignorable.

In order to derive \( \Delta x_i \) from \( p_0 \) and \( p_i \), which can be obtained from the navigator echo, matching of the distributions of \( p_0 \) and \( p_i \) is performed. Let the function that evaluates the degree of matching between \( p_0 \) and \( p_i \) be \( f(p_0, p_i) \). Then the calculation of \( \Delta x_i \) is represented as

\[
\Delta x_i = \text{arg max}_{\Delta x} f(p_0, p_i)
\]

(5)

In this study, the cross-correlation function is used as \( f \).

Practically, \( p_i(x) \) is the sampled distribution. Let \( x \in [-x/2, x/2] \), and let the number of samples be \( N_x \). Let \( x_n = (n_i - N_i/2) \times x/N_i \), \( n_i = 0, \ldots, N_i - 1 \) and \( p_{i,x_n} = p_i(x_n) \). Then the cross-correlation function is calculated as

\[
f(p_0, p_i) = \frac{\sum_{n_x} (p_{0,n_x} - \bar{p}_0)(p_{i,n_x} - \bar{p}_i)}{\sqrt{\sum_{n_x} (p_{i,n_x} - \bar{p}_0)^2 \sum_{n_x} (p_{i,n_x} - \bar{p}_i)^2}}
\]

(6)

Here \( \bar{p}_i \) is the mean of the distribution \( p_{i,x_n} \). In the optimization of Eq. (5), the golden section algorithm is used.

### 2.2. Pulse sequence

In order to incorporate projection profile matching in MR imaging, a pulse sequence is developed based on the gradient echo method. In this pulse sequence, the navigator echoes used to determine the movement in the \( x \) and \( y \) directions, respectively, are acquired alternately between the echoes for image reconstruction (Fig. 1). Because of this configuration, the time resolution of object tracking by projection profile matching depends on the imaging parameter TR, and is \( 1/\text{TR} \times 4 \) [Hz] in each direction.

#### 2.3. Real-time tracking/fast image registration system

In order to perform the above processing online, we implemented a real-time tracking/fast image registration in a clinical MRI system. In a 1.5-T MRI Signa MR/i system (GE Yokogawa Medical Systems Ltd.), the program of the data processing computer (Transfer, Processing and Storage: TPS) was modified so that the echo data could be sent immediately after acquisition without being stored in the image reconstruction buffer. This was done in order to minimize the delay before the echo data were processed.

The sent data pass through the host computer (SGI Octane), which is connected to the TPS through a BIT-3 interface, and are transferred to the real-time echo processing computer (CPU: Intel Pentium 4 2.8 GHz, RAM: 512 MB PC1066 RIMM and OS: RedHat Linux 7.3) via Ethernet (100 Base-T). Processing such as image reconstruction which was originally implemented in the MRI system was bypassed. Figure 2 shows the configuration of the system incorporating these components.

Figure 3 shows the processing flow in the real-time echo-processing computer. In this processing, the received echo data are discriminated as the imaging echo or navigator echo, based on the header tag in the transfer, and the subsequent processing is selected. In the case of the navigator echo, the current position information of the imaging object is updated after projection profile matching. Based on the result, image registration is performed. In the case of the data for image reconstruction, the signal is corrected

![Fig. 1. The sequence for acquiring imaging echo and navigator echo.](image-url)
on the basis of the current position information, so that the motion artifact is reduced.

The motion artifact accompanying translation of a rigid body is due to the phase-shift error of the data in each row of the $k$ space, produced by movement of the object during imaging. Consequently, the artifact can be reduced if the shifted phase is corrected. The phase shift is determined by the translation of the object. For a translation of $(\Delta n_x, \Delta n_y)$ pixels, the shifts are expressed as follows [15]:

$$\phi_x(n_x) = \frac{2\pi n_x \Delta n_x}{N_x}$$  \hspace{1cm} (7)

$$\phi_y(n_y) = 2\pi \Delta n_y \left[ n_y - \left( \frac{N_y - 1}{2} \right) \right] \frac{1}{N_y}$$  \hspace{1cm} (8)

By substituting the corrected data into the corresponding line in the $k$ space data, and applying the 2D fast Fourier transform, a 2D image expressing the latest echo data can be output. After correction, the image appears at the position of $t = t_0$. Then, by utilizing the position information at the time when the navigator echo is acquired, the image is aligned to the actual position of the object (registration).

Of course, the method can be applied not only to the corrected image, but also to the registration of the high-quality preoperative image. The above procedure makes it possible to present an image with the latest positional information to the surgeon at intervals of several tens of milliseconds, while reducing motion artifacts. Thus, the three items described in the previous section—(1) reduction of delay, (2) improvement of time resolution, and (3) image registration and correction—are realized.

3. Experiment

3.1. Real-time tracking of phantom

In order to investigate the performance of the proposed real-time tracking/registration system, tracking by the proposed method was performed while a phantom was moved in the MRI scanner gantry. The accuracy of displacement measurement of the phantom, and the delay until the actual phantom movement was reflected in the final result, were evaluated. As a gold standard for this evaluation, phantom tracking was also performed using video camera images. The effect of image correction on the motion artifact was also evaluated.

A uniform cylindrical phantom with a diameter of 160 mm was used, as shown in Fig. 4. The phantom was placed in the gantry so that the axis was oriented in the direction of the static magnetic field. This cylindrical phantom was moved during the scan so that the axis was always...
in the same direction. A wedge-shaped mattress was set to provide a slope in the gantry, as shown in Fig. 4, generating the vertical component (y direction) of the phantom movement. The imaging plane was set as axial, parallel to the bottom face of the cylindrical phantom, so that the circular phantom image moved in the horizontal and vertical directions on the 2D image.

The MR imaging parameters were TR: 100 ms, TE: 8.4 ms and FOV: 350 mm, and the resolution of the image was 256 × 128 pixels. The scan time for an image was approximately 25 seconds. The time resolution of position detection was 2.5 Hz (period 400 ms) in each direction. In real MRI-guided surgery, TR is 20 to 30 ms. In the implementation in this experiment, however, TR was set to more than 80 ms, since TR shorter than 80 ms causes transfer errors due to hardware limitation in this implementation.

In order to measure the locus of the phantom to be used as the reference for evaluation, a digital video camera (Sony DCR-PC110) was set outside the gantry, so that the cylindrical phantom could be imaged from the bottom (Fig. 4). The image was taken with a resolution of 720 × 480 pixels and a frame rate of 29.97 fps. When the phantom was at the reference position (center of the camera view field), the distance between the camera and the phantom was 5 m. After the experiment, a template image of the phantom bottom was constructed from the phantom image at the initial position in the series of acquired images. Template matching was applied to the other images in the time series, and the time course of the phantom position was calculated.

The delay until the change of the phantom position was expressed in the corrected image was estimated from the output time difference between the video camera image and the corrected MR image. For this purpose, the video camera was connected to the real-time echo-processing computer by an IEEE1394 connection, so that the real-time tracking data and the video image could be recorded simultaneously in the same PC with time of data arrival.

The delay is the sum of the times required for data acquisition, transfer, data processing, and display output, as shown in Fig. 5. It was noted in the evaluation that a similar kind of delay existed in the video image used as the gold standard. That is, letting the delay in the real-time tracking be \( T_a \) and the delay of the video camera image be \( T_b \), the time difference obtained from the records in the real-time echo-processing computer is \( T_c = T_a - T_b \). Consequently, \( T_b \) must be measured beforehand in order to determine \( T_a \).

\( T_b \) was determined before the experiment as follows. The video camera was directed toward the display of the real-time echo-processing computer. The object was moved between the display and the camera. Then, the delay between the directly taken object image and the object image taken again by the camera from the image displayed on the screen was determined (in frames).

Fig. 5. The relation between the delay of projection profile matching and video tracking with respect to the time point of the data acquisition.

\( T_c \) can be calculated from the time shift between the time series of the phantom’s displacement obtained by real-time tracking and video tracking. In this study, the shift in the time direction was calculated by matching the respective time-displacement data. The matching was performed by maximizing the cross-correlation coefficient.

### 3.2. Real-time tracking of liver of volunteer

In order to investigate whether the real-time tracking/fast image registration system could be used under clinical conditions, an in vivo experiment on the tracking of the liver moving as a result of respiration was performed. The liver movement accompanying respiration is dominated by parallel translation in the S-I direction [16]. Consequently, it is assumed in this study that the liver movement can be approximated as a 2D motion in the sagittal plane. The slice was set in the sagittal plane, and the liver movement was determined from the movement in the imaging plane.

The imaging parameters were the same as in the phantom experiment, being TR: 100 ms, TE: 8.4 ms, FOV: 350 mm and an image resolution of 256 × 128 pixels. The baseline image was taken at the beginning with breath-holding. Then the subject was instructed to breathe freely and real-time tracking/registration was performed.

In this experiment, there is no means of measuring the precise position of the liver inside the body at a higher rate than the acquired MR images. Consequently, an abdominal circumference sensor was attached to the volunteer subject to monitor the respiration phase. A qualitative evaluation was then performed using the acquired respiration phase curve and the result of liver tracking and measurement by the proposed system. At the same time, reduction of the artifact in the corrected image was investigated in order to validate that the liver displacement data were acceptable for image correction.
Based on the above evaluation, together with the results of the phantom experiment described in the previous section, the feasibility of the proposed system to the clinical site was investigated.

4. Results

4.1. Real-time tracking of phantom

Figure 6 shows the time course of the phantom position, derived from the real-time tracking and video tracking. Figure 7 shows a series of image updates in the real-time echo-processing computer in which the images are extracted at intervals of 9 frames and arranged in the time direction. The images in row (a) are without correction and registration. The images in row (b) are after real-time correction and registration. The figure shows the change in a period shorter than the acquisition of a single image in the conventional system.* Thus, it is shown that information concerning the position and character of the imaged object can be provided with higher time resolution than in the conventional method. The effectiveness of motion artifact suppression is also observed.

The delay $T_b$ of the video camera image itself is 3 frames (100 ms). The delay of real-time tracking with respect to the camera image is $T_c = 100$ ms. Thus, it is evident that the delay of the real-time tracking is $T_a = 200$ ms.

On the other hand, the error of the video tracking and the projection profile matching is 1.5 mm in the $x$ direction and 2.0 mm in the $y$ direction (as RMS values). These values correspond to 1.1 and 1.5 pixels, respectively, in the MR image. The standard deviations of the absolute error are 0.94 and 1.17 mm, respectively.

4.2. Real-time tracking of liver of volunteer

Figure 8 shows the time course of the respiratory phase and liver position, as determined from the projection profile matching. The liver repeats two-way motion in the S-I direction together with the diaphragm, as respiration proceeds. It is evident from Fig. 8 that the liver motion was tracked by the proposed method, and is aligned to the respiration phase.

Figure 9 shows the updating of the liver image. Panels (a), (b), and (c) are the baseline images, the real-time images without correction, and the real-time images with correction and registration, in which the edge of the liver extracted from the baseline image is superimposed at the estimated position of the actual liver, based on the result of projection

*For image taken with the same parameter values.
profile matching. Panel (a) shows how the edge of the liver deviates from the reference position; edges are superimposed on the same baseline image. We see from the figure that the motion artifact of the liver is reduced, and that the movement data obtained by projection profile matching are also adequate in the in vivo observation. For organs other than the liver, the assumption of translation described above does not apply, and the motion artifact is not eliminated.

5. Discussion

In this study, a real-time tracking/fast image registration system using the projection profile matching was implemented in a clinical MRI system. An evaluation experiment was performed using a phantom, and it was shown that tracking with a time resolution of 2.5 Hz in each direction and with a delay of 200 ms could be achieved, with the error of 1.5 mm (1.1 pixels) in the frequency encoding direction and of 2.0 mm (1.5 pixels) in the phase encoding direction. It was also shown that the motion artifact can be effectively corrected. A tracking experiment was performed on the liver of a volunteer, and it was verified that the observed movement data of the liver were synchronized with the respiration. These results show that the proposed method is feasible in a clinical environment.

The tracking period depends on the imaging parameters. In this study, TR was set as 100 ms, considering the system’s constraint of echo data transfer, which resulted in a time resolution of 2.5 Hz (corresponding to a period of 400 ms). It is possible to set TR as 20 to 30 ms by optimizing the system, and it is expected that time resolutions greater than 10 Hz will be achieved.

In intraoperative imaging, information has been obtained in intervals no shorter than 1 to 2 seconds. Compared to this, tracking directly utilizing MRI echo data has very high time resolution. This property offers promise as non-invasive real-time organ motion tracking for automated targeting, especially in MRI-guided robot [17] and focused ultrasound surgery (FUS) [18].

The reasons for the different matching accuracies in the x and y directions are considered to be the video tracking accuracy used as the evaluation reference, and the difference in the maximum displacements of the phantom in the x and y directions. The authors attempted more precise measurement with an experimental 2.0-T MRI and a CCD laser displacement sensor, which indicated an error of 0.3 mm for a FOV of 100 mm [19].

This study has considered only translation on a plane, and it is left as a future topic to increase the number of degrees of freedom in tracking. As an extension of the navigator echo, a spherical navigator echo has been proposed [20], in which movement with six degrees of freedom is detected by acquiring the navigator echo in such a way that points on a spherical surface in three-dimensional k space are sampled. Although there remain many problems related to use under surgical conditions, it is expected that the technique will be applied to real-time tracking. It will also be necessary in the future to address organ deformation.

6. Conclusions

This study has considered a method of tracking the movement of an internal organ in MRI guided surgery, and has proposed a real-time tracking/fast image registration system. A system that extracts and sequentially processes echo data from the clinical MRI system was experimentally constructed. Using this system, the imaged object can be tracked with an error of 2.0 mm and a delay of 200 ms. Furthermore, an in vivo experiment showed that the liver could be tracked at high speed without incision.

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APPENDIX

Principles of MRI

In MRI, the nuclear spin of water protons in a high magnetic field is excited by electromagnetic pulses, and the echo signal produced on relaxation is received by a coil and its distribution is imaged. The echo signal is produced by the spin precession, with the angular velocity \( \omega \) being proportional to the magnetic field intensity \( B \), represented as follows:

\[
\omega = \gamma B
\]  
(A.1)
where $\gamma$ is the gyromagnetic ratio.

In order to excite only the spins in the imaging slice plane, the gradient field is applied in the direction perpendicular to the slice ($z$), and an electromagnetic pulse with an angular velocity $\omega$ is applied. Thus, only the spins in the plane in which the magnetic field intensity is $B = \omega/\gamma$ resonate and undergo excitation. Below, only this plane is considered.

The echo signal received by the coil is the sum of the echo signals from the whole covered range. In order to image the spin distribution, the gradient magnetic field is introduced, and the echo signal is decomposed into components with frequency and phase depending on the spin position. The phase difference is generated by applying the gradient magnetic field in the $y$ direction $G_y$ for a specified period before the echo signal is acquired. The frequency is varied by applying a gradient magnetic field $G_x$ in the $x$ direction during signal acquisition. $G_x$ is called the frequency encoding gradient magnetic field, and $G_y$ is called the phase encoding gradient magnetic field. The $x$ and $y$ directions are called the frequency encoding direction and the phase encoding direction, respectively.

Let the distribution of the spin (actually the magnetization, which is the vector sum of the magnetic moment vectors of the nuclear spins) be $M(x, y)$, and let the period of application of $G_y$ be $T_y$. Then the acquired signal is a complex number proportional to

$$S(t) = \int_x \int_y M(x, y) e^{-i(G_x xt + G_y y T_y)} \, dx \, dy \quad (A.2)$$

Letting $k_x = \gamma G_x t$ and $k_y = \gamma G_y T_y$ in the above expression, Eq. (1) is obtained.

In order to derive the original distribution $M(x, y)$ from the acquired $S(k_x, k_y)$, the data for $S(k_x, k_y)$ must be acquired for the whole range, and the inverse Fourier transform must be applied. This space of $(k_x, k_y)$ is called the $k$-space. $k_x$ depends on the echo acquisition time $t$, and the signal data can be acquired by a single echo acquisition for the whole range in this direction. However, since $k_y$ is constant during single echo acquisition, acquisition must be repeated by varying $G_y$. Consequently, in ordinary MRI, echo acquisition is repeated a number of times corresponding to the image resolution in the $y$ direction.

Provision is made so that the peaks of the magnetization phase are aligned for all data in the course of data acquisition. For this purpose, before applying the frequency encoding gradient magnetic field, the gradient magnetic field in the same direction is applied for phase inversion. The echo signal obtained by this process is called the gradient echo. Imaging by the gradient echo is called the gradient echo method. The time between excitation by the electromagnetic pulse and the occurrence of the echo peak is called the time of echo (TE), and the interval of electromagnetic pulse application in repeated echo acquisition is called the time of repetition (TR).

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