**Technical Note**

**Rapid Tip Tracking with MRI by a Limited Projection Reconstruction Technique**

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To rapidly track invasive devices within MRI systems, a novel approach using a limited projection reconstruction technique is presented. Our method exploits the difference between images reconstructed from a limited number of projections and serves to depict the tip of a needle during its advancement. This method was implemented on a standard MRI system with a radial fast spin-echo sequence and examined in phantom studies. We demonstrated that the proposed method could track the tip within 300 msec and the tip depicted by the present technique was consistently displaced along the needle by a small distance (5 mm).

**Index terms:** Device tracking • Rapid imaging • Projection reconstruction • Interventional MRI

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**Abbreviations:** FOV = field of view. rFSE = radial fast spin echo.

RECENTLY DEVELOPED open-configuration MRI systems allow direct access to the patient during interventional and surgical procedures (1,2). During most intraoperative applications, interactive control of scan planes is necessary for localization and targeting. This requires the tracking of various flexible and rigid instruments and the acquisition of images that correspond to the position of these devices.

For tracking invasive devices within MRI systems, two approaches have been proposed: an active method and a passive method. In the active method, a small receiver coil is built into the tip of the device to obtain three-dimensional spatial coordinates of the tracked device (3–6). These techniques require continuous use of readout magnetic field gradient pulses and use the specialized hardware. In the passive method, direct visualization of the device in MR images is attempted on the basis of the associated signal void and susceptibility effects (7,8). The dedicated devices with paramagnetic components (9,10) allow conspicuous and consistent visualization, although they produce large artifacts, depending on orientation and imaging parameters. In this method, device tracking is accomplished by updating images and, therefore, fast scanning techniques are necessary to achieve sufficient speed for tracking. To address this problem, the keyhole technique (11) and its variants (12,13) have been exploited for MR-guided catheter tracking (7,8).

Recently, projection reconstruction techniques have been applied successfully for abdominal and thoracic imaging as fast motion-insensitive imaging sequences (14–16). In this work, we propose a rapid projection reconstruction approach that requires no additional hardware for device tracking. We demonstrate its feasibility on a standard MR imager using a phantom.

**METHODS**

Initially, a high resolution static baseline image is acquired and reconstructed. This image may be acquired with any standard sequence. It serves as a baseline image upon which dynamic needle placement related changes will be superimposed. Then, a low resolution second static baseline image is acquired using only a few projections with a radial fast spin echo (rFSE) sequence (15). During a dynamic process, for example needle placement, additional images are acquired using a few projections with the same rFSE sequence. The difference between the images acquired during the dynamic changes and the static baseline image should only contain information about the needle advancement. When the image is reconstructed from a few projections, the difference between these images shows streaking artifacts caused by angular undersampling. Using these artifacts as markers, the position of the needle tip can be identified clearly as a strong signal intensity spot as shown in Figure 1. Then, this image is superimposed on the high resolution baseline image. Because acquisition of only a few projections takes a much shorter time compared with that required for the full resolution image, considerably reduced imaging times, applicable to rapid imaging of dynamic processes, can be achieved.

This method was implemented on a commercially available MRI system (1.5-T Signa, General Electric Medical Systems, Milwaukee, WI). For the high resolution image, a spin-echo image was acquired using the following parameters: TR = 150 msec; TE = 20 msec; field of view (FOV) = 180 mm; slice thickness = 10 mm; image matrix size = 256 × 256 pixels. For the acquisition of projections, we applied an rFSE sequence with an interleaved approach as described by Rasche et al (15). The pulse sequence used for the study is illustrated in Figure 2. Imaging parameters were as follows: TR = 150 msec; TE = 20 msec; number of echoes in one echo train = 4; total number of shots = 24; FOV = 180 mm; slice thickness = 10 mm; number of data points per projection = 256. Only a single slice was acquired. The projections were acquired for the following angular order. For odd shots, (π/4) × n for even shots, (π/4) × n + π/8; where n is the echo number. In this way, we exploited the great flexibility of MRI in choosing the geometry for acquiring projections, in contrast to the mechanical restrictions of modalities such as CT scanners. Using only eight projections, an image update was obtained every 300 msec with only two shots of the rFSE sequence.

To examine our device tracking method, phantom studies were performed. We fabricated a cage for placement of both a phantom and a long bar attached to an 18-gauge MR-compatible biopsy needle (E-Z-
In the correlation study between the actual tip location and the location measured with the rapid projection reconstruction technique or the standard high resolution spin-echo sequence, we found that the positions depicted by these two methods were almost identical. The imaged tip location obtained with these two sequences, however, was consistently displaced along the axis of the needle, slightly behind the actual tip location, by approximately 3 mm.

**Discussion**

In theory, the radial streaks caused by angular undersampling do not have strong signal intensity (17) and, therefore, do not cause serious image degradation to obscure the potential target lesion, where attention should be paid. In an interventional procedure such as biopsy, the image plane must contain both an instrument and a target lesion. To accomplish this task, the needle tip should be monitored carefully to avoid maneuvering it out of the plane. Nevertheless, in practice, multiplane imaging will be necessary to complete this procedure. These images should cover the entire volume within which the procedure is to be performed. The expected course or needle path can be used to limit the imaging volume to be covered.

The method described here merely spots the tip of the needle, which is then superimposed on the previously acquired baseline image. This procedure works as long as the baseline image has remained unchanged. Therefore, after any gross motion, the baseline image must be refreshed. The present technique has achieved the imaging time of 3.6 seconds for 12 frame updates, indicating its potential for abdominal applications in which breath-holding is required.

The spatial accuracy of the tip depicted by the present technique compared to its actual location should be considered carefully. First, we compared these two locations in the correlation study. We found that the tip depicted by the present technique was consistently displaced along the axis of the needle by a small distance (5 mm). This type of displacement may be well compensated by a physician, or a simple display technique for compensation is available. The other issue is the difference between the actual tip location and the tip depicted by the present technique during the needle advancement. This accuracy will be determined when our method is integrated with real-time reconstruction, image display, and an independent position-tracking system (1,2). However, the temporal resolution of 300 msec achieved with the present technique can largely remove blurring during the dynamic process.

The active MR-tracking approach, in which a small receiver coil is incorporated into the device, uses a nonselective radiofrequency pulse and a gradient-echo signal to provide three-dimensional coordinates that can be mapped onto the arbitrary MR images of different planes (3-6). This method requires special hardware and the installation of a miniature coil at the tip of catheters and flexible endoscopes. Therefore, it is an expensive solution to the problem of tip tracking. In the active tracking method, a physician can choose whether to track the device on the previously acquired images of single or multiple planes or to have continuous serial images to monitor the progress of tip position. This active tracking method requires additional time to obtain localizing information using the pulse sequence. In contrast, our method can rapidly track the location of the tip as image updates. These images are believed to be sufficient for monitoring the needle just before reaching the target lesion, as long as gross motion does not occur.

To date, there are some limitations with passive techniques (7,8), in which visualization of the device is achieved by means of the susceptibility effects of paramagnetic materials (9,10). The appearance of the device based on susceptibility effects is not sufficiently reliable; the size and shape of device-induced signal void depends on device orientation in the magnetic field. In addition, the position of the device is updated only as fast as new images are acquired and reconstructed, resulting in the inability to make real-time adjustment of the imaging plane. In MR-
Figure 4. Needle tip tracking by the proposed method. Shown are three frames of a dynamic series of images during the needle advancement: (a), at .9 second; (b), at 1.8 seconds; and (c), at 3.6 seconds. The needle tip is clearly visualized as a white spot with streaking artifacts.

guided endovascular interventions (7,8), keyhole technique (11) and its variants (12,13) have been used to improve a temporal resolution. The use of this technique derives from the fact that the paramagnetic ring attached to the device causes a signal loss that is not sharply defined. When the keyhole technique is applied to monitor the needle, several considerations must be taken to meet the requirements of acceptable needle tip location and needle width with minimal blurring artifacts (18). To our knowledge, a dedicated imaging technique for passive tracking of the needle in real time within interventional MR guidance has not been established. Our method has successfully achieved passive tracking of a standard MR-compatible thin needle. This is accomplished by a simple strategy to depict only the tip using a projection reconstruction scheme with a few projections, avoiding serial acquisition of updated images with full spatial resolution.

In this work, we used an rFSE sequence to implement our method. Radial gradient-echo sequences are also applicable (16), with a concomitant increase in the rate of image update. At present, reconstruction in real time has not been implemented; however, it can be achieved because each reconstruction takes only eight filtered back projection steps. Although current implementation of our method does not directly offer the three-dimensional position of the device, dedicated software for calculation of positional information using the selected slice and the projections could be developed. Two tip positions can define the needle trajectory. An arbitrary plane containing this line can be displayed after multiplanar reformation of the multislice and/or volumetric images. A new baseline image can also be acquired along the predicted trajectory. Following these procedures, the dynamic imaging protocol can be repeated in that scan plane. For tracking the tip of flexible instruments, a more elaborate image acquisition and processing scheme should be designed. Because our method is implemented on a commercially available MRI system with standard hardware and does not require a special tracking device, combined with an integrated scan and display technique, it may provide a practical approach to the tracking problems.

References