Accelerated susceptibility-based positive contrast imaging of MR compatible metallic devices based on modified fast spin echo sequences

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Abstract
This study aims to develop an accelerated susceptibility-based positive contrast MR imaging method for visualizing MR compatible metallic devices. A modified fast spin echo sequence is used to accelerate data acquisition. Each readout gradient in the modified fast spin echo is slightly shifted by a short distance $T_{\text{shift}}$. Phase changes accumulated within $T_{\text{shift}}$ are then used to calculate the susceptibility map by using a kernel deconvolution algorithm with a regularized $\ell_1$ minimization. To evaluate the proposed fast spin echo method, three phantom experiments were conducted and compared to a
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spin echo based technique and the gold standard CT for visualizing biopsy needles and brachytherapy seeds. Compared to the spin echo based technique, the data sampling speed of the proposed method was faster by 2–4 times while still being able to accurately visualize and identify the location of the biopsy needle and brachytherapy seeds. These results were confirmed by CT images of the same devices. Results also demonstrated that the proposed fast spin echo method can achieve good visualization of the brachytherapy seeds in positive contrast and in different orientations. It is also capable of correctly differentiating brachytherapy seeds from other similar structures on conventional magnitude images.

Keywords: positive contrast, susceptibility mapping, metallic devices, magnetic resonance imaging

(Some figures may appear in colour only in the online journal)

1. Introduction

Magnetic resonance (MR) compatible metallic devices, such as brachytherapy seeds and biopsy needles, have been widely used in clinics. These metallic devices, typically composed of paramagnetic materials, e.g. titanium, have high magnetic susceptibility (~182 ppm for titanium) that often leads to susceptibility artifacts in conventional MR images (Ladd et al 1996, Glowinski et al 2000, Whitehead and Ji 2010). The susceptibility artifacts cover an area several times larger than each metallic device on the MR image, which limits the capability of conventional MR imaging in assessing these metallic devices (Moerland et al 1997, Ménard et al 2004). To address this issue, two new MR techniques, slice encoding for metal artifact correction (SEMAC) and multi acquisition variable resonance image combination (MAVRIC), have been introduced to minimize the metallic artifacts near a metallic implant device (Koch et al 2009, Lu et al 2009). These two techniques are developed to mitigate or correct the susceptibility artifacts in the vicinity of the large metallic devices, not for visualizing the devices themselves (Seevinck et al 2011, Lee et al 2013). Thus, x-ray computed tomography (CT) remains the standard imaging modality to assess these metallic devices (Nag et al 2000, De Brabandere et al 2006, De Brabandere et al 2013). However, due to the exposure to ionizing radiation and the limited soft tissue contrast of CT (Roach et al 1996, Fazel et al 2009, Griffey and Sodickson 2009), a reliable MR technique for visualizing such devices remains of great interest to the imaging community.

In recent years, several positive contrast MR techniques have been developed for visualizing metallic devices (Varma et al 2011). One category of these methods uses special novel pulse sequences to collect data for positive contrast MR imaging. In 2006 Mani et al achieved positive contrast using GRE acquisition for superparamagnetic particles (GRASP) by applying a decreased rephasing gradient to compensate for the slice-selection area. However, this sequence needs to be adjusted to the slice thickness to have optimal signal gain. It is also sensitive to magnetic field inhomogeneity (Mani et al 2006a, 2006b). In 2007 a novel sequence, inversion recovery with on-resonant water suppression (IRON), was developed to visualize off-resonance tissue areas based on a spectrally-selective on-resonant saturation pre-pulse (Stuber et al 2007). Recently, a new sequence, a center-out radial sampling with off-resonance reception (co-RASOR) method, was proposed to provide positive contrast MR for small paramagnetic objects by shifting the signal toward the object’s center using off-resonance reception (Seevinck et al 2011, De Leeuw et al 2013). Although the IRON and
co-RASOR techniques can create positive contrast MR images without further data processing, they require significant alternation to optimize the pulse sequences for different devices, which limit their practical applications (De Leeuw et al. 2013). In another approach, conventional pulse sequences are used for data acquisition, and the positive contrast MR images are produced using a post-processing algorithm. For example, the susceptibility gradient mapping (SGM) (Dahnke et al. 2008, Varma et al. 2011) applies a short-term Fourier transform (STFT) or a truncated filter to a localized region. The result reflects the echo shift in k-space due to the local susceptibility and produces positive contrast images. However, SGM is highly variable as it can depend on the phase of field perturbation due to spreading of the metallic devices, and the extension of the positive contrast regions created by SGM can lead to potential localization errors or misinterpretations of the metallic devices (Dahnke et al. 2008, Dong et al. 2014b).

Another typical technique of this category is the quantitative susceptibility mapping (QSM) method, which has been extensively studied in recent years. Although QSM has been used for the visualization of super paramagnetic iron oxide (SPIO) or naturally occurring iron-induced susceptibility (de Rochefort et al. 2008, de Rochefort et al. 2010, Acosta-Cabronero et al. 2013), QSM cannot be straightforwardly applied to create positive contrast image for metallic devices because conventional QSM techniques are based on gradient recalled echo (GRE) sequences, and the metallic devices are of high magnetic susceptibility, leading to severe signal loss and rapid phase wrapping.

Recently, a novel susceptibility-based positive contrast MR technique that combines sequence development and post-processing methods was introduced for the visualization of brachytherapy seeds (Dong et al. 2014a, 2014b). This technique uses an equivalent short echo time (TE) to acquire MR data based on a modified spin echo (SE) sequence, which is less sensitive to signal loss and rapid phase wrapping. Phase differences induced by susceptibility are used to derive the magnetic field offset. Susceptibility of the devices and tissues at their exact locations are then inferred using a kernel deconvolution algorithm with a regularized ℓ1 minimization (Dong et al. 2014b). Therefore, the susceptibility-based technique can highlight the devices due to their high susceptibility values. However, the SE sequence used in this technique is rather slow which makes it difficult to acquire high-resolution images in a clinically practical time.

To address this issue, this study assesses an accelerated method based on a modified FSE (fast spin echo) sequence for a susceptibility-based positive contrast MR technique. To analyze and validate the proposed method, phantoms with embedded titanium biopsy needle and/or brachytherapy seeds were imaged. The images were then compared to the original susceptibility-based positive contrast MR technique (referred to as the SE-based technique hereafter) and standard CT images.

2. Theory

2.1. Summary of the SE-based technique

Based on the QSM theory, the SE-based technique uses a constrained method to reconstruct the positive contrast MR image for the visualization of brachytherapy seeds. If the local field variation \( \Delta B(r) \) is known, then the susceptibility mapping \( \chi(r) \), i.e. the positive contrast MR image, can be reconstructed by solving a ℓ1 norm optimization problem in equation (1). This is because the susceptibility mapping \( \chi(r) \) is expected to be sparsely represented in the spatial image gradient domain.

\[
\chi = \arg \min_{\chi} \|W(D\chi - \Delta B)\|_2^2 + \lambda \|MG\chi\|_1 \quad \text{with} \quad G = [G_x, G_y, G_z]
\]
where $D = F \cdot F(C) \cdot F^{-1}$ and $F(C) = 1/3 - k^2/j^2$. $W$ is a weighting matrix that normalizes the magnitude image. $G$ is the gradient operator to obtain susceptibility gradients in all three dimensions. $M$ is the mask matrix to avoid the over smoothing of small structures of the susceptible objects. $\lambda$ is the regularization parameter that trades off data consistency and spatial smoothness (Dong et al 2014b). A nonlinear conjugate gradient algorithm can be used to obtain the final susceptibility mapping, and thus the positive contrast images of the metallic devices can be obtained (Lustig et al 2007).

A modified SE sequence was proposed to measure the local field variation $\Delta B(r)$. Compared to the conventional SE sequence, the 180° refocusing pulse is shifted toward the 90° excitation pulse with a short time $T_{\text{shift}}$. As an object with magnetic susceptibility can create a local inhomogeneous magnetic field, we just assume that a constant susceptibility gradient existed during the entire imaging period at each location in order to simplify the calculation. A higher-order susceptibility gradients induced by magnetic susceptibility variations are not considered in this study. Then the local field variation $\Delta B(r)$ can be calculated by equation (2), and the positive contrast image can be reconstructed by equation (1).

$$\Delta B(r) = \frac{G_{\text{sup}}}{B_0} = \frac{\Delta \Phi}{2\gamma B_0 T_{\text{shift}}} = \frac{\Phi(T_{\text{shift}}) - \Phi(0)}{2\gamma B_0 T_{\text{shift}}}$$

where $\Delta \Phi$ is the phase difference between the image phases (i.e. $\Phi(T_{\text{shift}})$ and $\Phi(0)$) obtained by the modified SE sequence with identical scan parameters except for the $T_{\text{shift}}$ (Dong et al 2014b). This method is time-consuming because only one phase line can be acquired in a repetition time (TR). For instance, if the TR is set to 2000 ms, it takes approximately 8.5 min to acquire an image with a data matrix of $256 \times 256$.

### 2.2. The proposed modified FSE sequence for accelerating positive contrast MR imaging

The modified FSE sequence is a direct extension of the SE-based technique. As shown in the diagram in figure 1, multiple echoes are acquired within one TR in FSE to improve data acquisition efficiency. Each readout of the modified FSE sequence is shifted by a short distance $T_{\text{shift}}$ (0.2–0.7 ms) to avoid phase wrapping due to the high susceptibility of the metallic devices. The local field variation $\Delta B(r)$ due to the susceptibility can be calculated from two datasets acquired by the modified FSE, with and without echo shifting, and thus the positive contrast image can be reconstructed by equation (1), which is similar to the SE-based technique.

The mathematical description of calculating the local field variation $\Delta B(r)$ is based on the theory that Fourier encoding is used for MR data acquisition. If other sources of phase distortions (e.g. coil sensitivity or eddy currents) are ignored, the measured signal of the FSE sequence can be written as:

$$S_0(t) = \int \rho(r)e^{i\Phi_{\text{sup}}(r,t)}e^{-i\Phi_{\text{img}}(r,t)}dr$$

where $\rho(r)$ is the spin density image, $\Phi_{\text{sup}}(r,t)$ is the concomitant phase generated by the susceptibility gradient $G_{\text{sup}}$, and $\Phi_{\text{img}}(r,t)$ is the Fourier encoding phase generated by imaging gradient. The $\Phi_{\text{sup}}(r,t)$ and $\Phi_{\text{img}}(r,t)$ can be expressed as:

$$\Phi_{\text{sup}}(r,t) = \gamma r \int_0^t G_{\text{sup}}(\tau)d\tau$$

$$\Phi_{\text{img}}(r,t) = \gamma \left( -\int_0^{T_{\text{shift}}} G_{\text{img}}(\tau)d\tau + \int_0^t G_{\text{img}}(\tau)d\tau \right)$$
If the modified FSE sequence is played out without echo shifting, then the total phase \( \Phi_{rt,0} \) during the data acquisition period can be described as

\[
\Phi_{rt,0} = \gamma \left( -\int_0^{T_{acq}/2} G_{\text{img}}(t) \, dt + \int_0^t G_{\text{img}}(t) \, dt + \int_0^t G_{\text{susp}}(t) \, dt \right) \\
\text{subject to } |t - nTE| \leq \frac{T_{acq}}{2}
\]

(6)

where \( n = 1, 2, 3, \ldots, L \), \( L \) being the echo train length in one TR of the FSE sequence. If the modified FSE sequence is played out with the echo shift \( T_{\text{shift}} \), then the phase \( \Phi_{rt,1} \) during the data acquisition period can be described as equation (7). Please note that the shifted time \( T_{\text{shift}} \) is defined as positive if the gradient is shifted away from the 90° excitation pulse.

\[
\Phi_{rt,1} = \gamma \left( -\int_0^{T_{acq}/2} G_{\text{img}}(t) \, dt + \int_0^t G_{\text{img}}(t) \, dt + \int_0^t G_{\text{susp}}(t) \, dt \right) \\
\text{subject to } |t - (nTE + T_{\text{shift}})| \leq \frac{T_{acq}}{2}
\]

(7)

If a constant susceptibility gradient was also existed during the entire imaging period at each location without considering the high order gradients, it is clear that the phase difference between \( \Phi_{rt,1} \) and \( \Phi_{rt,0} \) does not depend on the echo numbers \( n \) since

\[
\Phi_{rt,1} - \Phi_{rt,0} = \gamma \tau G_{\text{susp}} T_{\text{shift}}
\]

(8)

Thus, the signal \( S(t) \) acquired without echo shifts can be expressed as

\[
S(t) = \int \rho(r) e^{i\gamma G_{\text{susp}} T_{\text{shift}} \phi_{rt,0} - i\phi_{rt,1}} \, dr
\]

(9)

Applying the inverse Fourier transform on equations (3) and (9), we obtain that

\[
\rho_1 = \rho_0 e^{i\gamma G_{\text{susp}} T_{\text{shift}}}
\]

(10)
where $\rho_1$ and $\rho_0$ are the images that are acquired by the modified FSE sequence with and without echoes shifted, respectively. Ignoring the data noise, the phase difference $\Delta \Phi$ between $\rho_1$ and $\rho_0$ is $\gamma r G^{\text{nop}T_{\text{shift}}}$. Therefore the local field variation $\Delta B(r)$ can be calculated by equation (10)

$$
\Delta B(r) = \frac{G^{\text{nop}T}}{B_0} = \frac{\Delta \Phi}{\gamma B_0 T_{\text{shift}}}
$$

(11)

Hence, the modified FSE sequence provides reliable field information needed for positive contrast imaging although multiple echoes are acquired in a TR. Please note that the $T_{\text{shift}}$ in the proposed method is two times the $T_{\text{shift}}$ in the SE-base technique.

3. Materials and methods

3.1. Data acquisition

To test the proposed method, three experiments were carried out on a 3 T whole-body MRI scanner (SIEMENS Tim Trio, Germany) with an eight-channel phased array coil. In all three experiments, the data was acquired using the modified FSE sequence with and without $T_{\text{shift}}$. In these studies, $L = 7$ (turbo factor) was used for the tradeoff between the signal-to-noise ratio (SNR) and sampling speed because $L$ is limited by the T2 value of the object being imaged.

In the first experiment, the object being imaged is a biopsy needle inserted into a water phantom. The biopsy needle is made of titanium and is 160.0 mm long and 2.0 mm in diameter. The phantom is constructed by filling a sealed plastic container with distilled water doped with 1.0 g l$^{-1}$ copper sulfate solution. The needle was positioned parallel to the magnetic field. Thirty-seven axial slices were acquired. Scan parameters were: FOV = $80 \times 80 \times 80$ mm$^3$, matrix size = $128 \times 128 \times 37$, TR = 2000 ms, TE = 18 ms, in-plane resolution = 0.625 $\times$ 0.625 mm$^2$, slice thickness = 1.5 mm, slice gap = 25%, bandwidth = 134 Hz/Pixel, $T_{\text{shift}}$ = 0.6 ms (with echo shift) and 0 ms (without echo shift). The total acquisition time was 4.0 min.

The second phantom experiment was designed to test the proposed method for imaging brachytherapy seeds. The phantom was constructed by placing five dummy brachytherapy seeds (Seeds Biological Pharmacy Ltd, Tianjin, China) with different spacing ($L_1 = 5$ mm, $L_2 = 15$ mm and $L_3 = 10$ mm) in a porcine tissue. The seed has a titanium capsule (4.5 mm in length and 0.8 mm in diameter) with hemispherical shaped ends. Inside the capsule, a silver rod (3.0 mm in length and 0.5 mm in diameter) is impregnated with iodine-125, as schematically demonstrated in figure 2(a). The susceptibility of the materials in the seed and the tissue are listed in figure 2(b). A plastic stick was inserted to the tissue to simulate a cavity, and a bamboo toothpick was inserted to simulate a capillary in human tissue. In addition, a small animal bone was put in the phantom to simulate a human bone. The seeds in this experiment were placed along the direction perpendicular to the $B_0$ field. Image slices were acquired in the coronal plane which is perpendicular to the longitudinal direction of the seed. The scan parameters are: FOV = $120 \times 120 \times 15$ mm$^3$, matrix size = $192 \times 192 \times 10$, TR = 2000 ms, TE = 18 ms, in-plane resolution = 0.625 $\times$ 0.625 mm$^2$, slice thickness = 1.5 mm, no slice gap, bandwidth = 134 Hz/Pixel. $T_{\text{shift}}$ = 0.6 ms (with echo shift) and 0 ms (without echo shift). The total acquisition time was 3.92 min.

Furthermore, a third phantom MRI experiment was carried out to investigate the feasibility of the proposed method in imaging the seeds placed at different orientations, which is a more realistic clinical situation since seeds can migrate and shift. Ten brachytherapy seeds were randomly placed into a gelatin phantom doped with 1.0 g l$^{-1}$ copper sulfate solution. The
orientations of ten seeds were random relative to $B_0$, presenting a challenging scenario for the proposed method as the magnetic field induced by the seeds are orientation dependent. Most severe seed migrations and shifting in clinical applications. The data were acquired using the same scan parameters as those used in the second MRI experiment.

For comparison, the first two phantoms were also scanned using the SE-based technique on the same 3 T scanner and a cone-beam dental CT scanner (NewTom, Manufacturer’s Model, VGI, Italy). All scan parameters for the SE-based technique were the same as those used in the modified FSE except for the turbo factor ($L = 1$). The acquisition time was 8.6 min for the first MRI experiment and 13 min for the second MRI experiment. The scan parameters of CT imaging are: tube voltage $= 110$ kV, x-ray tube current $= 1$ mA, exposure time $= 15$ s, 3D scan with a spatial resolution of $0.24 \times 0.24 \times 0.15$ mm$^3$. The total CT acquisition time was 18 s.

3.2. Performance evaluation

Multiple regularization parameters were tested and the corresponding reconstructions were visually compared to choose the optimal parameter. After obtaining the positive contrast images, a measurement called the half-intensity region (Dong et al 2014b) was calculated for the results from the first experiment to evaluate the effect of positive contrast imaging. The half-intensity region was defined as the number of pixels that differ from the background tissue by more than a threshold, which was defined as $(I_{\text{max}} - I_{\text{min}})/2$, where $I_{\text{max}}$ and $I_{\text{min}}$ are the maximum and minimum intensities of the pixel in the area respectively. The half-intensity regions in the ordinary negative-contrast magnitude images and those in the susceptibility-based positive contrast images were compared. Meanwhile, the reduction of the half-intensity region indicates an improved visualization of the metallic devices, and thus more accurate localization.

To compare the imaging results of the MR techniques and CT, all reconstructed images were loaded onto a workstation (SIEMENS Tim Trio, Germany) for further imaging analysis. The maximum intensity projection (MIP) of each image set was reconstructed for visual comparison. In addition, for the second experiment, 2D images were retrospectively reconstructed from the images obtained by CT to match the location and slice thickness of the images obtained using MR techniques. The distances $L_1$, $L_2$, and $L_3$, between the seeds were then measured from a slice that was perpendicular to most of the seeds. All image reconstruction and analysis were performed offline in the MATLAB R2011b (Math Works, Natick, MA, USA) environment.

| Material      | Susceptibility(ppm) |
|---------------|----------------------|
| Titanium      | 182                  |
| Silver        | -24                  |
| Air           | 0.36                 |
| Capillary     | 0.38                 |
| Bone          | -2.93                |
| Water         | -9.05                |
| Human tissues | -11--7.0             |
4. Results

Figures 3 and 4 show the representative results of positive-contrast MRI of the biopsy needle. As seen from the conventional magnitude images, the presence of the biopsy needle induced severe distortions, which manifest as the large dark spots near the position of the needle on the magnitude image (figures 3(b) and (f)). The dark spot occupies approximately 45 pixels or 17.6 mm$^2$, which is much larger than the biopsy needle itself. The field map (i.e. the local field variation) (figure 3(e)) is calculated from the phase images (figures 3(c) and (d)) using the proposed technique, which is also similar to the field map from the previously proposed SE-based technique (figure 3(i)). This result demonstrated that the proposed FSE method can consistently acquire the field map information at high turbo-factors.

The positive contrast MR images of the biopsy needle reconstructed by the proposed method are shown in figure 4(a). The number of pixels occupied by the needle are dramatically reduced from 45 to 8 pixels on the axial positive contrast image (figure 3(b) versus figure 4(a)), which is close to the real size of the needle (8 pixels at a plane resolution of 0.625 × 0.625 mm$^2$).

Table 1 shows the positive contrast imaging results with the proposed method and with that of the SE-based technique. As shown, even though the SNR of the proposed method is slightly lower (85.2 versus 92.7). The maximum intensity projection (MIP) reconstructions from the images obtained by the proposed method, the SE-based technique, and the CT provide similar renderings of the needle (figures 4(d)–(f)). This similarity demonstrated that both the proposed method and the SE-based technique can provide good visualization of the needle and precise needle localization. However, the proposed method is 2–4 times faster.

Representative results of the second experiment are shown in figures 5 and 6. Similar to the results from the first experiment, on the conventional magnitude images, a dark spot of up to 20 pixels (8 mm$^2$) is shown at each brachytherapy location due to its high magnetic susceptibility. The spot is much bigger than the real size of the seed. In contrast, the images of the plastic stick, bamboo toothpick, and bone in the magnitude images are close to their real sizes. In addition, it is hard to differentiate the seeds from the plastic stick, bamboo toothpick, and bone on the magnitude and the phase images (see figures 5(b)–(d)). In this experiment, the seeds are perpendicular to the $B_0$ field, resulting in a spindle field distribution around the

![Figure 3. Representative results of an axial slice from the biopsy needle phantom.](image)
seeds (Whitehead and Ji 2010), which can be seen from the field map (figure 5(e)). Although obvious differences between the seeds and other structures can be seen in the field map, the locations of the seeds cannot be precisely defined. In addition, as porcine tissue is less homogeneous than the water/gelatin phantom, the field map is much more inhomogeneous and noisier. This increased noise represents a more challenging scenario to distinguish the seeds from the tissues.

Figure 6 shows that the proposed method can successfully generate the positive contrast MR images of the brachytherapy seeds. The number of pixels occupied by each seed are reduced from around 20 pixels (~8 mm$^2$) to 3 pixels (figure 5(b) versus figure 6(a)), which is close to the physical size of the seed. Figure 6 also shows that the positive contrast images from the proposed method (figure 6(a)) are comparable to those from the SE-based technique (figure 6(b)). In addition, compared to the standard CT images (figure 6(c)), the contrast

| Table 1. Half-intensity regions and reduction rates of the proposed method and the SE-based technique$^a$. |
|---------------------------------------------------------------|
| On magnitude images | On positive contrast images | Reduction rate | SNR | Scan time |
|---------------------|-----------------------------|----------------|-----|-----------|
| The proposed method | 45 ± 2.5 (44–46) | 8.3 ± 1.4 (8–9) | 79%–82% | 85.5 | 4.0 min |
| The SE-based technique | 43 ± 2.8 (42–45) | 5.7 ± 1.5 (5–7) | 83%–89% | 92.7 | 8.6 min |

$^a$ Values are given in mean ± standard deviation (inter-quartile range). 37 slices included in the analysis of the two techniques.
Figure 5. Representative results of the tissue phantom experiment. (a) A photo of the tissue phantom: the upper red arrow indicates the position of the bamboo toothpick, the middle blue arrow indicates the position of the plastic stick, and the lower green arrow indicates the animal bone. In addition, the locations and the spacing between the seeds are also labelled with $L_1 = 5\, \text{mm}$, $L_2 = 15\, \text{mm}$ and $L_3 = 10\, \text{mm}$; (b)–(e) are images generated by the proposed method, and (f)–(i) are images generated by the existing SE-based technique. Magnitude images ((b) and (f)), phase images without $T_{\text{shift}}$ ((c) and (g)), phase images with $T_{\text{shift}} = 0.6\, \text{ms}$ ((d) and (h)), and calculated field maps ((e) and (i)) are shown.

Figure 6. Representative positive contrast images of the brachytherapy seeds. (a)–(c) The positive contrast images generated by the proposed method, the SE-based technique, and CT, respectively. (d)–(f) MIP reconstructions from the images obtained by the proposed method, the SE-based technique, and CT. The brachytherapy seeds for the proposed method and SE-based technique were clearly distinguished from the bamboo toothpick (red), plastic stick (blue), and bone (green), which were hard to differentiate by magnitude.
between the seeds and tissue is visually lower in the positive-contrast MR images. However, the MIPs reconstructed from the positive contrast MR images demonstrated similar renderings of the seeds (figures 6(d)–(f)), providing good visualization of seeds. It is noted that on the images reconstructed by the proposed method (figure 6(d)) and by the SE-based techniques (figure 6(e)), the brachytherapy seeds were clearly distinguished from the bamboo toothpick (red), plastic stick (blue), and bone (green), which were hard to differentiate on magnitude images. Note that the bone shows as a high-contrast feature on the CT image due to the bone’s high x-ray attenuation coefficient, which makes it appear similar to the seeds. This is not the case on the positive-contrast images because the bone has a smaller magnetic susceptibility. The images with the proposed method also show less metallic artifacts as compared with the CT image.

Table 2 shows the distance between the two seeds measured from the reconstructed images with the proposed method, the SE-based technique, and the CT, as well as a comparison between the SNR and scan time. Clearly, the distances between two seeds measured on the positive contrast MR images are very close to the real values and the gold standard CT. The small difference that exists between the measured distances can be attributed to the partial volume effect, distortions between MRI and CT scans, and measurement errors. Results in this experiment demonstrate that the proposed method can accurately locate the seeds at distances as close as 5.0 mm, which is important for clinical applications.

Figure 7 shows the representative results from the third experiment where the brachytherapy seeds are placed in random directions. As shown, the proposed method accurately depicted the seeds in positive contrast with much improved definition as compared with the magnitude images. This set of results demonstrates the robustness of the proposed method for the visualization of the brachytherapy seeds.

5. Discussion

The visualization and localization of metallic devices are very important in clinical applications, especially in surgical procedures. However, due to the susceptibility of the metallic devices, artifacts with dark spots generally appear in conventional MR images. The dark spots are difficult to differentiate from natural cavities, bones, or other low signal areas (figures 5(b) and (f)). The proposed method addresses this problem by producing images of the metallic devices in positive contrast based on their high susceptibility. This technique works because metallic devices exhibit much higher magnetic susceptibility than typical biological tissues. Therefore, its applications are limited to devices with relatively higher magnetic susceptibility.

Generally, the orientation of the metallic devices with respect to the main magnetic field can affect the appearance of positive contrast images because these devices induce greater field perturbation when the angle with $B_0$ increases (Frahm et al 1996, Whitehead and Ji 2010). In the first and second experiments, the biopsy needle and brachytherapy seeds were
positioned perpendicular to $B_0$. Thus, sufficient phase changes induced by local field around the devices can be acquired using a short $T_{\text{shift}}$. The susceptibility was then calculated, based on which the positive contrast images of metallic devices are created. Although the perturbation of the field induced by the devices is small when the main axis of the devices are parallel to $B_0$ (Whitehead and Ji 2010), the proposed method can still achieved good visualization of the seed in positive contrast. In fact, results in figure 7(d) show that the proposed method worked even when the small seeds were placed with random orientations. This demonstrates the robustness of the proposed FSE method for the visualization of the brachytherapy seeds.

Compared to the standard SE-based technique, the proposed modified FSE sequence can acquire multiple echoes in a single TR. Although these echoes are encoded with different spatial frequency (in $k$ space) and have different echo times, the phase change between the scans with and without echo shifting is only induced by the susceptibility gradient in a period of $T_{\text{shift}}$. Therefore, the local field variation $\Delta B$ can be calculated from FSE regardless of the turbo factor. A turbo factor of 7 was used in our studies, which allows for a tradeoff between the SNR and sampling speed because the value of $L$ is limited by the T2 value of the object being imaged and because low SNR can affect the accuracy of $\Delta B$. Thus, the field map obtained by the proposed method is similar to that of the SE-based technique when an appropriate turbo factor is used, resulting in comparable positive contrast images of MR compatible metallic devices. It should be noted that a factor of $L$ improvement in data sampling speed cannot be obtained by the modified FSE sequence over the SE-based technique in multi-slice imaging. This is because a slice interleaved sampling scheme was used in practice, which leads to the sampling efficiency of FSE approaching that of SE as the number of acquired slices increases. Thus, 2–4 times of data sampling efficiency can be improved by the proposed FSE method, compared to the SE-based technique (tables 1 and 2). Besides, FSE has higher specific absorption rate (SAR) than the SE sequence. This may restrict parameter settings of the modified FSE sequence to maintain the SAR under the limit for patient safety consideration. Nevertheless, there are several approaches that have been developed to reduce the SAR for FSE. For example, parallel acquisition imaging and parallel excitations are believed to be effective at avoiding the RF heating issue in FSE sequence (Liu et al 2008, Song et al 2013). Besides, replacing the standard excitation and refocusing pulses with variable rate selective excitation (VERSE) pulses can also reduce the peak power in the SE and FSE pulse sequences (Zur et al 2003).

$T_{\text{shift}}$ is a critical parameter to the proposed technique. An optimal value of $T_{\text{shift}}$ should make the phase difference between the scans with/without $T_{\text{shift}}$ as large as possible and avoid phase wrapping due to the high susceptibility of the metallic devices. Generally, the range of $T_{\text{shift}}$ can be determined before MR experiment if the magnetic susceptibility of the metallic
device is known. This is because a resonant frequency offset map of the metallic device can be calculated according to its magnetic susceptibility, and then the range of $T_{\text{shift}}$ can be predicted (Whitehead and Ji 2010). In this work, $T_{\text{shift}}$ of 0.6 ms was used for both biopsy needle and brachytherapy seed experiments. This is because the biopsy needle and brachytherapy seeds are both made of the titanium that has magnetic susceptibility of ~182 ppm. The range of $T_{\text{shift}}$ between 0.2–0.7 ms can avoid severe phase wrapping according to the calculation. Thus, an optimal value of 0.6 ms was determined experimentally according to the output positive contrast image quality.

There are several other limitations to this study. First, interleaved multi-slice 2D FSE scans were used for data acquisition with a slice gap set to zero in the brachytherapy seed experiment due to the small size of the seeds. A slice gap of zero could lead to crossover between adjacent slices, which may cause artifact or SNR loss (Frahm et al. 1996). However, according to the results, reliable positive contrast images still can be achieved with a relatively long TR (~2000 ms). Second, the image reconstruction is time consuming because the phase of the two different $T_{\text{shift}}$ need to be unwrapped before calculating the field map. Otherwise, the field map cannot correctly reflect the phase changes caused by the perturbation of the local field. In the experiments presented in this study, it took approximately 50 s to reconstruct 10 slices using MATLAB on a 6 core workstation. All image reconstruction and analysis were performed offline in the MATLAB R2011b (Math Works, Natick, MA, USA) environment without any compiled C elements or parallel processing. Algorithm optimization and using graphics processing unit (GPU) or parallel processing methods (Kressler et al. 2006, Fox et al. 1994) should help speed up the reconstruction. Third, similar to previous works (Reichenbach et al. 1997, Dahnke et al. 2008, Dong et al. 2014b), a constant susceptibility gradient of each location was assumed to simplify the calculation of the local field variation $\Delta B(r)$. If the susceptibility gradient is not constant, the calculation of $\Delta B(r)$ become much complex because the susceptibility gradient is hard to be modeled. Thus, we only focus on the analysis to constant susceptibility gradient only. We have assumed that only a linear gradient is presented for imaging. This hypothesis seems to be sufficient as the results indicated. Last but not least, a proper regularization parameter was manually tuned and empirically determined according to the output image quality. Although the parameter is rather robust to both biopsy needle and brachytherapy seeds imaging, the empirical choice of regularization parameter is still a barrier to clinical adoption. Determining the optimal regularization parameter can be addressed using the method presented in (Khare et al. 2012). Iterative algorithms for selecting the optimal regularization parameter may help to address this problem (Daubechies et al. 2004, Lustig et al. 2007). Alternatively, one could establish the optimal parameter using an image quality metric using fully sampled image as a reference (Hollingsworth et al. 2014, Mann et al. 2016). We will investigate these parameter optimization methods is our future work. Nevertheless, the preliminary feasibility and reliability of the proposed method was demonstrated using the phantom experiments only. In our future studies, in vivo experiments will be conducted to fully evaluate the proposed method for clinical applications.

6. Conclusions

An accelerated susceptibility-based positive contrast imaging technique based on a modified FSE sequence was developed. The phantom experiment results demonstrated that the proposed fast imaging method can accelerate data acquisition by a factor of 2–4. Just as with the existing positive contrast imaging method, the proposed method can not only provide positive contrast MRI of brachytherapy seeds and biopsy needles, but can also correctly differentiate
the devices from other structures. Comparisons with the CT images verified the performance of the proposed method for visualization of MR compatible metallic devices. The data acquisition acceleration offered by the proposed FSE-based method is expected to significantly facilitate the utilization of the susceptibility-based positive contrast MR imaging in clinical applications.

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