Dynamic MRI has become increasingly feasible over the past several years because of technological advances, especially the increased data acquisition speed allowed by higher-performance hardware systems. However, in a number of dynamic applications the currently achievable temporal resolution may still be unsatisfactory. Several approaches have been developed to further increase imaging speed. Multiple-echo data acquisition strategies, such as echo-planar imaging (EPI) (1), rapid acquisition with relaxation enhancement (RARE) (2), and gradient and spin-echo (GRASE) (3), increase imaging speed by acquiring multiple k-space lines after each excitation. Complementary approaches to enhance imaging speed include methods whereby images are reconstructed using only a reduced set of the k-space data. Examples of such methods include partial-Fourier imaging (4,5), keyhole techniques (6,7), MR fluoroscopy (8,9), feature-recognizing MRI (10), multiple-region MRI (11), reduced-encoding MRI with generalized-series reconstruction (RIGR) (12,13), and reduced field-of-view (FOV) methods (such as that described by Hu and Parrish (14)), unaliasing by Fourier-encoding the overlaps using the temporal dimension (UNFOLD) (15,16), and 2D spatially-selective RF excitation (17,18). The performance of such methods depends on how well the assumptions upon which they are based are justified in a particular application. The present work focuses on combining two of these approaches, UNFOLD and 2D spatially-selective RF excitation, which are believed to be highly compatible.

In dynamic imaging applications, such as cardiovascular MRI, interventional MRI, or, in some cases, functional MRI (fMRI), only a small portion of the FOV may be of clinical interest. In cardiac imaging, for example, the entire chest is typically within the FOV even though only the heart is of special interest. In MR temperature mapping to monitor procedures such as focused ultrasound ablation of tissue, only the small portion of the FOV in which heat is concentrated may be of special interest. In fMRI for presurgical planning, the region around a tumor may be the only part of the FOV that has real clinical value. With a reduced FOV, fewer k-space lines are needed to achieve the same spatial resolution, leading to faster imaging. However, simply reducing the imaging FOV is normally unacceptable because it typically results in aliasing artifacts (also called “wraparound” artifacts). A straightforward way to avoid such artifacts is to avoid exciting any signal outside the desired ROI, as can be done using 2D spatially-selective RF excitation pulses (17,18). Such pulses perform a spatial selection along the slice direction in addition to selecting a limited region in the phase-encode direction.

Along the phase-encoding direction, a selection profile with sharp edges is needed to completely avoid any aliasing while the full extent of the reduced FOV is used for imaging. Usually, such sharp profiles require RF pulses that are very long in duration. For example, Reiseberg et al. (19) reported the application of 2D RF excitation in EPI using a 26-ms pulse to achieve a reduced FOV in the phase-encode direction. Even with this relatively long RF pulse, edges for the 40-mm FOV extended an additional 10 mm. If one used a shorter, more practical RF pulse for 2D excitation, the corresponding excitation profile would become even wider. Such a compromised profile would result in a severe aliasing artifact from the excited material located outside the boundaries of the reduced FOV.

In a preliminary study, Zhao and Panych (20) combined a dynamic rFOV method similar to that reported by Hu and Parrish (14) with a short-duration 2D spatially-selective RF excitation. Typically, neither method alone (whether the 2D RF excitation or the rFOV method) was sufficient to satisfactorily eliminate the aliasing artifacts. The rFOV method eliminated aliasing from static signal components outside the reduced FOV, but not dynamic signal components. The RF excitation method significantly suppressed signal outside the reduced FOV, but left aliased signals near the boundaries. Although the two methods combined performed significantly better than ei-
ther did individually, this hybrid approach failed in cases in which dynamic material was found outside and near the reduced FOV boundaries. Furthermore, the reference image required by this approach sometimes proved difficult to obtain in practice.

On the other hand, the UNFOLD method (15,16) does not require a reference image, it can handle dynamic variations in the aliasing artifacts to be suppressed, and it is optimal in terms of SNR. In the current work, we used 2D spatially-selective RF excitations to reduce the excited FOV and speed up the acquisition process, and used UNFOLD to suppress any leftover aliasing artifact. We found the combination of 2D RF excitation and UNFOLD to be a very reliable hybrid fast imaging approach. The application of a 2D excitation pulse can reduce the FOV many fold, and the addition of UNFOLD serves to relax the requirement for a sharp RF excitation profile, thus allowing for an RF pulse of more practical duration. As reported here, this hybrid method was successfully implemented in both dynamic phantom and in vivo imaging. Artifact-free images with a fourfold improvement in temporal resolution were obtained that are suitable for MRI-based temperature mapping studies.

THEORY

2D RF Pulse Design

We used the small-tip-angle model developed by Pauly et al. (17) in this study for the design of our 2D selective RF pulse. An on-resonance RF field with envelope \( B_1(t) \) is applied in conjunction with a magnetic field gradient \( G(t) \) to generate an excited profile \( M_{x_0}(r) \). With \( W(k(t)) = B_1(t)/|\gamma G(t)| \) and \( k(t) = -\gamma \int T G(s)ds \), where \( T \) is the time at which the RF excitation pulse ends, it can be shown that (17):

\[
M_{x_0}(r) = i\gamma M_0 \int W(k) S(k)\exp(ir\cdot k)dk. \tag{1}
\]

where \( S(k) \) represents the sampling function in excitation \( k \)-space, and \( W(k) \) represents the spatial frequency weighting function, which is the Fourier transform of the desired spatial profile, \( w(r) \). The domain of integration extends over the central region in excitation \( k \)-space, where the nonzero (i.e., sampled) locations of \( S(k) \) are found.

To reduce pulse duration, it is advantageous to choose a \( W(k) \) that is as compact as possible in excitation \( k \)-space. Of course, the need for a compact \( W(k) \) conflicts with the need for a compact profile \( w(r) \), i.e., one that has sharp transitions between excited and non-excited regions. Because of the conflicting need for compactness in the reciprocal \( k \) and \( r \) spaces, a 2D Gaussian shape for the targeted spatial profile, \( w(r) \), appears to be a good compromise.

In the reduced-FOV application, the slice thickness along \( z \) will be much smaller (<1 cm) than the desired excited region along the phase-encoded direction \( y \) (>>1 cm). Thus, the sampling function \( S(k) \) must cover a larger extent along \( k_z \) than along \( k_y \). For this reason an echo-planar trajectory, as shown in Fig. 1, would be appropriate for \( S(k) \). Figure 2 shows the spatially-selective RF pulse and the gradient waveforms that can be used to generate a Gaussian-shaped 2D profile. The slice (\( G_z \)) and phase (\( G_y \)) gradients produce the echo-planar \( k \)-space trajectory \( S(k) \), while the RF pulse (of magnitude \( |B_1(t)| \)) and phase \( \varphi(t) = angle(B_1(t)) \) produces the spatial frequency weighting \( W(k) \).

The sampling function \( S(k) \) must be designed to not only give an RF pulse of practical duration (<5 ms), but to also ensure that the periodicity of the excitation in the \( y \) direction is greater than the expected object size in that direction. The number of \( k_y \) lines in the trajectory (i.e., the sampling density in the \( k_y \) direction) affects both the pulse duration and the excitation periodicity along \( y \). This number must be small enough to keep the pulse duration short, and at the same time be large enough to ensure that the distance between the excited lobes is greater than the object size.
Figure 3 shows an excitation profile in the y direction (obtained using a 2D RF pulse as described in the Materials and Methods section) along with a number of convenient parameters to describe the profile. Along the y direction the excitation periodicity is inversely proportional to the increment between k lines in excitation k-space, and we define the distance between excited lobes as Dlobe. The distance at 5% and 2% of the maximum signal amplitude is defined as D5Gap and D2Gap, respectively. For an object larger than D2Gap, wraparound artifact from adjacent lobes in the excitation profile will result if the reduced FOV is set based on the width of the primary lobe of the excitation profile.

The profile widths at 90%, 50%, 5%, and 2% of the peak amplitude are defined in Fig. 3 as D90, D50, D5, and D2, respectively. The reduced FOV can be set anywhere between D90 (which would give significant aliasing within the reduced FOV) and D2 (which would give only a small amount of artifacts from outside the reduced FOV). Using the 2D RF excitation without UNFOLD, the minimum reduced-FOV size that can be reasonably used is about equal to D2, and the whole object should be less than about D2Gap in size. Accordingly, the maximum acceleration factor achievable with this pulse is D2Gap/D2 (or about 2.5 for the profile shown). For greater acceleration factors without significant wraparound artifacts, we used the UNFOLD method in combination with 2D RF excitation. The addition of UNFOLD (described in the next section) allows the acquired FOV to be further reduced while the aliasing artifacts that would normally originate from the decaying tails of the excited profile are suppressed.

UNFOLD Combined With 2D RF Excitation

UNFOLD is a trick used to manipulate how an aliased signal behaves in time (15). By forcing an aliased signal to behave in an unusual way from time frame to time frame, UNFOLD can effectively label it so it can be easily identified and corrected as part of the image reconstruction. The k-space sampling function is shifted by a fraction f of a line, and this shift varies from one time frame of the dynamic acquisition to the next. (Note that this shifting of the sampling function occurs during the acquisition, not during the excitation.) The effect of the shift is to apply a modulation through time to any aliased signal:

\[
T_i(t) = \exp(i2\pi l \cdot f(t)),
\]

where l is the order of aliasing for a given layer of overlap (e.g., \(l = 0\) for nonaliased material, \(l = 1\) or \(-1\) for the first layer of overlap in the positive (or negative) direction along the phase-encoding axis, etc.). Preferably, f(t) is chosen to be a linear function of time, so the modulation is a linear phase ramp through time, which produces a simple shift in the temporal frequency domain (15). Equivalently, this sampling scheme has been described as a “sheared grid” in k-t space (21), where the sampled k-space locations are changed by a fixed increment from one time point to the next (i.e., f(t) proportional to time (15)). In practice, only the fractional part of f(t) is used, as shifts by an integer number of lines introduce phase shifts that are multiple of \(2\pi\), and have no effect on the modulation being applied to the aliased signal.

UNFOLD is able to displace aliased signals with respect to each other in the temporal frequency domain. In applications such as fMRI (15) and non gated cardiac imaging, the expected signal is periodic, and UNFOLD can lead to large acceleration factors (a factor as high as 8 has been reported for fMRI in (15)). This is because periodic signals may populate only a small number of frequency locations in the temporal bandwidth (DC, fundamental and harmonics), and displacing the aliased signals with respect to each other may allow many of these sparse signals to be interleaved, leading to a tight packing of information in the temporal frequency domain (15).

In the present work, we used UNFOLD to displace aliasing artifacts as far as possible from the desired signal, all the way to the Nyquist frequency, where they could be filtered out (16,22). These aliasing artifacts are caused by the 2D RF excitation profile being too smooth spatially, and they are forced here to reverse their phase from time point to time point (i.e., to “flicker” in time at the Nyquist frequency). This is done by selecting f(t) = t/2, where t is the time-frame number, giving \(T_i(t) = \exp(i\pi t)\) in Eq. [2] for \(|l| = 1\) and t being an integer. A filter with a full width at half maximum (FWHM) of about 10% of the full temporal frequency bandwidth and centered at the Nyquist frequency is used to remove most of the aliased material left over by the imperfect 2D RF excitations. Because there is only one region in the bandwidth where artifacts can be safely stored and disposed of, i.e., near the Nyquist frequency, UNFOLD cannot correct here for aliasing more severe than a twofold overlap (i.e., \(|l| \leq 1\) (15,16). In other words, the acquired FOV should not be any smaller than half of the width of the excited profile shown in Fig.
3. For example, to avoid having aliased material of intensity greater than 2% of its fully-excited magnitude, one would select an acquired FOV no smaller than D$_{50}$/2 (see Fig. 3). Clearly, a larger (or smaller) acquired FOV would lead to a more benign (or more severe) level of residual artifacts after the application of UNFOLD.

**MATERIALS AND METHODS**

All experiments were performed using a modified version of a gradient-echo sequence, implemented on a 1.5 T GE Signa scanner (General Electric Medical Systems, Milwaukee, WI) with echo-speed gradient set (40 mT/m maximum and a slew rate of 150 T/m/s). As described below, experiments were performed in a static water phantom, in a dynamic water phantom, in a piece of meat while heat was applied, as well as in vivo without applying heat.

Figure 2 shows the RF and the gradient waveforms designed to generate a Gaussian-shaped 2D profile. The slice ($G_z$) and phase ($G_y$) gradients were chosen to produce an echo-planar $k$-space trajectory (Fig. 1), while the RF produces the spatial frequency weighting. The waveform was further adjusted to make maximum use of the available slew rate of the system in order to minimize the RF pulse duration. We also minimized the pulse duration by applying a Gaussian weighting to the width of the $k_z$ coverage as a function of $k_y$ (Fig. 1). This $k_y$ modulation of the $k_z$ coverage resulted in a reduction of the RF pulse duration by more than 50%.

The RF pulse design features nine segments along the $k_y$ direction, with gradient pulses applied at the end of the excitation to rephase the magnetization. The RF and gradient waveforms were designed for a system with a maximum gradient strength of 4 mT/m and a maximum slew rate of 150 T/m/s. The RF pulse waveform consisted of 1054 points with 4 μs dwell time, for a total duration of 4.2 ms.

A large homogeneous doped water phantom was imaged using the body coil, and a large FOV was set so that the full RF excitation profile (including its periodic lobes) could be visualized. The excitation profile shown in Fig. 3 was obtained with the 2D selective excitation pulse described above.

**Dynamic Phantom Experiments**

Figure 4 shows an image and schematic of a phantom used to demonstrate the combination of RF excitation and UNFOLD ex vivo under controlled dynamic conditions. The phantom consists of a fixed cylinder and two moving syringes filled with doped water, one located at the side of the cylinder and the other on top of the cylinder. The two syringes were connected to a cord that was attached to a wheel such that when the wheel was turned manually, the syringe on the side of the cylinder moved up and down and the syringe on the top of the cylinder moved from side-to-side (Fig. 4b). One full cycle of the syringe movement (up-down-up or right-left-right) was timed to be equal to 64 TR periods.

An interleaved EPI pulse sequence was modified to include a 2D RF excitation pulse to image the phantom during movement of the syringes (flip angle = 20°, echo train length = 8, bandwidth = 125 kHz, TE/TR = 10/22 ms, FOV = 240 mm). The arrow in Fig. 4a shows the FWHM (D$_{50}$) of the 2D RF excitation. Dynamic series of 128 × 64 images were acquired both with and without 2D RF excitation. The FOV in the phase-encoding direction was approximately equal to twice D$_{50}$. A full data matrix of 128 × 64 was acquired every eight shots, giving a temporal resolution of approximately six images per second (eight images per movement cycle of the syringe). However, phase-encode ordering was arranged so that enough data to reconstruct reduced-FOV images (32 lines) were acquired once every four shots, giving a temporal resolution of close to 12 reduced-FOV images per second (16 images per movement cycle of the syringe).

**Pulse Sequence for Temperature Mapping**

We modified a dual gradient-echo imaging sequence, which has been used extensively for temperature mapping using the proton resonant shift method, to create a hybrid RF-UNFOLD MR temperature mapping sequence. The modification of the sequence consisted of an RF modification and an UNFOLD
modification. The RF modification involved replacing the slice-selective RF pulse with a 2D RF pulse as described above. The UNFOLD modification involved shifting the acquisition $k$-space trajectory by $\Delta k/2$ on every second acquisition in the dynamic imaging series, where $\Delta k$ is the increment between acquisition $k$-space lines.

Abdominal Imaging Experiments

We tested the modified imaging sequence described in the previous section in vivo by acquiring several dynamic series of abdominal images from a human subject. Heating was not performed in the in vivo imaging experiments. The goal of the first set of experiments was simply to test the feasibility of using the new method in the presence of physiological motion to obtain images free of both motion and wraparound artifact. The abdominal imaging series were acquired 1) at full FOV without 2D RF excitation or UNFOLD, 2) at half FOV with 2D RF excitation but without UNFOLD, and 3) at quarter FOV with both 2D RF and with UNFOLD. The full FOV in both the phase- and frequency-encoding directions was 25 cm. The image acquisition matrix for full-FOV imaging was 192 in the frequency-encoding direction, and 128, 64, and 32 in the phase-encoding direction for full-, half-, and quarter-FOV imaging, respectively. Therefore, with a TR of 50 ms, the temporal resolution for full-, half-, and quarter-FOV imaging was 6.4 s, 3.2 s, and 1.6 s. The duration of the 2D RF pulse was 4.2 ms. The acquisition bandwidth was 16 kHz, and the TEs of the two gradient echoes were 5.6 ms and 20 ms. Images were acquired using an eight-channel flexible phased-array abdominal RF coil. The subject breathed freely during the acquisitions. Informed consent was obtained before each subject study under an approved IRB.

Temperature Mapping Experiments

In a separate set of experiments, the hybrid RF-UNFOLD sequence was applied for temperature mapping during heating of a phantom. In these experiments, a gel pack (Rapid Aid Reusable Cold & Hot Compress, Oakville, Canada) was heated in a microwave and then placed on top of a large piece of meat, which was then moved to the MRI scanner for imaging. Dynamic sets of images were acquired during three separate periods: 1) a period immediately after placement in the scanner, 2) a period beginning 5 min after the end of the first imaging period, and 3) a period beginning 15 min after the end of the second period. Each imaging period was approximately 100 s in duration. The imaging parameters were the same as those used in the abdominal imaging experiments with a quarter FOV (i.e., acquisition matrix = 192 × 32, TR = 50 ms), and thus the temporal resolution for the dynamic imaging was 1.6 s per scan.

The complex data were processed to produce maps of the change in phase over each dynamic series. Phase wraps in the data were removed voxel by voxel along the time dimension using the “unwrap” function in Matlab. The slope of phase change over the heating period was then computed from the unwrapped data by least-square fitting. Change in the water resonant frequency is temperature-dependent with a change of 0.01 PPM (6.4 Hz at 1.5T) per degree Celsius of temperature change. Thus, using this conversion factor and the TEs of the sequence, we obtained maps of temperature change directly from the phase difference maps.

RESULTS

Dynamic Phantom Experiments

Figure 5 shows reduced-FOV images of the dynamic phantom that were acquired using different combinations of 2D RF excitation and UNFOLD. Figure 5a was obtained with the conventional sequence alone, without 2D RF excitation or UNFOLD. Significant aliasing can be seen from both static (thin arrow in Fig. 5a) and dynamic (wider arrow in Fig. 5a) material outside the reduced FOV. By including 2D RF excitation, aliasing was significantly reduced (Fig. 5b), although artifacts are still visible near the boundaries of the reduced FOV image (arrows in Fig. 5b). Figure 5c was generated with UNFOLD alone, without 2D RF excitation. Again, there is significant reduction of aliasing compared to Fig. 5a, although some dynamic signal from outside the reduced FOV can still be seen (arrows in Fig. 5c). The signal from the moving syringes was sufficiently intense and dynamic that a non-negligible amount of artifact power “overflowed” beyond the limits of the 10%-wide region of the bandwidth dedicated for artifact suppression. When both 2D RF excitation and UNFOLD were used together, as shown in Fig. 5d, the aliasing artifact appeared to be completely eliminated.

FIG. 5. Reduced-FOV images were generated using an interleaved EPI sequence with phase encoding in the vertical direction. Each image corresponds to a single time frame of its corresponding dynamic image set: (a) using the conventional slice-selective method, both static (thin arrow) and dynamic (fat arrow) object wraparound is visible; (b) using 2D RF excitation only, both outside-of-FOV static and dynamic objects are suppressed, however, suppression is incomplete at the boundary of the FOV (see arrows); (c) using the UNFOLD method alone, outside-of-FOV dynamic object wraparound cannot be completely eliminated (see arrow); (d) a combination of both 2D RF excitation and UNFOLD essentially eliminates both static and dynamic object aliasing.
2D RF Excitation and UNFOLD Reduced-FOV MRI

Subject.) Reference boxes showing the limits of the half-

Figure 6 presents a more extreme example than Fig. 5, to show more clearly the potential advantages of combining the two approaches. In Fig. 6, the reduced FOV is further reduced by a factor of 2 (matrix = 128 x 16). The images in Fig. 6 were obtained using the same method as Fig. 5 (including the same 2D RF excitation) except that only every second k-space line was used for reconstruction. As expected, Fig. 6a (without UNFOLD and RF excitation), Fig. 6b (with RF excitation alone), and Fig. 6c (with UNFOLD alone) all show significant increases in the amount of artifacts as compared to their counterparts in Fig. 5. Note that the result with UNFOLD only (Fig. 6c) also has aliasing due to static signal contributions (thin arrow). This is because, as described in the Theory section, the UNFOLD method used here is not able to correct for an overlap factor of more than 2. Using both 2D RF excitation and UNFOLD gives a result (Fig. 6d) in which aliasing artifacts are essentially eliminated. (Note that the artifact along the vertical direction of the moving syringe, which is indicated with an arrow in Fig. 6d, is due to unresolved motion and is not an aliasing artifact.)

Abdominal Imaging Experiments

Figure 7 shows results of the abdominal imaging experiments for full-FOV imaging, half-FOV imaging with 2D RF excitation, and quarter-FOV imaging with RF-UNFOLD. In the time that one full-FOV image is acquired, two half-FOV or four quarter-FOV images can be acquired. To emphasize this difference in temporal resolution, variable numbers of time frames are shown for the three cases such that the total imaging time for each of the acquisitions in Fig. 7a–c is constant.

Figure 7a, which we refer to as the “full-FOV” image, was obtained with an FOV of 25 cm, the minimum FOV necessary to avoid serious wraparound artifact. (In fact, there is some wraparound of signal at the top of the image from a heating pad that was placed under the back of the subject.) Reference boxes showing the limits of the half-

Figure 7 shows results from the phantom heating experiments using the RF-UNFOLD acquisition. Results from the three heating periods are presented as temperature change contours, with separate contours A–K in steps of 0.4° (centigrade) per minute. The contours are overlaid on a gray-level image of the meat phantom. Figure 8a is from the first heating period and, as expected, shows the greatest rate of temperature change, up to 4°/min in the hottest region (bounded by the “K” contour). This region coincides with the location at which the heated gel pack was placed. Figure 8b and c show results from the second and third heating periods, respectively. These results also...
show the greatest rate of heating at the location at which the gel pack was placed. As expected, the rate of heating is much lower in the second and third periods than in the initial period. The lowest rate of temperature change is in the third heating period (Fig. 8c), rising only to 0.4°F/min in the region of maximum temperature rise (bounded by the “A” contour in the figure).

**DISCUSSION**

When a 2D RF excitation pulse is designed for reduced-FOV imaging, there is a tradeoff between spatial selectivity and pulse duration. Usually a small width and sharp edge excitation profile are desired, but they would require a dense sampling of a large region of excitation k-space. Such a sampling trajectory usually requires very long RF excitations. In practice, long-duration RF excitations are often not acceptable because they prolong the TE, resulting in loss in signal from the k-space. To prevent side lobe aliasing, the object size must be smaller than D5Gap or D2Gap (Fig. 3). To avoid aliasing from the vanishing tails of the excitation profile and a reduction in the distance between side lobes (due to periodicity of the excited profile). The spatial excitation profile in our abdominal imaging experiments, D5Gap was set around 37 cm, setting a maximum allowable size of 37 cm.

The slice thickness for a non-rectangular (Gaussian weighted) 2D RF excitation was approximately 6 cm, i.e., D5Gap/(D5/2) = 6.25. The maximum value of 6 is greater than the fourfold acceleration we actually obtained in our studies, because the temporal resolution of 1.6 s per image acquisition. The theoretical maximum acceleration factor, which preserves the SNR per unit time, is reduced by a factor of TR/(TR + Δ).

For the reasons discussed above, the principal goal of our RF excitation design was to keep the 2D RF excitation RF pulse as short as possible. The 2D RF pulse designed for this study had a total pulse length of only 4.2 ms. The reduced in RF pulse duration was due in part to the use of a non-rectangular (Gaussian weighted) 2D RF excitation pulse. The addition of UNFOLD allows the acquired FOV to be reduced from about D4 down to D2/2.

The spatial excitation profile in our abdominal imaging and temperature mapping experiments had a width at 5% of the maximum (D4) of 12.5 cm. We assume that all spins outside the 5% range are completely suppressed by this 2D RF excitation, then the minimum reduced FOV size must be no less than D4, or D2. A sharper profile (i.e., smaller D2 and D3) or a larger distance between lobes (i.e., large D2Gap and D5Gap) would come at the cost of a longer RF pulse. The addition of UNFOLD allows the acquired FOV to be reduced from about D4 down to D2/2.

The slice imaging time for full-FOV vs. reduced-FOV imaging is R(TR)/(TR + Δ). When Δ is insignificant with respect to TR, the acceleration factor due to reduced FOV imaging is equal to the FOV reduction factor. In the case in which Δ would be equal to TR, however, the acceleration factor becomes only R/2. Even in this case, if R is 4 or 8, there would still be a significant acceleration, although the SNR per unit time might suffer. Clearly, Δ should be kept as short as possible to derive the maximum benefit from reduced-FOV imaging. The theoretical maximum acceleration factor, which preserves the SNR per unit time, is reduced by a factor of TR/(TR + Δ).

FIG. 8. Temperature mapping results are shown for the heating of a meat phantom. RF-UNFOLD method was used for all acquisitions with a temporal resolution of 1.6 s per image acquisition. a: Temperature change contours for the first heating period overlaid on an image of the phantom. b: Temperature change contours for the second heating period, which began 5 min after the end of the first period. c: Temperature change contours for the third heating period, which began 15 min after the end of the second heating period.
As discussed above, the maximum acceleration factor of \( D_{\text{Gap}}/(D_y/2) \) is based on an assumption that the object size is equal to \( D_{\text{Gap}} \). It is also based on the assumption that the excited ROI is at the center of the object. If it is desired (as in the case of our abdominal imaging experiments) to excite a region off-center (see Fig. 7a), the profile shown in Fig. 3 must be shifted to the right or left with respect to the object in the FOV. The maximum allowable object size is then reduced from \( D_{\text{Gap}} \) by the amount that the excited region is shifted with respect to the center of the object. In our experiments, the offset was 5.5 cm from the center of the object. Thus, in this case the maximum allowable object size for a \( D_{\text{Gap}} \) of 37 cm is only 31.5 cm (i.e., 37–5.5). The maximum acceleration factor, assuming an object size of 31.5 cm, is now only 5 (i.e., 31.5/6.25).

In the present implementation of our 2D RF pulse, we use an echo-planar excitation \( k \)-space trajectory. One might expect spiral trajectories to be a better choice because spirals are generally efficient and spiral-in trajectories leave the magnetization refocused at the end of the pulse. In our application, however, a spiral trajectory was not favored because of the large asymmetry of \( k \)-space in the two dimensions (i.e., the span in \( k_x \) is much greater than in \( k_y \)). The sampling density in excitation \( k \)-space depends on the number of spirals, which must be a small number to ensure an RF pulse of short duration. In the \( k_x \) direction the sampling density would be similar to the density for an echo-planar trajectory. In the \( k_y \) direction, however, the density would be much lower and the period of the excitation in the \( z \) direction would be too short.

As a potential alternative to 2D RF excitation with UNFOLD for reduced-FOV imaging, one could select a rectangular box using standard slice selection along one dimension and a refocusing pulse along a second orthogonal direction. This approach has been used in the past in interventional MRI for MR-guided biopsy (23) and for cryotherapy monitored by MRI (24). The approach is inherently spin-echo based, however, and is somewhat limited for dynamic applications because the same volume cannot be excited rapidly without saturating the spins. The RF-UNFOLD method presented here can be implemented with either spin-echo or gradient-echo sequences.

In conclusion, the combination of 2D RF excitation and UNFOLD can allow significant reductions in the acquired FOV and thereby increase temporal resolution while keeping the RF pulse to a practical duration. In the proposed hybrid method, 2D spatially-selective RF excitation can be seen as a means of accelerating the acquisition by reducing the FOV, while UNFOLD can be seen as a way to relax the requirement for a sharp excitation profile and allow RF pulses to be shorter. With the combination of 2D excitation and the UNFOLD technique, it becomes possible to obtain essentially artifact-free, high-speed dynamic and even real-time images for clinical applications. To even further reduce the image FOV, it may be possible to form combinations with other methods, such as parallel imaging, that are compatible with RF excitation and UNFOLD. We are investigating optimal ways of combining parallel imaging with 2D RF excitation and UNFOLD, and our initial results show that, as could be expected, accelerations greater than those obtained by 2D RF excitation and UNFOLD, or by parallel imaging alone, are possible (25).

ACKNOWLEDGMENTS

The authors thank Dr. Christopher Hardy of GE Global Research for providing pulse sequence support. This work was supported in part by bioengineering research grants from the Whitaker Foundation to B.M. and L.P.P.

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