RESOLUTION ENHANCED T1-INSENSITIVE STEADY STATE IMAGING (RE-TOSSI)

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Abstract

Resolution Enhanced TOSSI is a new MRI pulse sequence for the generation of rapid T2 contrast with high spatial resolution. TOSSI provides T2 contrast by using non-equally spaced inversion pulses throughout a bSSFP acquisition. In RE-TOSSI, these energy and time intensive adiabatic inversion pulses and associated magnetization preparation are removed from TOSSI after acquisition of the data around the center of k-space. Magnetization evolution simulations demonstrate T2 contrast in TOSSI as well as reduction in the widening of the point spread function width (by up to a factor of 4) to a near ideal case for RE-TOSSI. Phantom experimentation is used to characterize and compare the contrast and spatial resolution properties of TOSSI, RE-TOSSI, bSSFP, HASTE and TSE and to optimize the fraction of k-space acquired using TOSSI. Comparison images in the abdomen and brain demonstrate similar contrast and improved spatial resolution in RE-TOSSI compared to TOSSI. Comparison bSSFP, HASTE and TSE images are provided. RE-TOSSI is capable of providing high spatial resolution T2-weighted images in 1 second or less per image.

Keywords

rapid imaging; pulse sequence; T2 contrast; spatial resolution

INTRODUCTION

MRI pulse sequences delivering T2-weighted contrast are used readily in clinical MRI. However, the traditional turbo spin echo (TSE) or RARE acquisition (1) can take several minutes to generate a T2-weighted image set. The single shot equivalent, or HASTE (2), produces rapid T2 contrast (in seconds) but suffers from severe spatial resolution degradation in the single-shot acquisition (3). A novel alternative method for producing rapid T2 contrast called T1-insensitive Steady State Imaging (TOSSI) has been developed by inserting nonuniformly spaced inversion pulses in a bSSFP imaging sequence in a manner which eliminates T1 contrast for all tissues (4,5). TOSSI has been used to generate multi-slice T2-weighted images in the head at 1-2 seconds per image (6). Single-shot TOSSI can also suffer from spatial resolution degradation due to magnetization decay during the acquisition, especially for tissues with short T2 values (or in acquisitions with a large
number of phase encoding lines). In this paper, a new pulse sequence called Resolution Enhanced TOSSI (RETOSSI) (7) is characterized, which can simultaneously improve the spatial resolution properties of TOSSI, reduce RF power deposition and further reduce image acquisition time while maintaining TOSSI image contrast.

BACKGROUND

Although TOSSI is composed of bSSFP imaging blocks, the magnetization does not approach a steady state value; instead the inversion pulses cause the magnetization to steadily decrease from an initial value toward zero. Despite the prolonged signal decay in TOSSI compared to HASTE (5), there is still magnetization decay in which leads to degradation of spatial resolution in the phase encoding direction. This decay becomes more problematic as the number of phase encoding lines increases and as the T2 value of the tissue decreases.

Hybrid imaging techniques have been proposed to improve spatial resolution and accelerate critical applications. The combined acquisition technique (CAT) method applies distinct imaging modules in succession to exploit the favorable qualities of the different techniques in different regions of k-space. CAT acquisitions have been described that either use different imaging sequences or the same sequence with different parameters (such as bandwidth, flip angle, etc.) (8-10). Another method called T2-TIDE, which exploits variable flip angles and the transient phase of bSSFP, provides initial HASTE or TSE-like T2 decay followed by transition to bSSFP signal levels (11). Using an appropriate K-space sampling strategy, T2-TIDE was shown to provide T2 contrast while reducing spatial resolution degradation compared to HASTE by gradually reducing the flip angle of bSSFP from 180 degrees to 60 degrees. In this work, it is hypothesized that removing the inversion pulses from the TOSSI acquisition during a second portion of the acquisition, coupled with an appropriate k-space sampling scheme, will allow T1 relaxation effects to be gradually reintroduced and used to improve spatial resolution properties of TOSSI while maintaining overall image contrast.

METHODS

Resolution Enhanced TOSSI (RE-TOSSI)

Figure 1 demonstrates the Resolution Enhanced TOSSI (RE-TOSSI) pulse sequence. Initially, balanced SSFP is performed between unevenly spaced adiabatic inversion pulses as in TOSSI (Fig 1a). The magnetization is stored and prepared with $\alpha/2$ at TR/2 pulses before and after each inversion pulse (12,13). Spoiling gradients are employed to dephase any remnant transverse magnetization after magnetization storage and inversion. After an initial TOSSI imaging block, the inversion pulses and associated magnetization storage and preparation pulses are removed (Fig 1b). A partial Fourier acquisition is used with the central region of k-space encoded first during the TOSSI portion of the acquisition and the outer region of k-space encoded second after the inversions are stopped (during the bSSFP evolution portion of the acquisition). As in TSE, the effective echo time (TEeff) is defined as the time that the central line in k-space is encoded. TEeff can be varied by the choice of partial Fourier factor as well as more fine control with added preparation pulses (PP).

Similar to the concept in T2-TIDE (11), such an acquisition will maintain T2 contrast since the data around the center of k-space is acquired using TOSSI, while spatial resolution will be improved due to the increased signal in the outer regions of k-space during the bSSFP portion of the acquisition. The resulting pulse sequence called Resolution Enhanced TOSSI (RE-TOSSI) is a new type of combined acquisition technique (CAT). The parameter $\lambda$
describes the fraction of the total k-space lines collected using TOSSI and has a range of 1 (all TOSSI) to 0 (all balanced SSFP).

Simulations

To test the RE-TOSSI hypothesis regarding signal decay, contrast and spatial resolution Bloch equation magnetization evolution simulations were performed in Matlab (The MathWorks Inc., Natick, MA). Six different tissues found in the abdomen (kidney cortex, spleen, subcutaneous fat, bone marrow, liver and paravertebral muscle were simulated. The T2 and T1 values used were taken from the literature and included T2/T1 (ms/ms) = 87/966, 79/1057, 58/343, 49/549, 46/586, 27/856 (14). TOSSI simulation parameters included FA = 60°, TR = 5.2 ms, T2opt (see below) = 90 ms, N_A = 2, 192 readouts. For fat, the simulation included 220 Hz off-resonance and the initial magnetization was set to zero, simulating an initial fat saturation pulse. The spacing of the inversion pulses was determined by calculating the longitudinal component of the magnetization for a specified T2 value (T2opt) (5):

\[ M_z = \cos(FA/2)^r \exp\left(-\sin^2(FA/2)^r t/T2\right) \]  

The ratio of time spent in the parallel state (T_p) to time in the antiparallel state (T_A) was then determined using (5):

\[ R = T_p / T_A = (1 + M_z) / (1 - M_z) \]  

For the TOSSI simulation, N_P was \{25,13,8,6,5,4,4,3,3,3,2,...,2\}. In order to determine the contrast mechanism in TOSSI, the simulations were repeated while setting all T1 values equal to the mean T1 value (726 ms). The simulation was then repeated while setting the T2 values equal to the mean T2 value (58 ms). For RE-TOSSI, the inversions were stopped in the TOSSI acquisition after 60 readouts. A zero-padded (8192 points) Fourier transform of the complex transverse magnetization was performed to obtain the point spread functions (PSF) of the two techniques for each tissue. This provides the point spread function under the assumption of linear k-space transversal, which was used throughout this paper. A zero-padded (8192 points) Fourier transform of a constant function was also calculated to demonstrate an ideal finite sampling PSF for comparison.

MRI Scanner

All experiments were performed on a clinical 1.5 T short and wide bore MR scanner with standard receiver coils (MAGNETOM Espree, Siemens Healthcare, Erlangen, Germany).

Phantom Imaging

Phantom experimentation was performed to determine the contrast and spatial resolution properties of RE-TOSSI. Eight 50-mL syringes (2.5 cm diameter) with different concentrations of gadolinium and agarose were used as well as a larger bottle provided by the vendor. A single channel head coil was used for signal reception. The T2 values of the phantoms where measured using a multi-contrast spin echo sequence (21 echo times between 18-388 ms, TR = 10,000 ms). The T2 value of each phantom was calculated using the scanner's software fitting function. Images were acquired using TSE (TE = 92 ms, TR= 10,000 ms, ETL = 18), HASTE (TE = 93 ms, TR = 10,000 ms), TOSSI (TE_{eff} = 456 ms, FA = 60°, TR = 4.6 ms, PF = 4/8, N_A = 2), bSSFP (FA = 60°, TR = 4.6 ms, PF = 8/8), HF-bSSFP (FA = 60°, TR = 4.6 ms, PF = 4/8), and RE-TOSSI (λ = 0.04, 0.09, 0.13, 0.22, 0.34, 0.45, 0.56, 0.67, 0.78, 0.89; other parameters same as TOSSI). All images had a 250 mm × 250 mm FOV, 256 × 256 matrix, and 7 mm slice thickness.
To assess the differences in contrast between the sequences, the SNR was calculated for each vial by taking the signal in each vial divided by the standard deviation of signal in a large ROI in air. To correct for sampling time, the SNR was corrected for bandwidth (BW) and partial Fourier (PF) according to Eq. [6] in (11):

$$c_{\text{SNR}} = \frac{\sqrt{BW \cdot \text{SNR}}}{PF}$$  \[3\]

The corrected SNR in HASTE, TOSSI and bSSFP were plotted as function of the corrected SNR in TSE. Linear regression analysis was performed to determine the slope, intercept and correlation coefficient of the linear regression. To assess the similarity of contrast in RE-TOSSI to TOSSI, linear regression was performed for the RE-TOSSI to TOSSI signal (for the ten values of $\lambda$ as well as 0 and 1). The intercept was normalized by dividing by twice the largest value. The slope, correlation coefficient and normalized intercept values were plotted as a function of $\lambda$.

To quantify the spatial resolution properties of the different sequences the modulation transfer function (MTF) was calculated for each sequence. The modulation transfer function provides a complete and quantitative description of spatial resolution (15). The MTF provides contrast as a function of spatial frequency and is related to the point spread function. As the point spread function becomes narrower (improved resolution), the MTF becomes broader. The edge response functions (ERF, a profile through the edge of the phantom in the phase encoding direction) were extracted from the phantom images and analyzed using transfer function methods (16). The ERF were smoothed using a $[1 2 1]$ kernel and differentiated to generate line spread functions (LSF). The MTF was calculated as the normalized magnitude of the Fourier transform of the LSF. The cutoff frequency ($f_c$) was determined as the frequency at which the magnitude of the MTF reached 1/10 of its peak height. A higher cutoff frequency implies improved spatial resolution. The area under the MTF curve (MTFA) was calculated as a global measure of spatial resolution (a sum over all spatial frequencies). The optimal frequency response, defined as the maximum of the $(\text{MTF}(k) \times k)$ as a function of $k$ (spatial frequency) was determined.

**Human Imaging Experiments**

After obtaining informed consent, images were collected in the abdomen and head of asymptomatic human volunteers as part of a study approved by the Institutional Review Board. The body and spine matrix coils were used for the abdominal imaging while the head matrix coil was used for the head imaging. For abdominal imaging, HASTE (TE = 77 ms, TR = 2000 ms), TOSSI (TEeff = 228 ms, PF = 5/8, N_A = 2, T2opt = 90 ms, FA = 60°, TR = 5.2 ms), RE-TOSSI ($\lambda = 0.31$, other parameters same as TOSSI) and bSSFP (FA = 60°, TR = 5.2 ms, PF = 8/8) acquisitions were used to acquire coronal images at the same locations with FOV = 350 mm × 350 mm, 256 × 256 matrix and 5 mm slice thickness. The volunteer was asked at the beginning of each scan to hold their breath. For brain imaging, axial slices were acquired using TSE (TE = 92 ms, TR = 4000 ms, ETL = 18), TOSSI (TEeff = 390 ms, PF = 6/8, N_A = 8, T2opt = 90 ms, TR = 5.2 ms, FA = 60°), RE-TOSSI ($\lambda = 0.42$, other parameters same as TOSSI) and bSSFP (FA = 60°, TR = 5.2 ms, PF = 8/8).

**RESULTS**

Simulation results are shown in Figure 2. Fig 2a shows the result of the transverse magnetization evolution during TOSSI. For the on-resonant tissues (all but fat), the magnetization levels throughout the acquisition demonstrate $T2$ contrast (i.e. simulated spins with a longer $T2$ value have higher signal). In the same curves, there are small oscillations in the transverse magnetization value during TOSSI. The simulated fat curve demonstrates
large oscillations due to off-resonance, and the overall signal level is below that of the other tissues. For comparison, the results of the simulation with the same T2 values but with the T1 value set equal to the mean T1 value are plotted in Fig. 2b. Note the similarity to the TOSSI evolution curves. The simulated fat curve has a lower average level since the mean T1 value (726 ms) is higher than the actual value (343 ms). Fig. 2c shows the same simulation when the T2 values are set equal to the mean T2 value (58 ms) (with T1 values equal to those in the original TOSSI simulation). The signals for all the on-resonance tissues (all but fat) collapse to almost a single curve. Fig. 2d shows the simulated transverse magnetization evolution during RE-TOSSI. For the first segment of the acquisition (repetitions 1-60), the RE-TOSSI signal evolution is identical to TOSSI. Note the elevated signal in the RE-TOSSI (Fig. 2d) simulation compared with traditional TOSSI (Fig. 2a) during the latter readouts (lines 61-192) after the inversion pulses were stopped in RE-TOSSI. During RE-TOSSI, the signals approach bSSFP values instead of continuing to decay as in conventional TOSSI. Fig. 2e shows the simulated TOSSI and RETOSSI point spread functions for liver (T2 = 46 ms, T1 = 586 ms). For comparison, the simulated PSF of a tissue with T2 = ∞ is also plotted (the Fourier transform of a boxcar function). The RE-TOSSI PSF has a full-width at half maximum (FWHM) that is 74% narrower than that of TOSSI; the first and second side lobe amplitudes are decreased by 45% and 42%, respectively. The RE-TOSSI PSF has a FWHM that is very close to the ideal PSF (1.08 pixels). Although reduced, RE-TOSSI sidelobes are still elevated compared to the ideal PSF. Fig. 2f shows the PSFs for subcutaneous fat (220 Hz off-resonant) during TOSSI and RE-TOSSI. The RE-TOSSI PSF has a full-width at half-maximum that is 56% reduced compared to TOSSI. The 1st and 2nd sidelobe amplitudes are reduced by 47% and 81% respectively. The RE-TOSSI PSF is very close to the ideal PSF.

Phantom images are shown in Fig. 3. The T2 weighted TSE image demonstrates T2 contrast and sharp edges. The T2 value (ms) for each vial is shown in the TSE image. The HASTE image has similar T2 contrast to the T2 TSE but has marked degradation of spatial resolution along the phase encoding (vertical) direction, which becomes more prominent as the T2 value decreases (closed arrow). The TOSSI image also exhibits T2 contrast; there is some spatial blurring, but much less compared to the HASTE image. There is some ghosting observable in the large phantom (black arrow). The bSSFP image does not have T2 contrast, as several vials appear bright, out of order, due to their high T2/T1 value (arrows). The half-Fourier bSSFP image also demonstrates this violation of T2 contrast. Both bSSFP images have sharp edges. The RE-TOSSI (λ = 0.13) image has T2 contrast as well as sharp edges.

Results obtained from phantom experiments are plotted in Fig. 4. In Fig. 4a, plots of the corrected SNR in the HASTE, TOSSI and bSSFP images are plotted as a function of the corrected SNR in the TSE image. Results of the linear regression fits are provided in the legend. There is excellent similarity of HASTE contrast to TSE. For bSSFP, the violation of T2 contrast is demonstrated by the points that fall off the line, resulting in a lower correlation coefficient (R² = 0.79). The TOSSI plot demonstrates monotonically increasing signal in the TOSSI image compared to the TSE value (i.e. T2 contrast); however, the slope of the fit (0.31) is reduced compared to HASTE (1.0). The Modulation Transfer Functions for the six pulse sequences are plotted in Fig. 4b. Note the widened MTF of RE-TOSSI (λ = 0.13) compared to TOSSI, and TOSSI compared to HASTE, demonstrating improved spatial resolution. Quantitative comparisons of the MTFs for the various techniques are provided in Table 1. These results quantitatively demonstrate the improved spatial resolution in RE-TOSSI compared to TOSSI and TOSSI compared to HASTE. The spatial resolution metrics of RETOSSI are very similar to TSE and bSSFP. Results of the RE-TOSSI to TOSSI signal linear regression analysis are plotted in Fig 4c. Here the slope, intercept and correlation coefficient are plotted as a function of λ. λ can be reduced from 1 to 0.13 without significantly changing the image contrast. Once λ is reduced beyond 0.13, the slope,
intercept as well as correlation coefficient change, indicating a change in image contrast (as can be observed in the HF-bSSFP image ($\lambda = 0$) in Fig. 3). Fig. 4d shows a plot of the area under the MTF curve (a measure of spatial resolution) as a function of $\lambda$. As expected, there is increased spatial resolution for smaller $\lambda$ values.

Comparison coronal images in the abdomen of an asymptomatic human volunteer are shown in Fig. 5. The RE-TOSSI image demonstrates T2 contrast and high spatial resolution. The TOSSI image has similar contrast to the RE-TOSSI image but has visible spatial resolution degradation, most notable in the muscle. The HASTE image also has T2 contrast but suffers from severe spatial resolution degradation throughout the image. The bSSFP image has high spatial resolution, but there is notable low signal in the bone marrow, violating T2 contrast. The effect of the non-uniform magnetic field can be seen in the bottom corners of the images. This gives rise to the well-known bands in the bottom corners of the bSSFP image. These bands are visibly widened in the TOSSI image (open arrow). The RE-TOSSI image has similar bands to the TOSSI image.

Figure 6 shows comparison images in the head of an asymptomatic volunteer. Note the similar contrast in the RE-TOSSI and TOSSI images. The grey-white differentiation is similar in the RE-TOSSI and TSE images. The RE-TOSSI image has improved spatial resolution compared to the TOSSI image, visible in the depiction of the extraocular muscle (open arrow). The bSSFP image demonstrates decreased grey-white matter differentiation, known in the steady state for bSSFP (17), as well as bright fat surrounding the optic nerve (closed arrow).

DISCUSSION

The simulation results demonstrate that the underlying contrast mechanism in TOSSI is differences in T2 value and that the magnetization decay during TOSSI leads to spatial blurring by increasing the FWHM of the PSF. By removing the inversion pulses from TOSSI, the magnetization is allowed to progress to the bSSFP value, leading to a PSF with near ideal width. The oscillations in the magnetization during TOSSI lead to elevated side lobes in the PSF which can manifest as ghosting. These are reduced in RE-TOSSI. The simulations also show the mechanism behind the low signal of fat in TOSSI as being due to off-resonance. This off-resonance leads to oscillation in the fat magnetization decay. An initial fat saturation was simulated and used in the in-vivo experiments in order to maintain the fat signal level at a near constant value throughout the acquisition.

The phantom results (Figs. 3, 4) allow comparison of the contrast and spatial resolution properties of the various techniques. HASTE is shown to maintain T2 contrast but suffer from severe spatial resolution degradation. Because of the prolonged decay in TOSSI at the lower flip angles compared to HASTE, there is less degradation of spatial resolution in TOSSI. bSSFP is shown to have good spatial resolution but lack T2 contrast. The results demonstrate that TOSSI has decreased T2 contrast compared to HASTE. This can be explained by the equation governing transverse magnetization in TOSSI (5):

$$M_{xy}=\sin(FA/2)^*\exp\left(-\sin^2(FA/2)^*t/T2\right)$$

This equation has a $\sin(FA/2)$ term modulating the exponential decay. Since the difference between two signals is also weighted by the same factor, contrast is also reduced by $\sin(FA/2)$. In the previous examples, a flip angle of 60° was used; this corresponds to a 50% reduction in predicted contrast. Comparing eqn 4 with simple exponential decay (neglecting the amplitude scaling factor addressed previously), a simple substitution can be made.
\[ \sin^2\left(\frac{\text{FA}}{2}\right) \times \text{TE}_{\text{tossi}} = \text{TE}_{\text{tse}} \]

Therefore, the effective echo time in TOSSI to produce similar contrast to TSE or HASTE is:

\[ \text{TE}_{\text{tossi}} =\text{TE}_{\text{tse}} \left[ \sin^2 \left(\frac{\text{FA}}{2}\right)\right]^{-1} \quad [5] \]

At a flip angle of 60°, this corresponds to a factor of 4 increase in the effective echo time. The phantom imaging allowed experimentation with RE-TOSSI over the range of \( \lambda \) values. A half Fourier acquisition was used to judge the extent around the center of k-space needed to maintain TOSSI contrast. In this experiment using 2.5 cm diameter vials, it was determined that 13% of the data around the center of k-space needed to be encoded using TOSSI in order to maintain TOSSI contrast and maximize spatial resolution. The resulting RE-TOSSI image (\( \lambda = 0.13 \)) demonstrates T2 contrast similar to TOSSI as well as sharp edges. These results further highlight the role of the center of k-space in determining image contrast.

The human abdominal imaging results are important for several reasons. They demonstrate that the contrast in RE-TOSSI is similar to TOSSI and that the relative signal of the various tissues agree in general with simulation predictions (kidney and spleen brightest, followed by liver and bone marrow and then by muscle). They also demonstrate the improved spatial resolution in RE-TOSSI compared to TOSSI, and TOSSI compared to HASTE. The bSSFP image has very low signal in bone marrow, which violates the T2 contrast. The images in the coronal orientation also allow visualization of the off-resonance properties of the various techniques. The widened off-resonance bands of TOSSI predicted elsewhere (18) are observed. However, there is still a large homogenous region for imaging even in a short and wide-bore scanner with a smaller region of high magnetic field homogeneity compared to standard clinical scanners. The in-vivo brain images again demonstrate similar contrast and improved spatial resolution in RE-TOSSI compared to TOSSI. The clear depiction of the extraocular muscles obtained with RE-TOSSI may be useful for clinical studies of the orbits, for example for the detection and evaluation of orbital infection (19,20). These can be life-threatening infections associated with sinusitis in children, who may benefit from rapid imaging and the lack of ionizing radiation in MRI.

The information gained in phantom experiments was used as a guide for in-vivo imaging. The largest partial Fourier factor compatible with the chosen effective echo time was used. In the abdominal imaging studies, a partial Fourier factor of 5/8 was used. TOSSI was performed during the acquisition of negative k-space (25% of negative k-space), through the center of k-space and for the central 13% of positive k-space lines. This resulted in \( \lambda = (0.25 + 0.13)/1.25 = 0.31 \). For the brain imaging, a longer effective echo time was used, which was compatible with a 6/8 partial Fourier acquisition. TOSSI was used to encode the negative k-space lines (50% of negative k-space), the center of k-space and the central 13% of positive k-space lines. This resulted in \( \lambda = (0.5 + 0.13)/1.5 = 0.42 \). Since the phantom experiments were performed using 2.5 cm diameter vials it is expected that TOSSI contrast will not be maintained for smaller sized objects. Therefore, using a partial Fourier acquisition to encode a larger extent of negative k-space prior to the effective echo time not only results in higher SNR (since more lines of k-space are acquired), but also ensures that T2 contrast is maintained for smaller sized objects as well. This strategy seemed to work well, as the in-vivo RE-TOSSI images appear to have identical contrast to the corresponding TOSSI images for all sized objects.

The off-resonance bands in TOSSI are widened compared to bSSFP due to the frequently applied \( \alpha/2-\text{TR}/2 \) preparation pulses, which are ineffective for off-resonance spins, associated with each inversion pulse. In RE-TOSSI the off-resonance bands are initially widened during TOSSI and then revert to the normal bSSFP width after the inversion pulses.
are stopped. It has been shown elsewhere (18) that the width of the stop bands in TOSSI and RETOSSI can be reduced using a set of 4 linearly increasing flip angle pulses at the expense of slower imaging time.

In RE-TOSSI, there is a large reduction in imaging time due to the removal of numerous inversion pulses and the associated steady state storage and preparation pulses and spoiling gradients, when compared to TOSSI. The increase in image acquisition rate in RE-TOSSI comes without a loss in SNR (unlike other methods of acquisition acceleration such as parallel or partial Fourier imaging). Another important benefit of RE-TOSSI is the large reduction in the average power deposition compared to TOSSI and HASTE (18). This decrease in average power deposition is caused by the elimination of many power intensive adiabatic inversion pulses in RE-TOSSI compared to TOSSI and imaging with small flip angles in the later portion of RE-TOSSI compared to the 180 degree refocusing pulses in HASTE.

Most previous combined acquisition techniques (8-10) used very different pulse sequence blocks to cover the various regions of k-space (e.g. gradient echo and EPI). Here, the fundamental repeating imaging block is balanced SSFP with the same parameters (e.g. same flip angle, bandwidth, etc.) throughout the sequence. This can be observed in the pulse sequence schematic (Fig. 1), where there is an overlap region between TOSSI and bSSFP imaging “blocks.” This overlap prevents discontinuities of magnetization amplitude and phase in k-space and therefore avoids the associated image artifacts or need to optimize the junction of the sequences.

As mentioned above, T2-TIDE has been shown to decrease the degradation of spatial resolution associated with HASTE images (11). In that method, T2 contrast is achieved by using bSSFP with 180° pulses around the center of k-space and then linearly decreasing the flip angle of the RF pulses over a transition region to ~60° while acquiring the rest of the lines of k-space. There are several possible advantages to using RE-TOSSI compared to T2-TIDE. First, since TOSSI is used to acquire the data around the center of k-space, which has a prolonged signal decay compared to HASTE, there will be less signal decay and thus less degradation of spatial resolution associated with RE-TOSSI (as seen in Figs. 3, 5). Secondly, there may be less RF energy deposition using RE-TOSSI since low flip angles are used to image throughout the acquisition.

RE-TOSSI was used with standard linear Cartesian data sampling in this paper. This is beneficial as all commercial scanners support online data processing for this trajectory. The trajectory also minimizes off-resonance and gradient timing offset effects. It may be possible to use other k-space data sampling strategies in which the center vs. the periphery of k-space are obtained at different times such as spiral (21) or concentric rings (22) trajectories. Even strategies such as radial imaging, which sample the center of k-space with each line, could possibly be used if a strategy such as KWIC is employed (23), wherein the central portion of k-space is not used for certain lines.

**CONCLUSION**

TOSSI, a novel way of generating T2 contrast, is shown to have improved spatial resolution compared to HASTE, but still suffers from substantial spatial resolution degradation in the single-shot acquisition. Resolution Enhanced (RE-)TOSSI is a new rapid imaging pulse sequence which modifies TOSSI by removing the inversion pulses after a user-selectable portion of the acquisition, leading to a unique TOSSI-bSSFP combined acquisition technique. RE-TOSSI is shown to simultaneously reduce the spatial resolution degradation associated with TOSSI while improving image acquisition time and RF power deposition.
In-vivo abdominal and brain RE-TOSSI images demonstrate T2-weighted contrast with inherently suppressed fat signal.

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Fig. 1.
TOSSI and RE-TOSSI acquisition schematics. (a) TOSSI block diagram depicting that TOSSI is composed of repeating blocks of bSSFP performed parallel and anti-parallel to the main magnetic field. Each bSSFP block is preceded by an $\alpha/2$-TR/2 preparation pulse and ended by an $\alpha/2$-TR/2 magnetization storage pulse. Adiabatic inversions are preceded and followed by gradients which spoil any remnant transverse magnetization. $N_P$ and $N_A$ denote the number of readouts in the parallel and antiparallel states, respectively. (b) RE-TOSSI schematic depicting the inversion pulses and k-space sampling during RE-TOSSI. The inversion pulses associated with TOSSI are removed at a user selectable point during the acquisition. This gives rise to overlapping blocks of TOSSI and bSSFP imaging. A partial Fourier acquisition is used with the center of k-space acquired during the TOSSI imaging block. The effective echo time ($T_{Eeff}$) is determined by the time the center of k-space is encoded. This can be adjusted using the partial Fourier factor and/or by using preparation pulses (PP) for more fine control. The fraction of readouts acquired during TOSSI is denoted by $\lambda$ while the remaining fraction of readouts during balanced SSFP is $1-\lambda$. Note that the time in the parallel state ($T_P$) varies during the acquisition while the time in the antiparallel state ($T_A$) does not.
Fig. 2. Simulation results. (a) TOSSI signal evolution for 6 tissues found in the abdomen. The T2/T1 values (ms/ms) are given in the legend (PV = paravertebral, SQ = subcutaneous). The simulation includes 220 Hz off-resonance for fat tissue and the fat signal is initially set to zero, simulating an initial fat saturation pulse. (b) TOSSI simulations performed while setting the T1 values equal to the mean T1 value. (c) TOSSI simulations while setting the T2 values equal to the mean T2 value. (d) RE-TOSSI signal evolution curves for the same T2/T1 values in (a) (inversions stopped after 60 readouts). (d) RE-TOSSI and TOSSI point spread functions for liver tissue; for comparison the point spread function of a constant function is also plotted. (e) PSFs for subcutaneous fat.
Fig. 3.
Phantom images. The T2 value of each vial is plotted on the TSE image. Note the T2 contrast in the TSE (TE = 92ms), HASTE (TE = 93ms), TOSSI (TEeff = 456ms, PF = 4/8) and RE-TOSSI images. There is worse loss of spatial resolution in the HASTE image compared to TOSSI (closed arrows). The vials in the bSSFP images do not follow the T2 signal progression (open arrows) due to the different T2/T1 values of the vials.
Fig. 4. Phantom imaging analysis. (a) Comparison of corrected SNR for each object in the HASTE, TOSSI and bSSFP images compared to the T2w TSE image. Parameters of linear regression are provided in the legend. (b) The modulation transfer function for all 6 techniques, obtained from the edge response functions from the short T2 vial. (c) RE-TOSSI to TOSSI linear regression parameters for 12 values of \( \lambda \) between 0 and 1. The intercept was normalized to twice the largest intercept (\( \lambda = 0 \)). (d) Area under the MTF curve as a function of \( \lambda \).
Fig. 5. Comparison coronal images in the abdomen of an asymptomatic volunteer. Note the similar contrast and improved spatial resolution in the RE-TOSSI ($\lambda = 0.31$, TEeff = 228ms, PF = 5/8) image compared to TOSSI. Open arrow depicts the widened off-resonance band in TOSSI compared to bSSFP. There is visible loss of spatial resolution in the HASTE (TE = 77ms) image. Note the low signal of bone marrow in the bSSFP image (closed arrow), violating T2 contrast. The image acquisition times are provided in the corner of each image.
Fig. 6.
Comparison axial images in the head of an asymptomatic volunteer. Note the similar contrast throughout the RE-TOSSI ($\lambda = 0.42$, $T_{\text{Eff}} = 390\text{ms}$, PF = 6/8) and TOSSI images and the improvement in spatial resolution evident in the extraocular muscles in the RE-TOSSI image (open arrow). There is reduced contrast between the grey and white matter in the bSSFP image as well as bright fat (closed arrow) compared to the FS TSE ($T_E = 92\text{ms}$), RE-TOSSI and TOSSI images. Image acquisition times are provided in the corner of each image.
Table 1

Summary of quantitative results of the phantom modulation transfer function (MTF) measurements$^a$

<table>
<thead>
<tr>
<th>Sequence</th>
<th>Cutoff Frequency$^b$ (c/mm)</th>
<th>MTFA$^c$ (c/mm)</th>
<th>OFR$^d$ (c/mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>RE-TOSSI ($\lambda = 0.13$)</td>
<td>0.35</td>
<td>0.25</td>
<td>0.27</td>
</tr>
<tr>
<td>TSE</td>
<td>0.34</td>
<td>0.24</td>
<td>0.25</td>
</tr>
<tr>
<td>bSSFP</td>
<td>0.36</td>
<td>0.23</td>
<td>0.17</td>
</tr>
<tr>
<td>HF-bSSFP ($\lambda = 0$)</td>
<td>0.33</td>
<td>0.19</td>
<td>0.14</td>
</tr>
<tr>
<td>TOSSI</td>
<td>0.23</td>
<td>0.13</td>
<td>0.10</td>
</tr>
<tr>
<td>HASTE</td>
<td>0.09</td>
<td>0.06</td>
<td>0.04</td>
</tr>
</tbody>
</table>

$^a$Nyquist limit for the experiment (0.98 mm pixel size) is 0.51 cycles/mm

$^b$Cutoff frequency = frequency at 1/10 of peak amplitude

$^c$MTFA = area under the MTF curve

$^d$OFR = optimum frequency response = max of (MTF(k) × k) curve