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Characterization and Reduction of Saturation Banding in Multi-Planar Coherent and Incoherent Steady State Imaging

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Abstract

Hypointense band artifacts occur at intersections of non-parallel imaging planes in rapidly acquired MR images; analysis of these bands and strategies to mitigate their appearance have largely gone unexplored. The magnetization evolution in the different regions of multi-planar images was simulated for three common rapid steady state techniques (FLASH, FISP, True-FISP). Saturation banding was found to be highly dependent on the pulse sequence, acquisition time and phase encoding order. Encoding the center of k-space at the end of the acquisition of each slice (i.e. reverse centric phase encoding) is demonstrated to be a simple and robust method for significantly reducing the relative saturation in all imaging planes. View ordering and resolution dependence were confirmed in multi-planar abdominal images. The added importance of reducing the artifact in accelerated acquisition techniques (e.g., parallel imaging) is particularly notable in multi-planar TrueFISP images in the brain.

Keywords

Steady State Imaging; Multi-plane MRI; Dark Band Artifact; View Ordering

INTRODUCTION

Intersecting imaging planes are sequentially acquired in multi-planar MRI. Diagnostic applications for multi-planar MRI include scout imaging (anatomical localizers) (1), repeated rapid imaging (a.k.a. fluoroscopy) for localization and studying movement (e.g. phonation) (2) as well as imaging freely moving subjects (e.g. fetal MRI) (3). Interventional applications include pre-stereotactic surgical planning to identify and avoid critical structures (4), real-time intra-procedural guidance in multiple orthogonal planes (5,6) as well as providing imaging feedback regarding device position (7). Orthogonal acquisition planes have also been used for system characterization (8). A well-known phenomenon includes the presence of dark “saturation” bands at the intersection of the imaging planes (1–3,8).

Figure 1a illustrates the intersecting plane dark band artifacts (arrows) in one of three orthogonal imaging planes acquired during fluoroscopic MR imaging in the face of a patient with an arteriovenous malformation (arrow heads). While sometimes useful, these hypointense artifacts can be extremely problematic as they obscure the underlying anatomy. Potential problems include decreased contrast for small anatomical structures, decreased contrast in areas of pathology as well as obscuring interventional instruments such as needles or laser fibers. In interventional MRI for example, the bands will cover the structures of interest as the planes are chosen so the targeted structures appear in all intersecting slices. Attempts to modulate or minimize the “saturation” effect have been made; an unobtrusive display overlay can then be used to show the relative position of the imaging planes. In echo planar imaging (EPI), Neustadter et al. studied reducing the dark band artifact in one (main) plane by reducing the tip angle during the acquisition of the two remaining ancillary orthogonal planes (3). However, this only reduced the artifact in one of the three planes at the expense of significantly reduced SNR in the remaining two orthogonal imaging planes.

Steady state sequences are attractive for multi-planar MRI due to their fast acquisition times and ability to generate different types of contrast. The most commonly employed sequences include fast low angle shot (FLASH) (9) and TrueFISP (a.k.a. balanced steady state free precession, bSSFP) (10,11). In FLASH, the transverse magnetization is spoiled (12,13) in each TR providing proton density, T_1 or T_2^* contrast depending on the TE, TR and tip angle. TrueFISP provides the highest possible SNR efficiency and T_2/T_1 contrast but is susceptible to off-resonance banding artifacts (distinct from the saturation bands addressed herein) in regions of magnetic field inhomogeneity (14). FISP (a.k.a. steady state free precession, SSFP) can be used to introduce T_2 -like contrast without off-resonance banding, albeit with lower SNR efficiency and lower contrast (15). The transient state of the magnetization in these sequences has been studied previously (16–18).

In this work, the behavior of the magnetization in the different regions of multiple intersecting imaging planes is characterized for each steady state sequence. Dynamic steady states are established, caused by the incomplete relaxation of magnetization between acquisitions and the transient response of the sequences. The expected banding is simulated as a function of base resolution in the three techniques. View ordering is investigated as a simple and effective way to minimize the observed banding artifacts. The importance of reducing the artifact in accelerated acquisition methods such as parallel imaging (19) is further highlighted.

METHODS

Computer Simulations

To characterize the saturation banding in orthogonal steady state acquisitions, magnetization evolution simulations were performed in Matlab (The MathWorks Inc., Natick, MA). In a tri-plane acquisition, four distinct regions of magnetization must be evaluated (Fig. 1b). The Bloch equation simulations followed the magnetization in the four regions during a sequential multi-plane MR acquisition, i.e. $\{Ax, Cor, Sag\}_n$, where n denotes the number of acquisitions; we assume that all phase encoding lines for each slice are acquired before moving to the next plane. The tip angle for RF excitation was set to zero in the regions not

experiencing RF pulses during the acquisition of a particular slice. The magnetization in all regions experienced relaxation and off-resonance precession during the simulated acquisition of all slices.

The Bloch equation simulations were performed in a similar fashion to Sekihara (20). The magnetization vector before $\mathbf{M}_n^-$ and after $\mathbf{M}_n^+$ each RF pulse was calculated in the following manner:

$$\mathbf{M}_n^+ = \mathbf{R}_x(\alpha) \mathbf{M}_n^- \quad [1]$$

where α denotes the tip angle, and $\mathbf{R}_x(\alpha)$ is the rotation operator about the x-axis. The following relaxation operator is defined:

$$\mathbf{S}(\text{TR}, T_1, T_2) = \begin{pmatrix} E_2 & 0 & 0 \\ 0 & E_2 & 0 \\ 0 & 0 & E_1 \end{pmatrix} \quad [2]$$

where $E_1 = \exp(-\text{TR}/T_1)$, $E_2 = \exp(-\text{TR}/T_2)$. The relationship between the magnetization at the end of the repetition $\mathbf{M}_{n+1}^-$ and the beginning of the repetition $\mathbf{M}_n^+$ is given by:

$$\mathbf{M}_{n+1}^- = \mathbf{R}_z(\theta_n) \mathbf{S}(\text{TR}, T_1, T_2) \mathbf{M}_n^+ + (1 - E_1) \mathbf{M}_0 \quad [3]$$

where θ_n is the dephasing angle, $\mathbf{R}_z(\theta_n)$ is the rotation operator about the z-axis and $\mathbf{M}_0 = (0, 0, M_0)$ is the thermal equilibrium magnetization vector. The dephasing angle θ_n is given by:

$$\theta_n = \theta_0 + (n - 1) \delta\theta. \quad [4]$$

θ_0 incorporates the dephasing per TR due to unbalanced gradients as well as magnetic field inhomogeneity. $\delta\theta$ is the phase increment in FLASH imaging (50° for RF spoiling, as implemented on the scanner), and is zero in FISP and 180° in TrueFISP (phase alternation). Eqs. 1 and 3 are used iteratively to follow the magnetization through the acquisition (from the initial thermal equilibrium).

Due to the non-zero gradient moment per TR in FLASH and FISP, the magnetization in a voxel $\tilde{\mathbf{M}}_n^+$ is calculated as the integral over all dephasing angles (assuming a uniform distribution of magnetization dephasing angles in the voxel due to gradient dephasing):

$$\tilde{\mathbf{M}}_n^+ = \int_0^{2\pi} \mathbf{M}_n^+(\theta_{n-1}) d\theta_{n-1}. \quad [5]$$

The observed magnetization is the transverse component of the magnetization vector $\mathbf{M} = (M_x, M_y, M_z)$:

$$M_{xy} = \sqrt{M_x^2 + M_y^2}. \quad [6]$$

In the TrueFISP acquisition, a linearly increasing set of 10 RF pulses was simulated to catalyze the steady state, and an $\alpha/2$ pulse was used at the end of the acquisition of each slice to store the magnetization along the z-axis (as is implemented on the scanner).

The relative saturation of the magnetization within the bands (regions A, B, and C) compared to outside the bands (region D) was calculated using the following equation:

$$\% \text{saturation} = \frac{M_{xy}(D) - M_{xy}(A, B, C)}{M_{xy}(D)} \times 100\%. \quad [7]$$

The predicted saturation observed in an image was estimated by the magnetization level at the time that the central line of k-space is encoded. For a linearly phase encoded acquisition, this corresponds to the repetition number given by the total number of phase encoding lines divided by two; in a reverse centrically phase encoded acquisition, it corresponds to the total number of phase encoding lines; for centric, it corresponds to the first acquisition.

Human Experiments

All human imaging experiments were performed on a 1.5 T short and wide bore MRI system used for both diagnostic and interventional clinical studies (MAGNETOM Espree, Siemens Healthcare, Erlangen, Germany). The vendor-supplied cardiovascular (CV) sequence can acquire linear and centrically encoded FLASH, FISP and TrueFISP images; it was modified to perform linear and reverse centric phase encoding. Experiments were performed as part of a study approved by the local institutional review board (IRB) after obtaining signed informed consent. Volunteers were placed in the scanner head first, in supine position. The body coil was used for RF transmission. In the first volunteer, multi-plane abdominal imaging was performed. The 6-element flexible body array coil was placed on the anterior thorax and abdomen and was used for signal reception along with the 24-element spine matrix coil in the patient table. Linear and reverse centric multi-planar (axial, sagittal, coronal) FLASH imaging was performed using five different phase encode resolutions: 33×256 (13%), 64×256 (25%), 128×256 (50%), 192×256 (75%), 256×256 (100%). In the second volunteer, multi-plane brain imaging was performed using a 32-channel head coil for signal reception along with parallel imaging; four GRAPPA acceleration factors were used: $R = 1, 2, 3, 4$ (with 24 reference lines) with both linear and reverse centric phase encoding.

Image Analysis

The images were analyzed in Matlab by averaging along a portion of the main saturation artifact in the image. Due to receive coil non-uniformity along the profile, the average profiles were normalized such that the average value of the beginning and ending pixels

along a short portion of the profile centered on the dark band artifact equaled 1. Therefore, the relative saturation in the band compared to adjacent tissue could be judged by the value at the center of the profile. The signal-to-noise ratio in the non-parallel images was calculated by dividing the mean signal in a homogenous ROI outside the saturation bands in tissue, divided by the standard deviation of an ROI placed in air.

RESULTS

Example dynamic steady states established during the three sequences are plotted in Fig. 2a–c. The magnetization evolution plotted is for simulated liver tissue ($T_1 = 590$ ms, $T_2 = 56$ ms) during an acquisition with 128 repetitions per slice (i.e. 128 phase encoding lines per slice). In the simulations, the number of grouped acquisitions was set to four (i.e. $n = 4$) to achieve a dynamic steady state of the magnetization in the different regions. The magnetization began at thermal equilibrium (M_0) and went through four iterations of Ax,Cor,Sag acquisitions with appropriate modification of the flip angle for the magnetization within each region during the acquisition of each slice (e.g. flip angle set to either 0° or 15° for FLASH based on whether the magnetization is within the excited slice). The plotted magnetization is from the fourth acquisition of the sagittal slice, showing the dynamic steady state established. The TE (2.2 ms) and TR (4.4 ms) were held constant for the three simulations to highlight differences in the dynamic and steady state magnetization levels in the three steady state techniques. The relative saturation in regions A, B and C (bands) compared to region D (outside bands) are plotted as a function of repetition number in Fig. 2d–f. Note the significant saturation ($> 60\%$) in regions A (intersection of all slices) and C (magnetization at intersection of current and previous slices) at the beginning of the acquisition. Region B (magnetization at intersection of current slice and slice acquired two images previously) is also saturated, but to a lesser degree ($\approx 20\%$). Note that the saturation is highly time dependent and generally decreases as the repetition number increases.

The predicted saturation in the three banding regions using linear and reverse centric phase encoded FLASH, FISP and TrueFISP are compiled and plotted in Fig. 3. The saturation is greater at the lower number of phase encoding lines and decreases as the number of phase encoding lines (and hence duration of acquisition) increases. Note that there can be a large level of saturation in a linearly phase encoded acquisition (e.g. $\approx 50\%$). Reverse centric phase encoding significantly reduces the banding magnitude compared to a standard linearly encoded acquisition for all three pulse sequences over a wide range in the number of phase encoding lines.

Coronal images from tri-plane FLASH acquisitions in the abdomen, at three different phase encoding resolutions, are shown in Fig. 4. The closed arrows denote the horizontal dark band artifact in the liver caused by the previous axial plane acquisition. The bands decrease in intensity as the phase encoding resolution increases (as predicted by the simulations). A fainter vertical banding artifact (open arrows) can also be seen in the lower resolution images (e.g. 64×256), caused by the sagittal slice acquired prior to the axial slice. Note the significantly reduced horizontal banding artifact intensity in the reverse centrically encoded images (middle row) compared to the linearly encoded images (top row) at all phase encoding resolutions. Average profiles through the main (horizontal) banding artifacts are

plotted in the bottom row. The relative saturation in the band was measured to be 41%, 30% and 17% (matrix = 64×256 , 128×256 , 256×256 , respectively) in the linearly encoded images; these values were reduced to 27%, 8% and 0% (respectively) using reverse centric phase encoding. For identical acquisition parameters (and the T1 and T2 values assumed for liver above), the simulations predicted 34%, 12%, and 0% banding (respectively) for linear encoding; while for reverse centric encoding the simulations predicted reduction to 10%, 0% and 0% (respectively). For the linear phase encoded images in figure 4, the liver SNR was calculated to be 47.2 (64×256), 25.9 (128×256) and 13.4 (256×256). Using reverse centric phase encoding, the SNR values were measured to be 31.2, 20.1 and 12.5, respectively. The corresponding change in SNR values from linear to reverse centric encoding were therefore -34%, -23% and -6%, respectively.

Axial images from the multi-planar TrueFISP acquisitions in the human brain using parallel imaging are shown in Fig. 5. Closed arrows denote the vertical banding artifact caused by the previous sagittal-plane acquisition which becomes more prominent as the acceleration factor increases. At the higher acceleration factors (e.g. $R = 4$), additional horizontal saturation bands are observed (open arrows) which were caused by coronal slices acquired prior to the sagittal slices (i.e. two slices previously). Note the reduction in saturation band artifact level in the reverse centrically encoded images (middle row, arrows) compared to the linearly encoded images (top row, arrows) at all acceleration factors. Average profiles through the main (vertical) saturation bands are plotted in the bottom row. The relative saturation in the main dark bands in the linearly encoded images was 15%, 31% and 37% ($R = 1, 2, 4$, respectively); the relative saturation in the bands was reduced to 0%, 13% and 22% (respectively) in the reverse centrically encoded TrueFISP images. Using identical acquisition parameters, the simulations predicted 9%, 23%, and 32% banding (respectively) for linear phase encoding; and reduction to 1%, 5%, and 15% (respectively) for reverse centric encoding.

DISCUSSION

That reverse centric phase encoding can reduce saturation band artifact magnitude compared to linear phase encoding in rapid multi-plane steady state acquisitions was anticipated. However, the quantitative relationships between saturation banding and number of phase encoding lines and pulse sequence as well as the extent of the difference between conventional and reverse centric phase encoding had not been previously characterized. The simulation results have demonstrated the nature of the magnetization evolution in the different regions of an image during a multi-plane steady state acquisition. The magnetization outside the bands (region D) experiences a dynamic state similar to that observed previously (17,18). Briefly, the FLASH magnetization evolution is jagged due to imperfections in spoiling the transverse magnetization; FISP magnetization experiences oscillations, while TrueFISP magnetization has a smooth evolution to the steady state. As expected, the steady state magnetization in FLASH was lower than that in FISP which was lower than TrueFISP. It is important to note that in the 128 line example presented in Fig. 2, there is enough time for the magnetization to reach steady state for the chosen parameters. However, the inadequate time for full relaxation, as well as acquisitions with fewer phase encoding lines, necessitate Bloch equation simulation to predict the dynamic steady state

established. Although there are time periods with a low degree of saturation created by the oscillations in the FISP evolution, it is difficult to exploit them in a view ordering method in vivo since the troughs may occur at different times for different tissues and the rapid changes in magnetization about the center of k-space would introduce artifacts. However, the overall decrease in saturation over time in all three techniques provides the motivation for encoding the center of k-space as far from the disturbance as possible (using reverse centric phase encoding).

The compilation of simulation results (Fig. 3) over a range of useful phase encoding lines per acquisition (from fully sampled to highly accelerated) shows the dramatic reduction in saturation banding that can be achieved by simply changing from linear to reverse centric phase encoding. Fig. 3 also quantitatively demonstrates the problem of increased saturation banding which will be encountered during accelerated imaging. The in-vivo liver results (Fig. 4) confirm the general trends seen in the simulations, i.e. lower artifact magnitude at higher resolution and significant decrease in banding using reverse centric encoding. The imaging results in the brain (Fig. 5) highlight the importance of reducing the artifact in accelerated methods such as parallel imaging. The saturation banding artifact magnitudes measured in vivo corresponded well in general with the simulation predictions, demonstrating similar trends. The expected saturations calculated via simulation take into account only the relative signal levels at the central k-space line, while the images are formed using the full k-space data. Also, some differences are expected due to uncertainties in the actual T_1 and T_2 values, motion, tip angle variation as well as other factors.

It is important to note that equal image quality and saturation banding artifact reduction is achieved in all of the imaging planes in this technique (images not shown); this is an important improvement over the tip angle variation method proposed by Neustadter (3) which compromises image quality in two planes for an improvement in the third. Other methods of accelerating image acquisition besides parallel imaging (e.g. partial Fourier, view sharing or any other technique that reduces the number of phase encoding lines acquired) will experience similar increases in saturation banding. Reverse centric phase encoding can also be used in those techniques to mitigate saturation banding, as was shown here for parallel imaging. It is possible that image contrast will be altered between linear and reverse centrically phase encoded images. However, reverse centric encoding approximates the steady state contrast levels more closely than linear encoding since the center of k-space is acquired closer to the steady state (while linearly encoded images are weighted to the dynamic state). Nevertheless, Figs. 4 and 5 demonstrate similar image contrast in the linear and reverse centric phase encoded images for two different sequences over a wide range of acquisition times.

Changing the phase encoding order from linear to reverse centric may alter the image SNR. As mentioned above, linearly encoded images can be weighted by the dynamic phase of the magnetization evolution while the reverse centric images are preferentially weighted by the steady state magnetization level. This has benefits as well as potential drawbacks. The benefits include signal levels that are closer to the well-known, predictable values in the steady state. On the other hand, as the results show, SNR may be reduced. The results also demonstrate that the loss in SNR is dependent on the phase encoding resolution, with higher

phase encoding resolutions having fairly small changes (i.e. <10%). It may be anticipated that in many cases the benefit of eliminating the saturation artifact and providing true steady state contrast will outweigh the loss of some transient signal SNR. For example, although centrically encoded images have much higher SNR than linearly encoded images (see figure 2a–c), linearly encoded images are used much more frequently in practice. In interventional MRI, where it is critical to determine the location of the device clearly, the loss of some overall image SNR can be justified in order to improve visualization of a device which would otherwise be obscured by the saturation band.

CONCLUSION

Dark band artifacts observed at the intersection of orthogonal acquisition planes in multi-planar MRI were characterized in short-TR steady state acquisitions. Significant temporal and resolution dependence was found. View ordering was investigated as one method to minimize the saturation banding artifacts. Reverse centric phase encoding is shown to be a simple and robust method for minimizing saturation banding artifacts in three types of multi-plane steady state sequences (FLASH, FISP and TrueFISP). No increase in acquisition time is incurred by changing the phase encoding order. The increased significance of reducing the artifact in highly accelerated image acquisition techniques such as parallel MRI is demonstrated.

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FIG. 1.

Dark band artifacts in multi-planar MRI. **(a)** Sagittal TrueFISP image in the head of a patient with a buccal arteriovenous malformation (arrow heads) prior to MRI guided sclerotherapy (S = superior, A = anterior). Note the vertical (closed arrow) and horizontal (open arrow) dark band artifacts obscuring the anatomy and pathology. **(b)** Schematic diagram highlighting four distinct regions of magnetization in a 3-plane acquisition: The magnetization in region A (magnetization at the intersection of two bands) experiences RF excitation pulses during the acquisition of all three slices (e.g. axial, coronal and sagittal). Region B (magnetization in the lighter band) experiences RF excitation pulses during the acquisition of the current slice as well as the slice acquired two images previously (e.g. sagittal and axial slices, respectively). Region C (magnetization in the darker band) experiences RF pulses during the acquisition of the current and previous slices (e.g. sagittal and coronal slices, respectively). Region D (the region outside the bands) experiences RF excitation pulses only during acquisition of the current slice (e.g. sagittal slice).

FIG. 2.

Transverse magnetization and relative saturation evolution in multi-plane steady state sequences. Simulated dynamic steady states established in (a) FLASH (FA = 15°), (b) FISP (FA = 15°), and (c) TrueFISP (FA = 70°) sequences. TE = 2.2 ms; TR = 4.4 ms; T1 = 590 ms, T2 = 56 ms (simulated liver tissue); 128 repetitions. (d–f) Relative saturation of the magnetization in the bands (regions A, B and C), compared to outside the bands (region D) as a function of repetition number (time), in the three steady state techniques. The magnetization in each region is plotted with a distinct line style (legend in (a)).

FIG. 3.

Banding artifact dependence on view ordering and base resolution. Relative saturation in the three regions of banding (A – C) for linear phase encoded (Lin; solid) and reverse centric encoded (RC; dashed) images in (a) FLASH, (b) FISP and (c) TrueFISP sequences. The figures represent a compilation of data from simulations with 32 to 256 repetitions per slice (in steps of 8). Other parameters same as those provided in Fig. 2.

FIG. 4.

Resolution and view ordering effects in multi-planar FLASH imaging. Coronal images in the abdomen acquired using linear (top row) and reverse centric (middle row) phase encoding. Closed arrows denote the horizontal saturation banding artifact in the liver from the previously acquired axial slice. Additional fainter vertical banding artifacts (open arrows) are seen in the low resolution images (e.g. 64×256) caused by the acquisition of the sagittal slice two slices previously. Average profiles through the main (horizontal) saturation banding artifact are plotted in the bottom row (Lin = linear, RC = reverse centric phase encoding). Each column corresponds to a different matrix size: **(a)** 64×256 , **(b)** 128×256 and **(c)** 256×256 . Acquisition parameters: FA = 15° , TE/TR = 3.26 ms/7.24 ms, FOV = 300 mm $\times$ 300 mm, slice thickness = 5 mm, BW = 250 Hz/pixel.

FIG. 5.

Effect of parallel imaging and view ordering in multi-planar TrueFISP. Axial images in the brain acquired using linear (top row) and reverse centric (middle row) phase encoding. Closed arrows denote the vertical banding artifacts caused by the previous sagittal plane acquisition. Additional fainter horizontal banding artifacts are seen in the highly accelerated images (open arrows), caused by the acquisition of the coronal slice two slices previously. Average profiles through the main (vertical) saturation banding artifact are plotted in the bottom row (Lin = linear, RC = reverse centric phase encoding). Each column corresponds to a different parallel imaging acceleration factor: (a) $R = 1$, (b) $R = 2$ and (c) $R = 4$.

Acquisition parameters: FA = 60°, TE/TR = 2.6 ms/5.2 ms, FOV = 200 mm × 200 mm, slice thickness = 5 mm, matrix 256×256, BW = 592 Hz/pixel.

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