Banding Artifacts Along the Frequency-encoding Direction in Multi-echo Magnetic Resonance Imaging: Origin and a Remedy

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Abstract

We report here a highly reproducible artifact seen in standard multi-echo spin-echo imaging, in which severe banding was seen along the frequency-encoding direction for all late echo images, but not for the first echo. A theoretical analysis showed that the banding resulted from the destructive interference effects of the stimulated echo in the presence of both an inaccurate calibration of the RF pulse flip angles and a gradient area offset. Experimental results from a water phantom verified the origin. A remedy method employing crusher gradients with alternating amplitudes was proposed and implemented, which successfully eliminated the artifacts as evidenced by in vivo imaging results from human brain.

Keywords: Magnetic resonance imaging, Multi-echo imaging, Stimulated echo, Banding artifacts

Introduction

Magnetic resonance (MR) imaging is now widely acknowledged as a useful tool in diagnostic radiology. Among the versatile soft-tissue contrast mechanisms in modern MR imaging, T2-weighted imaging is particularly one of the most important MR modalities in terms of showing conspicuous contrast between healthy tissues and pathological lesions [1,2]. In certain diseases, quantitative T2 measurements have been shown to be effective in distinguishing minor functional deficiency by reflecting, for example, perfusion changes or gliosis [3,4]. Obtaining diagnostic information from T2-related imaging examinations is therefore essential in biomedical research as well as in clinical practice.

No matter for qualitative T2-weighting or quantitative T2 measurements, the T2-related information is often obtained from multi-echo spin-echo imaging experiments (so-called the Carr-Purcell-Meiboom-Gill, or CPMG experiments) [5], or its associated variants such as dual-contrast fast spin-echo imaging. Compared with repeated single-echo experiments, the higher efficiency and ease of implementation of the CPMG experiment has made it attractive in basic science research and in clinical applications. We have recently observed, however, a highly reproducible phenomenon showing severe banding artifacts from a manufacturer-supplied standard multi-echo sequence (Fig.1), where the direction of banding variations was always found along the frequency-encoding direction (i.e., dark and bright bands along the phase-encoding direction). In addition, images reconstructed from the first echo were always free of such artifacts no matter how severe the later echoes were. The presence of this phenomenon could certainly hamper the usage of late-echo images for routine diagnosis.

In this article, we used both theoretical derivations and experimental validations to verify the origin of the banding artifacts. A remedy method that successfully eliminated the banding artifacts was also addressed.

Theory

Figure 2 shows a simple RF pulse sequence diagram of a multi-echo experiment where two echoes are to be sampled with TE2 = 2*TE1, as often encountered in MR imaging sequences provided by many MR manufacturers. With flip angle of the first excitation RF pulse being α and that of the latter two refocusing RF pulses being β, respectively, five echoes would appear in general [6], with their time of
Figure 1 Banding artifacts observed from images acquired from a bottle of water using a manufacturer-supplied standard multi-echo sequence. Shown are four images obtained with TE = 200 (top-left), 400, (top-right), 600, (bottom-left), and 800 msec (bottom-right), respectively. The artifacts were highly reproducible, with the first-echo image always free of banding. Frequency encoding direction was vertical.

One first note here that for the image with TE1, the received signals originate from the first spin-echo SE1 (Hahn echo) [8], whereas for the image with TE2, the received signals consist of SE2 (CPMG echo) [5] and SE3 (stimulated echo) [9]. In the ideal situation when $\alpha = 90^\circ$ and $\beta = 180^\circ$, SE3 has zero amplitude and thus SE1 and SE2 follow the exponential decaying curve characterized by T2. On the other hand, if the flip angles deviate from the ideal situation, SE3 would increase while both SE1 and SE2 decrease in amplitude. Note that not only the image with TE2 gets contaminated from the stimulated echo, but the signal reduction from TE1 to TE2 also no longer follows the original T2 relaxation decay.

The situation becomes complicated with consideration of the spatial encoding gradients. Fig.3 shows one readout scheme popularly employed in standard multi-echo sequences. With the area of the prephasing gradient canceling half that of the readout gradient, as desirable in the ideal situation, all three echoes occur at the expected time locations as illustrated in Fig.2. In reality, however, the area of the prephasing gradient may be different from its expected value. Fig.4 shows an example where the prephasing gradient has a slightly lower amplitude, such that it cancels slightly less than half of the readout gradient area. SE1 therefore gets shifted toward the front while SE2 toward a later time by the same amount. SE3, being from the composite refocusing action formed by RF2 and RF3, shows no transverse magnetization between RF2 and RF3. Thus SE3 gets shifted toward the front in a way analogous to SE1.

Figure 2. A simple RF pulse sequence diagram of a two-echo experiment where TE2 = 2*TE1. With flip angle of the first excitation RF pulse being $\alpha$ and that of the latter two refocusing RF pulses being $\beta$ respectively, five echoes would appear in general, with their time of occurrence and echo amplitudes listed in Table 1. For typical scanning parameters seen in clinical practice, TE1 is on the order of some tens to hundreds of milliseconds while the data readout window is about or less than 10 msec. In this study we shall focus ourselves on the first three echoes only, because with typical parameter ranges as stated above, echoes SE4 and SE5 commence at times far from the readout window for the two expected echoes and hence are not sampled into the receiver.
Table 1. Time, amplitudes (neglecting relaxation), and origins of the five echoes in a three-pulse experiment as shown in Fig.2 with TE2 = 2*TE1.

<table>
<thead>
<tr>
<th>echo</th>
<th>time</th>
<th>Amplitude</th>
<th>origin</th>
</tr>
</thead>
<tbody>
<tr>
<td>SE1</td>
<td>TE1</td>
<td>$M_0 \sin \alpha \sin^2 \frac{\beta}{2}$</td>
<td>excitation: RF1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>refocusing: RF2</td>
</tr>
<tr>
<td>SE2</td>
<td>2*TE1</td>
<td>$M_0 \sin \alpha \sin^4 \frac{\beta}{2}$</td>
<td>excitation: RF1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>first refocusing: RF2</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>second refocusing: RF3</td>
</tr>
<tr>
<td>SE3</td>
<td>2*TE1</td>
<td>$\frac{1}{2}M_0 \sin \alpha \sin^2 \beta$</td>
<td>excitation: RF1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>refocusing: RF2+RF3 composite</td>
</tr>
<tr>
<td>SE4</td>
<td>2.5*TE1</td>
<td>$M_0 \cos \alpha \sin \beta \sin^2 \frac{\beta}{2}$</td>
<td>excitation: RF2</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>refocusing: RF3</td>
</tr>
<tr>
<td>SE5</td>
<td>3*TE1</td>
<td>$M_0 \sin \alpha \cos \beta \sin^2 \frac{\beta}{2}$</td>
<td>excitation: RF1</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>refocusing: RF3</td>
</tr>
</tbody>
</table>

Figure 3. A readout scheme popularly employed in standard multi-echo sequences. With the area of the prephasing gradient canceling half that of the readout gradient, all three echoes occur at the expected time locations. Different shading shows gradient area cancellations.

Figure 4. Timing of three echoes with area offset in the prephasing gradient. If the prephasing gradient has a slightly lower amplitude, such that it cancels slightly less than half of the readout gradient area, SE1 gets shifted toward the front while SE2 toward a later time by the same amount. SE3 gets shifted toward the front analogously to SE1 because no transverse magnetization exists between RF2 and RF3 for the stimulated echo. Different shading shows gradient area cancellations.

The effects of the opposite shifting for SE2 and SE3 can be formulated as follows. Let $g(x,y)$ be the distribution of the magnetization (including the relaxation effects) in the two-dimensional image space to be obtained. The k-space data set is then the Fourier transform of $g(x,y)$, $G(k_x,k_y)$, and the reconstructed MR image is given by the absolute value of the inverse Fourier transform of $G(k_x,k_y)$. Since the signals near TE2 consist of both the second echo and the stimulated echo, albeit shifted in time differently, the k-space raw data $G(k_x,k_y)$ can be expressed as:

$$G_y(k_x,k_y) = G(k_x - \Delta k, k_y) + \delta G(k_x + \Delta k, k_y)$$  \hspace{1cm} (1)

where $\Delta k$ is proportional to the time shift caused by the percentage gradient area offset, assuming $x$ to be the direction of frequency encoding, and $\delta$ is a constant standing for the ratio of the amplitudes of the stimulated echo to that of the second spin-echo:

$$\delta = \frac{\sin^2 \beta}{2 \sin^4 (\beta/2)} = \frac{2}{\tan^2 (\beta/2)}$$  \hspace{1cm} (2)

The reconstructed image before taking absolute value is then given by:

$$g_y(x,y) = IFT(G_y(k_x,k_y)) = IFT(G(k_x - \Delta k, k_y) + \delta G(k_x + \Delta k, k_y))$$  \hspace{1cm} (3)

$$= (e^{i2\pi k_x x} + \delta e^{-i2\pi k_x x}) g(x,y)$$

where $IFT$ stands for inverse Fourier transform. Therefore the final MR image becomes:

$$|g_y(x,y)| = \left| e^{i2\pi k_x x} + \delta e^{-i2\pi k_x x} \right| \left| g(x,y) \right|$$

$$= \left| (1 + \delta^2) \cos^2 \frac{2\pi k_x x + (1 - \delta)^2 \sin^2 \frac{2\pi k_x x}{1 + \delta^2} \left| g(x,y) \right| \right|$$  \hspace{1cm} (4)

The derivations above indicate that in the simultaneous presence of inaccurate flip angle calibration and imperfect gradient area cancellation, the images at TE2 would show banding artifacts along the frequency-encoding direction, with the frequency of the sinusoidal banding shown as $2\Delta k$, proportional to the percentage of gradient area offset as predicted in Eq.(4). This could account for the banding artifacts shown in Fig.1. Furthermore, the amplitude of the sinusoidal banding is related to flip angle offset, which can show bright-to-dark intensity ratio of 1.12 with only 10% of
Figure 5. Suggested remedy scheme using crusher gradient pairs, immediately before and after the $\beta$ pulses, with alternating amplitudes. The crusher gradient pairs act to diphase the stimulated echo without substantially affecting the primary echo signals. The crusher gradients can be applied along any direction.

Figure 6. The in-phase channel (or the “real part”) of the k-space raw data for the second-echo image (TE = 400 msec) of a water phantom. In this figure, gray color stands for zero in value. White and black represent positive and negative values, respectively. The presence of two echoes with comparable amplitude evidenced possible interference of the primary second echo from the stimulated echo. Frequency encoding was along the horizontal direction.

Methods

Imaging experiments were performed on a 1.5 Tesla MR system (Siemens, Erlangen, Germany) using a circularly polarized head coil, even though the original artifacts were found on an MR system from another manufacturer. A multi-echo imaging sequence was used to acquire four images from a bottle of water at TE = 200, 400, 600, and 800 msec, respectively, to demonstrate the banding artifacts. Flip angles of the RF pulses were pre-calibrated using manufacturer-supplied pre-scan package and were not deliberately changed to any extent. But since all the RF pulses were slice-selective, their flip-angle profiles along the slice selection direction cannot be perfectly homogeneous due to the non-linearity of the Bloch equation [10]. In other words, we expected that the inaccuracy of flip angles must be present in almost all daily routine practice and there was thus no need to deliberately vary the RF flip angles. The area of the prephasing gradient was altered by reducing its amplitude through pulse sequence programming performed in-house, which allowed an accurate assessment of the gradient area offset. The k-space data were retrieved together with the inverse Fourier-transformed images.

For the remedy method as suggested in Fig.5, one set of in vivo images was acquired to demonstrate its efficacy. The crusher gradients were applied along the slice selection direction. Transaxial images of human brain were obtained using the multi-echo imaging sequence as mentioned above, at TE = 20, 40, 60, and 80 msec from a healthy subject who had no history of neurological diseases. The frequency-encoding direction was chosen to be antero-posterior due to the generally elongated shape of the head. Images acquired before and after adding the crusher gradients with alternating amplitudes were shown and compared, with gradient area offsets in the prephasing gradient deliberately kept identical.

Results

Figure 6 shows the in-phase (or “real”) channel k-space data obtained from the water phantom at TE = 400 msec. The presence of two echoes with comparable amplitude can be
Figure 7. Multi-echo MR images obtained from a water phantom with three different values of gradient area offsets. From left to right: images acquired with TE1 = 200, TE2 = 400, TE3 = 600, and TE4 = 800 msec, respectively. (a-d; top row) Gradient area offsets equal to zero. (e-h) Slight gradient area offsets. (i-m) Severe gradient area offsets. Frequency encoding was along the horizontal direction. The k-space data shown in Fig.6 correspond to the image in (j). It is also seen that that no matter how severe the later echo images were, images obtained from the first echo were always free of the banding artifacts.

Figure 8. Images of human brain obtained from the second echo (TE = 40 msec) of a multi-echo spin-echo imaging sequence before (a) and after (b) the addition of crusher gradients with alternating amplitudes. All other parameters, including the gradient area offset and the RF pulse flip angle calibrations, were kept identical. Banding artifacts were eliminated by the addition of the crusher gradients. Frequency encoding direction was vertical.

clearly seen. Fig.7 shows the multi-echo MR images obtained with three different values of gradient area offsets, of which the k-space data shown in Fig.6 correspond to the image in Fig.7j. It is seen that even we did not change the value of the flip angle calibration, the interference effects from the stimulated echo were always prominent, leading to obvious banding whenever gradient offsets were present. One further notice that no matter how severe the later echo images were, images obtained from the first echo were always free of the banding artifacts.

The effects of the crusher gradients with alternating amplitudes are demonstrated using in vivo images from the human head as shown in Fig.8. Figs.8a shows the image obtained with TE = 40 msec (i.e., the second echo), showing obvious banding artifacts. After the addition of the crusher gradients, the banding was eliminated (Fig.8b).

Discussion

In this article, we have reported the banding artifacts specifically found in standard multi-echo MR imaging sequences. The banding is always along the frequency-encoding direction, and always found in only the later echo images. In addition, we have analyzed its origin using theories, plus experimental verification evidencing our inferences. One remedy method was proposed, which was shown to successfully remove the artifacts. In fact, we were surprised at the initial observation of this phenomenon, because multi-echo imaging has been used for more than two decades as one of the most frequently used routine sequences clinically. In other words, a manufacturer-supplied sequence
should not present similar pitfalls in such an obvious manner. After retrieving the k-space data we realized that two echoes were present, spaced at some time apart from each other. Subsequently, a series of analyses were performed, which led to the eventual theories as stated here.

In reality, the presence of flip angle inaccuracy is always present for slice-selective RF pulses [10]. Particularly at the slice boundaries, it has been shown that the non-linearity of the Bloch equation itself could lead to flip angle errors as large as 50%, not to mention other imperfections due to hardware problems such as the insufficient coverage of the tissue of interest by the excitation RF coil [7]. Therefore, the existence of stimulated echo is quite common whenever there are three or more RF pulses in an imaging sequence. Note that although better RF pulse profiles could be obtained using more complicated designing techniques such as the Shinnar-LeRoux transform or the “spinor” theories [11,12], the resulting long RF pulse duration is generally not desirable for the refocusing pulses in standard multi-echo sequences.

On the other hand, offsets in the gradient area as large as those shown in Figs.1 and 6 are unlikely to be frequently encountered in daily practice. Gradient area offsets often arise due to eddy current effects, which lead to only slight deviations in the amplitudes and durations for all gradient waveforms. Even with older type of unshielded gradient coils, eddy current effects are generally small. In other words, the banding artifacts in late echo images, even if present, are usually of low frequency, manifesting itself as non-uniformity of the signals throughout the entire image field-of-view along the frequency-encoding direction [7].

Clinical image interpretations in such a case are usually not affected, unless an accurate quantitative T2 measurement is needed throughout the entire image field-of-view [4]. For the case shown in Fig.1, we strongly suspect that the manufacturer-supplied sequence never went through a gradient calibration procedure before releasing to its customers.

But even without a pre-release gradient calibration, the banding artifacts can be effectively eliminated using the crusher gradients with alternating amplitudes as shown in our study. Other possible remedy schemes exist, including one using prephasings gradients before every readout gradient for all echoes, plus the necessary rewinding gradient accompanied after the readout window [13]. This scheme has the advantage of guaranteeing constructive interference of the second spin-echo and the stimulated echo (and in fact for all higher outer echoes), somewhat similar to the balanced steady-state-free-precession imaging method [14,15]. However, this technique is not a preferential choice in standard multi-echo imaging, due to necessarily lengthening of the minimum echo spacing. In contrast, the use of the crusher gradients along the slice selection direction as shown in Fig.5 allows effective elimination of the banding artifacts without using too long gradient durations, hence the concomitant increase in echo spacing should be minimal.

In conclusion, we have reported the stimulated-echo-induced banding artifacts in multi-echo MR imaging, with theories and experimental results both supporting our reasoning. The remedy method using crusher gradients with alternating amplitudes was able to dephase the stimulated echo without affecting the other primary echoes. Hence it is an effective approach and should be suitable for routine use in clinical practice to enhance diagnostic accuracy.

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References


