Published in final edited form as:

Magn Reson Med. 2016 February ; 75(2): 537–546. doi:10.1002/mrm.25595.

Gradient-Modulated SWIFT

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Abstract

Purpose—Methods designed to image fast-relaxing spins, such as sweep imaging with Fourier transformation (SWIFT), often utilize high excitation bandwidth and duty cycle, and in some applications the optimal flip angle cannot be used without exceeding safe specific absorption rate (SAR) levels. The aim is to reduce SAR and increase the flexibility of SWIFT by applying time-varying gradient-modulation (GM). The modified sequence is called GM-SWIFT.

Theory and Methods—The method known as gradient-modulated offset independent adiabaticity was used to modulate the radiofrequency (RF) pulse and gradients. An expanded correlation algorithm was developed for GM-SWIFT to correct the phase and scale effects. Simulations and phantom and in vivo human experiments were performed to verify the correlation algorithm and to evaluate imaging performance.

Results—GM-SWIFT reduces SAR, RF amplitude, and acquisition time by up to 90%, 70%, and 45%, respectively, while maintaining image quality. The choice of GM parameter influences the lower limit of short T_2^* sensitivity, which can be exploited to suppress unwanted image haze from unresolvable ultrashort T_2^* signals originating from plastic materials in the coil housing and fixatives.

Conclusions—GM-SWIFT reduces peak and total RF power requirements and provides additional flexibility for optimizing SAR, RF amplitude, scan time, and image quality.

Keywords

SWIFT; frequency sweep; gradient modulation; SAR; GOIA; fast relaxing spins; ultrashort T_2 ; VERSE

INTRODUCTION

The time elapsing between spin excitation and signal acquisition in gradient-echo and spin-echo MRI is typically too long to visualize nuclei with ultrashort effective transverse relaxation times (T_2^*) in the range of 1 millisecond or less. Such ultrashort relaxation times

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are found in highly ordered tissues (e.g., tendons and knee meniscus), highly mineralized tissues (e.g., bone and teeth), lung (1,2), and tissues containing a high concentration of magnetic nanoparticles (3). Ultrashort- T_2^* -sensitive sequences typically acquire the free induction decay (FID), and the k-space consists of frequency-encoded FIDs arranged in a radial manner. Common ultrashort- T_2^* -sensitive sequences include the ultrashort echo time (UTE) sequence (4,5), zero echo time sequences (6–8), and sweep imaging with Fourier transformation (SWIFT) (9,10). In addition, there are single point and hybrid methods (11–13).

SWIFT uses swept radiofrequency (RF) excitation and nearly simultaneous signal acquisition in a time-shared mode. SWIFT captures signals from spins with ultrashort T_2^* down to the microsecond range and has found numerous applications (14–21). A current limitation, however, is the high specific absorption rate (SAR) when using large excitation bandwidth (B_w) to maximally preserve signal from spins with ultrashort T_2^* and/or when using a large excitation flip angle to maximize signal from spins with short T_1 .

Attempts to enhance RF pulse performance using modulated gradients have been relatively successful as a means to reduce RF power deposition (22–24). One example is the gradient-modulated offset independent adiabaticity (GOIA) approach, which transforms RF amplitude or frequency modulation to a corresponding gradient modulation while maintaining constant adiabaticity factor over the entire spectral B_w (23). Other approaches such as variable-rate selective excitation (VERSE) (22) can also be applied to modify the RF pattern in order to achieve the desired flat excitation profile. The implementation of such approaches in SWIFT is not trivial because the gradient modulation influences both the excitation and the phase evolution of the spins. As a consequence, the correlation procedure used in the original SWIFT method (9) is not capable of resolving the spatially and temporally varying phase of the spins that accumulates during the changing field gradients.

In this paper, we present an extension of the SWIFT method, which uses time-varying gradients during read-out. We call it gradient-modulated SWIFT (GM-SWIFT). We present theory, numerical simulations, and phantom and in-vivo human experiments. Image quality, RF amplitude, and SAR are compared between the standard SWIFT (constant gradient) and GM-SWIFT sequences.

THEORY

Generation of a GOIA Pulse

According to the GOIA approach (23), for given RF amplitude-modulation function $F_1(t)$ and gradient-modulation function $G(t)$, the RF frequency-modulation function $F_2(t)$ can be solved from the constraint:

$$\frac{d}{dt} \left(\frac{AF_2(t)}{G(t)} \right) = \frac{2\pi(\gamma B_1^0 F_1(t))^2}{KG(t)}, \quad [1]$$

where the adiabatic factor K is held constant over time and frequency, B_1^0 is the amplitude of the RF field, A is the amplitude of the frequency modulation, and c is the gyromagnetic ratio. $G(t)$ can be any function without zero point and that is not linearly proportional to $F_2(t)$. Figure 1a schematically presents the standard SWIFT sequence, implemented with a constant gradient and a stretched hyperbolic secant pulse (HS2) (25–28). Figure 1b shows the sequence diagram of GM-SWIFT with the frequency-modulation function calculated using Eq. [1], the amplitude-modulation function of the HS2 pulse, and the below HS-shaped gradient-modulation function:

$$G(t) = 1 - g_m \cdot \text{sech} \left[\beta \left(\frac{2t}{T_p} - 1 \right)^{n_p} \right], \quad [2]$$

where n_p is the HS modulation order ($= 4$ here for Fig. 1b), g_m is the gradient-modulation factor, T_p is pulse length, and $\beta = 7.6$ is the truncation factor for truncation of the value of F_1 to 0.001 at temporal edges. Although many different gradient shapes can in principle be used, the HS-shaped gradient-modulation function was chosen here based on its smoothness and continuity of derivatives.

The GOIA principle was developed originally for inverting spins adiabatically. When exciting spins in a subadiabatic manner, as done in standard SWIFT and GM-SWIFT, the GOIA principle still produces a flat excitation profile (i.e., the flip angle is constant in the range of the frequency-swept bandwidth). The value of K needs to be adjusted in order to allow the frequency sweep given by $AF_2(t)$ to fall in the interval $[-A, A]$ (23).

Correlation Procedures

In the SWIFT sequence, excitation and acquisition are nearly simultaneous. The acquisition happens in the gaps of the frequency-swept excitation pulse. The field gradient during this procedure plays the role of frequency encoding or readout. In standard SWIFT, G and B_w are constant, and projections of the spin density can be obtained using the correlation method described in the original SWIFT paper (9). In GM-SWIFT, G and B_w are time-dependent. As a result, spins excited at different times experience differing segments of the time-varying readout gradient and accumulate phase in a spatially and temporally dependent manner. Thus, the existing correlation procedure for standard SWIFT does not remove these additional phase variations; a different correlation procedure is needed for GM-SWIFT.

The first step is to derive the signal equation. The SWIFT sequence is operating in the linear, subadiabatic region (29). With the small tip-angle approximation (30), the evolution for a single isochromat at position x at time τ , $M_{xy}(x, \tau)$, can be derived by solving the Bloch equations (31) in the rotating frame

$$M_{xy}(x, \tau) = i\gamma M_0 \int_0^\tau B_1(t) e^{-i\gamma x \cdot \int_t^\tau G(t') dt'} dt, \quad [3]$$

where $B_1(t) = B_1^0 F_1(t) e^{-i \int A F_2(t) dt}$ is the modulation function of the RF pulse, and the bold $\mathbf{G}(t) = G(t) \hat{G}$, represents a vector; therefore, it specifies the modulation function of the field gradient as well as its direction. Although Eq. [3] represents the solution for a continuous B_1 field, it is also suitable for gapped excitation for the given bandwidth as long as the RF power and sampling requirements are met (32). The integral of $M_{xy}(x, \tau)$ over all spins at different positions gives the total signal response of the entire object,

$$r(\tau) = i\gamma M_0 \int_x \left\{ \rho_0(x) \int_0^\tau B_1(t) e^{-i\gamma \mathbf{x} \cdot \int_t^\tau \mathbf{G}(t') dt'} dt \right\} dx, \quad [4]$$

where $\rho_0(x)$ is the spin density. Recovering the spin density from the acquired signal is an inverse problem. The excitation k-space concept (31), which is defined as the integral of the time-varying gradient over time, can be used here to solve and provide further insight into the problem. By defining the excitation k-space $\mathbf{k}(t)$, the spatial frequency weighting function $w(\mathbf{k})$ and the spatial frequency sampling function $s(\mathbf{k})$ as follows:

$$\mathbf{k}(t) = -\gamma \int_t^{T_{acq}} \mathbf{G}(t') dt', \quad [5]$$

$$w(\mathbf{k}(t)_\tau) = B_1(t) / |\gamma \mathbf{G}(t)|, \quad [6]$$

$$s(\mathbf{k}, \tau) = \int_0^\tau \delta(\mathbf{k}(t) - \mathbf{k}) |\dot{\mathbf{k}}(t)| dt, \quad [7]$$

the signal response can be rewritten as:

$$r(\tau) = i\gamma M_0 \int_x \left\{ \rho_0(x) \int_k w(\mathbf{k}) s(\mathbf{k}, \tau) e^{i\gamma \mathbf{x} \cdot [\mathbf{k} - \mathbf{k}(\tau)]} d\mathbf{k} \right\} dx. \quad [8]$$

Both integrals of x and k are over the whole domains $([-\infty, \infty])$ in Eq. [8] and in the rest of the equations in this work. The integral in Eq. [5] starts at t , not zero, because spatial encoding does not commence until a spin isochromat has been excited by the frequency-swept pulse (29). T_{acq} is the acquisition time in one repetition period, TR (Fig. 1). Note that the spatial frequency sampling function $s(\mathbf{k}, \tau)$ is also explicitly time dependent through the acquisition time parameter τ . For almost all hardware designs, τ will be at equally spaced positions in the time domain, and $s(\mathbf{k}, \tau)$ maps τ to corresponding sample positions in k space. Both standard SWIFT and GM-SWIFT are radial k-space sampling sequences using

the projection method so the excitation k -space trajectory follows center out “spokes” in 3D spherical k -space. More specifically, $s(\mathbf{k}, \tau) = \text{step}(\mathbf{k}(\tau) - \mathbf{k})$ is a Heaviside step function describing signal existence only after a spin has been excited. Note that k -space sampling is uniform for standard SWIFT, but nonuniform for GM-SWIFT. After substituting $s(\mathbf{k}, \tau)$ into Eq. [8] and integrating over $\mathbf{x}$ the signal response becomes

$$r(\mathbf{k}(\tau)) = i\gamma M_0 \int_{\mathbf{k}} \{h(\mathbf{k}(\tau) - \mathbf{k})\} w(\mathbf{k}) d\mathbf{k}, \quad [9]$$

with

$$h(\mathbf{k}(\tau) - \mathbf{k}) \equiv \text{step}(\mathbf{k}(\tau) - \mathbf{k}) \int_{\mathbf{x}} \rho_0(\mathbf{x}) e^{i\gamma \mathbf{x} \cdot [\mathbf{k} - \mathbf{k}(\tau)]} d\mathbf{x}. \quad [10]$$

Here, h is the impulse response function (33). In the $\mathbf{k}$ domain, from Eq. [9] it can be seen that the signal response is the convolution of $h(\mathbf{k})$ and $w(\mathbf{k})$. By using the Fourier convolution theorem, the Fourier transform (FT) of $r(\mathbf{k}(t))$ becomes:

$$R(\mathbf{x}) = H(\mathbf{x}) \cdot W(\mathbf{x}), \quad [11]$$

where

$$H(\mathbf{x}) = \rho_0(\mathbf{x}) \otimes \mathcal{F}\{\text{step}(t)\} \quad [12]$$

and

$$W(\mathbf{x}) = \int_{\mathbf{k}} w(\mathbf{k}(t)) e^{-i\gamma \mathbf{x} \cdot \mathbf{k}} d\mathbf{k} = \mathcal{F}\{w(\mathbf{k}(t))\}. \quad [13]$$

Note that the signal response $r(\tau)$ acquired with uniform sampling interval in the time domain τ should be resampled to uniformly sample in the domain of $\mathbf{k}(\tau)$ (34). Then, the discrete FT can be used. The same applies for $w(\mathbf{k}(t))$. The spectrum of the spin system $H(\mathbf{x})$, which is actually the convolution of the spin density and the FT of the Heaviside step function as shown in Eq. [12], can be retrieved by

$$H(\mathbf{x}) = \frac{R(\mathbf{x}) \cdot W^*(\mathbf{x})}{|W(\mathbf{x})|^2}. \quad [14]$$

In a full set of radial acquisition data, each projection will have another corresponding projection acquired with the opposite gradient direction. In that case, the corresponding projections can be summed and the Heaviside step function will become a constant, with $H(\mathbf{x})$ becoming the real valued spin density $\rho_0(\mathbf{x})$.

With constant gradient, $G(t) = G_0$, the general analysis in equations [5–14] above, suitable for GM-SWIFT, reduces to the case of standard SWIFT. In that case, $k(t) = \gamma G t$ and $k \propto t$. The FT of $r(k(\tau))$ and $w(k(t))$ in the domain of k in Eq. [11] and Eq. [13] reduces to the direct FT on the signal $r(t)$ and the RF pulse $B_1(t)$ in the familiar time domain t (9).

The generalized correlation step can also be treated as a time retransformation or variable substitution in which the k -space is transformed to a special time-modulated frame. In this case, the gradient appears “constant,” and B_1 must be transformed into an effective B_1 in this frame. Then the properties of linear response theory, as used for the time-independent system or standard SWIFT case, can be applied in this special time-modulated frame.

METHODS

Simulations of a one-dimensional (1D) object and experiments on phantoms and a human subject were performed. In both simulations and experiments, the acquisition was prolonged after the RF pulse to increase the k -space radius and thus the resolution. The fraction of signal acquired during the RF pulse is defined as the ratio of pulse duration and acquisition time, $f_{\text{RF}} = T_p / T_{\text{acq}}$ (Fig. 1a) (10). Before performing correlation and reconstruction, the signal acquired after the RF pulse were accordingly weighted to compensate for the difference in receiver duty cycle. All GM-SWIFT excitation pulses used in the simulations and experiments were based on the HS-shaped gradient modulations, as shown in Eq. [2] using $n_p = 4$ with different modulation factors g_m . As mentioned, the excitation bandwidth B_w is also time dependent when using the time-dependent gradient. Thus, the parameter $B_{w-\text{max}}$ is used to describe the maximum excitation bandwidth as per TR period.

Simulations

Numerical simulations were done based on the Bloch equations (35). The simulated 1D object was composed of 1,000 isochromats distributed evenly. The gapped HS2 RF pulse shape was oversampled by a factor of 16, as typically done in SWIFT (32). The frequency modulation of the RF pulse was calculated according to Eq. [1] for the given $F_1(t)$ and $G(t)$. Simulations were carried out for different g_m (from 0.1 to 0.9) and compared for g_m equal to zero (standard SWIFT). The power spectral density of the pulses was calculated as the square of the FT of the RF pulses.

Simulations of the 1D point-spread function (PSF) were also performed. 1D spectral line broadening due to transverse relaxation was modeled in the simulations as an exponential decay between sampled data points. The procedures introduced in the Theory section were then used to transform the k -space data to the image domain to obtain the PSF.

Experiments

To validate the theory presented above and to evaluate image quality, GM-SWIFT with different gradient modulation factors were performed on a resolution phantom. The resolution phantom is a custom machined plastic (acrylic) container filled with MnCl_2 solution ($T_1 \sim 280$ ms). Images of the resolution phantom with different modulation factors ($g_m = 0, 0.8$ and 0.9), different $B_{w-\text{max}}$ (62.5 kHz and 31.25 kHz), and different f_{RF} (0.5 and 0.25) were acquired.

A T_2^* phantom was also imaged. The T_2^* phantom was made of seven water-filled tubes, each doped with a different concentration of MnCl_2 , immersed in deionized water in a bigger container. The T_2^* values of the phantom were then determined individually by spectroscopy. To further evaluate the performance of GM-SWIFT on ultrashort T_2^* spins, a single human molar tooth was imaged. The tooth was immersed in fluorine fluid (Fluorinert, FC-770, 3M, St. Paul, MN) for matching magnetic susceptibilities.

Human studies were performed according to the procedure approved by the institutional review board of the University of Minnesota Medical School. The brain and ankle of a healthy volunteer were scanned using $B_{w\text{-max}} = 62.5$ kHz and 96 kHz, respectively, and at different g_m and flip angles. Additionally for the ankle imaging, an off-resonance saturation pulse applied at +1 kHz from the central frequency was used as a magnetization preparation pulse to saturate signal from short T_2^* spins (10). Subtracting images acquired with and without the off-resonance pulse would yield an image dominated by short T_2^* signals.

Tooth imaging was performed using a 31-cm 9.4 T magnet interfaced to a DirectDrive console (Agilent Technologies, Santa Clara, CA) and a homemade surface coil. All other experiments were performed using a 90-cm 4T scanner (Oxford Magnet, Siemens SC72 gradient, Erlangen, Germany) interfaced to a DirectDrive (Agilent Technologies, Santa Clara, CA) console and a quadrature transverse electromagnetic volume coil.

The sound pressure level was also measured using an integrating sound level meter (model 2239A, Bruel & Kjaer, Naerum, Denmark).

RESULTS

Simulations

By visual inspection, negligible differences can be seen between simulated projections obtained with different levels of the gradient modulation (Fig. 2a). The case of $g_m = 0$ corresponds to standard SWIFT. The power spectral density (Fig. 2b) shows that the power was distributed uniformly across the pulse excitation bandwidth in the $g_m = 0$ case. As g_m increases, more power is concentrated in the center frequency area where most spins are excited. The relationships between g_m versus normalized SAR, RF amplitude, and acquisition time are shown in Figure 2c. SAR, RF amplitude, and T_{acq} all decrease as g_m increases, while other sequence parameters are held constant. For example, when $g_m = 0.9$, GM-SWIFT reduces SAR, RF amplitude, and T_{acq} by 90%, 70%, and 45%, respectively, as compared to standard SWIFT. The ability to decrease T_{acq} with GM-SWIFT is a direct consequence of the improved RF power efficiency of GOIA. That is, T_p decreases as g_m increases, with any fixed set of values for flip angle, RF amplitude, and spatial resolution.

Resolution Phantom Experiments

Images of the resolution phantom were acquired for the purpose of comparing the quality of images acquired with standard SWIFT versus GM-SWIFT. As shown in Figure 3a, GM-SWIFT images acquired with large gradient modulation in general have similar quality compared to standard SWIFT images. Intensity profiles (Fig. 3b) of the selected area confirm that standard SWIFT and GM-SWIFT provide similar image resolution. Notably,

image intensity arising from the plastic container (red arrows in Figs. 3a,b) is more intense in the standard SWIFT image as compared to the GM-SWIFT images. The latter is due to the different sensitivity of the methods to the ultrashort T_2^* of the plastic container, as will be discussed later.

GM-SWIFT with $g_m = 0.8$ can save about 50% of the RF amplitude, which can be accomplished in standard SWIFT by a 50% reduction in bandwidth. However, when comparing standard SWIFT images acquired with $B_w = 31.25$ kHz with GM-SWIFT images acquired with $B_{w-\max} = 62.5$ kHz at same flip angle, the GM-SWIFT images appear sharper, and susceptibility artifacts from plastics and air bubbles are reduced (yellow arrows in Fig. 3c).

T_2^* Sensitive Range

Figure 4 presents plots of the signal-to-noise ratio (SNR) versus T_2^* for different g_m values, along with the corresponding images of the T_2^* phantom. The noise signal was measured outside of the object (background signal). In general, the SNR is lower for shorter T_2^* s. The GM-SWIFT image with $g_m = 0.8$ (blue line), when compared to the standard SWIFT image (see green line), has higher SNR for $T_2^* > 400$ μ s and lower SNR for $T_2^* < 400$ μ s, whereas the SNR is nearly zero for $T_2^* < 100$ μ s. The SNR improvement of GM-SWIFT for tubes containing spins with $T_2^* > 400$ μ s occurs due to reduced background signal from the poorly resolved tubes containing spins with extremely short T_2^* (< 100 μ s). Although GM-SWIFT lacks sensitivity to spins with $T_2^* < 100$ μ s (marked as red region), it still preserves signal from spins with $T_2^* \sim 200$ μ s (marked as blue region). This point is exemplified by the tooth images shown in Figures 5a–d, in which the T_2^* of dentin is known to be about 200 μ s (36). GM-SWIFT with $g_m = 0.8$ can still preserve the signal from dentin and can delineate the shape of dentin to a comparable level as SWIFT, although with slightly increased blur (yellow arrows). To demonstrate the insensitivity of GM-SWIFT to extremely short T_2^* spins, a piece of silicone rubber with $T_2^* < 50$ μ s was placed on top of the phantom. As can be seen in Figures 5e and 5f, the rubber is visible and poorly resolved in the SWIFT image but is invisible in the GM-SWIFT image (red arrows). Due to the reduced visibility of the plastic containers in the GM-SWIFT case, the background is also much cleaner in the GM-SWIFT image than in the SWIFT image.

Results from simulations investigating the PSF of GM-SWIFT are shown in Figure 6. The amplitude of the PSF decreases as g_m increases due to the fast signal decay (Fig. 6a), whereas the full width at half maximum of the PSF increases only slightly as g_m increases, despite the ultrashort T_2^* (400 μ s) (Fig. 6b). When comparing the PSF of GM-SWIFT with that of standard SWIFT at lower B_w with the same RF power efficiency, it can be seen that GM-SWIFT has better resolution and higher signal level. This agrees well with the results of the phantom experiments (Fig. 3c).

Human In Vivo Experiment

Human brain images from standard SWIFT and GM-SWIFT are shown in Figure 7. The images in Figures 7a and 7b appear more like proton density-weighted images due to the low flip angle (2° for Fig. 7a; 4° for Fig. 7b) used. In comparison, the GM-SWIFT images

acquired with matched RF power exhibit good T_1 -weighted contrast. The RF power delivered in these two cases (Figs. 7b,c) was identical. According to the Bloch simulation, the flip angle for GM-SWIFT with g_m equal to 0.9 is about three-fold higher than in the standard SWIFT case. This high flip angle is frequently needed to obtain a good T_1 contrast between the white and gray matter in the brain.

Human ankle images acquired by standard SWIFT and GM-SWIFT are shown in Figure 8. In tendon, T_2^* is ~ 2 ms and T_1 is on the order of several hundred milliseconds (37). The Achilles tendon usually appears dark in gradient echo images (Fig. 8d). As in the phantom experiments, image quality of the ankle in Figure 8 is similar with GM-SWIFT and standard SWIFT acquisitions when performed with the same flip angle. The Achilles tendon is visible in all SWIFT and GM-SWIFT images. However, it is better delineated in the GM-SWIFT image with matched power than in the standard SWIFT image (red arrows) because the higher flip angle achieved is closer to the Ernst angle for tendon. With an off-resonance saturation pulse, the tendon spins with short T_2 are saturated, whereas the surrounding slowly relaxing spins are minimally affected. Subtracting GM-SWIFT images acquired with and without the off-resonance pulse yields an image dominated by short T_2^* signals (Fig. 8e). In the subtraction image, a low level of intensity is also seen in the muscle, which is most likely due to a magnetization transfer effect.

The A-frequency-weighted equivalent continuous sound pressure level (L_{eq}) of GM-SWIFT at different g_m values is shown in Figure 9. Standard SWIFT without any gradient modulation is quiet and has L_{eq} at around 52.5 dB. With gradient modulation, L_{eq} increases with g_m , but remains less than 75 dB. From Figure 9, it can be seen that the absolute sound pressure level is approximately linear with the gradient modulation. As compared with the typical L_{eq} values of common fast MRI sequences in the range of 100 to 110 dB (38), GM-SWIFT is still much quieter.

DISCUSSION

In this work, a modified SWIFT sequence with gradient modulation called GM-SWIFT is introduced. The corresponding generalized correlation procedures to reconstruct images for GM-SWIFT are developed. Bloch simulations and experimental data verified the theory and demonstrate the imaging capabilities of GM-SWIFT.

The signal acquired in a SWIFT acquisition is contaminated by the RF pulse; therefore, it needs the correlation step to resolve the resonating signal from the RF pulse signal. As mentioned, some extra steps are needed in the correlation procedures for GM-SWIFT over the standard one, because in the case when the field gradient is time-varying, the original correlation procedure is not applicable. The changing gradient has an influence on the time when the spin gets excited and the evolution after excitation. The accumulated phase of a given spin is not only dependent on the time interval that it evolves but also on the specific time point when it is excited. Because the excitation phase is mixed along with the acquisition phase in the SWIFT scheme, the concept of excitation k-space as defined in Eq. [5] (31) is a good tool for dealing with the phase problem here.

The ability of GM-SWIFT to reduce SAR is beneficial for imaging of humans. Achieving T_1 -contrast requires high flip angle; therefore, it can be limited by hardware or SAR or requires a magnetization preparation module that reduces acquisition efficiency. Due to its increased flexibility in setting the flip angle, GM-SWIFT is capable of producing T_1 contrast without magnetization preparation. Also, for imaging of fast relaxing spins, high excitation bandwidth is required but may not be attainable with standard SWIFT due to hardware (peak RF power) and SAR constraints. As a result, GM-SWIFT is a promising variation of SWIFT, particularly for human applications, by allowing reduced SAR, overcoming peak power limitations of the hardware, and producing often needed T_1 -weighted contrast. Furthermore, GM-SWIFT still produces much reduced sound level than conventional sequences (Fig. 9) because the gradient polarity need not be inverted quickly but instead is modulated smoothly below the gradient slew rate limit. Also, although the GOIA approach was used in this paper to generate RF pulses, other gradient modulation approaches such as VERSE (22) can also be applied to modify the RF pattern as long as they can achieve the desired flat excitation profile.

The simulations and phantom studies showed GM-SWIFT is able to achieve comparable image quality as standard SWIFT, even when using a gradient-modulation factor as high as 0.9. One consequence of gradient modulation is the reduced sensitivity of GM-SWIFT for ultrashort T_2^* spins as the gradient-modulation factor increases. However, the ability to alter the T_2^* sensitivity range with GM-SWIFT can be a benefit in certain cases. Standard SWIFT with constant gradient and large band-width is able to capture signal from spins with T_2^* as short as several microseconds; and as such, essentially all spins are present in SWIFT images, including plastic materials commonly used in coil construction. However, due to limited bandwidth, in practice these signals cannot be resolved and they manifest themselves as a halo over the whole image. With large gradient modulation, GM-SWIFT is less sensitive to spins with extremely short T_2^* (several to tens of microseconds); as a result, it generates images with a cleaner background. At the same time, GM-SWIFT with high $g_m = 0.9$ still retains the ability to resolve spins with T_2^* longer than 200 μ s but at the expense of some loss of image resolution (Fig. 6). The latter occurs in GM-SWIFT with HS-shaped gradient modulation because the fast decaying spins experience the low bandwidth portion of the GOIA pulse immediately after excitation. Compared to the constant gradient condition, some k-space information is lost in this process, which causes reduced sharpness in images of ultra-fast relaxing spins. Although most demonstrations in this article utilized high gradient modulation factor ($g_m = 0.8$ or 0.9), GM-SWIFT can also work with a medium or small g_m value. In practice, by increasing g_m just enough to satisfy SAR, peak RF amplitude, and/or T_1 -weighting requirements, undesirable effects due to an increased T_2^* -sensitivity can be minimized.

In addition to reducing SAR, peak RF amplitude, and acquisition time over standard SWIFT, GM-SWIFT provides other flexibilities. For example, besides the HS-shape used herein, other gradient modulation shapes (e.g., a linearly increasing function) can be used in GM-SWIFT for other purposes, such as increasing the bandwidth of the acquisition period after the RF pulse. The generalized correlation procedures introduced here should also be suitable for other types of gradient modulation.

Of note, in the limit $g_m \rightarrow 1$, GM-SWIFT resembles UTE. However, the gradient in UTE always starts from zero amplitude, whereas GM-SWIFT offers flexibility to operate with an optimal g_m value (i.e., the minimum value) that just satisfies some practical constraints (e.g., SAR, tolerated sound level, and unresolved background signal suppression), while minimizing image blur and SNR loss for short T_2^* spins. Compared with UTE, the requirement to switch fast between transmit and receive modes is more stringent with SWIFT, which currently limits its achievable bandwidth. However, GM-SWIFT reduces the bandwidth during excitation, which makes an increased switching time permissible such as in UTE. Hence, GM-SWIFT should be easier than SWIFT to implement on clinical scanners that cannot switch rapidly between transmit and receive modes.

As with most sequences that use gradient modulation, GM-SWIFT is sensitive to gradient performance. The gradient needs to be calibrated (39) so that the gradient timing error in different directions needs to be under several microseconds. A global delay in all directions can be compensated in the correlation step, but the occurrence of different delays in each direction introduces distortion on the shape of the gradient modulation, which produces bright/dark line artifacts at the edges of an object. A detailed discussion about the influence of gradient performance on GM-SWIFT is beyond the scope of this paper.

CONCLUSION

In summary, GM-SWIFT can achieve reductions of SAR, RF amplitude, or acquisition time, and provides control over the T_2^* -sensitive range for producing a clean background in images. The tradeoff is some reduction in the resolution of images of ultrafast relaxing spins, depending on the level of the gradient modulation used. By manipulating the gradient, GM-SWIFT can maximize the efficiency of RF power and provide a way to optimize the settings that balance RF amplitude, SAR, scan time, and image quality considerations.

Acknowledgments

Grant sponsors: National Institute of Biomedical Imaging and Bioengineering (NIBIB), National Center for Research Resources (NCRR), and National Center for Advancing Translational Sciences (NCATS) of the National Institutes of Health; Grant numbers: P41EB015894, S10RR023730, S10RR0027290, and KL2TR000113; Grant sponsor: WM KECK Foundation.

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FIG. 1. Pulse diagrams of (a) standard SWIFT and (b) GM-SWIFT with HS-shaped gradient modulation.

FIG. 2.

Simulation results. (a) Reconstructed 1-D images (object profiles) for different g_m values with the generalized (for $g_m = 0$) or original (for $g_m = 0$) correlation procedures. (b) Power spectral density of the RF pulses with different g_m . (c) Simulation results showing the relationships between the gradient-modulation factor g_m and SAR, RF amplitude, or acquisition time. The flip angle was fixed in the simulation investigating the relationship between SAR or RF amplitude with g_m . The RF amplitude, flip angle, and resolution were fixed in the simulation investigating the relationship between T_{acq} with g_m .

FIG. 3.

Images and intensity profiles of the resolution phantom. (a) Phantom images obtained with different g_m using $f_{RF} = 0.25$ and $B_{w-max} = 62.5$ kHz. (b) Intensity profiles corresponding to the yellow dash lines in (a). (c) Phantom images with different g_m and B_{w-max} using $f_{RF} = 0.5$. Here, flip angles for all images were set to 4° . GM-SWIFT images acquired with large gradient modulation have comparable appearance when compared to standard SWIFT images. Standard SWIFT at $B_w = 31.25$ kHz has the same power efficiency with GM-SWIFT images with $B_{w-max} = 62.5$ kHz and $g_m = 0.8$. When comparing the images of these two conditions, GM-SWIFT images appear sharper and susceptibility artifacts from the plastic and air bubbles are reduced (yellow arrows).

FIG. 4. (a) Plot of SNR (signal to background noise ratio) dependence on T_2^* using different modulation factors. (b) and (c) are images of the T_2^* phantom.

FIG. 5.

(a, b, c, and d) Images of a tooth. (b) and (d) are images with increased intensity scaling. The signal from the dentin is sufficient in both SWIFT and GM-SWIFT cases. GM-SWIFT with $g_m = 0.8$ is slightly more blurred (yellow arrows). (e) and (f) are images of the resolution phantom with a piece of silicone rubber placed on top of it (red arrows). The rubber is visible in the SWIFT image; however, due to the ultrashort T_2^* value, the rubber is not resolved well. Also, a halo over the whole image can be observed which comes from the plastic materials of the phantom. The image for GM-SWIFT (f) on the other hand is cleaner; the rubber and the container are both invisible.

FIG. 6.

(a) PSF for standard SWIFT and GM-SWIFT with $T_1 = 1000 \text{ ms}$ and $T_2 = 0.4 \text{ ms}$ at different B_w or gradient modulation factor g_m . (b) PSF in (a) with normalized amplitude. The standard SWIFT with $B_w = 62.5 \text{ kHz}$ has the same power-to-flip-angle efficiency as GM-SWIFT with $B_{w\text{-max}} = 125 \text{ kHz}$ and $g_m = 0.8$.

FIG. 7.

In vivo human head images. (a) and (b) are standard SWIFT images with flip angle 2° and 4° . (c) is GM-SWIFT image with factor $g_m = 0.9$. The RF power delivered in (b) and (c) are the same. Better T_1 contrast is observed in the GM-SWIFT images because the flip angle achieved could be increased to as high as 12° .

FIG. 8. In vivo human ankle images. (a, b, and c) are standard SWIFT or GM-SWIFT with fat saturation. (d) GRE image. (e) Subtraction result between GM-SWIFT image without and with off resonance saturation.

FIG. 9.

The frequency-weighted equivalent continuous sound pressure level L_{eq} of GM-SWIFT with different gradient-modulation factors, g_m .