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RF Phase Encoded Half-pulses in Simultaneous Multi-slice UTE Imaging

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Abstract

Purpose: To describe a simultaneous multi-slice (SMS) ultra-short echo time (UTE) imaging method using RF phase encoded half-pulses in combination with power independent of number of slices (PINS) inversion recovery (IR) pulses to generate multiple-slice images with short T_2^* contrasts in less than 3 min with close to an 8-fold acceleration compared to a standard 2D approach.

Methods: RF phase encoding is applied in a simultaneous multi-slice (SMS) ($N_{SMS} = 4$) excitation scheme using “sinc” half-pulses. With the use of coil sensitivity encoding (SENSE) and controlled aliasing in parallel imaging (CAIPI) in combination with a gradient echo 2D spiral readout trajectory and IR PINS pulses for contrast enhancement a fast UTE sequence is developed. Images are obtained using a model-based reconstruction method. Sequence details and performance tests on phantoms as well as the heads of healthy volunteers at 3T are presented.

Results: An SMS UTE sequence with an undersampling factor of four is developed using RF phase encoded half-pulses and PINS IR pulses which enables the acquisition of 8 slices at 0.86^2 mm² resolution in less than 3 min scan time. UTE images of the head are obtained showing highlighted signal of cortical bone. Image quality and T_2 contrast are comparable to the one obtained by corresponding single slice acquisitions with only minor deviations.

Conclusions: The method combining RF phase encoded SMS half-pulses and PINS IR pulses presents a suitable approach to SMS UTE imaging. The usage of coil sensitivity information and increased sampling density via interleaved slice group acquisitions allows to reduce the total scan time by a factor close to eight.

Keywords

Half-pulse; Simultaneous Multi-Slice excitation; Ultra-short echo time imaging; UTE; SMS; CAIPI and PINS

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INTRODUCTION

Conventional MRI methods usually target protons of unbound water that can move relatively freely in larger compartments like cells or extracellular spaces and, hence, possess a rather long transversal relaxation time (T_2). Often times, however, important structural features such as in collagenated tissue of the musculoskeletal system, cortical bone, or cartilage and tendons contain tissue with encapsulated water molecules that possess T_2^* values of a few milliseconds or less (1). Imaging based on these short T_2 protons can be very insightful, however, they do not significantly contribute to the MR signal of standard acquisitions (2,3).

In recent years, several techniques classified as zero and ultra-short echo time (ZTE and UTE) imaging as well as sweep imaging with Fourier transformation (SWIFT) methods have been developed to target short T_2 protons (4–8). Depending on the T_2 domain of the protons of interest different approaches are used. High-resolution images with good image quality have been obtained for cases focusing on tendons and cartilage using a slab selective 3D spiral acquisition (9). Imaging the more extreme range of T_2^* values well below 1 ms as in cortical bone or myelin, however, is even more challenging and demands alternative approaches to keep the excitation, readout and echo time (TE) as short as short as possible (10–12). Furthermore, specific/selective UTE imaging requires methods to suppress the much more intense long T_2^* signal from human tissue (13). The latter is often achieved by spin preparation techniques such as saturation (14) or adiabatic inversion pulses (15,16), T_2 selective excitation schemes (17), and/or by the subtraction of in-phase gradient-echoes (18). Depending on the method used a “short T_2 ” contrast is generated that is to some degree (often considerably) influenced by the T_1 characteristics of the tissue.

Slice-selective UTE sequences that target the extreme range of short T_2 protons typically make use of half-pulses, which do not require a refocusing gradient and are, therefore, compatible with minimal TE acquisitions (19). However, the summation of two full half-pulse measurements with opposite gradient polarity is required to obtain a single-slice image, which adds to the general problem of relatively long acquisition times for a single slice or yet a full coverage of a desired volume. Another drawback is the relatively low signal strength of slice-selective acquisitions for specific short T_2 imaging, which is related to the fast signal decay (transversal relaxation) as well as the imperfection of long T_2 suppression methods that partially attenuate the desired short T_2^* signal as well (13). Although progress has been made, it remains very challenging to obtain images of sufficient quality for clinical application in a reasonable time frame.

Simultaneous multi-slice (SMS) imaging has proven to be a powerful method to reduce scanning time and is used in a range of different MRI techniques today (20). UTE imaging, however, has not benefited from it so far even though the concept of combining RF phase encoding with UTE half-pulses for SMS excitations has been described before (21). The RF phase encoding approach is based on early SMS methods such as Hadamard encoding (22) and phase-offset multiplanar (POMP) (23) techniques that solely relied on RF encoding to separate simultaneously acquired slices. In these early methods the acquisition of a full set of N_{SMS} phase encoding repetitions was required and no acceleration was achieved directly.

However, the increase in volume that was excited during each shot lead to an improvement of the signal-to-noise ratio (SNR).

In modern parallel imaging techniques coil sensitivity information is typically combined with partially encoded k-space data to maximize performance. The additional use of coil sensitivity information allows one to obtain images with minimal aliasing artifacts from undersampled k-space data and as a result the scan time of the acquisition is reduced directly. Efficient (under-)sampling strategies such as controlled aliasing in parallel imaging (CAIPI) further enhance the performance of the of slice separation and higher undersampling/acceleration factors can be achieved (24). In these cases partial slice encoding is realized either via z-gradient or RF phase encoding. In view of an UTE imaging application RF phase encoding is advantageous since it does not require any additional time to play out z-gradients between excitation and readout, which would lead to an increase in the minimal echo time achievable.

In this work the aim was to reduce the acquisition time of an inversion recovery (IR) UTE half-pulse sequence using SMS excitation with RF phase encoding and coil sensitivity information for multi-slice coverage using an interleaved spiral multi-shot acquisition.

THEORY

Sequence Design using PINS Pulses for IR SMS UTE Imaging

The main aspect of this approach was to combine (A) the use of parallel imaging in a CAIPI type undersampled SMS acquisition using RF z-phase encoded half-pulses with (B) an interleaved sequence timing in which consecutive SMS slice groups are being acquired within the inversion recovery period of the sequence. The combination of both demands the use of multi-slice-selective inversion recovery pulses that only invert the slice regions of interest to ensure homogeneous contrast. Therefore, power independent number of slices (PINS) pulses were used as IR pulses (25,26). The use of a multi-leaf spiral readout trajectory further improves the efficiency of k_{xy} -space coverage.

For a single SMS slice group, a fully sampled RF phase encoded SMS k-space consists of N_{SMS} pairs of half-pulse excitations for each spiral leaf ($2 \times N_{SMS}$ excitations). The sequence timing was structured in a way that both half-pulse (hp) excitations were acquired immediately after another, followed by the remaining ($N_{SMS}-1$) RF z-encodings to complete a dataset for one single spiral leaf (see Figure 1, A). This pattern was repeated for all spiral leaves to give the entire data for one slice group. To reduce the scan time, these slice group acquisitions were interleaved within the inversion time of the PINS pulse. Figure 1, A shows the case for two interleaved slice groups marked in red and blue. The amount of slice groups that can be interleaved depends on several sequence parameters. Among them are the inversion time (TI), the time-to-repetition (TR) as well as the slice thickness and number of slice groups to be measured, but excitation profiles and SAR considerations are limiting factors as well.

Simultaneous Multi-Slice Imaging Using RF Phase Encoding in 3D k-space

In single-slice imaging the k-space encoding problem is reduced to the two in-plane dimensions (x, y) by the slice-selective excitation, since it removes the ambiguity of signal origin in the slice-dimension (z). However, with a simultaneous excitation of multiple slices additional information is necessary to detangle the signal from the individual slices. Several different approaches can be used to overcome this problem, which either use RF phase encoding (POMP), gradient encoding, coil sensitivity information (SENSE, GRAPPA) (27–29) or combinations of thereof (CAIPI) (30–32).

The RF encoding scheme used in this work is based on POMP imaging but differs in a few aspects. The encoding used in POMP sequences distributes the simultaneously excited slices in an extended field-of-view of the y-phase (FOV_y) direction and, thereby, remains in terms of reconstruction a 2D acquisition. Instead of shifting slices in y-phase direction, a 3D treatment of SMS RF encoding as proposed previously presents a more straightforward and general approach (33). As simultaneously excited slices actually do differ in their location in the slice direction, SMS imaging can be described in simple analogy to “true” volumetric 3D imaging. At Nyquist sampling the FOV, number of slices N_{SMS}/N_{3D} , the k increment and the slice separation $\Delta z_{\text{SMS}/3D}$ are related by:

$$\Delta z_{\text{SMS}/3D} \cdot \Delta k = \frac{1}{N_{\text{SMS}/3D}} \text{ and } FOV = \Delta z_{\text{SMS}/3D} \cdot N_{\text{SMS}/3D} \text{ and } k_{\text{max}} = \Delta k \frac{N_{\text{SMS}/3D}}{2}$$

Conventional 3D imaging methods use z-gradients at the beginning of the readout to obtain z-phase encoding information. However, in UTE imaging z-gradient encoding is undesirable due to the time required to play out the z-gradients after excitation during which the short T_2 signal would have declined significantly. Alternatively in analogy to POMP imaging, the slice encoding can be incorporated into the excitation using RF phase encoding. Each individual slice is excited with a specific phase simultaneously in a way that the phase pattern of the excited slices corresponds to the desired z-phase encoding (33). This way, in combination with intrinsically refocused half-pulses, 3D encoding of the simultaneously acquired slices can be realized without the penalty for longer TEs. A fully encoded measurement requires a minimum of N_{SMS} different encoding repetitions. In the case of UTE imaging using half-pulses each phase encoding excitation is further split up into N_{SMS} pairs of half-pulse excitations ($2 \times N_{\text{SMS}}$) (21).

In the case of multi-leaf spiral readout acquisitions a 3D stack-of-spirals k-space dataset is generated for each of the slice groups. The use of coil sensitivity information and incoherent undersampling strategies can be applied to reduce the acquisition time of the method while controlling aliasing artifacts to maintain a high image quality. Therefore, a fraction of spiral leaves are selected in each k-space plane to form a rotated 3D CAIPI type undersampling pattern (see Figure 1, B).

Pulse Design of RF Phase Encoded Half-pulses

In view of the small tip angle approach a simple way to generate these phase encoded multiband pulses is by summing each single slice excitation with its given slice offset and phase modulation (z and φ_n) as well as the Wong phase cycling scheme (ϕ_{Wong}) (34),

$$RF_{\text{MB}}(t, m) = \sum_{n=1}^{N_{\text{SMS}}} A(t) \cdot e^{i\{\Delta\omega(n)t + \varphi(n, m) + \phi_{\text{Wong}}(n)\}}$$

$$\Delta\omega(n)t = k_z \Delta z(n)$$

$$\varphi(n, m)_n = \text{rem} \left[\frac{2\pi \left(n - \frac{N_{\text{SMS}}}{2} \right)}{N_{\text{SMS}}} \cdot \left(m - \frac{N_{\text{SMS}}}{2} \right), 2\pi \right], \quad \text{for } m = 1, 2, \dots, N_{\text{SMS}}$$

where N_{SMS} denotes the number simultaneously excited slices, $A(t)$ the full sinc RF pulse envelope and m the counter for the different phase encodes. The multiband pulses are then separated into two halves, which are further modulated to match the corresponding trapezoid z -gradient to give the variable-rate selective excitation (VERSE)(35) half-pulses (see Figure 2). In order to reduce the RF peak power of the relatively short and highly modulated half-pulses, the Wong phase cycling scheme (ϕ_{Wong}) was added as phase modulation in addition to the slice phase encoding. Since the same phase cycle scheme is used for each of the phase encoding half-pulses it does not interfere with the RF phase encoding. The use of Wong phases reduces the peak amplitude of the RF pulses considerably even though it may not be optimized for the case of additional RF phase encoding (Figure 2 shows the RF half-pulses without Wong phases to display the RF phase encoding pattern).

METHODS

Pulse Design of RF Phase Encoded Half-pulses

A series of phase modulated SMS half-pulses with $N_{\text{SMS}} = 1, 2$ and 4 were designed using the small tip angle approximation. For each SMS excitation profile ($N_{\text{SMS}} = 1, 2$ and 4) the corresponding number ($1, 2$ and 4 , respectively) of pairs of differently phase modulated half-pulses was generated according to Equation 1. In each of these excitations (pairs of half-pulses) the individual slices are excited with different phases corresponding to the phase encoding pattern of the desired k_z -space plane. A representative set of half-pulses with $1160 \mu\text{s}$ duration and a set of *sinc* pulses ($3410 \mu\text{s}$) for comparison were used in this study (ramp-sampled VERSE pulses for total number of 3 mm thick slices $N_{\text{tot}} = 32$, max. gradient power = 2 G/cm , max. slew rate = 13000 G/cm/s and a sampling period = $1 \mu\text{s}$). Figure 2 shows an example of all four pairs of half-pulses for an $N_{\text{SMS}} = 4$ pulse (*half sms4*) including the corresponding excitation profiles and the phase encoding scheme obtained by Block equation simulations in Matlab (Mathworks, Natick, MA). For demonstration

purposes the Wong phase modulation has been omitted from pulses shown in Figure 2. A complete depiction of the final pulse and gradient shapes is given in the Supporting Information Figure S1. In order to account for gradient as well as RF/G_z timing imperfections of the scanner system, a basic gradient correction and adjustment of RF/G_z timing was performed (for more details see Supporting Information Figure S2) (36).

Pulse Design of Hyperbolic Secant PINS Pulses

Two Hyperbolic Secant PINS pulses *PINS1* and *PINS2* were designed that lead to multi-slice excitation with varying slice thickness with A_0 being the maximum B_1 field, β the modulation angular frequency and μ a dimensionless parameter used to adjust the adiabatic threshold of the pulse. (25,26) Parameters for *PINS1* (A): $A_0 = 0.0328 \mu\text{T}$, $\beta = 18688 \text{ rad/s}$, $\mu = 5$, pulse length = 11.8 ms, 40 rectangular pulses giving a slice width of 0.5 cm and slice separation of 2.4 cm; *PINS2* (B): $A_0 = 0.0325 \mu\text{T}$, $\beta = 14637 \text{ rad/s}$, $\mu = 8$, pulse length = 11.8 ms, 40 rectangular pulses giving a slice width of 0.6 cm and slice separation of 2.4 cm. The excitation profile was evaluated using a Bloch equation simulation and further investigated experimentally.

MRI Measurement General Information

MRI measurements were performed on a Siemens MAGNETOM Prisma 3T (Erlangen, Germany) scanner equipped with a 52-channel head coil for phantom and head images.

Sensitivity Map Prescan

A standard low-resolution (64×64 pixels) gradient echo scan was performed (14–16 s duration) prior to the corresponding SMS UTE scans, which was used in the SENSE reconstruction to generate a coil sensitivity map using the SenseMap Matlab code (37). The low-resolution sensitivity maps were up-scaled to fit the 256×256 matrix size of the UTE images.

Slice Profile Measurements

RF excitation as well as IR PINS pulse profiles were measured using a modified standard 2D fast low angle shot (FLASH) (38) on a uniform spherical phantom. In the case of the IR pulses, the second excitation was realized with a sinc pulse (slice thickness = 5 mm). The IR pulses to be analyzed were set to excite a y-z plane (G_x gradient blips) while encoding was performed in the x-y plane. For Half-pulse profiles encoding was done in the y-z plane. Data were acquired with a 60° excitation angle (90° for water/fat phantom) for PINS pulse testing and with a 30° excitation angle for half-pulses, over a 22 cm FOV_{xy}, a 512×512 image matrix size, TE = 5.1 ms, TR = 1500 ms and gradient spoiling. For UTE phantom and in vivo slice profile measurements a single slice spiral UTE sequence with an optional IR PINS pulse (TI=100 ms), a 22 cm FOV_{xy}, a 256×256 image matrix size, TE = 100 μs , TR = 300 ms and gradient spoiling was used. Images were reconstructed offline using a FFT or the model-based reconstruction described below in Matlab.

Structural Phantom Measurements

Data were acquired applying a 30° excitation using the different sinc and sinc half-pulses described above ($N_{\text{tot}} = 16$, slice thickness $w = 3$ mm, slice gap of 100%) with a 2D spiral trajectory (256 leaves, max. slew rate = 13000 G/cm/s, max. gradient power = 2.3 G/cm for the spiral readout and 2.5 G/cm for the rewinder) over a 22 cm FOV_{xy}, a 256×256 image matrix size, TE = 100 μ s (for half-pulses), TR = 500 ms and gradient spoiling. The sequence timing shown in Figure 1 was applied without the use of an IR pulse.

UTE Phantom Measurements

Data were acquired using the same sequence as for the structural phantom images with the addition of an inversion recovery (IR) PINS pulse (TI = 100 ms). Due to the interleaved slice acquisition the IR were played out staggered in time to match the slice acquisition timing. In order to avoid slice cross-talk, a slice gap of 167% and 300% was applied. The sequence was tested on a phantom containing differently concentrated aqueous MnCl₂ solutions (208.0, 104.0, 52.0, 26.0, 13.0, 6.5, 3.2, 1.6, 0.8, 0.4, 0.2 mM) with corresponding T₂ values of 0.041, 0.082, 0.160, 0.330, 0.660, 1.3, 2.643, 5.3, 10.6, 21.1, 42.3 ms (39).

In Vivo Imaging of the Head

In vivo MRI scans were performed on healthy volunteers. The internal review board approved the protocol and written consent was obtained for each participant. Data were acquired applying a 30° excitation using half-pulse (*half sms4*, $N_{\text{tot}} = 8$, slice thickness $w = 3$ mm, slice gap of 300%) with the same 2D spiral trajectory as above but two gradient echoes at TE = 100 μ s and 2.3 ms (256 leaves, max. slew rate = 13000 G/cm/s, max. gradient power = 2.3 G/cm for spiral and 2.5 G/cm for rewinder) over a 22 cm FOV_{xy}, a 256×256 image matrix size, TR = 300 ms and gradient spoiling. For contrast enhancement, IR pulses *PINS1* and *PINS2* described above were applied with TI = 100 ms. The sequence timing shown in Figure 1 was applied. A short (14 s) gradient echo prescan was used to generate a sensitivity map for SENSE reconstruction as well as a mask (thresholding).

Image Reconstruction

The final images were obtained using a generalized reconstruction method using the Michigan Image Reconstruction Toolbox (MIRT) (<https://web.eecs.umich.edu/~fessler/code/>). The k-space trajectories and data were inserted directly into the signal equation and solved numerically using a non-uniform Fast Fourier Transform (FFT) within a regularized conjugate gradient algorithm (40,41). All reconstructions were performed offline using Matlab. The 3D spiral k-space was retrospectively undersampled for $N_{\text{SMS}} = 4$ using a CAIPI type sampling pattern (R = 2, 4, 8, 16, and 32). Images were compared to those obtained by a 2D single slice $N_{\text{SMS}} = 1$ acquisition reconstructed with SENSE and without (coil by coil). Coil sensitivity information was used from a gradient echo prescan.

RESULTS

Half-pulse Design and Excitation Profile

A set of RF encoded half-pulses was designed according to the method described above. Pulse profiles and corresponding Bloch equation simulation for pairs of $N_{SMS} = 4$ pulses are shown in Figure 2. The Wong phase cycling scheme is omitted to demonstrate the underlying RF phase encoding pattern. Only minimal out-of-slice excitation and deviations in phase is observed in the simulated profiles. The excitation profile of the corresponding half-pulses (no RF phase encoding) was further tested in phantom scans using a spherical water phantom and compared to the one of an $N_{SMS} = 4$ sinc pulse (see Figure 3). The experimentally obtained profile of the half-pulse pair shows noticeable deviations from the simulations and considerable out-of-slice excitation occurs while the sinc pulse excitation profile is comparable to the simulated one. The real and imaginary components of individual SMS half-pulse excitation profiles are shown in the Supporting Information Figure S3. Further investigations on the slice profile were carried out on a UTE phantom with $T_2 = 660$ μ s and 2.6 ms and the head of a healthy volunteer and are shown in Supporting Information Figures S4 and S5. A broadening of the slice profile is observed for protons with $T_2 = 660$ μ s which is significantly lower than the duration of the half-pulse itself. In case of the head scans the use of an IR PINS pulse leads to the suppression of the majority of signal in the regions of the inversion bands (long T_2 proton pool). Signal from out-of-slice excitation into non-attenuated regions, therefore, contributes significantly more to the overall signal than in non-IR acquisitions, which is largely enhancing problem out-of-slice artifacts in the IR UTE images.

Structural Phantom Tests

In a primary step, structural phantom images were obtained to test the performance of the RF phase encoding method using half-pulses leaving the challenges related to rapid signal decay of short- T_2 protons aside. Figure 4 shows the images of single-slice as well as simultaneous multi-slice acquisitions. Reference images were obtained using ramp-sampled sinc pulses for $N_{SMS}=1$ and 4 and were compared to images obtained with a series of 1.16 ms long half-pulses for $N_{SMS} = 1, 2$ and 4 using SENSE reconstruction. Overall, the image quality is good with only minor differences between the single-slice *sinc sms1* and SMS *sinc sms4* images and acceptable out-of-slice excitation/slice cross-talk in the case of the half-pulse excitations, even with an increasing SMS factor (see Figure 4 B, C). The addition of RF phase encoding and half-pulse repetitions leads to an increase in signal intensity as can be seen by the increasing difference of absolute magnitude images going from *half sms1* to *half sms4* (see Figure 4, B). Generally high noise levels can be observed in the region outside and around the head coil cage where the sensitivity map breaks down and correlate with the increase in signal intensity from *half sms1* to *half sms4*. The normalized difference images (see Figure 4, B) show larger deviations on the edges of the phantom's fine structure which is rapidly changing in respect to z . This can be explained by out of slice excitation as well as the generally thicker slice profile (see Figure 3). It is to be noted that some of the differences shown in B and C result from desirable short T_2^* signal of the phantom, which is only visible in the UTE half-pulse images.

In view of reducing the acquisition time of the UTE scan the undersampling behavior of *half sms1* and *half sms4* acquisitions were investigated. Therefore, both datasets were reconstructed using only a fraction ($R = 1, 2, 4, 8, 16, 32$) of the fully sampled spiral k-space leaves. In the case of the 2D single slice acquisition (*half sms1*) reconstruction was performed with the use of coil sensitivity information (*sms1_SENSE*) and without (*sms1*, coil-by-coil). For the SMS acquisition (*sms4_SENSE*) the 3D k-space was undersampled using a CAIPI type strategy (see Figure 1, B) combined with SENSE reconstruction. Results for one selected slice are shown in Figure 5, A–C. As anticipated, the use of coil sensitivity information in the reconstruction drastically increases the robustness of image quality towards undersampling (see Figure 5, A compared to B and C). A significantly smaller but considerable enhancement in image quality is observable going from row B to C for values for $R = 4$ and larger. For the SENSE reconstructed images (rows B and C) an undersampling factor of $R = 2$ leads to hardly any visible decline in image quality. Even at $R = 4$ only small streaking artifacts become apparent which were considered to be tolerable especially in the case of *sms4_SENSE*. Slice-leakage evaluation (42) and G-factor simulations (43) shown in the Supporting Information Figures S6 and S7 further support these findings.

PINS Pulse Design and Phantom Tests

In order to generate the desired short T_2^* contrast PINS pulses were designed that enable a multi-slice inversion recovery (IR) sequence with an interleaved slice acquisition. Therefore, two hyperbolic secant PINS pulses of 11.8 ms length were generated (see Figure 6): *PINS1* (left) and *PINS2* (right). Both pulses were tested regarding their excitation profile on water and fat phantoms. In the case of the spherical water phantom, a near full inversion over the region of interest with a good slice profile was obtained for both pulses (see Figure 6, B, C). *PINS1* was designed to generate a narrow inversion band to enable the acquisition of a larger amount of interleaved slice groups (N_{SG}) using a minimal slice gap. However, the slice shift caused by the frequency offset of the fat signal was found to be too severe so that the pulse is not capable of inverting both the fat and water spins simultaneously over the entire slice region of the SMS acquisition (red arrow Figure 7, A). Additional means of fat suppression may, therefore, be necessary for *in vivo* imaging using *PINS1*. *PINS2* on the other hand generates an inversion band wide enough to cover the slice profile for both fat and water signal and can be used in *in vivo* imaging at the cost of larger gaps between consecutive slices (see Figure 6, right side). Both PINS pulses were tested on a UTE water phantom in the UTE half-pulse sequence. The phantom consists of aqueous $MnCl_2$ solutions with T_2^* values of a few seconds (water) down to $41 \mu s$. Selected slices are shown for single-echo $N_{SMS} = 4$ UTE scans with *PINS1* and *PINS2* (see Figure 7, B). Images were reconstructed without the use of coil sensitivity information. Overall the results demonstrate the method's ability to image fast relaxing protons down to about $160 \mu s$ and that a suppression/attenuation of a specific T_2 -pool of protons is achievable using the IR pulses, *PINS1* and *PINS2*. It should be noted that the both, T_1 and T_2 values of the $MnCl_2$ solutions, decrease linearly with the $MnCl_2$ concentration and, therefore, low T_2^* values go in hand with low T_1 values in the UTE phantom which is generally not the case for *in vivo* imaging. Furthermore, the reason for the relatively long TE of $100 \mu s$ is mainly due to the shape of the 'corrected' input gradient waveform that contains a tail of $100 \mu s$. Shorter TEs can be achieved by truncating the tail at the expense of the quality of the slice profile.

In Vivo Images

Based on the phantom results a spiral dual echo IR SMS UTE sequence protocol was set up for a total of eight slices using the RF pulses *PINS2* and *half sms4* ($N_{\text{SMS}} = 4$). The sequence was tested in head imaging and compared to the corresponding eight sequentially acquired spiral single slices ($N_{\text{SMS}} = 1$) using a non-selective hyperbolic secant inversion pulse.

Figure 8 shows results of head images of a healthy male volunteer. Both gradient echoes as well as a magnitude difference are depicted for $N_{\text{SMS}} = 4$ (sms4_SENSE) with an undersampling factor of $R = 4$. Additionally, the magnitude difference images obtained with and without SENSE reconstruction are shown for the fully sampled reference scan (sms1_SENSE and sms1, respectively). Generally high contrast for cortical bone of the skull is obtained. However, the suppression of fat tissue by the *PINS2* is slightly less effective than in the reference scans and a relative strong fat signal surrounding the skull can be observed in the first echo image of sms4_SENSE. Besides, residual signal likely due to out-of-slices excitation occurs in the brain region as well. Echo subtraction reduces the undesired soft tissue signal but some residual fat signal remains. Generally the signal intensity is similar for the $R = 4$ undersampled sms4_SENSE and the reference images and apart from small deviations in the uppermost slices of the skull, the image quality is comparable.

The undersampling behavior of the head images for the SMS scan (sms4_SENSE) as well as the reference scan, reconstructed with and without SENSE (sms1_SENSE and sms1) is shown in Figure 9. The performances are very similar to the ones previously described for the phantom scans. Both SENSE reconstructions improve the stability towards undersampling (rows B and C compared to A). In the *in vivo* images, however, undersampling also leads to an increase in noisiness of the SENSE reconstructed images because of the generally low signal intensity. The effect is slightly less pronounced in the SMS images due the increase of signal strength ($\sqrt{N_{\text{SMS}}}$) compared the reference images.

Again, at an undersampling factor of four, images with tolerable loss of quality are obtained. The duration of the IR SMS UTE sequence with $R = 4$ is about 2:50 min (including sense map prescan) compared to about 20:30 min for the fully sampled reference scan (no sense map) or around 5:30 min when undersampled by a factor of four (incl. sense map prescan).

DISCUSSION

The first strategy presented to reduce the overall scan time of slice-selective UTE imaging was a highly condensed timing scheme that acquires spiral k-space data in an interleaved manner with a slice-selective IR pulse. Because of the high sampling density in combination with the large peak power of the highly modulated half-pulses and the IR PINS pulse, however, specific absorption rate (SAR) limits are readily reached setting a limit for the acceleration. With the parameters of the sequence at hand, mainly the short TR and flip angle, only two slice groups could be interleaved without frequently running into SAR violations.

A major challenge is that the condensed design of the sequence relies on well-defined slice/gap profiles for the inversion and excitation to produce high image quality. As shown in Figure 3 the slice profile of the half-pulse excitation suffers from out-of-slice contamination. The deviation between Bloch equation simulation and experimentally determined profiles indicate that technical challenges known to half-pulse excitations such as incorrect timing between RF and gradient pulse and inconsistency of RF power output between pairs of half-pulse excitations are the main source of error (44). Some of these issues are complex and difficult to address, nevertheless, especially in this SMS approach the half-pulse excitation profile has a large effect on the image quality. The high modulation of the SMS half-pulses may further intensify the problem. The issue of out-of-slice excitation is a particularly crucial in this interleaved sequence design using IR PINS pulses. In contrast to single slice acquisitions using non-selective IR pulses, out-of-slice excitation may occur in slice regions outside of the inversion band of the PINS IR pulse (see Supporting Information Figure S5). In these regions the suppression of long T_2 spins, which in *in vivo* acquisitions usually dominates short T_2 signal, is ineffective and, therefore, enhance slice contamination significantly. A stronger emphasis on the correction of RF and gradient waveform inaccuracies of the SMS half-pulses (45) that would lead to significant improvements of the excitation profiles is, therefore, strongly desirable to further enhance this methodology.

While the degraded excitation profile of the half-pulses (Figure 3) appears to be a major weak spot of the method used, the inversion bands of the PINS pulses are well defined. The main criteria for the design of the hyperbolic secant PINS IR pulses was set to be the slice profile and uniformity over the region of interest which was tested on long- T_2 spins (Figure 6). However, hyperbolic secant IR pulses are frequently used in UTE imaging because of their robustness towards B_1 inhomogeneity and especially because their ability to specifically invert long- T_2 spins while leaving short T_2 spins largely unaffected as long as the adiabatic condition applies (13). A stronger focus in this work onto the T_2 response in the design of PINS IR pulses may, therefore, lead to further improvements of the short T_2 contrast. Finally a minor effect of the interleaved sequence timing in case of the IR acquisition is an attenuation of short T_2 signal in consecutive slice groups. Since the excitation profile of the IR PINS pulse for short T_2 signal is ill defined (see Block equation simulation in the Supporting Information Figure S8) a saturation effect due to out-of-slice excitation as seen in the head images of Figure 8 occurs.

The second strategy to reduce scan time is to use coil sensitivity information in combination with an undersampled SMS k-space acquisition. The structural phantom images in Figure 4 demonstrate the ability of the basic SMS method to acquire images of high resolution. Although issues with the slice profile of the SMS half-pulses remain the results are satisfactory. A crucial finding for achieving acceleration is that the image quality is very robust for low undersampling factors ($R = 2$ and 4). This is partly due to the multi-leaf spiral sampling that similarly to radial sampling oversamples the inner area around the k-space center when fully sampled. The comparison with the single slice reference images ($N_{\text{SMS}} = 1$) stress the value of using additional coil sensitivity information in the context of undersampling. SENSE reconstruction heavily depends on an artifact free coil sensitivity map. We observed that low-resolution gradient echo scans of only about 16 s were generally sufficient to generate reliable sensitivity maps resulting in a small increase in the total scan

time (46). However, these maps tended to break down in areas too close to the coils and too far outside the boundaries of the object (long- T_2 protons) and image masking was applied. In the case of IR UTE imaging extreme differences in contrast occur between image obtained in the gradient echo prescan and the actual UTE image. This is generally not a problem for *in vivo* imaging since the different T_2 proton pools (short and long) reside in relative close proximity within the body so that the sense map will still cover dark areas of the prescan such as regions of cortical bone. However, for the UTE phantom in which the different T_2 pools are widely separated (see Figure 7, B) the method fails.

In contrast to most SMS applications that intrinsically have a high signal-to-noise ratio (SNR), IR UTE acquisitions usually suffer from low signal intensity, as is the case for images obtained in this study (see Figure 8). The SMS acquisition benefits from the increase in signal strength ($\sqrt{N_{\text{SMS}}}$) compared to the single slice acquisitions. The loss of signal intensity due to fourfold undersampling ($R = 4$) is therefore being compensated in the SMS case with $N_{\text{SMS}} = 4$ (see Figure 8, 9).

CONCLUSIONS

The SMS approach using RF phase encoded half-pulses in combination with PINS IR pulses described in this study presents a promising method to reduce scan time for slice selective 2D UTE acquisitions with IR spin preparation. In this study the method was tested with sequence parameters optimized for cortical bone imaging but could in principle be used for other 2D UTE imaging application as well. Images are of slightly reduced quality, mainly due to out-of-slice excitation by the SMS half-pulses.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1:

(A) Sequence timing diagram of SMS half-pulse sequence: slice groups (blue and red) are acquired in an interleaved fashion; the acquisition of the half-pulse pairs, RF phase encodings and spiral leafs occur consecutively in this order from the lowest to highest structuring level. (B) Schematic presentation of the 3D CAIPI type undersampling strategy. Red dots mark the outer most part of selected spiral leaves to indicate the staggered orientation of consecutive k-space planes.

Figure 2:

Magnitude of a set of half-pulse pairs (A, B) with $N_{\text{SMS}} = 4$ (*half sms4*) as well as the corresponding sum of the simulated magnitude (C) and phase (D) of the half-pulse excitation profiles in the slice direction. The second half-pulse is played out with a negative z-gradient (see Supporting Information Figure S1) in order to traverse k-space from the opposed end to the center. The slice phase encoding pattern for each excitation is highlighted in D (red). Wong phase cycling scheme was not applied to the pulse shape shown in this figure in order to make the RF phase encoding scheme apparent. Further information including Wong phase half-pulse and z-gradient shapes as well as simulated individual half-pulse excitation profiles are given in Supporting Information Figure S1.

Figure 3:

Comparison of the experimentally determined slice profile (projections of spherical water phantom) between the sinc pulse (*sinc sms4* in red) and half-pulse excitation (*half sms4* in blue) is given in (A). The half-pulse profile depicted in (A) represents the sum of the individual half-pulse excitation profiles shown in (B). The half-pulse profile suffers from considerable out-of-slice excitation while the sinc pulse excitation has a well-defined profile. The variation in intensity between slices largely derives from the different curvature at each slice location of spherical shaped phantom. Profiles (sum of half-pulse excitations) in (C) and (D) are measured using a UTE phantom with $T_2 = 660 \mu\text{s}$ and 2.6 ms , respectively. Slice profile broadening is observed going to short T_2 component for which the relaxation time is well below the duration of the half-pulse ($T_2 < T_{\text{RF}}$). Additional information regarding the slice profile characterization are given in the Supporting Information Figures S2–S5.

Figure 4:

Comparison of structural phantom images obtained with sinc full and half-pulses using RF phase encoded SMS imaging. Four slices (# 1, 5, 9, 13) out of a total of 16 slices acquired with a gap of 100% are shown for the *half sms1*, *half sms2* and *half sms4* pulse as well as the *sinc sms1* and *sinc sms4* pulses (Figure 3 A). Both sinc pulse images were scaled by a factor of $\sqrt{2}$ for windowing purposes. Images were otherwise windowed identically. Absolute differences for slices 1 and 5 are shown in B and differences after normalization (division by mean value) for slices 9 and 13 in C. Intensities were scaled relative to the maximum value of *sinc sms1* images with an additional scaling factor of x5 for the normalized difference images in C. No masking was applied.

Figure 5:

Comparison of undersampled half-pulse images for $N_{SMS} = 1$ and 4. For $N_{SMS} = 1$ the images were reconstructed with coil sensitivity information (sms1_SENSE, row B) and without (sms1, row A). The images for $N_{SMS} = 4$ were obtained using 3D CAIPI type undersampling pattern (see Figure 1, B) in combination with SENSE reconstruction (sms4_SENSE, row C). Images are shown for an undersampling factor $R = 1, 2, 4, 8, 16, 32$. Images were normalized (divided by mean value).

Figure 6:

Pulse profile and excitation characteristics of two inversion-recovery PINS pulses used in this study. *PINS1* (left column): pulse length = 11.8 ms, $\mu = 5$, 40 rectangular pulses, slice width = 0.5 and slice separation = 2.4 cm; *PINS2* (right column): pulse length = 11.8 ms, $\mu = 8$, 40 rectangular pulses, slice width = 0.6 and slice separation = 2.4 cm. The magnitude, phase and z-gradients of both pulse designs are shown on top (A). Single slice FLASH images with the PINS pulse gradient blips played out in x-direction with TI = 1 ms and 150 ms are shown below (B), as well as the corresponding intensity profiles through the center of the phantom at TI = 1, 50, 100 and 150 ms (C).

Figure 7:
The profile images of both IR pulses, *PINS1* and *PINS2* (Figure 6), on water/fat phantom TI = 1 ms are overlaid (upper half) with the excitation profile of the *half sms4* profile in A. The red arrow indicates the area in the fat phantom where the excited slice location is not completely covered by the inversion band of the *PINS1* pulse due to the spatial dislocation of the latter caused by the frequency offset of the lipid protons. On the bottom (B) selected slices of aqueous MnCl₂ solution UTE phantom images are shown with T₂ values of a few seconds (water) down to about 40 μs with TE = 100 μs and TI = 100 ms.

Figure 8:
UTE head images: Two sets (eight slices) of echo images and magnitude differences obtained by the IR UTE SMS sequence using *PINS2* with $N_{SMS} = 4$, $R = 4$, $TR = 300$ ms, $TE = 20/TE_2 = 2.3$ ms, $TI = 100$ ms, a 40° flip angle, 256 spiral leafs in less than 3 min scan time (sms4_SENSE). Fully sampled reference images reconstructed with and without SENSE using a corresponding single-slice IR UTE sequence (20:30 min scan time) are also presented (sms1_SENSE and sms1; nonselective *IR* pulse played out at a frequency offset of -270 Hz). Images were windowed and masked equally.

Figure 9:

Comparison of undersampled half-pulse images for $N_{\text{SMS}} = 1$ and 4. As above the images were reconstructed with coil sensitivity information (sms1_SENSE, row B) and without (sms1, row A) for $N_{\text{SMS}} = 1$ and obtained using 3D CAIPI type undersampling pattern (see Figure 1, B) with SENSE reconstruction for $N_{\text{SMS}} = 4$ (sms4_SENSE, row C). Images are shown for an undersampling factor $R = 1, 2, 4, 8, 16, 32$. All images were normalized (divided by mean value) and are masked equally. The boxes mark the images shown in Figure 8.