Multislice 2D Spin-Echo Imaging Using Frequency-Swept Pulses

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Introduction: When frequency-swept pulses, which are typically known as adiabatic pulses, are used for refocusing in spin echo imaging due to their B_1-insensitivity, non-linear phase variations appear across a slice, causing a signal loss. In order to overcome this problem, a pair of identical frequency-swept pulses have been used to make a non-linear phase compensation across the slice (1). An alternative approach has also been proposed using frequency-swept pulses for both excitation and refocusing, where the non-linear phase profile produced by one pulse can be compensated by the other pulse under specific conditions (2,3). Here, new and general conditions are proposed for compensating the non-linear phase variation across the slice when hyperbolic secant (HS) pulses (4), which provide an excellent slice profile, are used for both excitation and refocusing. The new conditions were derived from the equation of the phase profiles produced by HS pulses in the presence of slice-selective gradients. Bloch simulations were performed to validate the analysis and multislice 2D spin-echo imaging of a water phantom and *ex vivo* cat brain were also performed satisfying one of the new conditions.

Theory: Suppose that HS pulses with the pulse length of $T_{p,1}$ and $T_{p,2}$ are applied for excitation and refocusing in the presence of slice-selective gradients G_1 and G_2 (Below, subscripts of 1 and 2 indicate excitation and refocusing, respectively). And, suppose that the rephasing gradient of $-G_1$ is applied for a half of the total length of G_1 ($=T_{p,1}/2$) immediately after HS excitation. Referring to our previous works (3,5) and using that the offset frequency $\Omega = \gamma G x$ in the presence of a gradient G, the phase accumulation after HS refocusing is written in terms of position x across the slice,

$$\phi(x) = \left[\frac{A_2 T_{p,2}}{\beta_2}\ln\left(\frac{A_2\cdot\mathrm{sech}(\beta_2)}{\sqrt{A_2{}^2-(\gamma G_2 x)^2}}\right) - \frac{A_1 T_{p,1}}{2\beta_1}\ln\left(\frac{A_1\cdot\mathrm{sech}(\beta_1)}{\sqrt{A_1{}^2-(\gamma G_1 x)^2}}\right)\right] - \frac{\gamma x}{2}\left[\frac{G_2 T_{p,2}}{\beta_2}\ln\left(\frac{A_2+\gamma G_2 x}{A_2-\gamma G_2 x}\right) - \frac{G_1 T_{p,1}}{2\beta_1}\ln\left(\frac{A_1+\gamma G_1 x}{A_1-\gamma G_1 x}\right)\right] - \frac{\pi}{2} \quad [1]$$

where A is the frequency-sweep amplitude in rad/s and β is a truncation factor (typically, sech(β)=0.01). From Eq.[1], the general condition can be derived for compensating non-linear phase profiles produced by HS excitation and refocusing across the slice. In other words, provided that $\beta_1 = \beta_2$, non-linear terms in Eq.[1] are cancelled out when both $A_1T_{p,1} = 2A_2T_{p,2}$ and $G_1T_{p,1} = 2G_2T_{p,1}$ are satisfied. According to the former condition, any pair of $T_{p,1}$ and $T_{p,2}$ can be chosen as long as AT_p of the HS pulse for excitation is twice as that for refocusing. And then, using the chosen $T_{p,1}$ and $T_{p,2}$, a pair of G_1 and G_2 is to be determined by the latter condition. One of the simplest conditions is [$T_{p,1} = 2T_{p,2}$, $A_1 = A_2$, $G_1 = G_2$], which is consistent with the previous results by Kunz (2) and Park et al.(3) in spin-echo imaging. Except for this condition, all the other conditions are new ones which have not been reported yet. Among the new solutions the simplest one is [$T_{p,1} = T_{p,2}$, $A_1 = 2A_2$, $G_1 = 2G_2$], which would be more attractive in the cases where reducing TE is critical and B_1-insensitivity is needed as well. All the conditions can also be applied to a chirp pulse.

Simulation: Bloch simulations were performed to verify the validity of theoretical analysis described above. EXOR phase cycling was performed to eliminate all undesired coherences due to imperfections in RF pulses. M_{xy} and phase profiles were obtained for the condition of [$T_{p,1} = T_{p,2}$, $A_1 = 2A_2$, $G_1 = 2G_2$] with a chemical shift offest $\delta\omega = 0$ or ±0.5 kHz. HS pulses with same T_p (= 4ms) were used for excitation and refocusing, having $A_1/2\pi = 2.5$ kHz and $A_2/2\pi = 1.25$ kHz, with $G_1 = 2G_2$. Excellent slice profiles are produced despite a shift in the slice position for $\delta\omega = \pm0.5$ kHz (Fig.1a). Complete rephasing occurs for $|\Omega| \leq A$ with $\delta\omega = 0$, but non-linear phase variations appear across the slice with $\delta\omega \neq 0$ (Fig.1b). According to another simulation, two chemical species of $\delta\omega = 0$ and 0.5 kHz with same number of spins may cause ~13% signal loss. The signal loss will reduced to ~3% with $\delta\omega = 0$ and 0.25 kHz. As $A >> |\delta\omega|$ or $T_{p,1}$ approaches $T_{p,2}$, the signal loss will become less.

Experiment: Multislice 2D spin-echo imaging were performed at 4T and 4.7T. At 4T, experiments were performed using a TEM head resonator and a cylindrical water phantom. HS pulses with $T_p = 7$ms were used having $A_1/2\pi = 1.43$ kHz and $A_2/2\pi = 0.715$ kHz. For comparison, the experiment was repeated using sinc pulses with $T_p = 7$ ms. FOV = 20×20 cm^2, matrix size = 256×128, TE/TR = 28ms/2s, and THK = 5 mm. The image obtained using HS pulses shows better image quality than that obtained using sinc pulses, especially near the periphery of the phantom where the RF field was weakest, because of the B_1-insensitivity of HS refocusing (Fig.2 and 3). The maximum SNR increase was ~15% in the periphery. At 4.7T, *ex vivo* cat brain imaging was performed with a transceive surface coil (12-mm diameter). HS pulses with $T_p = 4$ms were used having $A_1/2\pi = 2.5$ kHz and $A_2/2\pi = 1.25$ kHz. Sinc pulses were also used with $T_p = 4$ ms. FOV = 3.5×3.5 cm^2, matrix size = 128×128, TE/TR = 20ms/2s, and THK = 2 mm. As shown Fig.4, the adiabatic property of HS refocusing significantly reduces the signal loss related to B_1 inhomogeneity.

Discussion: The new conditions with $T_{p,1} \neq 2T_{p,2}$ for non-linear phase compensation across the slice offer an opportunity of reducing TE more than the condition with $T_{p,1} = 2T_{p,2}$. Although the conditions with $T_{p,1} \neq 2T_{p,2}$ may cause some signal losses due to chemical shifts or B_0 offsets when compared to the condition with $T_{p,1} = 2T_{p,2}$, it would be negligible (< 3%) in many clinical applications because, for example, fat is chemically shifted from water by about 3.5 ppm which amount to 0.22 kHz at 1.5T. Furthermore, they still provide better image quality and SNR due to its adiabatic refocusing than conventional pulses (e.g. a sinc). Therefore, when considering its B_1-insensitivity and excellent slice profile, it would be very promising to use frequency-swept pulses, specifically, HS pulses, in multislice 2D spin-echo imaging.

References: (1) Conolly S et al, MRM(18) 1991:28-38 (2) Kunz D, MRM(4) 1987:129-136 (3) Park J-Y et al, ISMRM 2006 (4) Silver M et al, JMR(59) 1984:347-351 (5) Park J-Y et al, MRM(55) 2006:848-857

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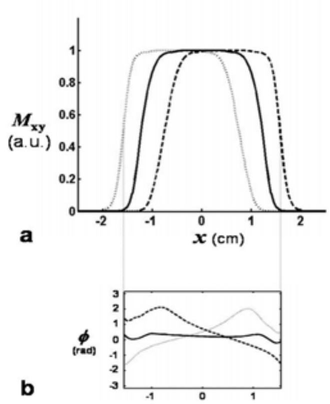

Fig.1 (a) M_{xy} and (b) phase profiles across the slice obtained from Bloch simulation with $\delta\omega$ = -0.5(--), 0(–), and 0.5(···) kHz.

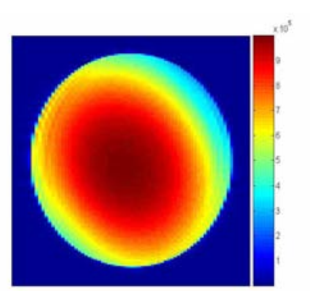

Fig.2 Multislice spin-echo imaging of a water phantom at 4T using sinc pulses.

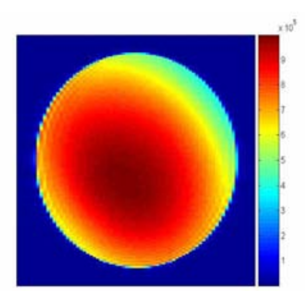

Fig.3 Multislice spin-echo imaging of a water phantom at 4T using HS pulses.

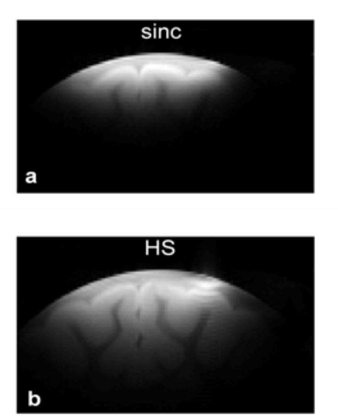

Fig.4 Multislice spin-echo imaging of *ex vivo* cat brain with a surface coil using (a) sinc pulses (b) HS pulses.