Spin-Echo MRI Using π/2 and π Hyperbolic Secant Pulses

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Introduction: Frequency-modulated (FM) pulses, including adiabatic full-passage (AFP) pulses, are commonly used in pulses sequences to invert spins because they are able to produce a broadband and B_1-insensitive π rotation. However, a single AFP pulse is not often used to refocus magnetization in spin-echo sequences because the phase of the magnetization following the pulse is a non-linear function of a frequency offset. As an alternative, two identical AFP pulses have been used to eliminate the non-linear phase profile produced by a single AFP pulse (1). Here, based on a theoretical analysis of the phase profile, general conditions to compensate the non-linear phase across the slice are proposed for the case of a hyperbolic-secant (HS) used for both π/2 excitation and π refocusing (i.e., the π/2 HS - π HS sequence) in multi-slice spin-echo imaging. The methods are demonstrated using Bloch simulations and by imaging experiments on a phantom and human brain at 4T.

Theoretical Description: In our previous work (2), the equation describing the phase profile across a slice was provided in terms of position *x*, and general conditions were derived for compensating the non-linear phase across the slice, in multi-slice 2D spin-echo imaging using a π/2 HS - π HS sequence. That analysis did not include the effects of a frequency offset δ that can arise from sources other than the gradient *G* (e.g., B_0 inhomogeneity, chemical shift, and magnetic susceptibility). When the phase equation is extended to include the effect of δ and constant phase terms independent of *x* are ignored, the more general equation of the phase profile across the slice can be written as

$$\phi(\delta,x) = \left[\frac{A_2T_{p,2}}{\beta_2}\ln\left(\frac{1}{\sqrt{A_2{}^2-(\gamma G_2x+\delta)^2}}\right)-\frac{A_1T_{p,1}}{2\beta_1}\ln\left(\frac{1}{\sqrt{A_1{}^2-(\gamma G_1x+\delta)^2}}\right)\right]-\left[\frac{T_{p,2}}{2\beta_2}(\gamma G_2x+\delta)\ln\left(\frac{A_2+\gamma G_2x+\delta}{A_2-\gamma G_2x-\delta}\right)-\frac{T_{p,1}}{4\beta_1}(\gamma G_1x+\delta)\ln\left(\frac{A_1+\gamma G_1x+\delta}{A_1-\gamma G_1x-\delta}\right)\right]. \quad [1]$$

Here, *G* is a slice-selective gradient, *A* (rad/s) is the frequency-sweep amplitude (= *BW*/2, where *BW* is a pulse bandwidth), T_p is a pulse length (s), and β is a dimensionless pulse truncation factor (Subscripts of 1 and 2 indicate π/2 excitation and π refocusing, respectively). If it is assumed that there is no frequency offset other than a gradient (i.e., $\delta = 0$), non-linear phase terms in Eq.[1] cancel each other in some specific conditions. Among all possible solutions, three simplest solutions are [$\beta_1 = \beta_2$, $T_{p,1} = 2T_{p,2}$, $A_1 = A_2$, $G_1 = G_2$], [$\beta_1 = \beta_2$, $T_{p,1} = T_{p,2}$, $A_1 = 2A_2$, $G_1 = 2G_2$], and [$\beta_1 = 0.5\beta_2$, $T_{p,1} = T_{p,2}$, $A_1 = A_2$, $G_1 = G_2$], which will be referred to as condition I, II, and III, respectively. In the case that $\delta \neq 0$, however, non-linear phase terms are non-zero with condition II, whereas they still cancel out with conditions I and III. Besides the remaining non-linear phase variation, the reduction of a slice thickness also takes place with condition II because $G_1 = 2G_2$, and thus, the slice selection of excitation and refocusing is mismatched. In the presence of $\delta (\neq 0)$, both the remaining non-linear phase profile and the reduced slice-thickness contribute to the signal loss when condition II is employed (Fig.1). On the other hand, the signal reduction with $\delta \neq 0$ offers a new possibility of spin-echo imaging with chemical-shift selection or susceptibility-weighted contrast.

Experiments: First, a multi-slice 2D spin-echo imaging of human brain was performed at 4T using the π/2 HS - π HS sequence with condition II (Fig.2). HS pulses were used with $T_{p,1} = T_{p,2} = 5$ ms, $A_1/2\pi = 2A_2/2\pi = 2$ kHz, and $\beta_1 = \beta_2 = 5.3$. For comparison, the experiment was repeated using sinc pulses with the same bandwidth. FOV = 30×30 cm^2, TE/TR = 22ms/0.5s. The power of the π HS pulse was set 3 dB higher than the adiabatic threshold to gain partial insensitivity to B_1 inhomogeneity. Because of the adiabatic property of the π HS pulse, images obtained using HS pulses display better image quality in terms of SNR, especially in the periphery of the brain. SNR improvement was ~7%, ~22%, and ~22% in the white matter, gray matter, and ventricle, respectively. Second, 2D spin-echo imaging satisfying condition II was performed on a water-silicon oil phantom, to show that the signal reduction with $\delta \neq 0$ can be used for spin-echo imaging with fat suppression (Fig.3). HS pulses with $T_{p,1} = T_{p,2} = 8$ ms, $A_1/2\pi = 2A_2/2\pi = 1.25$ kHz were used. For comparison, imaging was also performed with condition I. HS pulses with $T_{p,1} = 2T_{p,2} = 16$ ms, $A_1/2\pi = A_2/2\pi = 0.625$ kHz were used. FOV = 20×20 cm^2, TE/TR = 24ms/2s. Figure 3b shows that the signal from silicon oil was suppressed when condition II was employed ($\delta \approx 4.5$ ppm) and the spectrometer frequency was set to the Larmor frequency of the water proton. On the other hand, the spin dephasing due the chemical shift of the silicon oil is completely rephased, and thus, no signal loss was observed in the silicon oil, in the case of condition I (Fig. 3a).

Discussion: As compared with the alternative way to compensate the non-linear phase using two identical π HS pulses, the π/2 HS - π HS sequence allows shorter TE and lower power deposition. Condition I has limited benefit in reducing TE because $T_{p,1} = 2T_{p,2}$, whereas with conditions II and III the minimum TE is restricted by the same factors that limit TE in conventional spin-echo sequences. Condition III has a disadvantage that the π HS pulse with 2β requires 2~3 dB higher power to reach an adiabatic threshold and the sharpness of the slice profile is reduced, compared to conditions I and II. A disadvantage of condition II is a potential signal loss in the presence of a frequency offset (e.g., chemical shift or change in magnetic susceptibility). On the other hand, as shown in Fig.3, the signal reduction that occurs with such a frequency shift provides a new possibility for chemical-shift selected imaging (e.g., fat suppression) and susceptibility-weighted imaging. Other examples of MRI applications likely to benefit from the π/2 HS - π HS sequence include multi-slice versions of spin-echo EPI and diffusion-weighted imaging.

References: (1) Conolly S et al., MRM 1991;18:28-38 (2) Park J-Y et al., ISMRM 2007.

Acknowledgements: This work was supported by NIH grants P41 RR08079 and P30 NS057091.

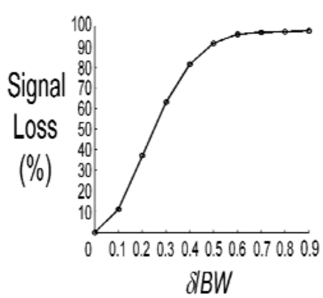

Fig.1 Signal loss obtained from Bloch simulations with condition II. It was assumed that T_2 is infinite and there is a single δ spin across the slice.

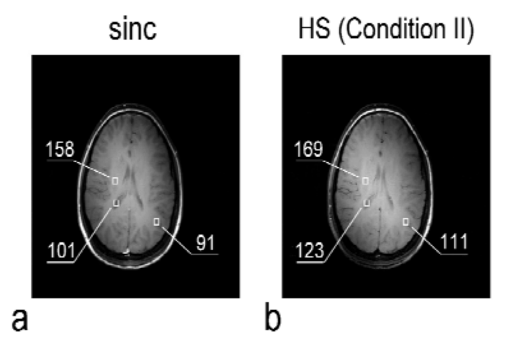

Fig.2 Multi-slice 2D spin-echo imaging of human brain using (a) sinc pulses and (b) HS pulses satisfying condition II. The numbers given in the figure represent SNR in white matter, gray matter, and ventricle.

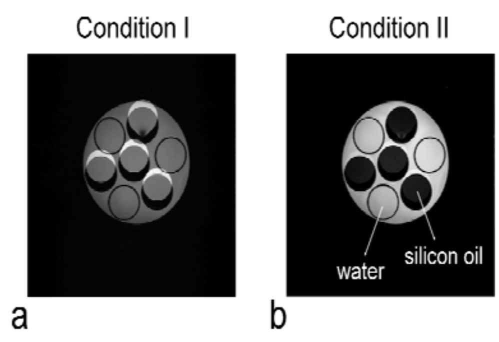

Fig.3 2D spin-echo imaging of a water-silicon oil phantom using HS pulses satisfying (a) condition I and (b) condition II.