Improved 3D Spin-Echo Imaging Using Frequency-Swept Pulses

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Introduction: Frequency-modulated (FM) pulses, including adiabatic full-passage (AFP) pulses, are sometimes preferred for selective refocusing in spin-echo sequences due to their tolerance of B1 inhomogeneity and ability to refocus broad bandwidths using only modest peak RF power. In such applications, typically a double spin-echo is produced and acquired, since the non-linear magnetization phase produced by the individual AFP pulses must be compensated by using two identical AFP pulses [1]. Recently, however, it was shown that a single slab-selective AFP pulse can be used for refocusing in 3D MRI (2, 3). In this case, the non-linear phase can be advantageous for reducing the dynamic range requirements placed on the analogue-digital-converter (ADC) and receiver because it spreads the signal energy in time and thus reduces the peak echo amplitude. Here, phase profiles produced by this sequence are analytically described. Based on this analysis, it is shown that this sequence produces a “pseudo-echo” in which isochromats have unique local refocusing points. Apodization using a sliding window is performed to avoid sacrificing signal arising from isochromats refocusing early or late in the pseudo-echo. The method is used to image Alzheimer’s plaques in vivo in the brain of a transgenic mouse.

Theory: Immediately after applying an excitation pulse, the magnetization vector is in the transverse plane and assumed to point along the x’ direction of the FM rotating frame. Then, provided the adiabatic condition is well satisfied, spins initially perpendicular to the time-dependent effective field vector of the AFP pulse \( (\mathbf{a}_{\text{AFP}}(t)) \) will remain perpendicular to \( \mathbf{a}_{\text{AFP}} \) throughout the adiabatic sweep. In the FM rotating frame, \( \mathbf{a}_{\text{AFP}} = \mathbf{a}_0(t|x'|) + j \mathbf{a}_0(t|x'|) \), where \( \mathbf{a}_0 \) is an RF amplitude and \( \Delta \phi(t) = \alpha \sigma - \alpha \omega \Delta t \) (\( \alpha \omega \); Larmor frequency of a given isochromat). Thus, during an AFP pulse, spins in the transverse plane experience a phase negation, precessing about \( \mathbf{a}_{\text{AFP}} \). When the hyperbolic secant (HS) pulse is used for refocusing, the entire phase accrued immediately following the AFP pulse (\( t = 0 \)) is given by

\[
\phi(\Omega, \Omega^0, t') = -\Omega \Delta - \frac{T_1}{2} \left[ \sqrt{\alpha_{\text{AFP}}^2 - \left( \Omega - \Omega^0 \right)^2} \right] \sech^2 \left( \beta \left( \Omega - \Omega^0 \right) t' \right) \left[ \Omega - \Omega^0 \right] d\Omega + \Omega^0 t' \]

where \( \Omega \) is an offset frequency, \( \alpha_{\text{AFP}} \) is the maximum RF amplitude (rad/s), \( \Delta \) is a time delay between the excitation and AFP pulses, and \( T_1 \) is the AFP pulse length. As shown in Fig. 1, \( \phi \) has a non-linear form similar to a quadratic function. In the presence of a readout gradient along the x axis (\( J(\Omega) = \gamma G \)), the vertex moves along the x-direction as the time progresses (\( t > 0 \)). It is also shown that the shape of \( \phi \) and its vertex position are approximately independent of \( \alpha_{\text{AFP}} \), except for a shift of the profile vertically in magnitude (Fig. 1). The isochromat corresponding to the vertex is locally rephased, so that it maximally contributes to the signal at a given vertex time (4).

The vertex time of an isochromat can be determined by taking the partial derivative of \( \phi \) with respect to \( \Omega \),

\[
\phi'(\Omega, \Omega^0, t') = \frac{\partial \phi(\Omega, \Omega^0, t')}{\partial \Omega} = \left[ \left( \Omega - \Omega^0 \right) \left[ \left( \Omega - \Omega^0 \right) t' \right] \left( \Omega - \Omega^0 \right) \right] \sech^2 \left( \beta \left( \Omega - \Omega^0 \right) t' \right) \left[ \Omega - \Omega^0 \right] d\Omega + \Omega^0 t' \]

Eq.[2] shows that each isochromat has a unique local rephasing time and thus the echo is a “pseudo-echo” (5). As expected, the vertex time is almost independent of \( \alpha_{\text{AFP}} \) (Fig. 2). The isochromats excited early and late will be refocused early and late, respectively. Images from such pseudo-echoes can be reconstructed using a fast Fourier transform (FFT). Given that the \( i \)th isochromat rephrases at time \( t_i \), a sliding window \( W(t - t_i) \) can be applied to the pseudo-echo for a proper apodization.

Methods: To verify the validity of theoretical derivation, a simulated 1D pseudo-echo was generated using the Bloch equations, and vertex times were determined and compared with Eq.[2]. The simulated echo was produced by a HS pulse using \( T_2 = 1 \text{ms} \) and \( \alpha / 2\pi = 5 \text{kHz} \). For in vivo 3D MRI experiments, the doubly transgenic Alzheimer’s mouse (APP-PS1 mouse) was used. In this spin-echo sequence, a single HS pulse was used for slab-selective refocusing in each of two directions (\( y \) and \( z \)). In this case, the non-linear phase can be advantageous for reducing the dynamic range requirements placed on the analogue-digital-converter (ADC) and receiver because it spreads the signal energy in time and thus reduces the peak echo amplitude. When imaging pseudo-echoes, images reconstruction using a sliding window provides better image quality as well as better SNR improvement than conventional apodization. Use of a single AFP pulse also enables shorter\( T_E \) and lower power deposition than required by multiple AFP pulses. Although only the HS pulse was evaluated here, this analysis can easily be extended to other types of AFP pulses.


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Fig. 1 Phase profiles (Eq.[1]) produced with two different \( \alpha_{\text{AFP}} \) values of HS refocusing at \( t' = 0.45T_2, 0.5T_2, 0.55T_2 \).

Fig. 2 Local rephasing times of isochromats: theory \( (\alpha_{\text{AFP}} = 7, 14 \text{kHz}) \) vs. simulation.

Fig. 3 The AD mouse brain images (a) without apodization and (b) using a sliding window.