DESIGNING A HYPERBOLIC SECANT EXCITATION PULSE TO REDUCE SIGNAL DROPOUT IN GE-EPI

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Target audience: This work will be of interest to researchers performing task-based and resting-state FMRI experiments who are investigating regions of the brain currently affected by susceptibility induced signal dropout in gradient-echo echo-planar images.

Purpose: Images acquired using gradient-echo echo-planar imaging (GE-EPI) suffer from signal-dropout in the orbitofrontal cortex (OFC) and temporal lobes (TL) [1]. This artifact can be reduced using RF excitation pulses with quadratic phase profiles [2] such as full-passage scaled-down Hyperbolic Secant (HS) pulses [3-5], which partially cancel the phase dispersion due to susceptibility gradients in the slice-selection direction, $G_{z,s}$ [2,5]. Here we determine by Bloch simulation the HS pulse parameters needed to give the most uniform signal response across the range of susceptibility gradients observed in the human head ($G_{z,s}$ ±300μTm^{-1}). From these simulations, and from phantom images, we show that previous predictions of the dependence of the voxel signal on the susceptibility gradient [2] are inaccurate. To ensure the slice has the required thickness we derive, for the first time, an expression for the bandwidth, Δf, of a HS pulse used for excitation; this is flip angle dependent, Eqn. 1. Finally using our optimised pulse we demonstrate recovery of signal in the orbitofrontal cortex of six healthy male volunteers.

Theory: Hyperbolic secant (HS) pulses have both amplitude $A(t)=A_0\text{sech}(\beta t)$ and phase $\phi(t)=\mu.\ln[\text{sech}(\beta t)]$ modulation. A_0, the maximum amplitude of the pulse is related to the desired flip angle, α, using Eqn. 2. β is the modulation angular frequency; for a fixed pulse duration, T, increasing β reduces the ripple in the stop-band of the slice profile. μ is a dimensionless parameter that determines both the sharpness of the slice profile [3] and the degree of quadratic phase [4,5]. It was previously assumed [5] that the bandwidth of an HS excitation pulse was $\mu\beta/\pi$ (as for inversion [3]), however we show that the bandwidth Δf (FWHM of the transverse magnetisation) is flip angle dependent, Eqn. 1 and Fig. 1.

$$\Delta f = \frac{\beta}{\pi^2}\cosh^{-1}\left[\frac{\cosh(\pi\mu)\left(\cos\alpha - 1/2\sqrt{3+\cos^2\alpha}\right)+\cos\alpha - 1}{1/2\sqrt{3+\cos^2\alpha}-1}\right] \quad \textbf{(1)}$$

$$A_0 = \frac{\beta}{\gamma}\sqrt{\left(\frac{\cos^{-1}\left[\cosh^2(\pi\mu/2)\cos\alpha+\sinh^2(\pi\mu/2)\right]}{\pi}\right)^2+\mu^2} \quad \textbf{(2)}$$

Methods: An HS pulse was designed such that the signal response for $G_{z,s}$ ±300μTm^{-1} was as uniform as possible for 3mm slices acquired with TR=2s and TE=30ms on a 3T GE MR750 system (General Electric, Waukshua, WI, USA) equipped with gradients with max amplitude 50 mTm^{-1}. The pulse had T=5ms (to match the SLR pulse excitation pulse being replaced such that the same number of slices could be imaged per TR), α=73° (Ernst angle for grey-matter at 3T assuming T_1=1.6s [6]), and β=3040Hz (to minimize the ripple in the slice profile whilst keeping the acoustic noise from gradient switching at an acceptable level). The optimal value of μ was determined by Bloch simulation* in MATLAB (The MathWorks Inc.). The signal response as a function of $G_{z,s}$ was also measured on the scanner by manually altering the shim gradient prior to acquiring images of a phantom. GE-EPI data with the SLR and the optimal HS pulse were acquired of six healthy male volunteers to determine the degree of signal recovery. Thirty-six slices were acquired top-down sequentially with a 0.3mm slice gaps, with an acceleration (ASSET) factor of two. The field-of-view was 21.2cm with a 64×64 matrix. The body coil was used for RF transmission and an 8-channel head coil was for signal reception.

Results: The bandwidth, calculated using Eqn. 1 (for μ=4.25 and β=3040Hz) is greater than the constant (inversion) bandwidth previously assumed to be valid for all flip angles α, Fig. 1. Bloch simulation, Fig. 2, shows that the optimal value of μ=4.25 results in a signal of ~45-50% (compared to a linear phase pulse when $G_{z,s}$=0) across $G_{z,s}$ ±300μTm^{-1}. In Fig.3 the simulated signal response from a linear phase pulse (red line) is shown along with the simulated (dark blue line) and measured signal response of the HS pulse (green crosses) and the previous theoretical predictions of Cho et al. [2] (light blue line). Previous theoretical predictions [2] do not match simulations and measurements. Relative to a linear phase pulse, signal is recovered when the optimized HS pulse is used for $G_{z,s}$<-140μTm^{-1} and >140μTm^{-1}, however the signal is reduced by up to 50% when -140μTm^{-1} <$G_{z,s}$<140μTm^{-1}. A representative example from one of the six subjects demonstrates the signal recovery in the OFC when the optimal HS pulse, Fig 4.(b), is used compared to the conventional SLR pulse, Fig 4.(a).

Figure 1

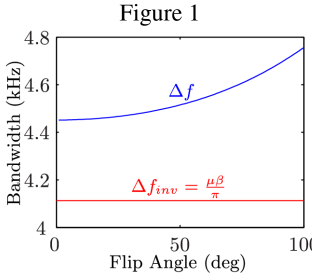


Figure 2

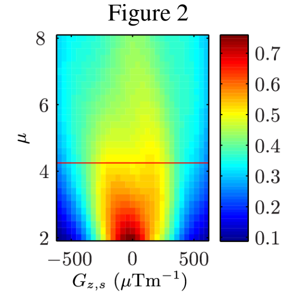


Figure 3

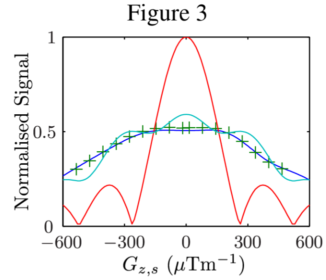
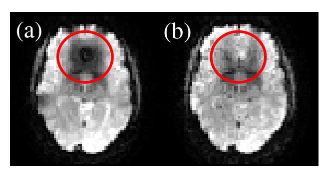


Figure 4



Discussion and Conclusions: We have derived an expression for the bandwidth of a HS pulse used for signal excitation; this varies with flip angle. If the inversion bandwidth, $\mu\beta/\pi$, was used to determine the slice selection gradient the excited slices would be thicker than prescribed. We show that previous predictions [2] of the dependence of the voxel signal on the susceptibility gradient are inaccurate; the discrepancy between theory and the simulated and measured data arises because the assumption of perfectly rectangular slice profiles, used to derive the theoretical signal response, is not achievable in practice. We have designed a HS pulse with a near uniform signal response across the range of susceptibility gradients observed in the human head. Using this pulse, FMRI data can be acquired at the same spatial and temporal resolution as the conventional SLR or other pulses, however signal recovery in the OFC and TL comes at the cost of signal losses of up to 50% in regions unaffected by susceptibility gradients.

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* Bloch simulation code written by Dr B. Hargreaves (www-mrsrl.stanford.edu/~brian/blochsim)