

Correction Method for Thin Slice spectral spatial RF pulses

Yuval Zur¹

¹GE Healthcare, Haifa, Israel

Introduction: Spectral-spatial (spsp) RF pulses are widely used for simultaneous fat suppression and slice selective excitation in fMRI and DWI echo planar imaging. There is a need to reduce the minimum slice width to less than 2 mm and/or improve the time-bandwidth product (TB) of the slice profile. At high field strength (3T) the water fat frequency separation Δ_{wf} increases and therefore the time $\Delta t = 1 / \Delta_{wf}$ between adjacent sub pulses must decrease. This requires very powerful gradients with high slew rates, which are limited by the hardware and by peripheral nerve stimulation.

To excite off center slices, a frequency waveform (Omega waveform) is played out in concert with the gradient, such that the RF pulses excite the desired slice position. As shown in Figure 1, there is a mismatch between the gradient and omega waveform due to eddy currents and gradient-RF delay, which creates a phase ϕ between even and odd sub-pulses and creates artifacts. A simple way to avoid these artifacts is to remove the even sub-pulses, such that the phase between sub-pulses is 0. However, this increases Δt by a factor of 2, which increases the minimum slice width by a factor of 2.5. The purpose of this work is to design reliable spsp RF pulses with even and odd sub-pulses by compensating the phase ϕ such that fat-free thin slices (≤ 1.5 mm) become feasible at 3T.

Theory: If $\phi \neq 0$ there are two artifacts: a) the fat signal is not suppressed and b) the water signal decreases. As shown in (1)

$$Fat\ signal = \frac{1 - \exp(i\phi)}{2} ; \quad Water\ signal = \frac{1 + \exp(i\phi)}{2} \quad [1]$$

The fat (water) signal in [1] is normalized such that the unsuppressed fat (water) signal is 1. ϕ is proportional to the gradient amplitude G and the distance z from the center, where the field due to G is higher. If we require that fat signal < 0.1 , then from Eq. [1], $\phi < 10^\circ$. It can be shown (1) that for this accuracy (assuming zero eddy currents) the RF-gradient delay must be set with a 0.1 μsec accuracy for $z = 7$ cm, which cannot be achieved in practice. Below we present a practical calibration method to measure ϕ accurately for any G , z and RF delay. When ϕ is known, we apply a phase shift of $-\phi$ to even RF sub-pulses to restore a coherent excitation and eliminate the artifacts created by ϕ .

Method: The phase ϕ , created by eddy currents and RF delay (2), can be written as

$$\phi = A + Bz + 2\gamma Gz \cdot T_{del} = A + Bz + Cz \cdot T_{del} \quad [2]$$

where T_{del} is RF-gradient delay in μsec . A , B and C are constants. B and C originate from eddy currents and RF-gradient delay and A is the z independent eddy currents ('b0' eddy currents). A , B and C are measured (see below) during system calibration. During scans ϕ is determined in real time using the prescribed slice width and slice positions z .

A , B and C are measured by running multi-slice single shot EPI scans with N different RF delays $T_{del}(n)$ where $n = 1$ to N . The normalized fat signal is measured at each slice and each RF delay. For each RF delay we run an EPI scan with 29 slices at $z = -7$ to 7 cm. We use $N = 7$ RF delays of $-14, -10, -6, 0, 6, 10, 14$ μsec . The measured fat signal magnitude at RF delay $T_{del}(n)$ is fitted to

$$normalized\ fat\ signal\ at\ RF\ delay\ T_{del}(n) = \frac{|1 - \exp[i\alpha(n) + i\beta(n)z]|}{2} \quad for\ n = 1\ to\ N \quad [3]$$

using a simplex non-linear fitting algorithm (3). The parameter $\alpha(n)$ is the constant A in [2]. By fitting $\beta(n)$ vs. $T_{del}(n)$ to a straight line we obtain B and C . In all cases the fit is very stable, which confirms the correctness of [1] and [2].

We measured A , B and C for many different RF pulses, and found that they are independent of gradient rise time, and proportional to gradient strength. Hence A , B and C are global parameters that can be applied for all spsp RF pulses and scaled in real time when the gradient amplitude and slice positions are known.

Results: Figure 2 show the normalized fat signal acquired with sagittal slices (slice select gradient is x) of a 17 cm diameter phantom for RF delay of 0 and 10 μsec (squares), and the simplex fit (full line). The fat signal after the phase compensation (full circles) is also shown. As can be seen, the phase compensated signal is independent of RF delay and is smaller than 8% for all z and RF delays. Similar results were obtained for the y and z gradients.

Conclusion: We have demonstrated a method to compensate spsp RF pulses by adding a phase $-\phi$ to the even sub-pulses. Stable fat suppression $\leq 8\%$ is obtained for all slice positions and RF delays. This method can also be used to correct 2D RF excitation with an EPI trajectory.

References: (1) Y. Zur, Design of improved spectral-spatial pulses for routine clinical use, Mag. Res. Med., 2000; 43, 410 - 420. (2) Y. Zur, ISMRM 2012 proceedings, p.3473. (3) W. Press, S. Teukolsky, W. Vetterling, B. Flannery. Numerical Recipes. Cambridge University Press; 2007. p 502 - 507.

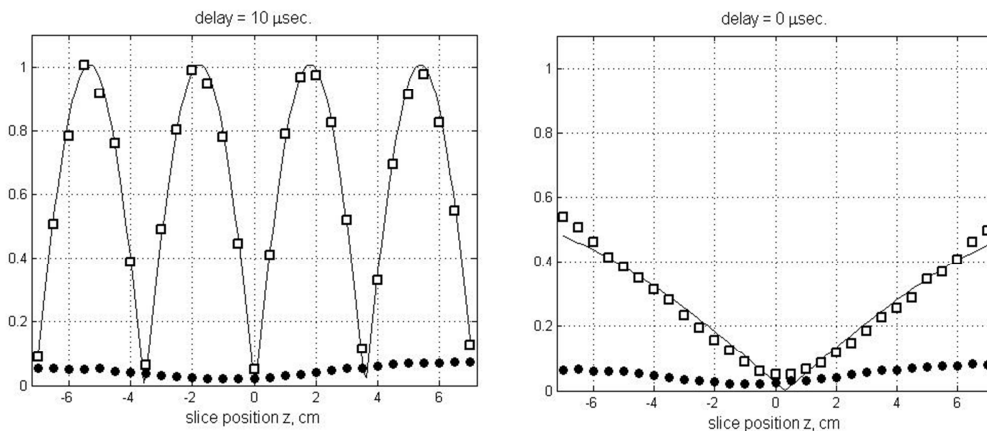
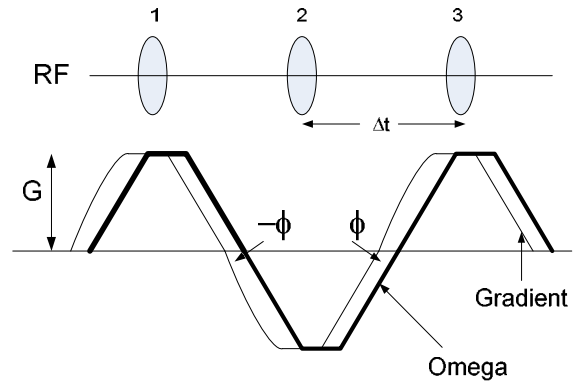


Fig. 2: Two data sets of a fat signal acquired at $z = -7$ to 7 cm with an RF delay of 10 μsec (left) and 0 μsec (right). The uncompensated data is represented by squares, the fit to Eq. [3] by a full line and the compensated data (after adding $-\phi$ to the even sub-pulses) by the full circles. Similar results were obtained for all gradients. After compensation, The fat signal is $\leq 8\%$ for all z and RF delays, and is independent of RF delay.