Bloch Simulations of UTE, WASPI and SWIFT for Imaging Short T2 Tissues

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Introduction: Clinical MR is predominantly geared towards the imaging of "long" T_2 species (T_2's greater than 10 ms). However, musculoskeletal (MSK) tissues of interest, such as ligaments ($T_2 \approx$ 4-10 ms), tendons ($T_2 \approx$ 2 ms) and cortical bone ($T_2 \approx$ 0.5ms) often exhibit low signal intensity in images acquired using clinical MR sequences [1]. Three specialized sequences designed to overcome these limitations are Ultrashort TE (UTE) imaging [1,2], Water And Fat Suppressed Projection Imaging (WASPI) [3], and SWeep Imaging with Fourier Transformation (SWIFT) [4]. We present theoretical work including Bloch simulations to investigate the T_2 blurring characteristics of these three techniques.

Theory: To simplify our analysis, we will constrain our focus on 1D objects along the x-axis with spin density ρ(x). UTE can be performed either in 2D (using slice selective half RF pulses) [1] or 3D (using non-selective RF pulses) [2], while WASPI and SWIFT are exclusively in 3D form (using short non-selective hard RF pulses). UTE and WASPI are based on acquiring the Free Induction Decay (FID) of the MR signal as soon after the end of the RF pulse as possible (e.g. minimum TE ≥ 8μs). This is typically accomplished by using a radial center-out k-space trajectory and data sampling time T_{AQ} on the order of a millisecond in duration (Fig.1a,b). In WASPI, the imaging gradient, G, is already applied during the RF pulse (Fig.1b). Therefore the RF pulse has to be kept short (~10μs, resulting in only a small flip angle) to avoid premature dephasing of the signal from the imaging gradient. In UTE the read gradient is zero during the RF pulse and is ramped up quickly during the readout, leading in part to radial ramp-sampling (Fig.1a). With either technique, magnitude images are reconstructed from the re-gridded k-space data. To model the effect of T_2 decay, the FID signal in the time domain is multiplied by an exponential decay function. The T_2 blurring in the image domain (1D) for WASPI and (to good approximation) for UTE is characterized by a convolution with a Lorenzian point-spread-function [2] (Eq.[1]).

$$P(x) = \frac{\dfrac{T_2}{T_{AQ}}}{1+\left(xk_{\max}\dfrac{T_2}{T_{AQ}}\right)^2} \quad where \quad k_{\max} = \frac{\gamma}{2\pi}GT_{AQ} \qquad (1)$$

The SWIFT technique can be regarded as a member of the general class of Simultaneous Excitation and Acquisition (SEA) pulse sequences, in which small excitation pulses and data acquisition intervals are rapidly interleaved (Fig.1c). For SWIFT the amplitudes $B_1(t)$ of the small RF pulses follow a fast-sweeping hyperbolic secant function with phase modulation Φ(t). This results in a flat frequency response if the magnetization is tipped by a small flip angle. The Bloch equations including T_2 decay ($R_2 \equiv 1/T_2$) can be solved for SWIFT in the small tip angle approximation using the following (standard) definitions:

$$\begin{matrix} \omega_1 \equiv \omega_x + i\omega_y = \gamma|B_1(t)|\exp\{i\Phi(t)\} \\ M \equiv M_x + iM_y \qquad \omega_z = \gamma Gx \end{matrix} \rightarrow \dot{M} + (R_2 - i\omega_z)M = i\omega_1 M_0 \quad \text{with solution :} \quad M(\tau) = iM_0 e^{-R_2\tau + i\gamma Gx\tau}\left(\int_0^\tau \omega_1(t)e^{R_2 t - i\gamma Gxt}dt\right) \equiv iM_0 e^{-R_2\tau+i\gamma Gx\tau}\,\Omega'(x)$$

where Ω′(x) is the spectral response function of the RF pulse. The total MR signal is a superposition of all spin isochromats within the spin density function $M_0 \rightarrow \rho(x)$:

$$S(\tau) = \int_{-\infty}^{\infty} M\,dx = i\int_{-\infty}^{\infty}\{\rho(x)\Omega'(x)\}e^{-R_2\tau+i\gamma Gx\tau}dx = iFT\{\rho(x)\Omega'(x)\}e^{-R_2\tau} \qquad (2)$$

In order to extract the spin density function ρ(x) (the image), one has to take the inverse Fourier transform of Eq.[2] and then divide by the complex spectral response function Ω′(x). From Eq.[2] the MR signal in the time domain can be seen to be multiplied by an exponential T_2 decay. Therefore, the SWIFT image would be expected to show a similar Lorenzian point-spread-function (Eq.[1]) and associated T_2 blurring as WASPI or UTE.

Simulation Results: Bloch simulations were performed to investigate the T_2 blurring of these techniques. The imaging sequences were designed to function within the hardware limits of current clinical scanners: G_{max} = 40mT/m and slew rate = 200mT/m/ms. For the simulated phantom shape we used a 1D step function similar to that used in the original SWIFT publication [4], with a spatial extent of 5cm. The 1D reconstructed image was obtained with a FOV = 20cm and resolution of 0.3mm. The reconstructed 1D images are shown in Fig.1 for various values of T_2.

Discussion: The simulated 1D images obtained with WASPI and SWIFT show comparable T_2 blurring as predicted by theory. The increased blurring of the UTE technique is related to the ramp-sampling, which means the acquisition time is slightly longer for the same resolution (see Fig.1a). The effect is pronounced only for T_2 ≤ 100μs. Therefore, either WASPI or SWIFT may provide advantageous for imaging extremely short T_2 tissues as encountered in dentine or trabecular bone.

Practical considerations: All of the techniques share one important hardware constraint - the minimum Transmit-Receive (T/R) switching time required to read the data quickly after the RF pulse(s). However, while the minimum TE in UTE and WASPI is mainly constrained by the switching time from transmit to receive (DAQ), SWIFT requires both fast switching from transmit to receive and back from receive to transmit. An advantage for UTE is that it can generate 2D images, which alleviates some of the scan-time constraints associated with 3D imaging. However, the rapid slewing of gradients employed by UTE can make this approach susceptible to degradation of image quality caused by eddy currents. Furthermore, gradient timing errors can corrupt the presumed k-space trajectory. Analogously, SWIFT can be susceptible to RF-DAQ switching delays due to the interleaved data acquisition, causing similar problems. A unique challenge for WASPI is that during the finite T/R switching time, the sequence does not acquire the center k-space region and hence requires a second (albeit shorter) acquisition at a lower gradient strength [3]. WASPI and SWIFT naturally operate at TE → 0, and therefore require additional magnetization preparation pulses to achieve T_2* contrast [3,4]. UTE images can be obtained at variable TE > 8μs which leads to flexible T_2* contrast and straightforward T_2* quantification.

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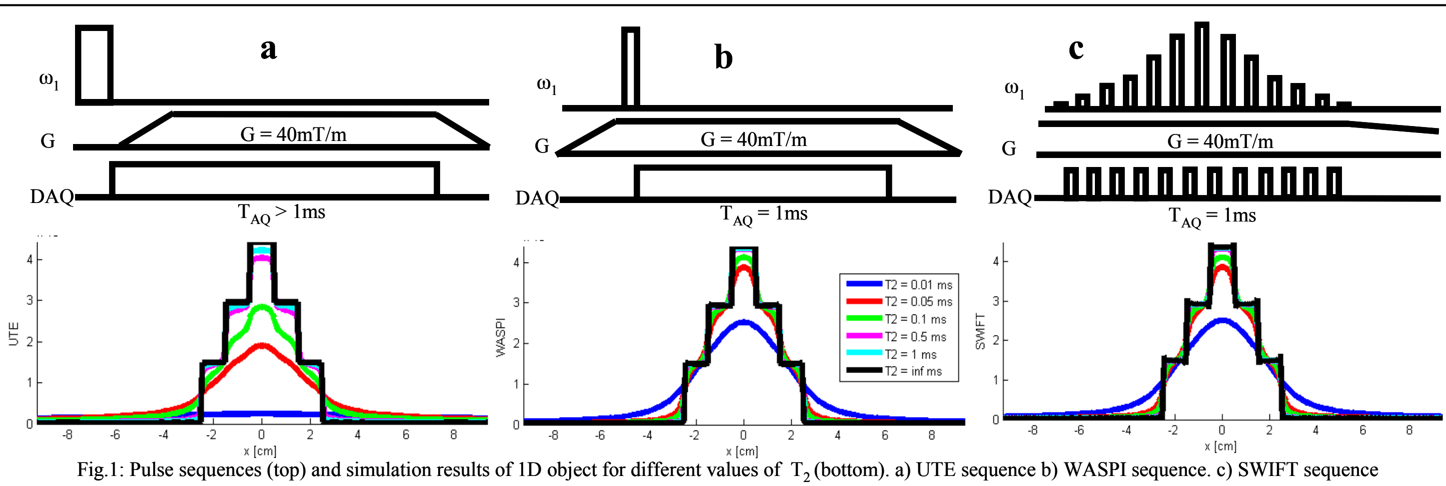

Fig.1: Pulse sequences (top) and simulation results of 1D object for different values of T_2 (bottom). a) UTE sequence b) WASPI sequence. c) SWIFT sequence