Simultaneous Short T2 Excitation and Long T2 Suppression RF Pulses

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Introduction: Ultrashort echo time (UTE) MRI requires specialized pulse sequences to overcome the short T_2 relaxation of the MR signal encountered in tissues such as ligaments, tendon or cortical bone. Imaging short T_2 tissues is achieved in UTE by acquiring the Free Induction Decay (FID) of the MR signal as soon after the end of the RF excitation pulse as possible. This is typically accomplished by using a radial center-out k-space trajectory and data sampling of only a few hundred microseconds in duration. Magnitude images are then reconstructed from the re-gridded k-space data. In order to achieve a better delineation of short T_2 tissues, several long T_2 suppression techniques have been developed, including dual echo subtraction techniques [1], and long T_2 preparation clusters using either long-duration hard pulses [2] or adiabatic pulses [3] to saturate or to invert and null long T_2 tissues. We present a specialized RF technique based on applying a 180° RF excitation pulse that can achieve short T_2 tissue excitation and long T_2 tissue suppression simultaneously.

Theory: The classical notion of a flip angle $\theta = \gamma B_1 \tau$ is derived from the Bloch equations while ignoring the T_2 transverse relaxation during the RF pulse. For tissues with rapid transverse relaxation, the intrinsic T_2 can be on the same order as the RF duration τ, so that the signal decay during the RF pulse may no longer be ignored, resulting in an altered magnetization trajectory [4]. This altered trajectory can be used to selectively excite only short T_2 tissues. Our technique is based on applying 180° RF pulses, which adequately invert only the longer T_2 tissues (which therefore generate no MR signal) while leaving short T_2 tissues partially in the transverse plane (generating MR signal) as illustrated in Fig.1. For a spoiled hard RF pulse train, the steady state transverse magnetization M_{ss} including T_2 decay is given by [5]:

$$M_{SS} = M_0 \frac{\theta(1-E_1)\exp\left(-\frac{\kappa}{2}\right)\sin\left(\sqrt{\theta^2-\frac{\kappa^2}{4}}\right)}{\sqrt{\theta^2-\frac{\kappa^2}{4}}\left\{1-E_1\exp\left(-\frac{\kappa}{2}\right)\left[\cos\left(\sqrt{\theta^2-\frac{\kappa^2}{4}}\right)+\frac{1}{2\sqrt{\frac{\theta^2}{\kappa^2}-\frac{1}{4}}}\sin\left(\sqrt{\theta^2-\frac{\kappa^2}{4}}\right)\right]\right\}} \quad with \quad \kappa \equiv \frac{\tau}{T_2} \quad and \quad E_1 = \exp\left(-\frac{TR}{T_1}\right) \quad (1)$$

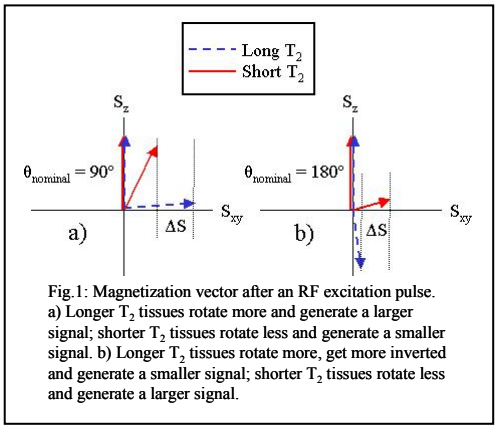

Fig.1: Magnetization vector after an RF excitation pulse. a) Longer T_2 tissues rotate more and generate a larger signal; shorter T_2 tissues rotate less and generate a smaller signal. b) Longer T_2 tissues rotate more, get more inverted and generate a smaller signal; shorter T_2 tissues rotate less and generate a larger signal.

Eq.[1] can be used to find the optimum value of κ to maximize the steady state signal numerically. (Alternatively, setting the derivative $dM_{ss}/d\kappa = 0$ yields a transcendental equation, which would also require a numerical solution). Using $\theta = 180°$, a plot of M_{ss} vs. κ for different values of TR/T_1 is shown in Fig.2. For this plot, M_{ss} was normalized by $1/\sqrt{TR}$ and therefore represents the SNR efficiency of the sequence. As expected, the steady state transverse magnetization goes to zero as κ goes to zero (full inversion). Therefore, long T_2 tissues ($T_2 >> \tau$) generates no MR signal and all that remains, is to maximize M_{ss} for the short T_2 tissues. Shown as dots are the locations of the peaks of $M_{ss}/\sqrt{TR}$. Fig.2 reveals that the optimum value of TR/T_1 to maximize the SNR efficiency lies near $TR/T_1 \approx 1$, which represents a readily achievable regime for MR imaging. The optimum values of κ as a function of TR/T_1 (for $\theta = 180°$) are shown in Fig.3. Superimposed as colored dots are the optimum values of κ evaluated at the discrete values of TR/T_1 shown in Fig.2. In the limit as the $TR >> T_1$ the optimum value of κ asymptotically approaches $\kappa \approx 7.5$. In the other limit as $TR << T_1$ the optimum values of κ rise sharply and enter an impractical regime ($\tau >> T_2$) for $TR/T_1 \rightarrow 0$. Hence, the optimum parameters to maximize SNR efficiency using $\theta = 180°$ excitation pulses for short T_2 tissues are $TR/T_1 \approx 1$ and (using Fig.3) $\tau/T_2 \approx 10$.

Experimental Verification: In order to verify the theoretical results, cortical bone specimen and phantom tests were performed. A cortical bone specimen ($T_2 \approx 0.4$ ms) and a saline ($T_2 \approx 125$ ms) filled syringe (with rubber stopper: $T_2 \approx 0.4$ ms) were arranged in a single plane. The imaging sequence used in these experiments consisted of a simple non-selective hard RF pulse excitation, followed by a 2D radial UTE k-space acquisition (TE = 12 μs), resulting in a projection image through the slice direction. Fig.4a shows the UTE image at a nominal flip angle of 30°. The image is dominated by the long T_2 saline signal, compared to the short T_2 cortical bone and the rubber stopper. The UTE image in Fig.4b was obtained using a 2ms RF pulse with nominal flip angle of 180°. The higher signal intensity of the short T_2 rubber and bone compared to the long T_2 saline in Fig.4b confirms that reversed T_2 contrast can be achieved from a 180° RF excitation pulse alone. Since the high spin density of the rubber stopper dominates the dynamic range of Fig.4b, Fig.4c shows the same image, re-windowed to better visualize the cortical bone. Note that the T_2 contrast within the cortical bone is inverted in Fig.4c, that is darker regions within the bone in Fig.4a are brighter in Fig.4c

Discussion: We experimentally tested a specialized pulse sequences using 180° RF pulses to achieve short T_2 tissue excitation and long T_2 tissue suppression simultaneously, which may open the possibility for direct excitation of only short T_2 tissues, in place of additional separate long T_2 suppression techniques. We found that the optimum parameters to maximize SNR efficiency using $\theta = 180°$ excitation pulses for short T_2 tissues are $TR/T_1 \approx 1$ and $\tau/T_2 \approx 10$. However, as Fig.2 reveals, for values of TR/T_1 that deviate by a factor of two ($TR/T_1 \approx 0.5$ or $TR/T_1 \approx 2$), the SNR efficiency is only slightly reduced. RF pulses exceeding a few milliseconds quickly become sensitive to off resonance effects during excitation. This puts a practical upper limit on the longest T_2 that can be readily excited by a 180° excitation pulse of a few hundred microseconds. Alternatively, using RF pulses with reduced pulse duration of $\tau \approx 5T_2$, causes only a small loss in SNR efficiency (see Fig.2). B_1 inhomogeneity may prove a particular challenge for our technique, causing degraded long T_2 suppression.

References: [1] Rahmer, et al. *MAGMA*, 2007. **20**(2): p. 83-92. [2] Wu, et al. *MRM*, 2003. **50**(1): p. 59-68.
[3] Larson, et al. *MRM*, 2007. **58**(5): p. 952-61. [4] Tyler, et al. JMRI 25:279 (2007). [5] Carl, et al. *ISMRM*. 2009, p. 4343. Honolulu, Hawaii, USA.

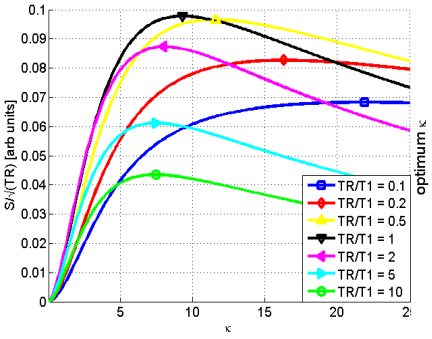

Fig.2: Transverse steady state signal for $\theta = 180°$ for different values of TR/T_1 as a function of $\kappa \equiv \tau/T_2$.

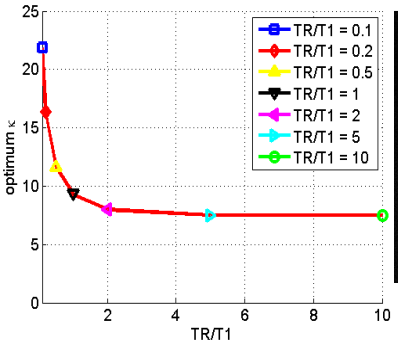

Fig.3: Optimum values of $\kappa \equiv \tau/T_2$ as a function of TR/T_1. Superimposed are the optimum values of κ evaluated at the discrete values of TR/T_1 shown in Fig.2.

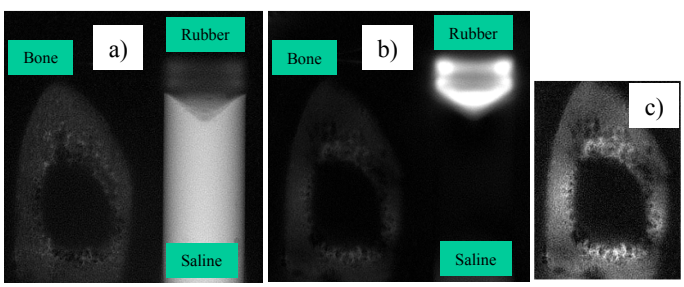

Fig.4: UTE image of cortical bone and saline filled syringe, along with rubber stopper. a) Image obtained at 30° flip angle. The image is dominated by the long T_2 saline signal, compared to the short T_2 cortical bone and the rubber stopper. b) Image obtained at 180° flip angle. Here the image is dominated by both the short T_2 cortical bone and the rubber stopper (high proton density) compared to the long T_2 saline. c) Same image as in b), re-windowed to better visualize the cortical bone.