Spectral Resolution versus SNR and Acquisition Duration: An Adversarial Balancing Act.

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BACKGROUND. The exquisitely high spectral resolution of MR spectroscopy (MRS) has always gone hand-in-glove with low signal to noise ratio (SNR). Good spectral resolution requires *long* signal acquisition times (T_{acq}) and therefore long repetition times (TR), which is antagonistic to better SNR, requiring *short* TRs [1]. These conflicting requirements suggest that there may exist a "sweet spot". To find it, a search criterion has to be formulated. The question of resolvability of two adjacent, equal intensity, spectral lines was addressed in optics by Rayleigh in the latter half of the XIX century and states that the lines are resolved when the amplitude at the center, or saddle point is $8/\pi^2 \approx 0.81$ of their maximum amplitude. Accordingly, two Lorentzians characterized by the decay time T_2^* are resolved if their frequency separation $\Delta\omega$ is more than $0.66\pi/T_2^*$ rad/sec. In MRS, however, the situation is more complicated than that in optics due to its low SNR. Indeed, the saddle may appear or disappear due to random noise fluctuations hiding its underlying spectral origin. Therefore, we extend the Rayleigh's criterion to measure the depth of the saddle in units of the standard deviation of the noise *i.e.* the spectral contrast to noise ratio (CNR) [2]. The larger the CNR, the smaller the chance of the random origin of the saddle and thus the better the two lines are resolved.

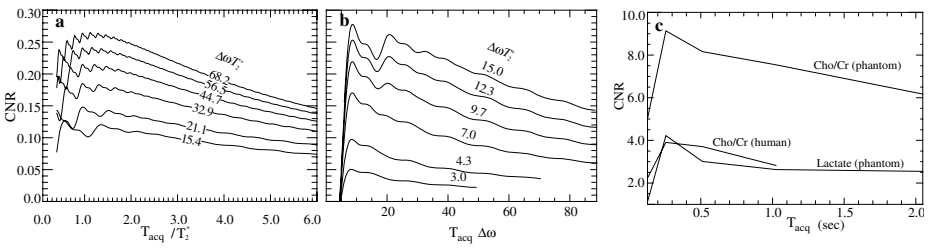

Figure 1. Theoretical CNR including spin excitation history effects as a function of acquisition duration (a,b). Optimal duration for well separated lines (a) is $1.5T_2^*$*, and* $2.5\pi/\Delta\omega$ *otherwise (b). Experimental CNR (c) attains maxima around* T_{acq}*=256 ms.*

THEORY. Assuming that the measurement noise is white and the sampling interval is much smaller than T_2^*, introducing $T = T_{acq}/T_2^*$, $x = \Delta\omega T_2^*$, and denoting $\Re$ the real part operator, the dependence of CNR on the readout duration, T_{acq}, given by

$$CNR(T) \propto \sqrt{\frac{x}{2T\Delta\omega(1-sinc(xT/2))}}\left(\Re\left(\frac{1-e^{-T}\,e^{+ixT}}{1-ix}\right)+1-e^{-T}-2\,\Re\left(\frac{1-e^{-T}\,e^{+ixT/2}}{1-ix/2}\right)\right),$$

is plotted in figure 1. The function exhibits a peak at $T_{acq}=1.5T_2^*$ for well frequency separated lines ($\Delta\omega T_2^* > 4.5\pi$) and $T_{acq} = 2.5\pi/\Delta\omega$ otherwise.

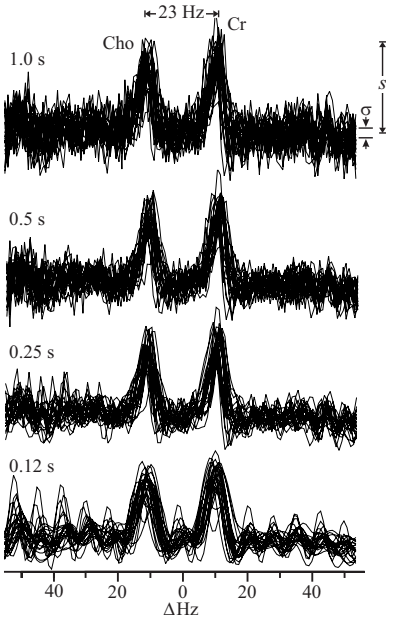

Figure 2. In vivo spectra from 24 voxels around Cho and Cr acquired from a volunteer with different readout durations. Note, maximal CNR is achieved with $T_{acq} \approx 1.5 \cdot T_2^* = 250$ *ms.*

DATA ACQUISTION and RESULTS. Experiments were performed in a Siemens Trio 3T full-body MRI scanner. Three dimensional proton (^{1}H) MRS with (7.5 mm)^3=0.42 cm^3 resolution [3], were acquired from the brain of a healthy volunteer who gave a written Institutional Review Board approved consent, and a standard brain-metabolites phantom. The MRS data were reconstructed offline with no spatial or temporal filters, using custom software. The five (four in human) different readout durations, 2048, 1024, 512, 256, 128 ms, were emulated by acquiring the ^{1}H-MRS signals for 2048 (1024) ms and correspondingly discarding trailing points of the free induction decay in post processing. The *in vivo* results are shown in Fig. 2. For each T_{acq}, the depth of the valleys in the spectra from all voxels were used to estimate the CNR of the lactate doublet (phantom only) and the choline-creatine group (figure 1c and 2). Note that the observed CNR (figure 1c) is in excellent agreement with theoretical prediction (figure 1a).

CONCLUSION. For a given fixed total experimental length, the conflict between good SNR *and* spectral resolution culminated in $T_{acq}=1.5T_2^*$ condition for the optimal CNR. Contrary to long acquisition ($T_{acq} \gg 5T_2^*$) with application of the T_2^*-matched filter [1], the advantages of the presented method are due to not acquiring mostly noise data at the tails of the free induction decay and by increasing the number of averages.

References. 1. Ernst RR, Bodenhausen G, Wokaun A. Principles of Nuclear Magnetic Resonance in One and Two Dimensions. Oxford University Press, 1987. 2. Haacke EM, Brown RW, Thompson MR, Venkatesan R. Magnetic Resonance Imaging: Physical Principles and Sequence Design. John Wiley & Sons, 1999. 3. Gonen O, Arias-Mendoza F, Coelman G. 3D localized in vivo 1H spectroscopy of human brain using a hybrid of 1D-Hadamard with 2D-chemical shift imaging. Magn. Reson. Med. 1997, 37:644-650.