Cross Peak Enhancement in Constant-Time COSY (CT-COSY)

S. Ramadan[1], and C. E. Mountford[1]

[1]Radiology, Brigham and Women's Hospital, Boston, MA, United States

INTRODUCTION

In vivo Localized Correlation Spectroscopy (L-COSY, 90° 180° Δt1 90° Acq, where all RF pulses are slice-selective) (1,2) is a developing technique that enables the researcher to un-scramble spectroscopic finding is a relatively narrow spectral bandwidth along a second dimension, and thus, facilitating analysis and improving reliability. However, the coexistence of in-phase diagonal peaks and anti-phase cross peaks in the same spectrum might lead to a situation in which anti-phase cross peaks canceling each other and in-phase diagonal peaks reinforcing one another. In constant time (CT)-COSY (3,4), however, the experimenter has the privilege to (a) increase effective resolution along F1 by eliminating peak splitting due to homonuclear coupling and (b) increase signal to noise ratio due to the collapse of many peaks into one peak along F1. We propose here a method by which amplitude of cross peaks can be improved and that of diagonal peaks reduced when the time delay between the leading 90° and terminal 90° RF pulses increases. Brain phantom and Muscle in vivo data are presented for illustration.

MATERIALS AND METHODS

The pulse sequence used in this work is shown in Figure 1, and is made out of [90° (CT-Δt1)/2 180° (CT+Δt1)/2 90° Acq]. Thus, the delay between the first and last RF pulses is constant and is equal to CT. Effectively, the first and second RF pulses generate a spin echo, which is then allowed to evolve before allowing for coherence transfer to occur by the terminal 90° RF pulse.

MR data was collected on a standard Magnetom Tim Trio system (Siemens AG, Erlangen, Germany) with a 60 cm diameter bore (software version VB15A) and a head matrix (Siemens AG, Erlangen, Germany) for phantom experiments or an 8 channel knee coil (In Vivo Corp., Gainesville, FL) for in vivo muscle studies.

Phantom spectra were acquired from a brain phantom that contains brain metabolites at physiological conditions.

Phantom experimental conditions were: RF carrier frequency at 2.4 ppm, TR 2 s, voxel size: 4x4x4 cm^3, water suppression was achieved with WET (5), spectral width=2000 Hz, increments size of 0.6ms was used in 64 Δt1 increments giving an indirect spectral width of 1666 Hz, 2 averages per increment, 1024 data points were acquired. Two CT-COSY experiment were acquired, one with CT=40.5 ms, and another with CT=58.5 ms.

In vivo muscle spectra were acquired from the soleus muscle of a consenting male volunteer. Experimental parameters were: RF carrier frequency at 2.4 ppm, TR=1.8s, Voxel size: 1.6x1.7x3.0 cm^3 , water suppression was achieved with WET (5), spectral width=2000 Hz, increments size of 0.6ms was used in 50 Δt1 increments giving an indirect spectral width of 1666 Hz, 8 averages per increment, 1024 data points were acquired. Two CT values were used 34 ms and 68 ms.

Two-dimensional spectral processing was equivalent for each pair of 2D spectra, and was done in Felix (6). Fourier transform was done after zero filling to double the acquired matrix size in both dimensions, and multiplying F2 and F1 by skewed $sine^2$ and $sine^2$ apodization functions, respectively. Peaks volume evaluation was done using standard Felix software features, and all values were normalized for comparison.

RESULTS AND DISCUSSION

As CT increased from 40.5 to 58.5 ms, phantom cross peaks volume increased variably from 16 to 51%.

In vivo, however, as CT increased from 34 ms to 68 ms, volume of cross peaks increased variably up to 1125 %, as can be seen in Figure 2. In practical terms, new cross peaks emerged as CT increased.

Assuming a two-spin, weakly coupled system and using product operator formalism to analyse the magnetization, we obtain two observable terms after the terminal 90 RF pulse:

$$\sin(w1\ t1)\ \sin(\pi\ J_{12}\ CT)\ 2\ I_{1z}\ I_{2y} - \sin(w1\ t1)\ \cos(\pi\ J_{12}\ CT)\ I_{1x}$$

The first term ($I_{1z}\ I_{2y}$) gives rise to an anti-phase cross peak with a single modulation frequency in F1 ($\sin(w1\ t1)$). The second term which gives rise to a diagonal peak, has also the same modulation frequency in F1 as the cross peak. However, the *amplitude* of cross and diagonal peaks are functions of sine and cosine of the same argument, respectively, and these functions change differently as their arguments vary. It can be easily seen that the cross peaks have a zero intensity when $CT=2n/(2\ J_{12})$, and maximum intensity when $CT=(2n+1)/(2\ J_{12})$ where n is a non-zero integer. Thus, in CT-COSY, the amplitude of cross peaks is a function of J value and CT, and for an unfortunate value of CT, some cross peaks might be missed. Due to the sine/cosine dependence, the diagonal peak amplitude decreases and cross peak amplitude increases as CT increases.

One disadvantage of CT-COSY is that since CT is constant, T2* losses in CT-COSY are usually higher than standard L-COSY where 90°-90° delay increases as Δt1 increases. This signal loss is critical for short T2 species and at high fields. CT choice must only be done after comparison with standard L-COSY (1) to make sure that all cross peaks are being detected at a particular CT.

CONCLUSION

Cross peak intensity of CT-COSY can be improved when CT is increased. Care must be taken, though, not to increase CT too long as T2 magnetization losses can reduce the overall sensitivity of the experiment.
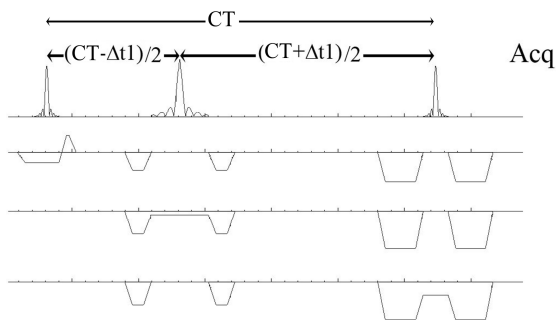

Figure1. CT-COSY pulse sequence. The sequence is made of three RF pulses: [90°-180°-90°-Acq] with 90°-90° delay equals to CT. Second and third RF pulses are surrounded by magnetic pulse spoilers, and a slice selective gradient is applied during each RF pulse.
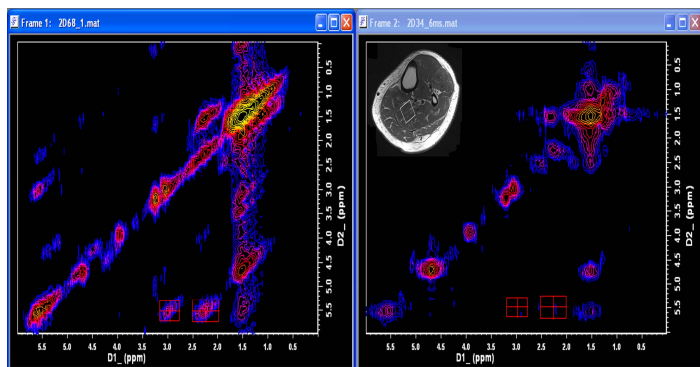

Figure2. CT-COSY spectra with CT=68 ms (left) and CT=34 ms (right). Improvement of cross peak intensity (in small read boxes) as CT increases is evident. Inset shows calf muscle axial image and position of voxel.

REFERENCES

1. Thomas MA, et al. Magn Reson Med 2001;46:58-67. **2**. Ziegler A, et al. Magn Reson Mater Phy 2001;14:45-49.
3. Berger S. Spectros Lett 2000;33(1):1-8**. 4**. Girvin ME. J Magn Reson Ser A 1994;108:99-102.
5. Ogg RJ. J Magn Reson Ser B 1994;104:1-10. **6**.Felix. Felix NMR. Version 2007; 2007.