Eliminating Spurious Lipid Sidebands in $^1$H MRS of Breast Lesions


Detecting metabolites in breast lesions by in vivo $^1$H MR spectroscopy can be difficult due to the abundance of mobile lipids in the breast which can produce spurious sidebands that interfere with the metabolite signals. Two-dimensional J-resolved spectroscopy has been demonstrated in the brain as a means to eliminate these artifacts from a large water signal; coherent sidebands are resolved at their natural frequencies, leaving the noncoupled metabolite resonances in the zero frequency trace of the 2D spectrum. This work demonstrates that using the zero frequency trace—or equivalently the average of spectra acquired with different echo times—can be used to separate noncoupled metabolite signals from the lipid-induced sidebands. This technique is demonstrated with simulations, phantom studies, and in several breast lesions. Compared to the conventional approach using a single echo time, echo time averaging provides increased sensitivity for the study of small and irregularly shaped lesions. 

Key words: sideband artifact reduction; echo-time averaging; breast cancer; choline compounds

$^1$H MRS studies of excised breast tissue (1–3) have shown that a resonance at 3.2 ppm, attributed to choline (Cho)-containing compounds, increases in malignancy. Several groups have published results showing that this peak can be detected using in vivo MRS and that its presence is highly correlated with cancer (4–8).

Typical breast MRS studies use a single-voxel localization technique with the intent of selecting signals from the lesion of interest and excluding signals from the surrounding adipose tissue and normal parenchyma. If the intent is to detect the presence of elevated Cho, it is acceptable to include some signal from other tissues as long as it does not interfere with the Cho signal from the lesion. The strong mobile lipid signal from adipose tissue is problematic because it can introduce spurious sidebands which may be indistinguishable from a Cho resonance. Planning a spectroscopic examination requires skill in selecting an optimal voxel size and position that includes as much of the lesion as possible to maximize the Cho signal while minimizing the presence of adipose tissue, which may create sideband artifacts. This can be difficult due to the heterogeneous distribution of tissue in the breast, the irregular shape of lesions, and the limitations on the minimum voxel size (typically 0.5–1 mL). In our experience of studying over 330 voxels in 70 patients with single-voxel $^1$H MRS, we have found that it is often these lipid-induced sidebands rather than signal-to-noise considerations that limit the sensitivity for detecting Cho. Reducing or eliminating these artifacts would permit the spectroscopist to plan larger voxels around lesions with irregular shapes without undue regard for exclusion of adipose tissue.

The source of these sideband artifacts is a modulation in the $B_0$ field caused by the pulsed gradients (9–11). This modulation causes the appearance of sidebands—satellite peaks distributed antisymmetrically about large resonances. In the breast, this sideband pattern is convolved with each of the broad lipid resonances in the spectrum, making it difficult to distinguish between real and artifact peaks. The conventional approach to reduce the artifact is to suppress the large lipid signal. This can be done most simply by using a long echo time, since the $T_2$ of Cho is longer than that of lipid (2). More direct approaches such as frequency-selective suppression, frequency-selective excitation (12), inversion recovery, or postprocessing techniques (13) can also be used individually or in combination. Selective suppression of the lipid signal is particularly difficult due to its multiple resonances, short spin-lattice relaxation rates ($T_1$), and highly variable concentration. These techniques can be effective but they have some disadvantages, such as reduction of the Cho signal, possible introduction of artifacts, and difficulty of implementation. Hurd et al. (14) proposed the use of 2D oversampled J-resolved spectroscopy in the brain to eliminate sideband artifacts from unsuppressed water signal. Since the sidebands are coherent and modulate periodically with echo time $TE$, they are resolved at their frequency of oscillation in the second ($F_2$) dimension and noncoupled resonances, such as Cho, are left intact at the zero-frequency trace ($F_1 = 0$) of the 2D spectrum. This experiment is performed by acquiring a number of 1D spectra with incrementally longer echo times and applying a Fourier transform on the set of spectra to produce a 2D spectrum. It can be shown that the zero-frequency trace of the 2D spectrum, previously termed the $J_0$ spectrum, can alternatively be produced by simply averaging the complex FIDs from each of the 1D spectra. For simplicity, this spectrum—produced either by averaging or via a second Fourier transform—will be referred to as the echo-time averaged or TE-averaged spectrum. Compared to a con-

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The TE-averaged spectrum has greatly reduced artifacts at the cost of a small signal loss due to $T_2$ decay.

The technique of averaging multiple echo times has been previously proposed under the names SIMPLE (15) and anti-editing (16). In each of these studies, spectra from several different echo times were averaged to cause cancellation of J-coupled resonances while causing constructive addition of noncoupled resonances. Since the frequencies of $^1$H J-couplings are limited to the 5–10 Hz range, only a few echo times need to be averaged to create cancellation of the coupled resonances. The use of TE averaging to eliminate sidebands is conceptually the same; artifacts add destructively while noncoupled metabolite signals add constructively. However, cancellation of sidebands requires averaging more TE values since the $B_0$ modulation typically has a wider range of frequencies than $^1$H J-couplings.

The goal of this article is to demonstrate the applicability of TE averaging in reducing lipid-induced sidebands in breast MRS. A model of the artifact is presented and verified via simulation. The ability of this technique to suppress this artifact is demonstrated with simulations, phantom studies, and in vivo experiments. The in vivo cases demonstrate how this technique can reduce the likelihood of false-positive and false-negative findings of Cho in voxels with high lipid content. Finally, implications of this development on voxel planning strategy for breast MRS are discussed.

**THEORY**

The original homonuclear 2D J-resolved spectroscopy experiment (17) used a 90°–$t_1$–180°–$t_2$–acquire pulse sequence, where $t_1$ is equivalent to the echo time $TE$, J-coupling of resonances evolves in the $t_1$ time period and chemical shift evolves in the acquisition period, called $t_2$.

The time-domain signal from a monoexponentially decaying singlet resonance is:

$$s(t_1, t_2) = M_0 e^{-t_1/T_2^*} e^{i2\pi f_0 t_2} e^{-\omega t_2^*},$$  

where $M_0$ is the signal intensity, $T_2^*$ is the transverse relaxation time, $T_2$ is the effective transverse relaxation time, and $\omega$ is the frequency of the resonance. The 2D experiment is performed by sampling the signal $s$ in both the $t_1$ and $t_2$ dimensions to produce a discrete 2D signal $s[n_1, n_2]$. A series of $N_2$ different 1D FIDs is acquired at echo times $t_1 = TE_0 + n_1\Delta TE$ to sample the $t_1$ dimension with a spectral width $SW_1 = 1/\Delta TE$. Each FID is sampled with $N_2$ points at times $t_2 = n_2\Delta t_2$, giving a spectral width $SW_2 = 1/\Delta t_2$ in the $t_2$ dimension. Applying a discrete Fourier transform on the chemical shift dimension ($F_2$) produces a set of spectra, $s[n_1, F_2] = DFT_2[s[n_1, n_2]]$. A discrete Fourier transform is then applied on the $n_1$ dimension to produce the discrete 2D spectrum:

$$s[F_1, F_2] = DFT_1[s[n_1, F_2]] = \sum_{n_1=0}^{N_1-1} s[n_1, F_2] e^{-i2\pi F_1 n_1/N_1}. \tag{2}$$

The $F_1 = 0$ trace, or so-called $f_0$ spectrum, of $s[F_1, F_2]$ can be found by setting $F_1 = 0$ in Eq. [2]:

$$s[0, F_2] = \sum_{n_1=0}^{N_1-1} s[n_1, F_2]. \tag{3}$$

The trace $s[0, F_2]$ is equal to the sum overall of the $N_1$ 1D spectra. Therefore, the 2D $f_0$ spectrum is equal to the average of the set of echo-time incremented 1D spectra $s[n_1, F_2]$, scaled by $N_1$.

For a single resonance without J-coupling or any other $t_2$-varying processes, a spectrum acquired at $TE = t_1$ can be modeled as an ideal spectrum acquired at $TE = 0$ multiplied by an exponential decay constant: $s[n_1, F_2] = s_0[F_2] e^{-t_1/\tau}$. Substituting this and $t_1 = TE_0 + n_1\Delta TE$ into Eq. [3] gives

$$s[0, F_2] = s_0[F_2] e^{-TE_0/\tau} \sum_{n_1=0}^{N_1-1} e^{-n_1\Delta TE/\tau}. \tag{4}$$

This equation gives a measure of the signal intensity of a single metabolite in a TE-averaged spectrum. A conventional 1D spectrum with NEX averages can be considered a special case of this experiment, where $\Delta TE = 0$, $TE_0$ is the echo time, and $N_1 = NEX$. Equation [4] can be used to determine the effect of acquisition parameters on metabolite signal intensity and to correct for $T_2$ relaxation in quantitative studies.

To model the artifact which produces the spurious sidebands, it is assumed that the sudden application of a field during acquisition produces sidebands at the frequency of false-positive and false-negative findings of Cho in voxels with high lipid content. Finally, implications of this development on voxel planning strategy for breast MRS are discussed.

**METHODS**

All measurements were performed with a hybrid 4 T system, consisting of a 90 cm bore magnet (model 4T-900, Oxford Magnet Technology, Oxfordshire, UK) interfaced with an imaging spectrometer (model Unity Inova, Varian, Palo Alto, CA). The gradient system (model Sonata, Sie-
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FIG. 1. LASER pulse sequence for single-voxel localization with adiabatic pulses. The adiabatic half passage (AHP) excites all magnetization into the transverse plane. Six equally spaced adiabatic full passage (AFP) pulses are then applied in three pairs, with each pair providing slab-selective refocusing in one Cartesian direction. Gradients shaded black are for spatial selection, those shaded gray are for crushing. Water suppression is performed using VAPOR. Signal is acquired during the $t_2$ period, and the $t_1$ time is varied by incrementing $\tau_{CP}$ to produce the $F_1$ spectral dimension or to accomplish TE averaging.

Spectra were acquired with a pulse sequence called LASER (localization by adiabatic selective refocusing), shown in Fig. 1. LASER is an adiabatic single-voxel localization method based on spatially selective 180° pulses similar to PRESS (19, 20), described in detail elsewhere (21). Briefly, a nonselective adiabatic half passage (AHP) pulse is used to excite all resonances. This is followed by a spin-echo train of three pairs of adiabatic full passage (AFP) pulses. Each pair selects a slab of spins while refocusing the effects of $B_0$ variation and chemical shift. LASERs high bandwidth adiabatic pulses provide both $B_0$ insensitivity and very accurate spatial localization and require less peak power than equivalent amplitude-modulated pulses. The echo time was varied by uniformly incrementing each interpulse time $\tau_{CP}$ so that the $t_1$ period corresponded to the full TE, lasting from the end of the AHP excitation pulse to the beginning of the acquisition. Water suppression was performed using VAPOR (22). All spectra were acquired with 4096 complex points and an $F_2$ spectral width of 6000 Hz.

A phantom study was performed using a 500-mL flask filled with tapwater, placed in a breast coil in the center of the magnet. A 10 $\times$ 10 $\times$ 10 mm voxel was placed in the magnet isocenter and spectra were acquired using the LASER sequence. Two series of conventional spectra were acquired using $TE = 45$ and 196.2 ms and $NEX = 64$. Another series was acquired on the same voxel using 64 different TE values varying incrementally from 45–196.2 ms, each with $NEX = 1$. This data series was processed in two ways: 1) averaging the complex FIDs and performing a single Fourier transform in the $F_1$ dimension to produce a 1D TE-averaged spectrum, and 2) applying a 2D Fourier transform on the FIDs to produce a 2D $F_1$-resolved spectrum with $F_1$ spectral width $SW_1 = 1/\Delta TE = 417$ Hz. These experiments were then repeated on the same voxel using a standard PRESS sequence to verify that the artifacts were not specific to the LASER sequence. The maximum amplitude of crushing gradients was 32 mT/m with a rise time of 400 $\mu$s in both sequences. The PRESS echo time was adjusted in the conventional asymmetric fashion by varying the delays around the second refocusing pulse and keeping the delays around the first refocusing pulse constant.

Synthesized spectra were generated in Matlab (MathWorks, Natick, MA) using a model similar to Eq. [5] and applying Fourier transforms to the resultant FIDs. Acquisition parameters were selected to match the experiments described above. Parameters describing the resonance ($T_1$, $T_2^*$) and artifacts ($f_g$, $A_g$, $T_g$) were adjusted to generate spectra matching those observed on our system. To simulate the artifacts generated in the LASER sequence, it was assumed that the effect of the crusher gradients surrounding each refocusing pulse could be modeled as a single $B_0$ perturbation occurring at the center of the pulse. Furthermore, only the perturbations associated with the final two refocusing pulses were modeled, since these are closest to the acquisition period and have the largest effect. As can be seen in Fig. 1, perturbations coincident with the final two pulses begin at a time $T_{1g} = TE/12$ and $T_{2g} = TE/4$ prior to the acquisition period. The modulation frequency, $f_g$, was set to 360 Hz, consistent with the dominant frequency observed on our system.

TE averaging was tested on a subset of patients participating in an ongoing study of breast MRI/MRS being conducted at our institution. Patients with either suspicious findings in other imaging modalities or those with biopsy-confirmed breast cancer were recruited for our study. Informed written consent was acquired prior to all human studies in compliance with our institutional review board. To identify lesions, a fat-suppressed 3D FLASH 256 $\times$ 256 $\times$ 64 image was acquired before and after intravenous administration of a 0.1 mmol/kg Gd-DTPA bolus. Using our own software written in Matlab, these images were processed and voxels for spectroscopy were planned while the patient remained positioned in the magnet. A power calibration, manual adjustment of the linear shims, and adjustment of water suppression was performed on each voxel. A series of 64 spectra was then acquired with TE varying incrementally from 45–196.2 ms, with $\Delta TE = 2.4$ ms. Using a 3-sec pulse repetition time gave a total acquisition time of $\sim 3.3$ min. As in the phantom studies, these data were processed by averaging each FID to produce the TE-averaged spectrum and also by applying a 2D Fourier transform to produce a 2D spectrum. To compare this technique to a conventional 1D spectrum, TE was fixed at a short echo time (39.4 or 45 ms) to maximize the Cho signal and the acquisition was repeated with
64 averages to keep the thermal noise contribution equal with both techniques.

Spectra were processed using Lorentzian line broadening (2–8 Hz), manual zero and first-order phasing, and frequency correction to account for $B_0$ field shifts due to respiration. 2D spectra are shown in absolute value mode with manually selected contour levels. Chemical shift referencing was based on water at 4.7 ppm. Spectral processing and fitting were done with software provided by the spectrometer manufacturer. Water/fat peak ratios are presented to provide an approximate measure of the lipid content in a voxel. These ratios were calculated by measuring the integral of the unsuppressed water and 1.3 ppm lipid peaks without correcting for $T_2$ relaxation.

RESULTS

The simulations show that a sinusoidal perturbation of the $B_0$ field produces distorted spectra with two side peaks distributed antisymmetrically around each true resonance. In the 2D spectrum the two sidebands occur at $(F_1,F_2) = (\pm f_g, TE/T_{m} \mp f_g)$ as shown in Fig. 2a. In the conventional 1D spectrum the sidebands appear at $\pm f_g$ with opposite phase. Since the phase of the sideband peaks modulates with TE, averaging spectra with varying TE’s causes cancellation of these peaks. Figure 2b compares the conventional spectra to the TE averaged spectrum using all 64 echo times. Even the longest TE spectrum still shows the sidebands, whereas the TE-averaged spectrum shows nearly complete removal of the artifacts and an undistorted resonance remains.

Figure 2c,d shows equivalent spectra from a LASER-localized voxel in a water phantom. There is good agreement between the simulated spectra in Fig. 2a,b and the phantom results, thus supporting the validity of the artifact model. The phantom spectra show a dominant frequency of ~360 Hz, but there are clearly more frequency components than are simulated in the model. The sidebands are distributed on axes corresponding to the last two refocusing pulses, as in the simulation. In these and other spectra not shown, it is typically the gradients around the 5th pulse which cause the greatest distortion, since these crusher gradients are three times larger than those around the last pulse. As in the simulated case, the TE-averaged 1D spectrum shows much less distortion than even the longest TE conventional spectrum.

The PRESS phantom spectra shown in Fig. 2e,f are also consistent with the model and show that the artifact is not particular to the LASER pulse sequence. The sidebands have a similar set of $f_g$ frequencies but are distributed along the $F_2$ and $F_3$ axes, corresponding to perturbations from the first and second refocusing pulses, respectively. Figure 2e shows the aliased sidebands in the $F_1$ direction. Again, the TE-averaged 1D spectrum has less distortion than the conventional spectra, although there is some residual artifact near +200 Hz, possibly due to the above-mentioned aliasing. As expected, the intensity of the maximum PRESS sidebands at the shortest echo time (~0.2% of the primary resonance) is smaller than that in LASER (~0.5%) since the final gradient pulse in LASER is nearer to the acquisition period.
To date, we have used TE averaging in vivo in 61 separate voxels in 15 breast cancer patients. In 12 instances we acquired both TE-averaged spectra and conventional spectra from the same voxel for comparison. In all cases, but most distinctly in those voxels that had high lipid content, the TE averaged spectra showed reduced artifacts. Two of these comparisons are presented here to demonstrate how TE averaging can help avoid misinterpretations of conventional spectra. Both of these cases are from voxels prescribed to enclose secondary lesions of uncertain pathology in breasts with a distinctly separate, biopsy-confirmed malignant lesion.

Figure 3 demonstrates an instance where Cho is not detectable in the spectrum collected at a single echo time, but is clearly discernable with TE averaging. Figure 3a shows a sagittal slice from a 3D fat-suppressed, $T_1$-weighted, contrast-enhanced image from a breast with biopsy-confirmed inflammatory breast cancer. The primary lesion is visible in the inferior portion of the breast. A $15 \times 16 \times 15$ mm voxel was placed around a roughly spherical lesion ($\sim 16$ mm dia.) that showed dynamic contrast uptake indicating possible malignancy. It was necessary to include some adipose tissue so the voxel could fully encompass the lesion. This led to a high lipid content in the voxel, with an unsuppressed water/fat peak ratio of 11. The upper plot in Fig. 3c shows the conventional spectrum ($TE = 45$ ms) with 64 averages. The broad side peaks in the 0 to $-2$ ppm region, and some of those in the 2 to 4 ppm region, are not actual NMR peaks but sideband artifacts attributable to the large lipid signal at 1.3 ppm. The TE-averaged spectrum shown beneath, acquired with 64 different echo times from 45–196.2 ms, shows greatly reduced artifacts in each of these regions. A close-up of the $2.4–4$ ppm region shown above demonstrates how it is difficult to distinguish between the true resonance at 3.25 ppm and the numerous sidebands in the region. The same data used for the TE-averaged spectrum was also converted into a 2D $J$-resolved spectrum, shown in Fig. 3b. This demonstrates that the pattern of gradient-induced sidebands in vivo is consistent with the simulation and phantom results in Fig. 2a,c.

Figure 4 demonstrates an instance where a peak near 3.2 ppm is visible in the conventional spectrum, but the TE-averaged spectrum reveals that this peak is likely a spurious sideband. This patient had a primary lesion of biopsy-confirmed ductal carcinoma, not visible in the slice shown in Fig. 4a. The spectra shown in Fig. 4c,d are from a small, spherical lesion ($\sim 7$ mm dia.), separate from the primary lesion, in which a measurement of Cho was attempted. The $8 \times 8 \times 9$ mm voxel placed around this lesion included significant adipose tissue, with an unsuppressed water/fat peak ratio of 0.4, so no water suppression was used. The conventional spectrum in Fig. 4c again shows sidebands on either side of the large 1.3 ppm lipid peak. There is a peak at 3.2 ppm which, based only on the spectrum acquired with $TE = 39.4$ ms, could be misinter-
preted as Cho. When TE averaging is used the baseline is undistorted and the peak disappears. This could also mean the peak is an actual NMR signal with a very short T2; however, the presence of an antisymmetric peak reflected about the 1.3 ppm lipid peak suggests the peak is artificial.

Figure 4b shows the 2D spectrum of the same data, demonstrating that the gradient-induced artifact is again present. Effective removal of this sideband avoids a false finding of high Cho; after applying TE averaging or 2D Fourier transform, the 3.2 ppm signal was below the thermal noise level.

DISCUSSION

Gradient-induced sideband artifacts are known to be problematic when trying to detect weak signals in the presence of very intense signals, as when performing brain MRS without water suppression (11). These artifacts can be even more problematic in breast spectra, even with good water suppression, because there may be several intense peaks from the different lipid moieties. Hurd et al. (14) have shown that these artifacts can be reduced or eliminated by using a homonuclear 2D J-resolved experiment. Artifacts are resolved in the F1 dimension because the gradient-induced perturbation of the B0 field evolves during a fixed proportion of the t1 period. This can be seen with the LASER sequence in Fig. 1. If a B0 perturbation with frequency f0 were to occur before the AHP, it would evolve for the entire t1 period, leading to an artificial resonance appearing at ±f0 in the F1 and F2 spectral dimensions. A perturbation originating at some point during the t1 period has less time to evolve, leading to a reduced frequency in the F1 dimension. Therefore, a B0 perturbation originating at a time Td prior to the acquisition period will produce artifact peaks occurring on the F2 = F1 · TE/Td axis.

It was shown above using Eqs. [2] and [3] that the F3 = 0 trace, or J0 spectrum, is equivalent to the result of averaging each of the spectra acquired at the various echo times. For the data presented here, these two methods of processing the data—TE averaging vs. 2D Fourier transform with extraction of the zero-frequency trace—are mathematically equivalent. If the intent is to observe singlet resonances, then TE averaging is quicker, can be performed with less sophisticated 1D software, and provides a more concise description of the processing technique.

The data shown in Figs. 2–4 clearly show how TE averaging can reduce sideband artifacts and improve the spectral baseline compared to a conventional single TE spectrum. In each of the comparisons shown, the same number of averages were used. Therefore, the amount of thermal noise is the same in each spectrum, but the TE-averaged spectra show reduced effective noise. This is because the sidebands contribute to coherent noise which is reduced by TE averaging.
The in vivo cases displayed compare a conventional spectrum acquired at short TE (39.5 or 45 ms) to a TE-averaged spectrum using much longer echo times (45–196 ms). These short echo times were selected for comparison to minimize $T_2$ losses of the Cho peak. This is not an ideal comparison since the lipid signals, and therefore the coherent noise, are more intense at the shorter echo times. Using conventional spectroscopy with a long echo time will reduce the lipid signal relative to the Cho signal because the $T_2$ of lipid (typically $\sim$100 ms) is much shorter than that of Cho (>350 ms) (2). This can lead to increased sensitivity compared to the short echo time spectrum since the coherent noise is reduced faster than the Cho signal. However, for voxels with high lipid content the coherent noise may limit sensitivity even at long echo times.

A comparison between a conventional spectrum and a TE-averaged spectrum, as in Fig. 3c, shows that both have equal thermal noise content and the peak shapes for non-coupled metabolites are the same. The TE-averaged spectrum shows two important differences: sideband artifacts and coupled resonances disappear and short $T_2$ components are greatly reduced in intensity. Signals such as Cho, which have a long $T_2$ compared to the TE range, will be reduced slightly. Assuming the singlet resonance has a monoexponential $T_2$ decay, the amount of signal reduction can be determined using Eq. [4]. For the sequence parameters presented in this work, and using a $T_2$ of 350 ms for Cho, the signal reduction is approximately 20%, which is consistent with our observations.

If there is little or no lipid in the voxel, there is no need for artifact reduction and a conventional spectrum with short TE will have a better SNR than the TE-averaged spectrum. As the amount of lipid increases, the SNR of the conventional spectrum decreases as the sidebands create coherent noise. Thus, TE averaging provides a robust method which reduces the sensitivity to lipid content at the expense of only a small decrease in SNR due to $T_2$ decay.

This insensitivity to lipid content changes the strategy used to plan voxels for breast spectroscopy. Using conventional spectroscopy, planning a voxel requires skill to minimize the adipose tissue included in the voxel while maximizing coverage of the lesion. If too much lipid is included the sideband artifacts can obscure real resonances (as in Fig. 3) or appear as peaks which could be misinterpreted as NMR resonances (Fig. 4). This could explain some of the instances reported in the literature, where a Cho signal was seen at some echo times and not others (6,8). With TE averaging, there is little penalty for including more adipose tissue in the voxel. This allows the planning of large voxels to fully enclose irregular lesions and provides greater sensitivity for studying small lesions surrounded by adipose tissue.

TE averaging can be used with localization schemes other than LASER and PRESS. Prior to implementation, it is advisable to perform a 2D J-resolved experiment with a phantom to determine the sideband frequencies, since they will vary with different hardware and pulse sequences. Due to the multiple acquisitions required, TE averaging is not readily extended into spectroscopic imaging applications.

**CONCLUSIONS**

TE averaging has been demonstrated to allow MRS detection of Cho in the presence of abundant lipid signal. Compared to conventional spectroscopy, TE averaging greatly reduces sideband artifacts caused by gradient-induced $B_0$ modulation, with a small penalty of reduced metabolite signal due to $T_2$ decay. Applied to breast spectroscopy, this technique has been shown to reduce the artifacts that can be the source of false-positive and false-negative findings of elevated Cho and allows the use of larger voxels when studying small or irregularly shaped lesions. Together, these improvements increase the sensitivity for Cho detection in vivo.

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