Applications of High-Resolution Echoplanar Spectroscopic Imaging for Structural Imaging

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Echoplanar spectroscopic imaging (EPSI) was introduced as a fast alternative for spectroscopic imaging and has been recently implemented on clinical scanners. With further advances in gradient hardware and processing strategies, EPSI can be used to obtain spectroscopic images whose spatial resolution parallels that of conventional anatomic images within clinically acceptable acquisition time. The present work demonstrates that high-resolution EPSI can be used to derive structural images for applications in which spectroscopic information is beneficial. These applications are chemical shift (fat-water) imaging, narrow-bandwidth imaging, and $T_2^*$ mapping. In this paper, the EPSI sequence design and processing strategies are detailed and experimental results in normal volunteers are presented to illustrate the potential of using EPSI in imaging anatomic structures. J. Magn. Reson Imaging 1999; 10:1–7.

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Index terms: echoplanar spectroscopic imaging; fat-water imaging; spectroscopic imaging; narrow-bandwidth imaging; $T_2^*$ mapping

To reduce the long acquisition time required for spatial encoding, fast spectroscopic imaging approaches were introduced. Among them was an echoplanar-based encoding technique first described by Mansfield (6). In echoplanar spectroscopic imaging (EPSI), an alternating gradient, which simultaneously encodes space and chemical shift (time), is applied along one of the spatial (readout) directions, thereby reducing phase-encoding dimension by one and increasing speed. Recent advances in gradient hardware, which have increased both the speed of gradient switching and gradient strength achievable and improved eddy current performance of gradients, led to successful implementations of EPSI (7–14). EPSI allows for a larger coverage of the k-space, thus permitting a relatively high spatial resolution within practically acceptable acquisition time. When signal-to-noise ratio (SNR) is not a limiting factor, EPSI can be used to obtain spectroscopic images with resolutions comparable to those of anatomic images in practical scan time. Although for most spectroscopic applications the improved spatial resolution afforded by EPSI is not utilized due to SNR limitations, it can be used for imaging applications in which spectroscopic information may be beneficial. In this paper, three such applications are described. A preliminary form of this work was presented previously (15).

The first application of EPSI described in this paper is to use EPSI to obtain fat and water images as introduced by Dixon (16). The spectroscopic imaging technique permits the separation of fat and water according to their spectroscopic signature. Because the fat and water peaks can be readily identified in the spectroscopic data, fat-water imaging based on EPSI has the advantage of being insensitive to field inhomogeneity, which is a major limitation of the two-point Dixon method (16–18). In addition, because images from spectroscopic imaging data are derived based on water and fat peaks having finite width rather than a single frequency point, the EPSI approach is not affected by the finite spread of the spectral peaks of interest, such as the broad peak of the lipid signal resulting from resonance frequencies of the methyl and ethyl groups.

Conventional MRI techniques for routine imaging applications suffer from chemical shift artifact (CSA), which arises from inconsistencies in the resonance
frequency due to chemical shift differences of tissues (or B₀ inhomogeneity). At 1.5 T the difference between the fat and water resonances is approximately 220 Hz (or 3.5 ppm). Due to this difference, images of the fat are shifted from that of the water, and areas with severe B₀ inhomogeneity are spatially distorted along the frequency-encoding direction. The extent of CSA depends on the sampling bandwidth (relative to the frequency differences), which is dictated by the strength of the frequency-encoding gradient. The larger the bandwidth, the smaller the chemical shift artifact. Thus, CSA can only be reduced by the use of a broad bandwidth. Since SNR in MR images is inversely proportional to the square root of the bandwidth, a compromise between SNR and chemical shift artifact has to be made in practice for conventional imaging sequences. In a fast spectroscopic imaging sequence, spatially encoded data acquired at different time points can be used to generate images that are virtually free of CSA. These images can be subsequently combined to produce a CSA-free image with an SNR corresponding to the total sampling time of the free induction decay (FID). In other words, CSA-free structural images having an effective narrow bandwidth can be derived from properly acquired EPSI data. Since T2-weighted images are the most SNR limited, the application of EPSI for CSA free imaging is demonstrated in this work for long-TE spin-echo imaging.

Recently MRI has been applied to characterize bone properties by transverse relaxation rate induced by local magnetic field variations (19–24). The trabecular bone network forms interfaces with the bone marrow, leading to susceptibility effects and increases of transverse relaxation rate (R₂*) of bone marrow, the amount of increase depending on the trabecular bone density and the bone architecture. The change of the magnetic properties of bone marrow by the bone structure can be used to assess changes in the structure of bone in diseases such as osteoporosis. The measured T₂* has been shown to be a good discriminator between osteoporosis patients and normals (22,23). Because EPSI samples T₂* decay curve directly, it can be readily used to derive T₂⁎ (or R₂*) maps. Therefore, the third application of EPSI in this paper is T₂* mapping. The advantages of using spectroscopic approaches for T₂* mapping are 1) the high resolution in resolving various T₂* and 2) the ability to obtain peak specific T₂*.

**MATERIALS AND METHODS**

**Acquisition**

The EPSI sequence was implemented on a 1.5 T SIEMENS Vision scanner (SIEMENS Medical Systems, Iselin, NJ), which is capable of achieving a maximum gradient of 25 mT/m with sinusoidal ramping in 300 μsec. EPSI data were acquired using a spin-echo excitation with phase encoding (Gpe) in one dimension and echoplanar encoding (Gepe) in the other. The sequence shown acquires only the second half of the spin-echo although full echo is acquired for narrow-bandwidth imaging.

320 points, which were interpolated to 232 (256) equidistant k-space points in processing for the 500 (240) mm FOV. During processing echoes arising from the positive gradient lobe and the negative gradient lobe are processed separately and averaged to obtain the final matrix. Considering each echo separately leads to time resolution, δt, equal to the duration of two gradient lobes.

For fat-water imaging a cross section of the pelvis was studied with an EPSI sequence (TR/TE 500/43 msec, slice thickness 10 mm. FOV 500 mm, number of excitations (NEX) 2, matrix size 232 × 256,δt 1.6 msec, a total of 64 echo pairs acquired in 102.4 msec for each phase-encoding step, and TA 256 seconds). The TR/TE were chosen for T₁ weighting in the image. To demonstrate the insensitivity of the EPSI-based approach to field inhomogeneity, the imaging data were collected without subject-specific localized shimming. A standard Siemens quadrature body coil was used for data acquisition. For comparison, fat-water images were also obtained with the two-point Dixon technique (16) using a spin-echo excitation (17,18) and comparable parameters. Based on the 220 Hz chemical shift between fat and water at 1.5 T, the out-phase image was acquired with a 2.27 msec TE shift in the Dixon technique.

Narrow bandwidth images of a normal volunteer were obtained in the pelvis using an EPSI sequence (TR/TE 2000/70 msec. slice thickness 10 mm. FOV 500 mm, NEX 1, matrix size 232 × 256,δt 1.6 msec, 32 echo pairs acquired in 51.2 msec for each phase-encoding, and TA 512 seconds) as well as a spin-echo sequence with comparable imaging parameters. Unlike the other two applications, the EPSI data acquisition window for narrow bandwidth imaging was centered about the spin-echo generated by the 90 and 180 pulses for optimal SNR. The TR/TE were selected for T₁ weighting appropriate in the body, and the long readout time (51.2 msec) was employed to achieve an effective narrow bandwidth. The standard Siemens quadrature body coil was used for this study. The effective bandwidth was 19.5 Hz/pixel.

T₂* mapping using EPSI was performed on a coronal slice of a knee in a normal volunteer with TR/TE 300/43 msec, slice thickness 10 mm, FOV 240 mm, NEX 1, matrix size 256 × 256,δt 3.2 msec, 64 echo pairs.
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acquired in 204.8 msec, and TA 78 seconds. The choice of TR/TE was made to acquire adequate data points in time for T2* calculation with the shortest possible TR as other types of contrast were not of interest here. A Siemens knee coil was used for this study.

**Processing**

**Pre-processing of EPSI Data**

Uniform analog-to-digital convertor triggering during the sinusoidal gradient ramping results in non-uniform samples in the k-space along the echoplanar encoding direction. In order to use the fast Fourier transform for samples in the k-space along the echoplanar encoding direction in the in-phase and out-phase images before further processing.

**Derivation of Structural Images From EPSI Data**

Subsequent to the sinc-interpolation in the k-space, EPSI data were Fourier transformed in the spatial dimensions, generating pixel-wise time domain data. For both fat-water imaging and narrow bandwidth imaging, the time domain data were extended in length by a factor of 4 with zero-padding and Fourier transformed into the spectral domain where peaks corresponding to the water and fat signal in the absolute spectrum were identified on a pixel-by-pixel basis. A multiresolution approach was used to obtain the peak locations for each pixel. By recursively averaging the EPSI data in non-overlapping blocks of 4 pixels, a pyramid (tree) representation of the data was created. Peak search was then applied to the pyramid in a top-down fashion with the analysis of each node guided by results of its parent node, ensuring the robustness of the procedure in the presence of B0 variation. Peak locations for the global spectrum were first obtained by searching for zero-crossing of the first derivative. At subsequent scales, search windows for the fat and water peaks, respectively, were determined by centering it at the corresponding peak location of its parent node and assigning it a width equal to the separation of the two peaks in its parent node and searches for the relevant peaks were performed in each window by identifying the zero-crossings in the first derivative and choosing the zero-crossing with the largest peak. Note that a search window can be wrapped around the spectral window when needed. The multiresolution approach and the wraparound accounted for possible foldover in the spectrum.

Upon the peak detection, each peak was phased by applying a zero-order phase correction to 100 Hz region around the peak. That is, \( \hat{S} = S \times e^{i\phi} \), where \( \hat{S} \) is the phased spectrum and \( \phi \) is the phase factor. The value of \( \phi \) was obtained by minimizing the following cost function:

\[
\left( \sum_{n=r}^{r_f} |S(n)| - \sum_{n=r}^{r_f} |\Re(S(n))| \right)^2 + \left( \sum_{n=r}^{r_f} |\Im(S(n))| \right)^2 + 2\max \left( \{|\Re(S)|\} \right)^2 + 2 \sum_{n=r}^{r_f} |\Re(S[r_p]) - \hat{S}[2r_p - r_\text{c}]|^2
\]

where \( r_\text{c} \) and \( r_p \) are the lower and upper limits of the region over which spectrum is phased, and \( r_p \) is the peak location of the absolute value spectrum in the region. In this cost function, the first term was intended to maximize the integral of the real part of the phased spectrum, the second term was for minimizing the integral of its imaginary part, the third term was incorporated to maximize the peak height of its real part, and the fourth term maximized the symmetry of its real part around the peak. It was noted that all four criteria were needed for robust phasing. The minimization of the cost function was achieved with a multi-scale search as follows. Starting from the full search range (i.e., \([- \pi, \pi]\)), the cost function is evaluated at eight values of \( \phi \), equally spaced in the search range, and the \( \phi \) corresponding to the lowest cost, \( \phi_0 \), is selected. Subsequently, the search is performed around \( \phi_0 \) with the search range narrowed by a factor of 4. The same procedure is iteratively repeated 12 times, resulting in a phase resolution dictated by the machine precision. This binary search method was justified because the cost function was found to be a smooth function with no local minima. The method is computationally efficient but requires a high signal-to-noise ratio (SNR). This requirement is met in this application because we are phasing only the water and fat peaks, which have sufficiently high SNR.

In order to generate the water/fat images the real part of the phased spectrum was integrated over the fat and water frequency ranges, respectively. These ranges were determined on a pixel-by-pixel basis using peak locations determined above and a width corresponding to one-tenth of the spectral width. For comparison, fat/water images were also obtained using the two-point Dixon technique as outlined by others (16–18). To avoid the effect of asymmetric echo position in deriving the Dixon images, a constant and linear phase correction was applied to remove echo offset along the readout direction in the in-phase and out-phase images before the subtraction of the two.

To obtain (effective) narrow bandwidth images based on EPSI data, it is necessary to combine images obtained at all points along the time axis of the EPSI data. Because phase accrual due to off-resonance occurs with time in these images, summation along time cannot be simply performed in complex domain. Although absolute value summation can be used, it sacrifices SNR in the resultant image. To avoid this problem, the narrow bandwidth image was derived using the following procedure. The peaks corresponding to the water and fat signals were detected, and phased with a zero-order phasing in the frequency domain as described...
above. To account for off-resonance effect, spectral segments containing corresponding peaks were Fourier transformed into the time domain where they were corrected by a phase factor linearly dependent on the time from the spin-echo, i.e., \( f(t) = f(t) \times e^{i(\delta + \pi - T_E)} \). This phase correction is equivalent to a shift in the frequency domain. The factor \( \delta \) is determined based on the criterion that minimizes

\[
\left( \sum \left| f(t) - \sum \text{Re} [\hat{f}(\Delta)] \right|^2 + \sum \left| \text{Im} [\hat{f}(\Delta)] \right|^2 \right).
\]

Subsequent to the linear phase adjustment, the real part of time domain signal was integrated to produce the corresponding pixel intensity. Two special situations were addressed in our implementation of this approach. The first concerns, pixels containing both fat and water exhibit two prominent peaks in the spectrum. Spectral segments corresponding to the two peaks were phased and integrated separately, and the resultant intensities were added to produce the final image intensity. This approach slightly increases the noise in these pixels. Another special case concerns background pixels containing only noise. To account for these pixels, they were identified by thresholding based on spectral energy in the water and fat regions and assigned the integral of the real part of their time-domain signal.

The \( R_2^* \) maps were generated by fitting the real part of the FIDs to an exponential function after processing as described above. Specifically, peaks were identified and phased in the frequency domain and Fourier transformed to the time domain where a linear phase adjustment was applied to remove the off-resonance phase modulation. Furthermore, to improve the sensitivity to noise, the FIDs were multiplied by an exponential line broadening function (4 Hz) before the fitting. A nonlinear least squares curve fitting routine in Matlab was subsequently applied, and the resultant \( R_2^* \) value was corrected by removing the exponential factor of the line broadening. In pixels in which both water and fat signals are present, the FIDs corresponding to the two are isolated and processed as described above and fitted separately. The resultant \( T_2^* \)s are combined by averaging.

**Figure 2.** Results of a fat/water imaging study. **a:** Conventional spin-echo image of the slice studied. **b:** The \( B_0 \) map of the slice derived as a by-product of EPSI processing. **c–f:** Fat and water images obtained using the EPSI approach (c, d) and Dixon’s method (e, f); (c) fat image obtained from EPSI data, (d) water image obtained from EPSI data, (e) fat image obtained using the two-point Dixon method, and (f) water image obtained using the two-point Dixon method.
RESULTS

Figure 2 shows the results for fat/water imaging. Figure 2a shows the conventional spin-echo images of the same slice, and Fig. 2b demonstrates the resonance frequency map derived from the EPSI data. Fat/water images derived from EPSI data are shown in Figs. 2c and d, respectively, and those obtained using the two-point Dixon technique are shown in Figs. 2e and f, respectively. In both the fat and water images obtained from the Dixon technique, substantial signal mixing arising from $B_0$ inhomogeneity is present, especially in areas having large off-resonance (Fig. 2b). This mixing is particularly evident in Fig. 2f, where the contamination from the fat signal is dominating what is supposed to be the water image. Such mixing is completely absent in EPSI-derived water/fat images (Fig. 2c,d). Note that the slight shading in Fig. 2c arises from the non-uniform coil sensitivity, which is apparent in Fig. 2a. Similar results were obtained in seven other studies. These results demonstrate that EPSI is insensitive to $B_0$ inhomogeneity for fat and water imaging.

Figure 3a illustrates the narrow bandwidth image of the volunteer obtained from EPSI data and Fig. 3b is the corresponding spin-echo image for comparison. The spin-echo image in Fig. 3b exhibits severe chemical shift artifacts, particularly of fat and bone marrow. This is evidenced by the apparent gap between the fat and muscle as indicated by the arrow. In contrast, the EPSI-derived image is free of chemical shift artifact. To demonstrate that an effective narrow bandwidth was experimentally achieved with the EPSI approach, SNR was calculated for both the EPSI-derived image and the spin echo image by taking the ratio of the mean in the region of interest (ROI) in the tissue to the mean in an ROI outside the tissue region. The results are tabulated in Table 1. These results show that the SNR in the two images are comparable. It should be noted that the non-uniform sampling in the k-space has two opposing effects on the SNR of the EPSI image. On one hand, the effective bandwidth is somewhat higher than determined based on the total sampling time, making the SNR lower than one would predict. On the other hand, the $\text{sinc}$-interpolation filters the noise in the k-space, making the SNR higher. The combination of these two factors makes the apparent SNR in the EPSI image slightly higher. This is supported by the result of SNR measurements shown in Table 1.

An $R_2^*$ map of a coronal section of a human knee is presented in Fig. 4b. The corresponding anatomic image is shown in Fig. 4a. Although the map is noisy due to the inherent low SNR of EPSI data, it is adequate for potential applications. Figures 4c and d illustrate profiles of $R_2^*$ values along lines (as indicated in Fig. 4a by arrows numbered 1 and 2) in the left-right direction and superior-anterior direction, respectively. As can be seen from these profiles, $T_2^*$ is approximately 45 msec for fat, and 25 msec for muscle. In agreement with studies reported by others (22), the profile in Fig. 4d clearly depicts the monotonic increase in $T_2^*$ (ie, decrease in $R_2^*$) from the epiphysis to the diaphysis along the femur. The $T_2^*$ value in the epiphysis is less than 15 msec, and that in the diaphysis is roughly 35 msec.

DISCUSSION

In this paper, echoplanar spectroscopic imaging, which was originally developed for spectroscopic studies, is applied for obtaining structural images. This is made possible by advances in gradient hardware, which allow the acquisition of high-resolution EPSI data. In applying EPSI to obtain fat and water images, spectral information was used to identify corresponding resonances, making the final result insensitive to $B_0$ inhomogeneity. This is in contrast to images obtained with the two-point Dixon method, which resulted in fat and water images with incomplete separation due to off-resonance effects. Another advantage of using EPSI for fat-water imaging

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arises from the fact that resonance peaks of interest, especially that of fat, are often broad. This broad line shape slightly degrades the SNR in images derived with Dixon’s method but is accounted for with the EPSI approach.

In the present work, EPSI is also used to derive narrow band images free of chemical shift artifact. With conventional spin-echo sequences, severe chemical shift artifact arises when imaging with a narrow bandwidth to improve SNR. In our approach, chemical shift information in the EPSI data is used to form images free of chemical shift artifact. Consequently, the SNR of the resultant image is comparable to that of the image obtained from the corresponding spin-echo sequence. When there is more than one peak above noise in the spectrum of a pixel, images are derived by summing up those corresponding to these peaks. In this case, these pixels may have a slightly degraded SNR due the summation of images.

Experimental results of using EPSI for T2* mapping demonstrate that the EPSI-based approach is capable of providing acceptable T2* maps even though the SNR is not optimized in our study. For applications that require better SNR, data averaging can be readily used. Because each NEX only takes 78 seconds in our present implementation, substantial averaging can be utilized within clinical practical time. Another advantage of using EPSI in T2* mapping is that it is possible to obtain resonance-specific T2* values for different components in individual voxels. This ability may be beneficial in applications in which the T2* is different between different tissues, such as water and fat, in individual voxels.

A major pitfall of using EPSI for structural imaging is the acquisition and processing of an extensive amount of data for the generation of final images. The size of the raw data imposes a high demand on disk storage and computer processing power. In our implementation, the size of the data matrix is typically \(256 \times 256 \times 128\), which requires approximately 67 MB of disk space. The processing is performed off-line on an SGI Octane workstation (Silicon Graphics, Palo Alto, CA) in a PV-Wave (Visual Numerics, Boulder, CO) environment, and takes approximately 5 minutes to complete. While these requirements are high compared with conventional imaging studies, with technical advances in computer hardware, they will not be the limiting factor in the future.

\[\text{Figure 4. a: Anatomic image of a coronal slice of a human thigh obtained using a spin-echo excitation. b: } R_2^* \text{ map of the same coronal slice. c: Profile of a cross section (transverse direction) of the } R_2^* \text{ map. The profile location is indicated by the number 1 in a and d. Profile of a longitudinal-section (sagittal direction) of the } R_2^* \text{ map. The profile location is indicated by the number 2 in panel a. The thick line in the profiles is obtained by averaging the actual } R_2^* \text{ values, which are shown by the thin line.}\]
Further improvements of the applications described can be achieved. First, the spectral bandwidth can be expanded by a factor of 2 by combining echoes arising from both the positive and the negative lobes using the interlaced Fourier reconstruction technique as described previously (14). Spectral bandwidth is constrained by the gradient performance of the system as well as physiological limitations. An expansion of bandwidth will avoid the aliasing of lipid peak, making the processing more straightforward. Second, as mentioned earlier, SNR can be improved by using more averages in data acquisition. Third, other strategies can be adopted to form structure images. For example, integration of the real part of the spectrum required spectral phasing, and can be substituted by integration of magnitude spectrum if T2* effect is not severe. Finally, a more sophisticated description of the local magnetic environment, may be devised to elucidate more detailed description of the local magnetic environment.

Several clinical applications of the techniques described here can be envisioned. For example, water-fat imaging can be used for breast imaging (27), quantitative measurement of body fat content (28), and prostate cancer (29). Narrow bandwidth imaging can be used for long-TE imaging in examining shoulders (30) and tendons (31). As indicated earlier, T2* mapping can be applied to studying bone structure in patients with osteoporosis (22,23). The utility of the EPSI method in these applications remains to be further demonstrated.

CONCLUSIONS

Echoplanar spectroscopic imaging was applied to obtain structural images in applications in which the use of chemical shift information can be beneficial. These applications include 3B inhomogeneity-insensitive chemical shift (fat-water) imaging, chemical shift artifact free narrow bandwidth imaging, and T2* mapping. Experimental results were obtained and presented to illustrate that EPSI is a viable technique for these imaging applications. The clinical utility of these approaches remains to be further evaluated.

REFERENCES