Comparison of Chemical Shift-Encoded Water–Fat MRI and MR Spectroscopy in Quantification of Marrow Fat in Postmenopausal Females

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Purpose: To validate a chemical shift-encoded (CSE) water–fat imaging for quantifying marrow fat fraction (FF), using proton magnetic resonance spectroscopy (MRS) as reference.

Materials and Methods: Multiecho $T_2$-corrected MRS and CSE imaging with eight-echo gradient-echo acquisitions at 3T were performed to calculate marrow FF in 83 subjects, including 41 with normal bone mineral density (BMD), 26 with osteopenia, and 16 with osteoporosis (based on DXA). Eight participants were scanned three times with repositioning to assess the repeatability of CSE FF map measurements. Pearson correlation coefficient, Bland–Altman 95% limit of agreement, and Lin’s concordance correlation coefficient were calculated.

Results: The Pearson correlation coefficient was 0.979 and Lin’s concordance correlation coefficient was 0.962 between CSE-based FF and MRS-based FF. All data points, calculated using the Bland–Altman method, were within the limits of agreement. The intra- and interrater agreement for average CSE-based FF was excellent (intrarater, intraclass correlation coefficient [ICC] = 0.993; interrater, ICC = 0.976–0.982 for different BMD groups). In the subgroups of varying BMD, inverse correlations were observed to be very similar between BMD ($r = -0.560$ to $-0.710$), T-score ($r = -0.526$ to $-0.747$), and CSE-based FF, and between BMD ($r = -0.539$ to $-0.706$), T-score ($r = -0.501$ to $-0.742$), and MRS-based FF even controlling for age, years since menopause, and body mass index. The repeatability for CSE FF map measurements expressed as absolute precision error was 1.45%.

Conclusion: CSE imaging is equally accurate in characterizing marrow fat content as MRS. Given its excellent correlation and concordance with MRS, the CSE sequence could be used as a potential replacement technique for marrow fat quantification.

J. MAGN. RESON. IMAGING 2016;00:000–000.

Quantification of marrow fat content is of considerable interest, especially because of its association with the pathophysiology of bone loss and cancer therapy-induced bone marrow damage.1–3 Although marrow biopsy is the reference standard method for quantification of fat content, this method is subject to sampling error owing to highly heterogeneous marrow fat distributions.4 Additionally, this procedure may result in bleeding or infection. Noninvasive methods for quantification of fatty marrow have been developed such as micro-computed tomography (CT), dual-energy CT, and magnetic resonance imaging (MRI).1,5–8 Among these imaging methods, magnetic resonance spectroscopy (MRS) has been found to be safe and accurate and is generally considered as an adequate reference standard.
noninvasive technique for fat and metabolite quantification. However, MRS has some disadvantages, such as being unable to cover a large area with a single voxel, long acquisition time, and relatively long and complicated postprocessing.

Quantitative chemical shift-encoded (CSE) water–fat imaging is an emerging method for the quantification of fat content. In order to obtain accurate fat fraction (FF) measurements, several confounding factors must be considered, including $T_1$ bias, noise bias, $T_2^*$ decay, spectral complexity of fat, eddy currents, and B0 field inhomogeneity. Several researchers have presented two- and three-point Dixon MRI to quantify marrow fat content. However, the algorithm was only introduced for a single peak fat spectrum.

Modeling of $T_2^*$ decay effects is of great importance when quantifying proton density FF either in the marrow in the presence of bone trabecula or in the liver with iron overload. Recent work by multiple groups has shown a good agreement between CSE MRI and MRS-based marrow FF measurements when confounders are taken into consideration, eg, single-$T_2^*$ effects and multipeak fat spectrum model. While correcting for a single-$T_2^*$ value has been demonstrated to improve FF estimation, a single-$T_2^*$ model may not accurately model the effects of the two separate signals of water and fat because the $T_2^*$ decay rates of water and fat are not equal. A dual-$T_2^*$ correction can reduce errors in FF estimates from single-$T_2^*$ reconstruction at high FF, but has inherently worse noise performance.

To minimize the aforementioned confounding factors, we aimed to 1) validate CSE imaging with a complex spectral model of fat and $T_2^*$-correction using a priori-known $T_2$ for quantifying vertebral marrow fat content in postmenopausal populations, using multiecho $T_2$-corrected MRS as reference; and 2) to examine relationships between bone mineral density (BMD), T-score, and marrow fat assessed by CSE imaging and MRS.

Materials and Methods

Subjects

In this cross-sectional study, 83 postmenopausal women were recruited from the community between January 2014 and September 2015. Inclusion criteria were: women were ambulatory; more than 50 years old; were at least 1 year postmenopausal; and able to give informed consent. Key exclusion criteria included any disease known to affect bone metabolism such as hypocalcemic or pituitary disorders, diabetes mellitus, vitamin D deficiency, impaired renal function, reported malignancy or exposure to radiation; use of medications known to influence bone metabolism such as estrogen, bisphosphonates, glucocorticoids, or regular use of aspirin; and any confounders that had the potential to interfere with the interpretation of the findings such as smoking history, drinkers, vertebral hemangiomas, or silent vertebral body fractures. Drinkers were those who drank an alcoholic beverage more than once per day during the past month. For every participant, age, years since menopause (YSM), height, body weight, and body mass index (BMI) were recorded. The local Ethics Committee Review Board approved this study, and written informed consent was obtained from all participants.

Dual-Energy X-Ray Absorptiometry

Areal BMD (g/cm$^2$) of lumbar spine (from L1 to L4) was assessed in all subjects using a Prodigy Lunar scanner (GE Healthcare, Waukesha, WI; v.enCORE.13.40). The scanner was routinely calibrated and quality control measures were followed as recommended by the manufacturer to control possible baseline drift. Subjects were grouped into three categories according to the T-score: normal bone mass defined as T-scores ≥ –1 SD; osteopenia as T-score between –1.0 and –2.5 SD; and osteoporosis as T-score ≤ –2.5 SD.

MRI Protocol

MRI exams were performed using a 3.0-T unit (Ingenia, Philips Healthcare, Best, the Netherlands) within 2 weeks after dual-energy x-ray absorptiometry scanning. A surface coil was placed under the lumbar spine region as the radiofrequency receiver and the body volume coil was used as the radiofrequency transmitter. The spine imaging protocol included standard clinical sagittal $T_1$- and $T_2$-weighted images for anatomical and morphological assessments of the lumbar region.

Single-voxel MRS of the third vertebral body was performed using a stimulated echo acquisition mode sequence without water suppression. After local shimming and gradient adjustments, data were obtained using the following parameters: volume of interest (VOI), 1.5 × 1.5 × 1.5 cm$^3$; TR, 6000 msec (a long TR was chosen to minimize $T_1$-bias effects); four TE values, 11/15/20/25 msec (short TE was to reduce J-coupling effects, and multiecho acquisition allowed the calculation of $T_2$-weighting effects and the $T_2$-corrected area of individual spectral peaks, as the different fat peaks have different $T_2$ values); and data points, 4096; bandwidth, 5000 Hz. Outer volume saturation bands were used to eliminate unwanted signal contamination from outside the voxel. These saturation bands typically covered the adjacent vertebral discs and cerebrospinal fluid. We set a distance (2–3 mm) between the VOI and saturation-band to avoid the missing signal from VOI.

CSE water–fat MRI was performed using an eight-echo 3D spoiled gradient echo sequence in the spine sagittal plane. The imaging parameters were: four echoes per TR, monopolar readouts and flyback gradients; TR, 10 msec; number of TEs, 8 (TE$_{min.}$ 1.48 msec; ΔTE, 1.2 msec); field of view, 400 × 400 mm; frequency direction, A/P (to minimize breathing artifacts); matrix, 256 × 256; slice thickness, 3 mm; flip angle, 3° (given the large $T_1$ difference between water and fat components in the bone marrow, a small excitation flip angle was used to minimize $T_1$-bias effects); bandwidth, 1.3 kHz/pixel; SENSE acceleration, 2; and number of signals averaged, 1.

Data Analysis for MRS

Theoretically, there are nine resonances in a typical marrow spectrum. CH$_3$ methyl protons (0.90 ppm), bulk CH$_2$ methylene protons (1.30 ppm), CH$_2$ methylene protons α- (2.25 ppm) and...
b- (1.59 ppm) to the carbonyl, methylene proton z to a double bond (2.03), dialllylic CH2 protons (2.77 ppm), glycerol (4.10–4.35 ppm), H2O (4.65 ppm), and olefinic protons (5.19–5.31 ppm). T2-corrected MRS spectra acquired from the third vertebra were exported and processed to estimate marrow FF by an MR physicist using the Advanced Method for Accurate, Robust and Efficient Spectral fitting of MRS data (AMARES) algorithm with use of prior knowledge in the jMRUI software (http://www.jmrui.eu). The rules applied in the fitting procedure were: the line-width of the water and methylene (1.3 ppm) peaks was

<table>
<thead>
<tr>
<th>TABLE 1. Baseline Characteristics of Participants in the Three Groups</th>
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<tbody>
<tr>
<td><strong>Normal BMD group</strong> <strong>(n = 41)</strong></td>
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<tr>
<td>--------------------------------</td>
</tr>
<tr>
<td>Age (years)</td>
</tr>
<tr>
<td>YSM (years)</td>
</tr>
<tr>
<td>Height (cm)</td>
</tr>
<tr>
<td>Body weight (kg)</td>
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<tr>
<td>BMI (kg/m2)</td>
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<tr>
<td>Vertebral BMD (g/cm2)</td>
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<tr>
<td>Femoral neck BMD (g/cm2)</td>
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<tr>
<td>Total hip BMD (g/cm2)</td>
</tr>
</tbody>
</table>

BMD, bone mineral density; BMI, body mass index; YSM, years since menopause. Data are presented as mean ± SD.

ªP < 0.001 and bP < 0.05 compared with the normal BMD group, cP < 0.05 compared with the osteopenia group using one-way ANOVA (multiple comparisons were made with the Bonferroni correction).

FIGURE 1: Two examples of marrow fat fraction (FF) results from chemical shift-encoded (CSE) water–fat imaging and MRS. CSE-based FF and MRS-based FF agree closely in both subjects [a subject with normal bone mineral density (a) and the other with osteopenia (b)]. Fat peak assignment: (1) –CH=CH– and –CH–O–CO– (5.19–5.31 ppm); (2) H2O (4.65 ppm); (3) –CO–CH2–CH2– (2.25 ppm) and –CH2–CH=–CH–CH2– (2.03 ppm); (4) –CO–CH2–CH2– (1.59 ppm); (5) –(CH2)n– (1.30 ppm); (6) –(CH2)n–CH3 (0.90 ppm); (7) –CH2–O–CO– (4.20 ppm); (8) –CH=CH–CH2–CH=CH– (2.77 ppm). Peaks 4 and 7 are buried within the methylene (1.3 ppm) and water peaks, respectively. Peak 8 is small and is rarely seen in the human marrow clinically.
not resolvable and appear as single peaks. Therefore, we evaluated 5.31 and 5.19 ppm; 2.25 and 2.03 ppm; 1.59 and 1.30 ppm) are the superimposed water peak. Additionally, several resonances (at clearly distinguishable from the 4.65 ppm water peak because of too weak to be accurately measured in the bone marrow at clinical at each TE by a nonlinear least-squares fitting algorithm, using the water and fat peaks were calculated from the areas of water and fat 100, where \( I_{\text{lipid}} \) is the sum of the area amplitudes of the resonances (at 0.9, 1.30, 1.59, 2.03, 2.25, 5.19, and 5.31 ppm) and \( I_{\text{water}} \) is the area amplitude of \( \text{H}_2\text{O} \) resonance.

**Data Analysis for Water–Fat Imaging**

The gradient echo imaging data were processed offline using in-house-built routines implemented in MatLab (v. 2014-64bit, MathWorks, Natick, MA). To increase the accuracy of the reconstruction, a region-growing algorithm was first applied to assess the field-map variation similar to that described in Yu et al.23 because the water–fat separation strongly depends on the initial guess of the B0 inhomogeneity map. A complex-based water–fat decomposition was then performed using a \( T_2^* \)-correction and a precalibrated eight-peak spectral model of fat, accounting for the presence of the multiple peaks in the fat spectrum.11 The precalibrated fat spectrum was modeled using the vertebral marrow fat spectrum characterized by Karlampinos et al.10 and Dieckmeyer et al. instead of the choice of the liver fat spectrum. A \( T_2^* \)-correction using a priori-known \( T_2 \) equal to the average previously measured values (considering women only, the corresponding mean values were determined to be \( T_{2\text{water, priori}} = 24.6 \text{ msec} \) and \( T_{2\text{fat, priori}} = 72.6 \text{ msec} \) for the vertebral bodies) was chosen because it can remove \( T_2^* \) bias without reducing noise performance in vertebral FF map, as described in detail by Karlampinos et al.10 Image reconstructions yielded coregistered a water-only, fat-only, and quantitative FF map. The FF map accurately reflected the underlying proton density ratio of the fat signal over the sum of water and fat signals in a range of 0–100%, when using a multiplex spectral model of fat and a small excitation flip angle. The multiecho water–fat algorithms can jointly quantify fat content and \( T_2^* \) per voxel.2,4 The FF maps were exported to an OsiriX DICOM viewer to manually draw regions of interest (ROIs) by two independent raters (X.L. and G.L., with 5 and 13 years of working experience with MRI, respectively). Three ROI measurements were made from the three most central slices depicting each vertebra (from L1 to L4, covering three-fourths of the vertebral height) and were excluded from the cortical bone and endplates when drawing ROIs. The FF values were obtained by averaging the values obtained from the three slices selected.

**Reproducibility Analysis for CSE Water–Fat Imaging**

Eight randomly selected participants from the study cohort were scanned three times with repositioning to assess the repeatability of FF measurements at CSE water–fat imaging. The repeatability was expressed as root mean square absolute precision error in [%] (absolute units) and as root mean square coefficient of variation (RMS-CV) in [%] (relative units), according to Gluer et al.24 To assess the interobserver reliability, two observers independently analyzed all the FF maps. In the intraobserver analyses, two measurements of the primary rater on two separate occasions 2 months apart were compared. Both the observers were blinded to the results obtained earlier.

**Statistical Analysis**

Mean ± standard deviation (SD) values were calculated for each variable. Differences between the groups were assessed using one-way analysis of variance (ANOVA) followed with Bonferroni post-hoc multiple comparisons and analysis of covariance (ANCOVA) when controlling for covariates. Correlation analyses were used to describe associations between BMD, T-score, and FF. Intrarater and interrater reliability were conducted using the intraclass correlation coefficient (ICC). To establish the level of agreement between the fat content with the CSE water–fat separation sequence and that with MRS, linear regression and Bland–Altman analysis were performed.25 Finally, Lin’s concordance correlation coefficient was calculated to describe the strength of agreement: >0.95 indicates almost perfect agreement; 0.80–0.95, substantial agreement; 0.70–0.80, moderate agreement; and <0.70, poor agreement.26 Data was analyzed using IBM SPSS v. 23.0 (IBM SPSS, Armonk, NY). Statistical significance was set to \( P < 0.05 \).
Results

Baseline Characteristics
Table 1 lists the demographic and clinical characteristics of the study population. The mean age of the 83 postmenopausal women included in this study was 62.8 ± 6.6 years. Figure 1 illustrates the FF results from two subjects: one with normal bone mass and the other with osteopenia. Excellent fat and water separation was achieved in both subjects. Associated MRS plots are also shown. Figure 2 summarizes the FF results from all the subjects. There were significant differences in the average fat content among the three groups (P < 0.001) after adjustment for age, YSM, and BMI.

Repeatability of CSE Imaging Scans
The RMS absolute precision error of the FF measurements at CSE imaging was 1.45%. RMS-CV indicates the average error of FF measurements. The RMS-CV of FF measurements was 2.50%.

### TABLE 2. Correlations Between Bone Mineral Density (BMD), T-score, and Marrow FF Measured by Chemical Shift-Encoded (CSE) Imaging and MRS

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Groups</th>
<th>BMD</th>
<th>T-score</th>
</tr>
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<tbody>
<tr>
<td></td>
<td></td>
<td>r</td>
<td>P-value</td>
</tr>
<tr>
<td>MRS-based FF</td>
<td>Normal bone mass</td>
<td>-0.555</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>Osteopenia</td>
<td>-0.539</td>
<td>0.005</td>
</tr>
<tr>
<td></td>
<td>Osteoporosis</td>
<td>-0.706</td>
<td>0.002</td>
</tr>
<tr>
<td>CSE-based FF</td>
<td>Normal bone mass</td>
<td>-0.571</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>Osteopenia</td>
<td>-0.560</td>
<td>0.003</td>
</tr>
<tr>
<td></td>
<td>Osteoporosis</td>
<td>-0.710</td>
<td>0.001</td>
</tr>
<tr>
<td>MRS-based FF</td>
<td>Normal bone mass</td>
<td>-0.536</td>
<td>0.001</td>
</tr>
<tr>
<td></td>
<td>Osteopenia</td>
<td>-0.536</td>
<td>0.005</td>
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<tr>
<td></td>
<td>Osteoporosis</td>
<td>-0.610</td>
<td>0.021</td>
</tr>
<tr>
<td>CSE-based FF</td>
<td>Normal bone mass</td>
<td>-0.545</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>Osteopenia</td>
<td>-0.591</td>
<td>0.003</td>
</tr>
<tr>
<td></td>
<td>Osteoporosis</td>
<td>-0.604</td>
<td>0.022</td>
</tr>
</tbody>
</table>

*P*-values were calculated by bivariate correlations and **P*-values were calculated by partial correlations after controlling for age, years since menopause, and body mass index.
Intra- and Interrater Reliability
For FF measurements at CSE imaging, the intrarater reliability for duplicate measurements was substantial agreement (ICC = 0.993, 95% confidence interval [CI] = 0.990–0.996; *P* < 0.001). Interobserver agreement for the average FF value (L1–L4) at CSE imaging was excellent (ICC = 0.982, 95% CI = 0.996–0.990 for normal bone mass; ICC = 0.976, 95% CI = 0.947–0.989 for osteopenia; and ICC = 0.979, 95% CI = 0.941–0.993 for osteoporosis, respectively).

Agreement between FF at CSE imaging and MRS-measured FF
Mean FF with MRS was 62.1% ± 11.1%, and mean FF with CSE imaging was 60.4% ± 10.1% (*P* > 0.05). Figure 3a shows the scatterplot between the marrow fat content with MRS and that with the CSE water–fat separation sequence, with linear regression. The Pearson correlation coefficient was 0.979 (*P* < 0.001). Lin’s concordance correlation coefficient was 0.962. Using the Bland–Altman plot, the CSE image method demonstrated good agreement with MRS, where all points were within the 95% limit of agreement (Fig. 3b).

Correlations Between Marrow Fat Content, BMD, and T-score
In the subgroups of varying BMD, inverse correlations were found to be very similar between BMD, T-score, and CSE-based FF and between BMD, T-score, and MRS-based FF. Adjusting for age, YSM, and BMI did not attenuate the relationships between BMD, T-score and FF (Table 2).

Discussion
The strength of our work was that initial clinical experience of CSE water–fat imaging using a multiplex spectral model of fat, suitable *T*₂*-corrected method, and a small excitation flip angle applied to marrow fat quantification. Further, a good sample size for the subgroups of varying BMD could enhance the statistical power, and the statistical analysis took most of the potential confounding factors into account. Our study showed that CSE imaging is equally accurate in characterizing marrow fat content as multi-TE *T*₂*-corrected MRS.

CSE water–fat MRI offers a quantitative and qualitative tool to quantify not only adipose tissue content, but also an index of the underlying tissue inhomogeneity. There are several studies using chemical-shift MRI on vertebrae. Ojanen et al.²⁷ reported that in-phase and out-of-phase MRI can provide similar vertebral marrow fat estimation as MRS. Similar results were observed in the study done by Regis-Arnaud et al.¹³ Conversely, Gokalp et al.²⁸ reported vertebral marrow FF calculated with a double-echo fast low-angle shot sequence is not a reliable parameter for predicting BMD in female patients. However, these studies neither corrected for any *T*₂*- decay effects nor modeled multiplex fat spectrum. Similarly, Takasu et al.¹⁴ used the IDEAL algorithm to measure FF and discriminate between patients suffering from symptomatic and asymptomatic myeloma but without employing *T*₂*-correction and the precalibrated multiplex fat spectrum. Our data suggest that vertebral marrow fat content could be reproducibly assessed using CSE water–fat imaging, with an absolute precision error of 1.45% as well as excellent intra- and interrater reliability. Interestingly, the most recent publication by Baum et al.⁸ indicates that whole spine marrow fat could be reproducibly (absolute precision error was 1.7% averaged over C3–L5) assessed by using CSE-based water/fat MRI with a *T*₂*-correction and vertebral marrow FF showed anatomical variations with increasing values from C3 to L5. Our results also support a previous study¹⁵ in which a six-echo gradient-echo imaging with *T*₂*-correction provided an accurate means of determining marrow fat content in the presence of trabecular bone.

Bone marrow is a complex connective tissue, consisting of both hematopoietic and fatty marrow. Bone marrow is characterized by a large difference in *T*₁ relaxation times of fat and water protons, which leads to bias during quantification of fat content.¹⁹ The choice of flip angle is critical in the CSE-MRI FF quantification. In relation to TR, the flip angle (3° used in our study) should be sufficiently low to minimize *T*₁-bias effects, but sufficiently high to maintain an adequate signal-to-noise ratio.⁹,²⁹

Fat has a number of spectral peaks. The presence of multiple peaks in the fat spectrum complicates the ability to correct for spatial chemical shift artifacts, particularly the spectral peak from olefinic protons at 5.3 ppm (being close to the water peak).¹¹ Indeed, marrow fat, subcutaneous and visceral adipose tissue such as liver fat exhibit distinct, tissue-specific properties and therefore different metabolic activities.³⁰ In comparison to liver and subcutaneous fat, red marrow has higher levels of unsaturated fatty acid.³¹ The fat spectrum varies from organ to organ and a fat spectrum suitable to liver may not give accurate values for bone marrow. In several previous studies,²,¹⁵,³² the fat spectrum was modeled using the liver fat spectrum characterized by Hamilton et al.³³ and only the three main peaks (0.9, 1.3, and 2–2.2 ppm) were considered.³² In our work, the presence of multiple peaks in the fat spectrum was considered using a precalibrated fat spectrum characterized in the red bone marrow.¹,³¹ Excellent fat and water separation was achieved in all the subjects. Since the use of an incorrect fat model may result in some errors in MRI estimation of FF, further studies are required to elucidate the difference of marrow proton density FF measurements using various multiplex fat models.

*T*₂*- correction is a crucial prestep for quantitative water–fat imaging, particularly in bone marrow because the
presence of bone trabeculae creates local heterogeneities in the magnetic field that shorten $T_2^*$. Unlike $T_1$ bias effects and accurate spectral modeling, $T_2^*$ correction can’t be achieved through choosing appropriate imaging parameters (such as low flip angle) and prior knowledge (relative amplitudes and position of multiple fat peaks). $T_2^*$ correction can be carried out by modifying the signal model to account for the signal decay due to $T_2^*$. The majority of CSE imaging that correct for $T_2^*$ decay to improve the accuracy of marrow fat quantification assume a common (single-$T_2^*$) for water and fat, which is valid for liver applications at relatively low fat content. However, water and fat signals have independent $T_2^*$ that may influence estimation of fat content, particularly at regions of high fat content and short $T_2^*$, which can be frequently encountered in bone marrow. In liver fat quantification, a single-$T_2^*$ correction was chosen because of the advantage of a range of clinically relevant signal-to-noise ratios and water/fat ratios at low fat concentrations compared to dual-$T_2^*$ correction. A dual-$T_2^*$ correction can correct for the marrow fat bias at nominal FF value close to 50%, but shows poor noise performance at low FF. A recent publication by Karampinos et al. indicated that using a priori-known $T_2$, a $T_2^*$-correction can remove the FF bias induced by the difference of $T_2^*$ between water and fat components without reducing noise performance in the FF map of the lumbar spine. Typically, based on such $T_2^*$-correction using a priori-known $T_2$, eight-echo acquisitions used in the present study have provided an excellent balance between accurate estimation of $T_2^*$ decay effects and acquisition time.

Several studies have documented inverse relationships between marrow fat content and bone mass. The correlation coefficients between marrow fat content and BMD that we observed in the subgroups of varying BMD are higher than those of prior reports (typical $r = -0.32$ to $-0.55$). A rational explanation is that multiple confounding factors, particularly the presence of multiple peaks in the fat spectrum and $T_2^*$ decay effects, were considered in our study. We also documented the inverse association of BMD with CSE-based FF or MRS-based FF was very similar, which further supports that CSE imaging is equally accurate in characterizing marrow fat content as MRS.

We also acknowledge that, because of the necessity of image registration and establishment of CSE-based FF maps, the whole analysis process was relatively time-consuming, which is not ideal in real clinical practice. More easily handled software or accelerating the method is greatly needed. Second, a comparison was performed between one vertebral body with MRS vs. four vertebral levels with CSI imaging, and the results may have interscan variability depending on the MR scanner. Finally, our study group only consisted of postmenopausal women.

In conclusion, CSE imaging with a complex spectral model of fat and $T_2^*$-correction is equally as accurate in characterizing vertebral marrow fat content as MRS. CSE water–fat MRI offers an additional advantage of inhomogeneous determination of marrow tissue in FF maps over MRS. Given its excellent correlation and concordance with MRS, the CSE water–fat separation sequence has the potential to replace MRS for fat quantification.

Acknowledgments

Contract grant sponsor: National Science Foundation of Shanghai Science and Technology Commission; contract grant number: 14ZR1442300; Contract grant sponsor: Shanghai Municipal Commission of Health and Family Planning; contract grant number: 201440387; Contract grant sponsor: National Natural Science Foundation of China; contract grant number: 81202809, 81373856.

The authors thank for Dongmei Wu, PhD, at MRI Laboratory of East China Normal University for fruitful discussions during preparation of the article and Yongming Dai, PhD, for help with pulse sequence optimization.

Conflict of Interest

The authors state no conflicts of interest.

References


