Assessment of vertebral fractures and edema of the thoracolumbar spine based on water-fat and susceptibility-weighted images derived from a single ultra-short echo time scan

Sophia Kronthaler1 | Christof Boehm1 | Georg Feuerriegel1 | Peter Börnert2 | Ulrich Katscher2 | Kilian Weiss3 | Marcus R. Makowski1 | Benedikt J. Schwaiger4 | Alexandra S. Gersing1 | Dimitrios C. Karampinos1

1Department of Diagnostic and Interventional Radiology, Technical University of Munich, Munich, Germany
2Philips Research, Hamburg, Germany
3Philips GmbH, Hamburg, Germany
4Department of Diagnostic and Interventional Neuroradiology, School of Medicine, Technical University of Munich, Munich, Germany

Correspondence
Sophia Kronthaler, Department of Diagnostic and Interventional Radiology, Klinikum rechts der Isar, Technische Universität München, Ismaninger Str. 22, 81675 Munich, Germany.
Email: sophia.kronthaler@tum.de

Funding information
The present work was supported by the European Research Council (grant agreement no. 677661, ProFatMRI). This work reflects only the authors’ views and the European Union is not responsible for any use that may be made of the information it contains. Finally, the authors acknowledge research support from Philips Healthcare

Purpose: To develop a methodology to simultaneously perform single echo Dixon water-fat imaging and susceptibility-weighted imaging (SWI) based on a single echo time (TE) ultra-short echo time (UTE) (sUTE) scan to assess vertebral fractures and degenerative bone changes in the thoracolumbar spine.

Methods: A methodology was developed to solve the smoothness-constrained inverse water-fat problem to separate water and fat while removing unwanted low-frequency phase terms. Additionally, the corrected UTE phase was used for SWI. UTE imaging (TE: 0.14 ms, 3T MRI) was performed in the lumbar spine of nine patients with vertebral fractures and bone marrow edema (BME). All images were reviewed by two radiologists. Water- and fat-separated images were analyzed in comparison with short-tau inversion recovery (STIR) and with respect to BME visibility. The visibility of fracture lines and cortical outlining of the UTE magnitude images were analyzed in comparison with computed tomography.

Results: Unwanted phase components, dominated by the $B_1$ phase, were removed from the UTE phase images. The rating of the diagnostic quality of BME visualization showed a high preference for the sUTE-Dixon water- and fat-separated images in comparison with STIR. The UTE magnitude images enabled better visualizing fracture lines compared with STIR and slightly better visibility of cortical outlining. With increasing SWI weighting osseous structures and fatty tissues were enhanced.
1 | INTRODUCTION

In patients with vertebral fractures or degenerative changes of the spine, computed tomography (CT) and MRI are often performed. A CT is preferred to assess the osseous status of the fractured vertebra as well as the adjacent segments for therapy selection and potential surgery planning. Acute vertebral fractures are often associated with bone marrow edema. To differentiate acute from chronic vertebral fractures MR imaging is the standard of reference for the evaluation of the bone marrow to see whether edema is present or not. CT examinations of the spine are associated with radiation exposure, additional examination time, and costs. To assess the discoligamentous aspects, the soft tissue structures and the osseous components of the spine, it would be desirable to acquire all information in a single MRI examination.

MRI provides different approaches for depicting both the soft-tissue and osseous components. First, ultrashort-echo time (UTE) imaging enables signal detection from tissues with short $T_2^*$ components, such as cortical bone. Recent studies suggested the use of UTE and zero TE (ZTE) sequences for the depiction of cortical and trabecular bone. It was further shown that UTE imaging enables the morphological assessment of fractures and degenerative bone changes in the spine and joints. In addition, susceptibility-weighted imaging (SWI), a technique that uses the different magnetic susceptibilities of tissues, is sensitive to tissues that distort the magnetic field by means of paramagnetic or diamagnetic effects. In the past, SWI was mainly used in brain imaging, e.g., to differentiate bleedings from calcifications. Recently, SWI was extended to other regions of the body: as calcified structures are diamagnetic, allowing SWI to visualize the calcified bone matrix. SWI could, thus, be a promising technique to enhance the contrast of osseous tissue components. Second, Dixon sequences are commonly used in the clinical routine either for fat suppression or to generate water- and fat-separated images and to assess soft-tissue components. Dixon techniques exploit the chemical shift-induced phase difference between water and fat signals and usually require the acquisition of multiple images at different TEs. However, due to the need for multiple echoes, Dixon sequences can prolong scan times.

The combination of UTE data acquisition with conventional water-fat separation (UTE-Dixon) enables the depiction of both, the soft-tissue and the osseous components. UTE-Dixon has applications in imaging and quantifying short $T_2$ tissues, eliminating the necessity for fat suppression pulses that directly suppress the long $T_2$ signals. Previous studies have shown that an extension of the well-known IDEAL algorithm can be used to generate water, fat, and quantitative susceptibility maps. Beyond that, the water- and fat-separated images were used to suppress long $T_2$ components which proves itself beneficial in pseudo-CT imaging or in the context of PET attenuation map generation. However, also typical UTE-Dixon imaging requires the acquisition of multiple echoes which prolong the repetition time (TR) and conventional Dixon does not consider the short $T_2$ decay of water signal.

Single-echo Dixon (sTE-Dixon) methods rely on a single complex TE image to decompose fat and water components directly from the complex MR signal. However, the presence of unwanted phase terms corrupts the simple sTE-Dixon approaches based solely on the complex MR signal. To remove unwanted phase terms different techniques were previously reported that either use additional reference scans or that use a region growing algorithm to estimate the unwanted phase terms. The acquisition of additional reference scans yields longer scan times and errors due to patient motion or other sources of inconsistencies. Thus, double echo steady state (DESS) acquisition was recently combined with UTE to perform sTE-Dixon. UTE-DESS acquires two complex signals and can solve for background phase terms in the sTE-Dixon processing. However, the acquisition of two complex signals prolongs the TR. In contrast, UTE imaging, based on single-echo spoiled gradient echo acquisition, acquires a single complex signal and has not yet been combined with sTE-Dixon.
methods to generate water- and fat-separated images from a single-TE UTE (sUTE) acquisition without additional calibration scans. A sUTE acquisition would also minimize the effect of $T_2^*$ decay on the water-fat separation.

The purpose of this work was to develop an approach to simultaneously performing single-UTE Dixon (sUTE-Dixon) imaging and SWI using one sUTE scan to assess the vertebral shape and bone marrow changes at the thoracolumbar spine in patients with vertebral fractures.

2  | METHODS

2.1 | Signal model

The complex water-fat signal $S(t)$ at time $t$ is comprised of the magnitude signal of water $W$ and fat $F$, which, if one neglects $T_2^*$ decay effects for short $t$ and assumes a multiplex peak fat spectrum, takes the form:

$$S(t) = (W + c(t)F) e^{\phi(t)} \text{with } c(t) = \sum_{p=1}^{P} \alpha_p e^{i2\pi \Delta f_p t}$$

(1)

The fat signal component consists of a fat spectrum $c(t)$ with $P$ spectral peaks with relative amplitudes $\alpha_p$ and chemical shifts $\Delta f_p$. The phase between the water and fat components is defined by

$$\theta(t) = \angle(c(t))$$

(2)

$\phi(t)$ accounts for all phase terms that water and fat experience as a common phase. $\phi(t)$ comprises contributions from spatially dependent field $B_0$ inhomogeneities, eddy currents, signal delays in the receiver chains, and phase contributions due to the $B_1$ transmit/receive phase. In this work, eddy currents and signal delays were addressed by means of the gradient impulse response function based on measurements with the thin slice method.

2.2 | Phase contributions

To achieve a water-fat separation based on a single TE, the chemical shift phase $\delta(t)$ and the unwanted phase terms $\phi(t)$ have to be known. At a UTE TE = 0.14 ms and $3T_1$, the chemical shift induced phase difference between water and fat, assuming a nine-peak fat-model is $\delta(TE) = 0.326$ rad. In UTE imaging, the phase contribution due to $B_0$ inhomogeneities was expected to be small with respect to the short TE. Those $B_0$ terms originate from the magnet inhomogeneity, the shim field, the object-based susceptibility, and residual terms from background fields. For the assessment, a Cartesian multi-echo reference scan was acquired to measure the $B_0$ fieldmap in the thoracolumbar spine (Figure 1B). The 90% percentile of the phase $\phi_{pm}$, that resulted from the Cartesian fieldmap at TE = 0.14 ms, was in the range between 0 rad and 0.12 rad (Figure 1C). This phase, induced by the fieldmap, was small compared with the chemical shift phase $\delta(TE)$. However, the UTE phase contained a strong contribution of the $B_1$ phase which varied slowly in the axial plane (Figure 1D–I), representing the dominant term in the UTE phase at TE = 0.14 ms when scanning the lumbar spine at $3T_1$. The $B_1$ phase is caused by the electric conductivity of tissue and has approximately a parabolic shape according to Maxwell’s equations.

2.3 | sUTE-Dixon processing

Under the assumption, that the unwanted phase terms primarily consisted of the $B_1$ phase, which varies smoothly over the field of view (FOV), we propose to solve the following smoothness-constrained non-linear inverse water-fat problem for the unwanted phase term $\phi$:

$$\phi = \arg \min_{\phi} \psi(\phi)$$

$$= \arg \min_{\phi} \frac{1}{2} \| (W + Fe^{i\delta(TE)}) e^{i\phi} - S_{exp}(TE) \|_2^2 + \lambda \| MV \phi \|_2^2$$

(3)

where $\phi$: optimal solution which contains all unwanted phase components, $W$: magnitude of the water signal, $F$: magnitude of the fat signal, $\delta(TE)$: phase difference between water and fat due to the chemical shift, $S_{exp}$: measured signal, $\lambda$: regularization parameter, $M$: mask which was derived from the magnitude images and $V = \left( \frac{\partial}{\partial x}, \frac{\partial}{\partial y}, \frac{\partial}{\partial z} \right)^T$ the $3D$ gradient in the coordinate system of the acquired image. The problem $\psi$ was linearized as follows:

$$\psi(\phi + d\phi) = \frac{1}{2} \left\| (W + Fe^{i\delta(TE)}) e^{i\phi} (1 + i d\phi) - S_{exp}(TE) \right\|_2^2$$

$$+ \lambda \| MV \phi + M V d\phi \|_2^2$$

(4)

The update $d\phi$ was found by solving:

$$0 = \frac{\partial}{\partial \phi} \psi(\phi + d\phi)$$

(5)

$$\left[ (W^2 + F^2 + 2WF \cos(\phi)) + 2\lambda V H M^2 V \right]$$

$$d\phi = - \text{Im} \left( (W + Fe^{i\delta(TE)}) e^{i\phi} S_{exp}(TE) \right) - 2\lambda V H M^2 V \phi$$

(6)

If $\phi$ is known, water and fat can be calculated from Equations (1) and (2) as follows:

$$F = \frac{\text{Im}(S)}{\sin(\delta(TE))}$$

(7)
where $S_{\text{exp}}(TE) e^{-i\phi}$ is the demodulated signal.

Finally, Equation (3) was solved using the conjugate gradient method as follows:

1. Unwrap the UTE phase in 2D
2. Choose a regularization parameter $\lambda$ and set initial conditions with $\phi_n$ being the unwanted phase term during the n-th iteration: $\phi_{n0} = 0; W = 0; I' = 0$
3. Calculate the update $d\phi_n$ according to Equation (6)
4. Update $\phi_{n+1} = \phi_n + d\phi_n$
5. Demodulate unwanted phase terms $S = S_{\text{exp}} e^{-i\phi_{n+1}}$
6. Calculate $F$ and $W$ according to Equations (7) and (8)
7. Repeat steps 2–6 until the relative update $\|d\phi_n\|_2^2$ is smaller than 0.01 with $\phi_n$ being the solution after the n-th iteration step

To prevent phase wraps during the update steps, the phase was scaled to be between $[-\pi, \pi]$ for steps 3–4 and rescaled for the demodulation step 5. A tissue mask was used in the regularizer to only perform the water-fat separation in areas with tissue components. Moreover, the mask prevents errors due to the abrupt increase at the border from signal to no-signal regions where the derivative

![Figure 1](image1.png)

**Figure 1** Analysis of unwanted phase term components in a sagittal UTE scan of a patient’s spine. (A) Magnitude image of the measured UTE signal $|S_{\text{exp}}|$. (B) $B_0$ fieldmap obtained from a six-echo Cartesian multi-echo scan. (C) Histogram of the expected phase contribution $\phi_{fm}$ to the UTE phase based on the Cartesian $B_0$ fieldmap shown in (B). 90% of the fieldmap-based phase values were observed between 0 and 0.12 rad. (D–F) Phase of the sagittal UTE scan reformatted coronally and axially. (G–I) Corresponding line profiles of the UTE phase images. An unwanted phase term was observed in the UTE phase which was larger than the expected phase term (C) due to local field inhomogeneities and susceptibility effects. The unwanted phase term was prominent in AP and RL direction (G, H)

$$W = \Re\left(\hat{S}\right) - |F|\cos(\theta(TE))$$  (8)

where $\hat{S} = S_{\text{exp}}(TE) e^{-i\phi}$ is the demodulated signal.
becomes very large. The signal magnitude was scaled between 0 and the 99th percentile of the maximum to remove outliers. The regularization parameter $\lambda$ was optimized by visual inspection of the water-fat maps, $\phi$ and the corrected signal phase. The problem was under-regularized if $\phi$ contained non-smooth components. The problem was over-regularized if the corrected phase still contained slowly varying unwanted phase terms. All water and fat maps were obtained with $\lambda = 1$. All sUTE-Dixon images were processed with the spectral fat model described in Ren et al.\(^{36}\) To investigate the influence of the spectral fat model on the water-fat separation, the proposed processing was also tested with a wide range of biologically plausible fat spectrum models\(^{40–44}\) (Supporting Information Figure S1, which is available online). The maximal RMSE, between the water image obtained with plausible fat spectrum models and water image obtained with the reference fat model,\(^{36}\) was 1.81% (Supporting Information Table S1).

### 2.4 | SWI-processing

For the generation of SW images, the UTE phase was unwrapped and unwanted phase terms were removed using the result from the sUTE-Dixon processing. The phase was then used to generate a phase mask $f$ as follows:\(^{45}\):

$$S^*_n = \left( f \left| S_{\text{exp}} \right| \right)^{-1} \text{with } f = \begin{cases} \frac{\pi}{\exp(-\phi)} & \text{for } -\pi < \phi \leq 0 \\ \frac{\pi}{\exp(-\phi)} & \text{otherwise} \end{cases}$$

(9)

To increase bone contrast, the phase mask was scaled between 0 and 1 in areas with a phase between $-\pi$ and 0 accordingly. In all areas with a large negative phase or a positive phase, means $|\phi| < -\pi$, the phase mask was set to 1 and thus there was no weighting added to the signal magnitude. For the generation of the SW image $S^*_n$ the phase mask was then $n$-times multiplied with the magnitude of the original UTE image $|S_{\text{exp}}|$. The contrast of the SWI was finally inverted such that bone appears bright.\(^{5,11}\)

### 2.5 | In vivo measurements

In vivo imaging was performed in the lumbar spine of nine patients (seven female, two male; mean age 65.9 ± 15.8 y) with spine fractures. Informed written consent and approval was obtained for each subject by the institutional review board (Klinikum rechts der Isar, Technical University of Munich, Munich, Germany). The spine fracture patients received an MR and a CT scan within 3 days after symptom onset. The CT scans were part of the clinical diagnostic work up.

For the UTE measurements, a stack-of-stars UTE was acquired on a clinical 3.0T MR system (Ingenia Elition X, Philips Healthcare, The Netherlands) using the built-in 12-channel posterior coil, a 12-channel anterior coil and the following parameters: TE 0.14 ms, TR 6.3 ms, flip angle 5°, in-plane resolution 0.45 × 0.45 mm\(^2\), slice thickness 3 mm, FOV 250 × 250 × 279 mm\(^3\), ramp length 0.08 ms, max. gradient strength 15.04 mT/m, sampling dwell time 3.12 μs with 568 samples, acquisition window 1.77 ms, 945 number of spokes, with radial percentage of 85%, partial Fourier with a factor of 0.6 in slice direction and a resulting scan time of 6.3 minutes. sUTE-Dixon processing was performed solving Equation (3). All sUTE-Dixon processing computations were performed in Python on the graphics card of a workstation with GPU 24GiB RAM, 24 core CPU (Intel Xeon Gold) and 768 GB memory. In average the water-fat separation of a full UTE spine data sets took 160 s using 22 iteration steps. SWI processing was also performed using the mask of Equation (7) multiplied up to three times.

For conventional Dixon imaging and for comparison, a six-echo 3D monopolar time-interleaved multi-echo gradient-echo sequence was used\(^{46}\) with following parameters: two interleaves with three echoes per TR and TR/TE/ΔTE: 8.2/1.3/1.1 ms, flip angle 3°, voxel size: 1.8 × 1.8 × 1.8 mm\(^3\), FOV: 626 × 511 × 102 mm\(^3\), receiver bandwidth: 1504 Hz/pixel, frequency direction A/P, scan time: 3.7 minutes. Water–fat maps were calculated using chemical shift encoding-based water-fat separation assuming a common T\(_2\) for water and fat and a multi-peak fat model, tuned specifically to bone marrow.\(^{36,47}\) Furthermore, a sagittal short-tau inversion recovery (STIR) was acquired which is used in the standard clinical routine to detect bone marrow edema.

### 2.6 | CT measurements

CT was performed on one of two CT scanners (Somatom Definition AS+, Siemens Healthineers, and IQon Spectral CT, Philips) with the following parameters, according to routine clinical protocols: Collimation, 0.6 mm; pixel spacing, 0.4/0.3 mm; pitch factor, 0.8/0.9; tube voltage (peak), 120 kV; modulated tube current, 102–132 mA. Images were reformatted with 3 mm slice thickness using a bone-specific convolution kernel (I70H/YB).

### 2.7 | Radiological reading

The visual image analysis of the 6TE-Dixon, STIR, and sUTE-Dixon images was performed by two musculoskeletal
radiologists separately and independently (each with 3 y of clinical experience and 5 y of experience in musculoskeletal research), blinded to clinical and all other information. Images of the nine scanned patients were graded for overall diagnostic image quality on a five-point Likert scale (score of 1, inadequate; 2, poor; 3, moderate; 4, good; 5, excellent). Cohen’s κ was used to determine the inter-reader agreement of the MR imaging findings. All patients were diagnosed with at least one acute vertebral fracture and showed a corresponding bone marrow edema. The radiologists rated STIR, inverted UTE magnitude, and sUTE-Dixon water-fat images with respect to image quality as well as assessment of bone marrow edema in comparison to STIR as standard of reference. The visibility of fracture lines and visibility of cortical outlining was compared with a corresponding bone marrow edema on a conventional CT scan. SW images were not included in the radiological reading.

3  |  RESULTS

3.1  |  sUTE-Dixon processing results

To optimize the regularization parameter, sUTE-Dixon water-fat separation was performed for different regularization parameter values (Figure 2). Figure 2 shows the estimated unwanted phase components ϕ after the 22th iteration, the UTE phase after the demodulation of these unwanted phase components and the resulting water and fat maps. Shown is a sagittal UTE scan of a patient with an acute fracture and edema in L2 (Figure 2 red arrows). For λ = 0.01 the unwanted phase term contained high spatial frequency features which means the smoothness constraint was under-regularized. Useful phase information was demodulated from the UTE phase but the contrast between water and fat was poor. As λ was increased, the contrast of the water and fat images increased as well, and unwanted phase components were demodulated without loss of UTE phase information. For λ = 10 the problem was over-regularized which resulted in maps that falsely identified subcutaneous fat as water (Figure 2, green arrow). The maps with the highest contrast between water and fat were obtained for λ = 1. The obtained phase ϕ contained all smoothly varying background phase components leaving a corrected UTE phase that depicted a contrast mainly driven by chemical shift (Figure 3). To assess the image quality of the sUTE-Dixon water- and fat-separated images, a low-resolution Cartesian 6TE-Dixon based water-fat separation and STIR was used for comparison (Figure 4). The patient showed an acute wedge compression fracture and bone marrow edema in L3. The STIR, added here as a standard of reference, shows the build-up of fluid with a bright signal (Figure 4, red arrows). In all water- and fat-separated images and with both methods, sUTE-Dixon and 6TE-Dixon, the compression fracture and the bone marrow edema were well depicted. Furthermore, the separation line between fluid and bone marrow was clearly visible in the fat images from both methods. Areas with prominent water or fat composition were correctly identified in the sUTE-Dixon maps when compared with the 6TE-Dixon maps. However, the anterior subcutaneous fat region was affected by abdominal breathing motion yielding errors in the sUTE-Dixon water and fat separation (Figure 4, green arrows).

3.2  |  UTE-SWI

Figure 5 shows the UTE-SWI results in comparison with a conventional CT. The patient had an acute compression fracture and bone marrow edema. The corrected phase image shows that tissue containing bone and tissue containing fat had a negative phase whereas tissues containing mainly water depicted a positive phase. The application of the phase masks decreased the signal amplitude in the UTE magnitude images in areas with a negative phase and increased the signal when the processed image was inverted. In the illustrated SW images $S'_0$, $S'_1$, $S'_2$, $S'_3$, the contrast of osseous structures increased, when the magnitude of the original UTE image |S(x)| was multiplied up to three times with the phase mask. The fracture line as well as the visibility of the cortical outlining increased in the fractured vertebra (Figure 5, red arrows). In contrast, the visibility of the cortical outlining in the vertebra below decreased. Due to the negative phase in fatty tissue regions, changes in the bone marrow composition were additionally highlighted next to the osseous structures (Figure 5, white arrows). The green arrow points at a circumscribed sclerotic bone region which became more visible after multiple applications of the SWI weighting. The same osseous structure was detectable in the CT images. Although, the osseous structures were accentuated SNR decreased slightly with each SWI weighting step. Small amounts of noise were propagated to the SW images as a result from noise in the phase images and in the applied phase masks.

3.3  |  Reading results

To determine the performance of the proposed method over a larger number of patients, Table 1 sums up the reading by two radiologists. The rating of the diagnostic quality of the edema visualization showed high ratings for the sUTE-Dixon water- and fat-separated images in comparison with STIR, which was used as the standard
of reference. The interreader agreement was moderate for the water images and substantial for the fat images.

The diagnostic quality of the sUTE-Dixon fat images with respect to edema visibility was rated equally high as the water-separated images (good by both readers). The UTE magnitude images enabled higher scores for visualizing fractures lines compared with STIR with a substantial interreader agreement. With respect to visibility of cortical outlining, the inverted UTE magnitude images ($S_τ'$) reached slightly higher scores ($3.1 \pm 0.3$ for reader 1 and $3.2 \pm 0.4$ for reader 2) compared with the STIR images ($2.9 \pm 0.6$ for reader 1 and $2.9 \pm 0.6$ for reader 2).

Figure 6 shows three representative scans of patients included in the reading. Each subject showed signs of an acute vertebral fracture and bone marrow edema (Figure 6, red arrows). The build-up of fluid in the edema was

---

**Figure 2** Tuning of the regularization parameter $\lambda$. Shown are the sUTE-Dixon results after solving the non-linear inverse W-F problem with four different regularization parameters. For $\lambda = 0.01$ the phase was under-regularized, which resulted in a solution $\phi$ after $n = 22$ iterations that contained non-smooth components. For $\lambda = 10$, the phase was over-regularized, which resulted in a corrected UTE phase that contained remaining unwanted smooth phase terms. In the over-regularized maps, the subcutaneous fat was falsely identified as water (green arrow). For a $\lambda$ between 0.1 and 1, the unwanted phase terms were removed best. A $\lambda$ of 1 gave the best water–fat separation and the visibility of an acute fracture with bone marrow edema (red arrows) was increased compared with the maps with $\lambda = 0.1$.
FIGURE 3  In vivo result of a patient’s spine after solving the non-linear inverse W-F problem with $\lambda = 1$. (A) Solution $\phi$ for the phase which contains all unwanted phase components after the 22-th iteration step. Phase after removing the unwanted phase terms $\phi$ (B) and the corresponding line profiles (C, D) drawn along the red line from anterior to posterior. The phase $\phi$ contained all smoothly varying components leaving a corrected UTE phase that depicted a contrast mainly driven by chemical shift.

FIGURE 4  Comparison of sUTE-Dixon, Cartesian 6TE-Dixon and STIR of a patient with an acute wedge compression fracture of L3 with bone marrow edema. The STIR showed an edema-equivalent signal alteration (red arrows). The UTE scan had a higher in-plane resolution and thicker slices compared with the STIR and the 6TE-Dixon scans. The bone marrow edema was visible in the sUTE-Dixon maps and the contrast between water and fat was comparable with the 6TE-Dixon maps. In the sUTE-Dixon water images, there was signal within the anterior subcutaneous fat region (green arrows), which is prone to artifacts due to abdominal breathing.
highlighted as bright signal in the STIR images. The edema was clearly visible in the sUTE-Dixon water-and fat-separated images. The sUTE-Dixon showed the vertebral fracture and the edema in the water- and fat-separated images.

4 | DISCUSSION

The proposed sUTE-Dixon-SWI methodology allows the removal of unwanted low-frequency background phases and enables simultaneous water-fat separation and SWI processing from a single echo complex UTE image. The formulated smoothness-constrained inverse problem solves the water fat problem while simultaneously removing the unwanted low-frequency phase terms. Therefore, no additional calibration scans are needed to remove unwanted phase components. Another novel aspect of the formulation is the use of a tissue mask in the regularizer and the scaling of the phase to prevent phase wraps during the unwanted low-frequency phase estimation update steps.

By analyzing carefully possible phase contributions, we have shown that the UTE phase in the present data acquired in the thoracolumbar spine at TE = 0.14 ms at 3T was mainly affected by the $B_1$ transmit and receive phase. This $B_1$ phase added in first order a low-frequency modulation to the UTE phase, which varied mainly in anterior-posterior (AP) and right-left (RL) direction. Caused by the electric conductivity of tissue, it has approximately a parabolic shape according to Maxwell’s equations, and a corresponding postprocessing of the $B_1$ phase yields quantitative values of the electric conductivity. A comparison of the obtained conductivity with literature values would in turn provide an additional criterium to identify the optimal value for $\lambda$. Phase contributions from $B_0$ inhomogeneities or the local field changes were small, due to the short TE used. Similar arguments were recently presented while performing sTE-Dixon processing of UTE-DESS data. UTE-DESS acquires two complex signals and can, therefore, solve directly for background phase contributions. The present work instead removed the unwanted low-frequency phase terms based on a single
The tuning of the regularization parameter showed that for $\lambda = 1$ high contrast water- and fat-separated images were obtained. The sUTE-Dixon methodology performed consistently and reliably, produced high quality water- and fat-separated images for all nine patients, involved in this study, regardless of the patient's size using a fixed $\lambda = 1$. Thus, the data post-processing is fully automated which yields an advantage over filtering approaches, where the kernel size and filter type has to be defined for each subject.

In the presented study, we showed that the proposed sUTE-Dixon-SWI methodology allows the simultaneous assessment of vertebral fractures and edema of the thoracolumbar spine from a single MR sequence. The radiological reading suggested that the derived water/fat-sUTE-Dixon and SWI-UTE images can potentially replace the clinical standard of reference, STIR and CT images, in assessing edema and fracture lines, respectively. Compared with the 6TE-Dixon water- and fat-separated images, the sUTE-Dixon maps showed a good agreement and high contrast between water and fat. Thus, the proposed sUTE-Dixon-SWI technique presents several advantages: First, UTE scans are available on most clinical MR systems and the proposed technique could be realized as a data post processing step. Second, the patient can be scanned with one imaging modality. Beyond that, several sequences can be replaced with a single 3D scan acquired in a scan time of 6.3 minutes. The protocol has a large FOV and can potentially be transferred to other body regions. Due to the radial acquisition, the technique shows reduced sensitivity to motion along the frequency, in plane, and encoding direction.

While the present study shows the benefit of a single scan for water-fat separation and SW imaging, it has several limitations. Some limitations were specific to the thoracolumbar spine protocol and may depend on the imaged anatomy. First, water-fat separation can be challenging in anterior fat regions affected by respiratory motion which depends on the anatomy and the scan duration. Second, the UTE images were subject to slight fat blurring (water-fat shift of 0.74 pixel) due to the radial k-space trajectory. The influence of the fat blurring depends on the readout bandwidth and the in-plane resolution of the scan protocol. Finally, the implemented partial Fourier imaging in the slice encoding dimension affected SNR and might add blurring in the slice encoding direction. However, the partial Fourier imaging can be replaced by a parallel imaging acceleration in the future.

Several limitations were specific to the proposed sUTE-Dixon methodology and are independent of the imaged anatomy. First, noise was propagated from the phase...
masks into the SW images and, therefore, SNR decreased slightly at each weighting step. Appropriate denoising of the phase could help to prevent noise propagation and might be subject of future investigations. Second, in the SWI the phase information was used as a weighting in the magnitude. Therefore, not only the contrast of osseous structures is manipulated but also areas with high fat content are weighted. It is important to note that the contrast in the SW-like images comprises both susceptibility and chemical shift effects. Due to the similarity between the weighting of osseous tissue and fatty tissue, SW-like images must be evaluated carefully and were, therefore, not included in the radiological reading. Third, the TE of 0.14 ms was chosen based on the available minimum TE of the pulse sequence and the clinical system presently used. The minimum TE depends on the switching time of the RF system between transmission and reception. Potentially, shorter TEs may be achievable with a different scanner system, yet the question remains whether shorter TEs are beneficial. For shorter TEs, the magnitude images include higher signal from short $T_2^*$ components; however, the phase difference between water and fat is smaller. If TE increases, signals from short $T_2^*$ components decrease, nonetheless the phase difference between water and fat increases and, thus, the contrast between water and fat increases. Estimating the optimal TE is a tradeoff between the short $T_2^*$ magnitude signal and the water-fat phase contrast. In our study, we showed that, for TE = 0.14 ms, high quality water- and fat-separated images can be obtained with magnitude images that include short $T_2^*$ signal components. However, future work is required to define the optimal TE. Finally, the sUTE-Dixon approach could be combined with a low-resolution calibration scan to estimate the fieldmap, which would involve only relatively little additional scan time. An a priori known low-resolution field-map could further improve quality of the water-fat separated images using the proposed processing and might enable quantitative applications using the proposed

FIGURE 6  In vivo UTE lumbar spine sagittal images of three patients with acute vertebral fractures (red arrows). The build-up of fluid in the edema was highlighted in the STIR images with a bright signal. The edema as well as the fracture line were clearly visible in the sUTE-Dixon water–fat images. In subject 2, the signal drops toward the anterior part because the subject was scanned without an anterior coil.
processing. However, additional work would be required to investigate the additional value of a low-resolution calibration scan to estimate the fieldmap and any implications of such a scan for quantitative imaging applications.

5 | CONCLUSION

We proposed a methodology for the removal of unwanted low-frequency background phases, simultaneous water-fat separation and SWI processing from a single echo complex UTE image. The proposed method enabled the simultaneous assessment of vertebral fracture and edema of the thoracolumbar spine from a single MR sequence.

ACKNOWLEDGMENTS

The present work was supported by the European Research Council (grant agreement no. 677661, ProFatMRI). This work reflects only the authors’ views and the European Union is not responsible for any use that may be made of the information it contains. Finally, the authors acknowledge research support from Philips Healthcare. Open access funding enabled and organized by ProjektDEAL.

CONFLICT OF INTEREST

Kilian Weiss, Peter Börnert, and Ulrich Katscher are employees of Philips Healthcare.

ORCID

Sophia Kronthaler https://orcid.org/0000-0001-7913-1238
Christof Boehm https://orcid.org/0000-0003-1321-5804
Ulrich Katscher https://orcid.org/0000-0003-1379-1115

REFERENCES

1. Shanachi AM, Kiczek M, Khan M, Jindal G. Spine anatomy imaging: an update. Neuroimaging Clin N Am. 2019;29:461-480.
2. Piazzolla A, Solarino G, Lamartina C, et al. Vertebral bone marrow edema (VBME) in conservatively treated acute vertebral compression fractures (VCFs): evolution and clinical correlations. Spine. 2015;40:E842-E848.
3. Mandalia V, Henson JH. Traumatic bone bruising—a review article. Eur J Radiol. 2008;67:54-61.
4. Weiger M, Fuessmann KP. Short-T2 MRI: principles and recent advances. Prog Nucl Magn Reson Spectrosc. 2019;114-115:237-270.
5. Ma YJ, Chen Y, Li L, et al. Trabecular bone imaging using a 3D adiabatic inversion recovery prepared ultrashort TE cones sequence at 3T. Magn Reson Med. 2020;83:1640-1651.
6. Lu X, Jerban S, Wan L, et al. Three-dimensional ultrashort echo time imaging with tricomponent analysis for human cortical bone. Magn Reson Med. 2019;82:348-355.
7. Wiesinger F, Sacolick LJ, Menini A, et al. Zero TE MR bone imaging in the head. Magn Reson Med. 2016;75:107-114.
8. Schweiger BJ, Schneider C, Kronthaler S, et al. CT-like images based on T1 spoiled gradient-echo and ultra-short echo time MRI sequences for the assessment of vertebral fractures and degenerative bone changes of the spine. Eur Radiol. 2021;31:4680-4689.
9. Argentieri EC, Koff MF, Breighner RE, Endo Y, Shah PH, Sneag DB. Diagnostic accuracy of zero-echo time MRI for the evaluation of cervical neural foraminal stenosis. Spine. 2018;43:928-933.
10. Breighner RE, Bogner EA, Lee SC, Koff MF, Potter HG. Evaluation of osseous morphology of the hip using zero echo time magnetic resonance imaging. Am J Sports Med. 2019;47:3460-3468.
11. Breighner RE, Endo Y, Konin GP, Gulotta LV, Koff MF, Potter HG. Technical developments: zero echo time imaging of the shoulder: enhanced osseous detail by using MR imaging. Radiology. 2018;286:960-966.
12. Haacke EM, Mittal S, Wu Z, Neelavalli J, Cheng YC. Susceptibility-weighted imaging: technical aspects and clinical applications, part 1. AJNR Am J Neuroradiol. 2009;30:19-30.
13. Böker SM, Adams LC, Bender YY, et al. Differentiation of predominantly osteoblastic and osteolytic spine metastases by using susceptibility-weighted MRI. Radiology. 2019;290:146-154.
14. Boker SM, Adams LC, Fahlenkamp UL, Diederichs G, Hamm B, Makowski MR. Value of susceptibility-weighted imaging for the assessment of angle measurements reflecting hip morphology. Sci Rep. 2020;10:20899.
15. Boker SM, Adams LC, Bender YY, et al. Evaluation of vertebral body fractures using susceptibility-weighted magnetic resonance imaging. Eur Radiol. 2018;28:2228-2235.
16. Adams LC, Bressem K, Boker SM, et al. Diagnostic performance of susceptibility-weighted magnetic resonance imaging for the detection of calcifications: a systematic review and meta-analysis. Sci Rep. 2017;7:15506.
17. Engel G, Bender YY, Adams LC, et al. Evaluation of osseous cervical foraminal stenosis in spinal radiculopathy using susceptibility-weighted magnetic resonance imaging. Eur Radiol. 2019;29:1855-1862.
18. Boker SM, Bender YY, Adams LC, et al. Evaluation of sclerosis in Modic changes of the spine using susceptibility-weighted magnetic resonance imaging. Eur Radiol. 2017;88:148-154.
19. Dixon WT. Simple proton spectroscopic imaging. Radiology. 1984;153:189-194.
20. Glover GH. Multipoint Dixon technique for water and fat proton and susceptibility imaging. J Magn Reson Imaging. 1991;1:521-530.
21. Glover GH, Schneider E. Three-point Dixon technique for true water/fat decomposition with B0 inhomogeneity correction. Magn Reson Med. 1991;18:371-383.
22. Su KH, Frield HT, Kuo JW, et al. UTE-mDixon-based thorax synthetic CT generation. Med Phys. 2019;46:3520-3531.
23. Jang H, von Drygalski A, Wong J, et al. Ultrashort echo time quantitative susceptibility mapping (UTE-QSM) for detection of hemosiderin deposition in hemophilic arthropathy: a feasibility study. Magn Reson Med. 2020;84:3246-3255.
24. Wang K, Yu H, Brittain JH, Reeder SB, Du J. k-space water-fat decomposition with T2* estimation and multifrequency fat
SUPPORTING INFORMATION

Additional supporting information may be found in the online version of the article at the publisher’s website.

FIGURE S1 (A) Representative fat spectrum models. Shown are on the left side the different models for one parameter set (cl of 17.55, ndb of 1.9 and nmdb of 0.9) and on the right side the Peterson 8 peaks model for all different values of cl, ndb and nmdb. (B) Water maps obtained with the Ren marrow reference spectral model (left) and the water map obtained for the spectral model that resulted in the highest RMSE of 1.8% (right)

TABLE S1 Simulation results using different spectral models. The root mean squared error (RMSE) was calculated between the water maps obtained by the Ren marrow reference spectral model and the specified spectral model. The maximal RMSE was 1.81% and is highlighted in red, the corresponding water map is shown in Figure S1B

How to cite this article: Kronthaler S, Boehm C, Feuerriegel G, et al. Assessment of vertebral fractures and edema of the thoracolumbar spine based on water-fat and susceptibility-weighted images derived from a single ultra-short echo time scan. Magn Reson Med. 2022;87:1771–1783. doi:10.1002/mrm.29078