Two-dimensional UTE overview imaging for dental application

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Purpose: To investigate the applicability of a 2D-UTE half-pulse sequence for dental overview imaging and the detection of signal from mineralized dental tissue and caries lesions with ultra-short $T_2^*$ as an efficient alternative to 3D sequences.

Methods: A modified 2D-UTE sequence using 240-µs half-pulses for excitation and a reduction of the coil tune delay from the manufacturer preset value allowed for the acquisition of in vivo dental images with a TE of 35 µs at 1.5T. The common occurrence of out-of-slice signal for half-pulse sequences was avoided by applying a quadratic-phase saturation pulse before each half-RF excitation. A conventional 2D-UTE sequence with a TE of 750 µs, using slice selection rephasing, was used for comparison.

Results: Quadratic phase saturation pulses adequately improve the slice profile of half-pulse excitations for dental imaging with a surface coil. In vivo images and SNR measurements show a distinct increase in signal in ultrashort $T_2^*$ tissues for the proposed 2D-UTE half-pulse sequence compared with a 2D-UTE sequence using conventional slice selection, leading to an improved detection of caries lesions.

Conclusion: The proposed pulse sequence enables the acquisition of in vivo images of a comprehensive overview of bone structures and teeth of a single side of the upper and lower jaw and signal detection from mineralized dental tissues in clinically acceptable scan times.

Keywords
2D UTE, caries, dental imaging, half-pulse, tune delay

1 INTRODUCTION

X-ray-based 2D projective imaging is the current gold standard for routine diagnostic procedures in dental practice. Intra-oral techniques, such as bitewing and periapical radiographs, provide detailed information of the crowns and roots of the teeth in an only very limited area of the dental arch, whereas extra-oral panoramic radiographs yield a complete overview of the upper and lower jaw including all teeth, temporo-mandibular joints, and maxillary sinuses. Unfortunately, the panoramic
radiograph can suffer from nonuniform magnification, geometric distortions, and lower resolution compared with intra-oral imaging methods.\textsuperscript{2} Cephalometry provides projections from the side of the head in its entirety, thereby superimposing the right and left side of the skull, not allowing for a detailed analysis of individual teeth.\textsuperscript{1} Distortion-free panoramic radiographs can be calculated from cone beam computed tomography data at the expense of a much higher radiation exposure.\textsuperscript{3}

Considering the potential risk from the exposure to ionizing radiation and the only limited visualization of soft-tissue structures, there has been a continued interest in alternative approaches for the generation of dental images. Previous studies have shown standard MRI sequences to be able to visualize and assess a wide range of dental conditions: pulp vitality,\textsuperscript{5,6} apical periodontitis,\textsuperscript{7} impacted teeth,\textsuperscript{8} trigeminal and mandibular nerve bundles,\textsuperscript{9-11} and function of the temporo-mandibular joint.\textsuperscript{12-15} The increased soft-tissue contrast in MRI also allows for further characterization of apical bone lesions, which is not possible with cone beam computed tomography.\textsuperscript{16}

Dental MRI, however, is challenging due to low hydrogen content and extremely short $T_2$ relaxation times of the highly mineralized tissues of interest. The relaxation times, well under 400 µs for dentine,\textsuperscript{17,18} cause faint magnetization and a quickly dephasing signal.

Sequences capable of capturing the signal before the rapid decay of the FID like UTE imaging,\textsuperscript{19,20} zero-TE imaging,\textsuperscript{21,22} and sweep imaging with Fourier transformation\textsuperscript{23-26} have been proven to be applicable to dental imaging. However, these 3D sequences rely on the intrinsic volumetric approach, which limits their application for providing a large FOV of a single side of the jaw at sufficiently high spatial resolution in clinically acceptable scan times. Translation to 2D-UTE imaging is also challenging because of the strong signal decay due to the mandatory slice-selective excitation with related prolonged TEs. Half-pulse excitation pulses, which do not require any rephasing gradient after the slice selection, have been previously suggested for 2D UTE\textsuperscript{27} but are known to be sensitive to eddy current effects. Several methods have been introduced for the compensation of these effects.\textsuperscript{28-33}

In this work, we apply a 2D-UTE approach using 240-µs half-sinc-pulse excitations for dental imaging in clinically acceptable scan times. To ensure proper slice selection, a previously proposed combination with a quadratic phase (QP) pulse\textsuperscript{33} was used to avoid unwanted out-of-slice signal, caused by the eddy current–induced slice-selection gradient distortions during the half-pulse excitations. This method possesses the benefits of X-ray projection techniques like a comprehensive side view of the oral cavity without the drawbacks of inherent geometric distortions and varying magnification, while combining it with the advantages of MR-based UTE sequences, such as enhanced soft-tissue contrast, signal detection from short $T_2$ tissues, and lack of ionizing radiation.

2 METHODS

In this study, we customized a conventional 2D-UTE sequence, which originally consisted of a full sinc-Gauss pulse with slice-selection gradient rephasing. Although the readout was left unchanged, short half-sinc pulses were designed and the coil tune delay was modified, as is described subsequently.

2.1 Slice excitation

As suggested earlier,\textsuperscript{27,34} by splitting the excitation into two subsequent acquisitions with inverted slice-selection gradient polarity, the use of a rephasing gradient can be avoided. However, half-pulse sequences are known to be susceptible to gradient waveform distortions and phase errors caused by eddy currents.\textsuperscript{30}

For the compensation of the gradient waveform distortions, the real slice-selection gradient waveform was approximated by the discrete convolution of the ideal trapezoidal gradient waveform with a mono-exponential model using an empirical time constant\textsuperscript{35} of $\tau = 39$ µs, applying a cutoff when $G_{real}$ reaches 1% of the peak gradient strength:

\[
G_{real} (t) = \sum_\theta G_{trapez} (t - \theta) e^{-\frac{|t - \theta|}{\tau}}. \tag{1}
\]

Applying the VERSE principle\textsuperscript{36} together with Equation 1, a precompensated pulse waveform, adapted to the gradient ramp-down, was calculated.

To avoid severe signal decay during excitation, the length of the half-pulse was chosen as short as possible for a desired slice thickness of 10 mm, resulting in a pulse duration of 240 µs and a peak $B_1$ of 23 µT. This short pulse length led to playing out most of the pulse during gradient ramp-down, causing improper off-center slice profiles, which were compensated by a modulation of the RF pulses as follows:\textsuperscript{36}

\[
B_{1comp} (t) = B_1 (t) \exp \left( -\int_{0}^{T} G_{real} (\tau) \, d\tau \right). \tag{2}
\]

for a slice offset $z_0$. Gradient system delays were examined using the method of Dyun et al.,\textsuperscript{37} and the timing between pulses and slice gradients was adjusted accordingly.

The $B_0$ eddy currents, induced by time-varying gradients in the cryostat, led to a difference in phase accumulation during excitation with slice-selection gradients of positive and negative polarity,\textsuperscript{30} and thus to the appearance of sidelobes and a spatial shift of the slice profile. For compensation, reference scans were performed before each in vivo measurement.
by acquiring FIDs for positive and negative slice-selection gradients at the center of the slice profiles. From the initial data points, the phase shift was evaluated, which was then corrected during image reconstruction by multiplication with a constant phase term before the combination of the individual half-pulse images, as previously described by Wansapura et al.30

Despite all of these measures, some tissue was still excited outside of the slice of interest, and a QP saturation pulse was designed as described by Schulte et al38 and applied before each half-pulse excitation.33 The QP saturation pulse was used to saturate a 50-mm-thick volume on either side of the imaging slice, thus compensating for slice profile distortions of the subsequently applied half-sinc pulses. A sequence diagram of the excitation part is provided in Figure 1.

2.2 | Tune delay minimization

Tune delay defines a short time period between the end of the RF pulse and the start of data acquisition that allows the receive coils to be properly tuned before signal acquisition and avoids an overlap between RF pulse and sampling event, and thus damages to the sensitive preamplifiers in the receive chain. For TE minimization, the tune delay for the four-element dental surface coil (NORAS, Hoechberg, Germany) was reduced from the default value of 300 µs as preset by the manufacturer to 10 µs by changing the tune-delay value in the corresponding coil configuration file on the MR system, ensuring safe switching to receive mode. Acquiring data with such a reduced tune delay led to the corruption of initial data points of the FID, as they were sampled during the transient phase of the finite impulse response filter and a still partially detuned coil. An analysis of the sampled FID shows that the error to the signal can be approximated by an $e^{-it}$ exponential function, behaving like a low-pass filter. The effect on the signal and the resulting image can then be described as

$$s(t) \cdot [1 - e^{-it}] \rightarrow S(x) - [S(x) \ast L(x)],$$

where $s(t)$ is the ideal uncorrupted signal; $S(x)$ is its Fourier transform; and $L(x)$ is a Lorentzian function (mathematical operations usually needed for non-Cartesian image reconstruction such as convolution kernel and density compensation are omitted for reasons of simplicity). Equation 3 shows that the Fourier transform of the approximated error on the FID signal describes the effect on the final image to be similar to the well-known unsharp mask filter, in which a smoothed version of the original signal is subtracted from the signal itself, theoretically causing minor edge enhancement, a slight DC offset, and reduced SNR in the images.39 The effect of a reduced tune delay on the FID signal and image quality was investigated and is provided in Supporting Information Figures S1-S3.

2.3 | Experiments

All data were acquired with a 1.5T whole-body imaging system (Achieva; Philips Healthcare, Best, the Netherlands)
using a four-element dental surface coil (NORAS) with the tune delay set to 10 µs. Before each half-pulse excitation, an 8 ms QP pulse with time-bandwidth product of 34, a stop band of 10 mm, transition width of 1.5 mm, and saturation bands of 50 mm on either side of the slice was played out with subsequent gradient spoiling. Data acquisitions of the same profile with inverted slice-selection gradient polarities were performed directly after each other to reduce the effect of patient motion on the images. The proposed half-sinc-pulse sequence was compared with a radial 2D-UTE sequence using conventional slice selection with gradient rephasing and full-sinc pulses for excitation, and is denoted as “conventional 2D-UTE” from here on. All experiments were performed with a radial center-out trajectory, slice thickness = 10 mm, FOV = 180 × 180 mm, in-plane resolution = 0.5 × 0.5 mm, TR = 120 ms, flip angle = 15°, and a scan time = 1:25 minutes for a single image and 2:50 minutes for a complete half-pulse image. Readout gradients for both sequences had a strength of 21 mT/m, ramp-up duration of 0.117 ms, and plateau length of 1.113 ms. During the half-pulse excitation, the gradient strength was 21 mT/m with a ramp-down time of 0.134 ms. A total of 407 data points were sampled from every FID with an acquisition bandwidth of 330.6 kHz and dwell time of 3.025 µs. For the half-pulse sequence, a TE of 35 µs and excitation bandwidth of 12.6 kHz was achieved. The TE was calculated from the end of the RF pulse to the beginning of data sampling and includes the tune delay of 10 µs as well as other hardware delays such as front-end switching times and control overhead. Measurements with the conventional 2D-UTE sequence were performed with an excitation bandwidth of 6.6 kHz and a minimum TE of 750 µs due to gradient rephasing. To allow for a better comparison, data were also acquired with a half-pulse scan with TE = 750 µs.

The performance of the QP saturation pulse was tested in a phantom experiment using a spherical phantom filled with doped water. Slice profiles were measured for the conventional 2D-UTE sequence and the half-pulse sequence with and without the saturation pulse. To verify proper off-center correction of the half-pulses, the position of the slice was chosen to resemble those in the in vivo scans, with an offset of 50 mm from the isocenter on the mediolateral axis and inclinations of 5° around the sagittal axis and 20° around the longitudinal axis.

In vivo scans of the left and right side of the jaw were acquired from 2 volunteers after obtaining written informed consent and were part of a dental MRI study approved by the ethics committee of the Ulm University Medical Center. To properly secure the mask-like dental coil on the faces of the volunteers, touch fastener strips were attached to the coil and a head-positioning cushion, which also helped to reduce involuntary head movements. Scanning was performed in oblique sagittal plane with slight medial inclination parallel to the ramus of the mandible. Care was taken to ensure that the following anatomical structures were incorporated: maxilla and mandible of one side of the jaw with all molars and premolars, the temporo-mandibular joint, and the maxillary cavity. Image reconstruction using gridding and further image processing was performed with in-house-developed software implemented in MATLAB (MathWorks, Natick, MA). After the B₀ eddy current correction and summation of the individual half-pulse images, the sum-of-squares approach was used to combine the individual coil images.

The SNRs were calculated on the reconstructed magnitude images according to

\[ SNR = \frac{\bar{I}_{ROI}}{\sigma_{BG}}, \]

where \( \bar{I}_{ROI} \) is the average intensity in the region of interest (ROI) and \( \sigma_{BG} \) is the SD of the background noise. A correction factor of 0.695 was used to adjust for the Rician noise distribution in the magnitude images, as well as the use of a four-element coil. To verify an expected SNR increase in tissues with very short T₂ for the half-pulse sequence, the dentin parts of the crowns of three teeth (16, 18, and 46 in ISO dental notation) in 1 volunteer were chosen as ROIs. Additionally, a caries lesion in tooth 17 and the lateral pterygoid muscle next to the temporo-mandibular joint were used for SNR measurements. To compensate for the double acquisition in the half-pulse images, the SNR values were divided by a factor of \( \sqrt{2} \) to allow for a proper comparison.

### 3 | RESULTS

Slice profiles acquired from phantom experiments are displayed in Figure 2. The profile created by the half-pulse sequence without the QP saturation pulse is wider and shows significantly more signal from outside of the desired imaging region than the profile of the conventional 2D-UTE sequence (dotted black line), and half-pulse sequence without (striped red line) and with (solid blue line) combination of the 8-ms quadratic phase (QP) saturation pulse. This demonstrates the well-defined slice selectivity due to suppressed out-of-slice signal.
sequence. Adding the QP pulse before the half-pulse excitations resolved these issues, creating a slice profile with desired FWHM of 10 mm and very narrow transition widths between slice and saturation band, thus avoiding any out-of-slice signal.

In vivo scans of 1 volunteer’s right jaw acquired with the conventional 2D-UTE sequence and the half-pulse sequence with TE = 750 µs and 35 µs are shown in Figure 3. Dental structures, including the position and orientation of the molars and premolars, as well as the temporo-mandibular joint and the maxillary sinus, are clearly visible in all images. No apparent obscuring of anatomical details by out-of-slice signal from the cheek or tongue is noticeable in the half-pulse images. The detection of signal in dentin (solid arrows in Figure 3) can be appreciated in the half-pulse image with a TE of 35 µs; furthermore, the proximal caries lesion in tooth 17 (dotted arrows Figure 3) appears larger and closer to the pulp.

In Figure 4, the results from the conventional 2D-UTE scan and the half-pulse scan with TE of 35 µs of the second volunteer’s left jaw are displayed. The half-pulse image shows a distinct increase in signal intensity in dentin (solid arrows in Figure 4). Distortion artifacts and signal loss due to the presence of a metallic implant as replacement for tooth 25 is clearly visible in both images (striped and dotted arrows). Signal loss from the implant screw in the maxilla is almost completely removed in the half-pulse scan (striped arrow Figure 4B), whereas artifacts in the region of the abutment and crown are still present (dotted arrow).

The SNR measurements made from in vivo scans are provided in Figure 5. Conventional 2D-UTE scan and half-pulse scan with an equivalent TE of 750 µs have similar SNR in all ROIs, whereas SNR values from the half-pulse scan with the shorter TE of 35 µs are substantially larger in all short $T_2^*$ ROIs.

4 | DISCUSSION AND CONCLUSIONS

A proper slice profile with very sharp transition width is essential for dental imaging when using surface coils, as superimposed signal from close-by tissues with slow $T_2^*$ relaxation such as cheek and tongue could easily decrease local contrast for close-by short $T_2^*$ tissues. Phantom measurements demonstrate the improved slice selectivity of the half-pulse sequence with the addition of the QP saturation pulse. Saturation bands of 5 cm appear to be adequate for half-pulse dental imaging, as no apparent out-of-slice signal was observed in the in vivo scans.

Because of the decrease in TE and pulse duration, in vivo scans acquired with the presented approach allow for signal detection in highly mineralized dental tissues such as dentin. This is reflected in distinctly larger SNR values in tissues with very short $T_2^*$ measured with the half-pulse scan at a TE of 35 µs. Breakdown of enamel and dentin by acid-producing bacteria cause an accumulation of liquid in demineralized structures.
Sequences with UTEs are especially well suited for an improved detection and visualization of signal from small amounts of acid and saliva in between hard dental tissues, leading to a visibly larger extent of the lesion, but has previously only been reported for 3D acquisitions.20,41-43 The largest increase in SNR was measured with the modified 2D-UTE half-pulse sequence in the caries lesion in tooth 17. Because ultrashort $T_2^*$ relaxation processes do not dominate the signal in the lateral pterygoid muscle, SNR values in this ROI are relatively similar for all three imaging sequences, with slightly higher value for the half-pulse sequence with TE of 35 µs. A comparison of the half-pulse sequence with a 3D-UTE sequence is provided in Supporting Information Figures S4 and S5.

No negative effects on image quality due to the reduced tune delay were observed in any of the scans (see Supporting Information Figures S1-S3). This suggests that the effect of signal corruption of a few initial data points, as described by Equation 3, is almost negligible for the UTE sampling scheme. Because the initial data points were acquired during the gradient ramp-up, k-space velocity is still relatively slow, with eight corrupted data points corresponding to 0.68 k-space points. Additional correction of the central k-space signal, similar to the reconstruction of missing k-space points in zero-TE imaging with algebraic methods,44,45 does not seem necessary, but may require further investigation.

Metallic materials like titanium or cobalt-chromium (CoCr) alloys, which are often used for dental implants, are a common cause for artifacts in MRI.46 Local field perturbations close to the implant lead to nonexcited off-resonant spins and signal loss due to rapid dephasing. A decrease in signal loss around an implant can be achieved by reducing the TE or increasing the excitation bandwidth, thus exciting off-resonant spins and acquiring the signal before the spins are completely dephased. Although a reduced artifact size of the dental implant screw can be appreciated in the in vivo half-pulse image, a possible influence on this effect from slice-profile distortions, in addition to the short pulse duration (high excitation bandwidth) and TE, cannot be ruled out. A similar area of signal loss around the implant crown in the half-pulse and conventional 2D-UTE images indicates that a material with a substantially higher susceptibility than the titanium of the implant screw, such as a CoCr alloy,47 was used as crown material. Although this demonstrates that metallic implants are no definite reason to exclude the use of half-pulse sequences for dental imaging, 2D-UTE sequences are generally not recommended for metal artifact reduction, as inherent slice selectivity and exclusive use of frequency encoding make them susceptible to off-resonances, causing geometric distortions and signal pile-ups.47

One limitation of combing half-pulse sequences with QP saturation pulses is an increased power deposition at higher field strengths as well as the inability to perform interleaved multislice acquisitions due to wide saturation bands around each slice.

In general, the use of thick-slice or projective techniques may limit the sensitivity due to superimposed or partly overlapping tissues like gingiva or dental pulp. However, it was shown earlier20 that MRI in principle is more sensitive to early caries lesions than X-ray-based techniques, most likely due to rapid increase of proton density (acid/fluid) with simultaneous increase of the $T_2^*$ (breakdown of mineralized structures), indicating an at least similar performance of the suggested approach as the current clinical standard of bite-wing X-ray. Furthermore, considering that caries most likely develops either from occlusal or approximal locations with no overlap, the technique has the potential for X-ray-free detection of caries lesions. Bone density changes, as visible in, for example, Figure 4B (dashed arrow), can clearly be appreciated in the projective image as well as the cortical bone border. Whether the delineation of the nerve channel is sufficient to identify, for example, inflammatory processes cannot be answered from the cases available.

In conclusion, the presented approach of using a modified 2D-UTE half-pulse sequence for dental imaging has
been demonstrated successfully with in vivo acquisitions at 1.5T field strength. The UTE sampling scheme allowed a reduction of the coil tune delay without any visible effects on image quality, and in combination with 240-µs excitation pulses lead to an effective TE of 35 µs. The presented methods allowed the acquisition of comprehensive overview images while providing improved signal detection from strongly mineralized dental tissues and caries lesions. It provides an efficient alternative to 3D-UTE methods requiring isotropic spatial resolution, with estimated scan times for a 3D-UTE sequence with kooshball trajectory, similar FOV (in-plane), and resolution of approximately 60 minutes, whereas other 3D-UTE sequences capable of anisotropic resolutions, such as 3D cones, might prove to be another viable alternative. Further research might explore the possibilities of acquiring panoramic images depicting the entirety of the jaw by combining images from several slices, which has previously been reported for 3D data sets. The suggested technique has the potential to provide the relevant information for initial dental diagnosis including caries diagnosis, assessment of periodontitis, and bone degradation. With the potential of identification of inflammation and the advantage of full-jaw coverage, including the teeth from crown to root plus full coverage of the bone structures of the jaw, it may eventually become a competitor to the well-established bite-wing X-ray technique. Sensitivity issues have to be worked out during larger-scale clinical studies.

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FIGURE S1 The first 20 data points of FID's acquired in a spherical phantom with the conventional 2D-UTE sequence with TE of 750 µs are shown for coil tune delays of 100 µs (violet, identical to blue line), 60 µs (blue), 50 µs (green), 10 µs (yellow), and 0 µs (red). If the tune delay is reduced, signal corruption due to the acquisition during the signal buildup in the digital filter and a still partly detuned coil can occur. Reduction of the tune delay from the vendor setting of 100 µs to 60 µs shows no visible influence on the FID. A slight effect on the FID can be seen with the tune delay set to 50 µs, with reduced signal intensity of the first four data points. Further reduction of the coil tune delay increases the strength of the effect and the amount of data points affected by it. A coil tune delay of 10 µs, as used for all in vivo acquisitions, shows corruption of the first eight data points. At the minimal tune delay of 0 µs, the initial FID signal intensity has been reduced significantly and the progression of the signal can be approximated with $1 - e^{-2t}$. The effect on the signal and the resulting image can then be described as $s(t) = [1 - e^{-2t}] \cdot S(x) - |S(x) \ast L(x)|$, where $s(t)$ is the ideal uncorrupted signal, $S(x)$ is its Fourier transform, and $L(x)$ is a Lorentzian function (mathematical operations usually needed for non-Cartesian image reconstruction such as convolution kernel and density compensation are omitted for reasons of clarity). The resulting effect on the final image is similar to an unsharp mask filter, theoretically causing minor edge enhancement, a slight DC offset, and reduced SNR in the images

FIGURE S2 Images of a spherical phantom acquired with the conventional 2D-UTE sequence for coil tune delays of 100 µs (A), 50 µs (B), 35 µs (C), 20 µs (D), 10 µs (E), and 0 µs (F). An impact on image quality can only be observed for a tune delay of 0 µs (F) with apparent signal surrounding the liquid (arrows) and reduced signal in the liquid

FIGURE S3 The SNR values calculated from image acquisitions shown in Supporting Information Figure S2. Apparent SNR values were calculated using the “difference method” described by Reeder et al.\(^2\) to reduce the effect of a possible DC offset. The effect of the unsharp mask filter-like behavior can be observed for a tune delay of 0 µs as distinctly reduced SNR in the center of the liquid and an increase in SNR above the liquid. Only a slight increase in SNR is visible in the region of interest (ROI) above the liquid for a tune delay of 20 µs and 10 µs. Keeping the coil tune delay values above a certain threshold, dependent on the coil and system hardware, the relatively slow k-space velocity during gradient ramp-up of the UTE sampling scheme leads to an almost negligible effect of the signal corruption of a few initial data points on image quality

FIGURE S4 Images from phantom measurements for a comparison between half-pulse and 3D-UTE sequences. A cube-shaped phantom (edge length = 100 mm), in which a pencil
eraser consisting of a soft ($T_2^* \approx 1.9$ ms) and hard rubber part ($T_2^* \approx 0.73$ ms) and a slice of a tangerine were immersed in agarose, was used for measurements. The $T_2^*$ of caries lesions with $T_2^* \approx 0.694$ ms is similar to the $T_2^*$ of the hard rubber part. Scans of the phantom were performed with the 2D-UTE half-pulse, a true 3D-UTE sequence with nonselective excitation and kooshball trajectory with isotropic resolution, and a 3D-UTE sequence with anisotropic resolution using a stack-of-stars (SOS) sampling pattern that requires phase encoding and does not allow similar UTE as 2D-UTE half-pulse or true 3D-UTE sequences. The half-pulse sequence was performed with the same parameters as in the manuscript, the true 3D-UTE sequence with equivalent resolution, and TE as the half-pulse sequence but with reduced FOV of 110 mm$^3$ and the 3D-UTE SOS sequence with equivalent in-plane FOV (minimum number of slices = 5) and anisotropic resolution as the half-pulse scan, but a minimum TE of 410 µs due to the phase-encoding steps. To make contrast and SNR measurements comparable, all sequences used the same TR as the 2D-UTE half-pulse scan in the manuscript. Scan times were 2:50 minutes for the 2D-UTE half-pulse sequence, 3:14 hours for true 3D UTE, and 7:27 minutes for 3D-UTE SOS. Similar contrast can be observed in the half-pulse (A) and true 3D-UTE image (B), whereas in the 3D-SOS-UTE image (C), less signal is visible in both eraser parts.

**FIGURE S5** The SNR measurements from three ROIs calculated from the images in Supporting Information Figure S4. After adjustment of the SNR values from the true 3D-UTE and 3D-SOS UTE images for the noise-averaging properties of 3D scans to the half-pulse sequence parameters, only 1.2% difference in SNR could be observed in the ultrashort $T_2^*$ hard rubber part for the half-pulse and true 3D-UTE sequence, whereas distinctly lower SNR was measured with 3D SOS UTE due to the longer minimal TE of 410 µs. Similar behavior was observed for the soft rubber part. Because ultrashort $T_2^*$ relaxation processes do not dominate the signal in the tangerine, SNR values in this ROI are relatively similar for all three imaging sequences.

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