Single point imaging with radial acquisition and compressed sensing

Serhat Ilbey1 | Pia M. Jungmann2,3 | Johannes Fischer1 | Matthias Jung2 | Michael Bock1 | Ali Caglar Özen1

1Department of Radiology, Medical Physics, Medical Center University of Freiburg, Faculty of Medicine, University of Freiburg, Freiburg, Germany
2Department of Diagnostic and Interventional Radiology, Medical Center University of Freiburg, Faculty of Medicine, University of Freiburg, Freiburg, Germany
3Department of Radiology, Cantonal Hospital Grisons, Chur, Switzerland

Correspondence
Serhat Ilbey, Department of Radiology, Medical Physics, Medical Center University of Freiburg, Faculty of Medicine, University of Freiburg, Freiburg, Germany. Email: serhat.ilbey@uniklinik-freiburg.de

Purpose: To accelerate the Pointwise Encoding Time Reduction with Radial Acquisition (PETRA) sequence using compressed sensing while preserving the image quality for high-resolution MRI of tissue with ultra-short $T_2^*$ values.

Methods: Compressed sensing was introduced in the PETRA sequence (csPETRA) to accelerate the time-consuming single point acquisition of the k-space center data. Random undersampling was applied to achieve acceleration factors up to $Acc = 32$. Phantom and in vivo images of the knee joint of six volunteers were measured at 3T using csPETRA sequence with $Acc = 4, 8, 12, 16, 24, \text{and} 32$. Images were compared against fully sampled PETRA data ($Acc = 1$) for structural similarity and normalized-mean-square-error. Qualitative and semi-quantitative analyses were performed to assess the effect of the acceleration on image artifacts, image quality, and delineation of anatomical structures at the knee.

Results: Even at high acceleration factors of $Acc = 16$ no aliasing artifacts were observed, and the anatomical details were preserved compared with the fully sampled data. The normalized-mean-square-error was less than 1% for $Acc = 16$, in which single point imaging acquisition time was reduced from 165 to 10 s, reducing the total scan time from 7.8 to 5.2 min. Semi-quantitative analyses suggest that $Acc = 16$ yields comparable diagnostic quality as the fully sampled data for knee imaging at a scan time of 5.2 min.

Conclusion: csPETRA allows for ultra-short $T_2^*$ imaging of the knee joint in clinically acceptable scan times while maintaining the image quality of original non-accelerated PETRA sequence.

KEYWORDS
compressed sensing, magnetic resonance imaging, musculoskeletal imaging, Pointwise Encoding Time Reduction with Radial Acquisition, short $T_2$, single point imaging, ultra-short echo time
1 | INTRODUCTION

MRI has a great potential for clinical assessment of tissues with $T_1$ in the sub-millisecond range, such as bones, tendons, and teeth. In musculoskeletal (MSK) disorders, MRI is an indispensable tool for the detection of pathologies of connective tissues such as menisci, rotator cuff, ligaments, and tendons. With conventional echo-based sequences direct imaging of tissues with sub-millisecond $T_2$ values is not feasible because the TEs are usually longer than a millisecond. However, FID-based projection techniques have been proposed that offer ultra-short acquisition delays: UTE imaging, back-projection low angle shot (BLAST), rotating ultra-fast imaging sequence (RUFIS), zero TE (ZTE), point-wise encoding time reduction with radial acquisition (PETRA), water- and fat-suppressed proton projection MRI (WASPI), and HYFI (hybrid filling of the dead-time gap). A more extreme version of 3D radial projection imaging has been implemented by simultaneous RF excitation and signal acquisition, such as in sweep imaging with Fourier transform (SWIFT).

In UTE, radial k-space data are acquired directly after a non-selective RF pulse. In ZTE, PETRA, WASPI, and HYFI, the frequency-encoding gradients are already switched on before the RF excitation. In all of these sequences, the finite RF pulse duration ($t_{RF} \geq 8$ µs for clinical systems) and acquisition delays between RF pulse and data acquisition (typically, $t_{gap} > 20$ µs) result in a sampling gap at the center of k-space. To recover the missing central k-space samples single points in Cartesian space (PETRA), or time-efficient radials (WASPI), or a combination of those (HYFI) can be acquired. Alternatively, in ZTE algebraic reconstruction is used to synthesize the missing k-space center, which is possible when dedicated hardware is used to limit the sampling gap to a few points. WASPI is the fastest technique that can retrieve the missing data, but it suffers from oscillatory image artifacts, and the HYFI technique trades-off the image quality against the reduction in acquisition time. PETRA has provided artifact-free images for various MSK imaging applications, including meniscus and articular cartilage. However, total acquisition times in PETRA are long for high-resolution protocols or when long acquisition delays are required due to the large sampling gap at the center of the k-space. Acquisition times between 8 and 13 min have been reported for spatial resolutions of 0.8 mm and higher.

Compressed sensing (CS) has been used to drastically accelerate MRI data acquisitions. In MSK, imaging CS was applied to reduce the scan times in metal artifact reduction, 3D relaxation mapping, and sodium imaging, or to improve the image quality as a complementary tool to the parallel imaging. In particular, the pure phase encoding in single point imaging (SPI) is perfectly suitable for a random undersampling procedure such as CS, because each data point in k-space is sampled independently from the others. So far, SPI with CS has only been presented for pre-clinical MRI applications to quantify $T_2$.

In this study, we applied a CS acceleration to the central SPI part of the PETRA sequence to reduce the acquisition times to clinically acceptable ranges. In knee images with sub-millimeter spatial resolution, different acceleration rates were applied and compared with a PETRA acquisition without acceleration.

2 | METHODS

2.1 | PETRA sequence

In general, a PETRA sequence consists of a radial and a Cartesian part (Figure 1): the outer k-space is acquired along k-space spokes with a radial acquisition, whereas in an inner k-space sphere, SPI is used to acquire Cartesian k-space points. In the radial acquisition, the amplitude of the readout gradient $|G| = \sqrt{G_x^2 + G_y^2 + G_z^2}$ is kept constant, but its direction is changed for every TR. For spoiling gradients are not ramped down after acquisition but rather ramped directly to the strength required for the next repetition. As the center of k-space cannot be acquired during radial readout due to the dead time between transmit and receive operations, additional SPI data need to be sampled to fill the inner k-space. The radius of the missing part of the k-space in PETRA is given by $k_{gap} = N_{gap} k_{dw}$, where $N_{gap} = \frac{t_{dw}}{t_{RF}}$, $k_{dw} = \gamma |G| t_{dw} = \gamma G_{dw}$ is the dwell time, $TE$ is the echo time, and $\gamma$ is the gyromagnetic ratio. Here, $TE$ is the time at which the first data point is acquired along a radial spoke. The required number of SPI samples in the sphere around the k-space origin is approximately

$$N_{SPI} = \frac{4}{3} \pi \left(N_{gap}\right)^3 = \frac{4}{3} \pi \left(\frac{TE}{t_{dw}}\right)^3 = \frac{4}{3} \pi \left(TE \cdot BR \cdot OS \cdot BW_{px}\right)^3$$

(1)

as $t_{dw}^{-1} = BR \cdot OS \cdot BW_{px}$, where BR is the desired base resolution, and OS is the oversampling factor. Thus, $N_{SPI}$ increases with the third power of the pixel bandwidth $BW_{px}$, BR, OS, and TE. In Table 1, measurement times for SPI acquisition with respect to BR, TE, and $BW_{px}$ are given with the corresponding maximum gradient amplitude ($|G|$). Note that for $TE = 50$ µs, $BR = 512$, OS = 2, $BW_{px} = 1000$ Hz, and TR = 2 ms, the acquisition of $N_{SPI}$ points can take as long as 18 min.
To reduce scan time for SPI acquisition, the random undersampling scheme is used over a spherical k-space shell, and only the data near k-space center are fully sampled. This full sampling pattern near k-space center was chosen to maintain a high SNR, which is represented in the central k-space points. The size of the fully sampled region was determined experimentally.

In Figure 2, the sampling patterns for the SPI section at \( k_z = 0 \) plane are presented for eight-fold acceleration (\( \text{Acc} = 8 \)). The undersampling schemes were generated with the following design parameters:

- Random versus Poisson-disk sampling: positions of SPI samples are fully randomly (Random) or pseudo-randomly (Poisson) determined using Poisson-disk distributions.\(^{38} \) 3D Poisson sampling patterns were generated using the sample elimination technique.\(^{39} \)

- Variable sampling density: Poisson sampling patterns were generated with a varying sampling density in radial direction. The density of samples was inversely proportional to the distance to k-space center.

- Shape of the fully sampled region: data near k-space center was fully sampled over a cubic volume of side length \( k_{\text{inc}} \) (Cube) or over a spherical volume of diameter \( k_{\text{inc}} \) (Sphere).

Using above design parameters five different undersampling schemes were generated: Random-Cube (RC), Poisson-Cube (PC), Poisson-Sphere (PS), variable-density Poisson-Sphere (vdPS), and variable-density...
Poisson-Cube (vdPC). These schemes with varying size of the fully sampled region were compared empirically from volunteer experiments relative to fully sampled PETRA data. The same undersampling scheme that results in minimum error was used for all subsequent csPETRA experiments.

For comparison, the total number of acquisitions for the complete inner k-space is illustrated for WASPI, PETRA, HYFI, and csPETRA methods in Supporting Information Figure S1, which is available online. The WASPI method is faster than PETRA and HYFI as it radially acquires data, whereas PETRA is the slowest among all other techniques, as each k-space point is sampled with SPI. The proposed csPETRA has the shortest acquisition time up to $k_{\text{gap}} = 48 k_{dw}$ for $\text{Acc} = 16$.

### 2.3 Image reconstruction

For image reconstruction, measured csPETRA data are represented by a set of linear equations:

\[
\begin{bmatrix}
    s_1 \\ s_2 \\ \vdots \\ s_C
\end{bmatrix} =
\begin{bmatrix}
    \mathcal{F} \Gamma_1 \\ \mathcal{F} \Gamma_2 \\ \vdots \\ \mathcal{F} \Gamma_C
\end{bmatrix} \begin{bmatrix}
    m+n
\end{bmatrix},
\]

where $s_c$ is the measured signal of the $c$'th coil, $m$ is the 3D image, and $n$ is the noise. For the accelerated data acquisition, the equation becomes:

\[
s = M \mathcal{F} \Gamma m + n,
\]

where $M$ is the SPI sampling mask. As $M$ is not a full-rank matrix, in general the equation is ill-conditioned. Therefore, images are reconstructed iteratively by solving the following minimization problem:

\[
\arg\min_{m} \lambda \text{TV}(m) + \alpha \|m\|_1 \\
\text{s.t.} \|M \mathcal{F} \Gamma m - s\|_2 < \varepsilon,
\]

where $\text{TV}(\cdot)$ is 3D total variation operator, $\| \cdot \|_1$ and $\| \cdot \|_2$ are the $\ell_1$ and $\ell_2$-norm operators, $\lambda$ and $\alpha$ are scalar weights of TV and $\ell_1$-norm regularization operators, respectively. $\varepsilon$ is the bound on the data fidelity error. $\ell_2$-norm and TV functions are selected as regularization terms.

In all subsequent measurements radial projection data were first re-gridded onto a Cartesian grid using Kaiser-Bessel convolution kernel and then combined with the SPI data after applying Hamming and density compensation filters.\textsuperscript{40,41} The iterative reconstruction was implemented using Alternating Direction Method of Multipliers (ADMM)\textsuperscript{42} of the BART toolbox.\textsuperscript{43} For multi-channel receive coils, data from each channel was reconstructed independently due to high memory requirements, then the images were combined with the sum-of-squares method. The regularization parameters were optimized over the range of $[10^{-4}, 5 \cdot 10^{-2}]$ with $10^{-4}$ steps with respect to nMSE. $\alpha$ was set to 0.0009, 0.001, 0.0014, 0.002, 0.0028, and 0.004 for $\text{Acc} = 4, 8, 12, 16, 24$, and 32, respectively. The $\lambda$ was set to $10^{-4}$ for each acceleration rate.
TABLE 2 Evaluation criteria for the semi-quantitative analysis of volunteer images using five-point Likert-scales (1 best, 5 worst)

| Artifacts and image quality                  | Scale (5 best, 1 worst)                                                                 |
|----------------------------------------------|----------------------------------------------------------------------------------------|
| Individual assessment for                    | 1 = severe artifacts, no diagnostic image                                               |
| Bone                                         | 2 = moderate artifacts, diagnostic quality limited                                      |
| Muscle                                       | 3 = moderate artifacts, but still sufficient diagnostic quality                         |
| Intra-articular structures                   | 4 = some minor artifacts, but diagnostic quality identical to Acc = 1                  |
|                                              | 5 = no additional artifacts, diagnostic quality identical to Acc = 1                   |

| Visual contrast and delineation               | Scale (5 best, 1 worst)                                                                 |
|----------------------------------------------|----------------------------------------------------------------------------------------|
| Individual assessment for                    | 1 = contrast markedly inferior to Acc = 1, delineation markedly inferior to Acc = 1     |
| Cartilage versus meniscus                    | 2 = contrast moderately inferior to Acc = 1, delineation moderately inferior to Acc = 1 |
| Cartilage versus fluid                       | 3 = contrast slightly inferior to Acc = 1, delineation slightly inferior to Acc = 1     |
| Cartilage versus subchondral lamina          | 4 = contrast slightly inferior to Acc = 1, delineation identical to Acc = 1             |
| Meniscus versus fluid                        | 5 = contrast identical to Acc = 1, delineation identical to Acc = 1                     |
| Subchondral lamina versus fatty bone marrow  |                                                                                        |
| Osseous trabeculae versus fatty bone marrow  |                                                                                        |

2.4 Phantom and knee MR image acquisition

To assess the csPETRA image quality, experiments were performed at a clinical 3T MRI system (MAGNETOM Prisma Fit, Siemens AG, Erlangen, Germany). A resolution phantom was measured with a single-channel receive loop coil (Ø = 11 cm, Siemens AG, Erlangen). In addition, images of the right knee of six healthy volunteers were acquired with a 15-channel transmit/receive knee coil (QED, Cleveland, OH).

Conventional turbo spin echo (TSE) images were acquired for reference. The TSE sequence parameters are presented in Supporting Information Table S1. PETRA data were acquired with a full SPI section: FOV = (180 mm)³, BW(px) = 625 Hz, TR = 2 ms, TE = 48 µs, tRF = 8 µs, flip angle = 6°, BR = 400, OS = 2, voxel size = (0.45 mm)³, radials: 150 000, SPI points: 82 519, T_acq = 7.8 mins. Note that, this acquisition is already undersampled in the azimuthal direction for the radial spokes. To fulfill the Nyquist criterion \( N_{rad} = \pi (400)^2 \approx 500,000 \) spokes are required. Undersampling with SPI was retrospectively applied for a direct comparison.

The acoustic noise next to the magnet and additive acoustic noise of csPETRA with different acceleration factors are measured using a digital sound level meter (DSL 331, Tecpel Co. Ltd., Taiwan) and compared to other rapid imaging techniques.

All methods were carried out in accordance with relevant guidelines and regulations, healthy volunteer scanning was approved by the Institutional Review Board (Ethikkommission) of the University Medical Center Freiburg (No. 160/2000), and informed written consent was obtained before imaging.

2.5 Quantitative analysis

The performance was evaluated by measuring the structural similarity (SSIM) index\(^4^4\) and the nMSE = \( \frac{\|x-x_{ref}\|^2}{\|x_{ref}-x_{ref}\|^2} \), where \( x \) is the reconstructed image, \( x_{ref} \) is the reference PETRA image with a full SPI section, and \( \gamma \) is the mean operator. For the calculation of SSIM index and maps, exponents for the luminance, contrast, and structural terms were set to 1, SD of isotropic Gaussian function was set to 1.5, and regularization constants \( C_1, C_2, \) and \( C_3 \) were set to \( 10^{-4}, 9 \cdot 10^{-4}, \) and \( 4.5 \cdot 10^{-4} \), respectively.

2.6 Qualitative analysis

PETRA and csPETRA images with acceleration factors of 1 (no acceleration), 4, 8, 12, 16, 24, and 32 were qualitatively assessed and semi-quantitatively analyzed independently by two radiologists specialized on musculoskeletal radiology (12 and 2 y of experience). The readers were blinded to volunteer and image details. All images were transferred to picture archive and communication server (PACS). Both radiologists described the artifacts visually. In addition (i) artifacts and image quality for bone, muscle, and intra-articular structures; and (ii) contrast between different adjacent structures and delineation of those were evaluated semi-quantitatively using five-point Likert scales (Table 2).\(^4^5\) Original PETRA images (Acc = 1) were used as the standard of reference (five points for all
parameters). Inter-reader agreement was calculated using Kappa with quadratic weighting. For comparison of different acceleration factors, paired t-tests were applied. Mean values of the two readers were used. All P-values < .05 were considered statistically significant.

3 | RESULTS

In Figure 3 images of the first volunteer which were retrospectively undersampled using the RC, PC, PS, vdPS, and vdPC schemes are presented together with the absolute difference images and SSIM maps for Acc = 8. The undersampling scheme PS, at which the samples are acquired according to the Poisson-disk distribution in 3D with a spherical fully sampled region with $k_{in,s} = 10k_{dw}$, outperformed other schemes in terms of nMSE and SSIM. For PS case nMSE of 0.5% and a SSIM of 0.99 (Table 3) were found.

In Figure 4, phantom images acquired with PETRA (Acc = 1) and csPETRA (Acc = 4, 8, 12, 16, 24, and 32) are presented together with the difference images and SSIM maps. A conventional T1-weighted TSE image is shown for reference. Because undersampling is done only for the low-frequency components, edges are preserved as can be concluded from the SSIM maps.

In Figures 5 and 6, a comparison of the high-resolution knee images of a volunteer are presented together with the difference images and SSIM maps—(raw) image data are made available at https://github.com/serhatilbe y/csPETRA. T1-weighted TSE images are also shown for reference. The error intensity increases with Acc, which is visible in structures that contain low frequency information such as muscles (see the Qualitative Analysis section below).

nMSE and SSIM index for csPETRA images presented in Figures 5 and 6 were compared with the original PETRA image (Acc = 1). Table 3 shows the nMSE and SSIM table of images with their acquisition times. The nMSE is less than 1% even for the acceleration factor of Acc = 16, in which the SPI acquisition time was reduced from 165 s to 10 s. The average SSIM of all volunteers for Acc = 16 was 0.97. SSIM maps suggest that structural information is preserved for intra- and periarticular structures including menisci, cartilage, ligaments, and periarticular tendons. A slight decrease in SSIM is observed particularly in periarticular muscles.

The reconstruction of a 3D data set took approximately 20 min for one channel using a workstation with 3.6-GHz six-core CPU and 128 GB RAM.

Acoustic noise next to the magnet was 58 ± 1 dBA, whereas the acoustic noise during the radial and SPI part of the PETRA (Acc = 1) measurements were 59 ± 1 dBA, and 60 ± 1 dBA, respectively. The noise level has increased about 2 dB, 5 dB, 7 dB, 9 dB, and 11 dB for the Acc = 8, 12, 16, 24, and 32, respectively, compared with Acc = 1. However, acoustic noise level of csPETRA with Acc = 16 was still approximately 23 dB lower than the typical acoustic noise levels of conventional techniques (e.g., TSE) and UTE. These results are illustrated in Supporting Information Figure S2.

![Figure 3](image_url)

**FIGURE 3** Images of the first volunteer, which were retrospectively undersampled using the RC, PC, PS, vdPS, and vdPC schemes are presented together with the four times up-scaled absolute difference images and SSIM maps for Acc = 8. The undersampling scheme Poisson-Sphere (PS), at which the samples are acquired according to the Poisson-disk distribution in 3D with a spherical fully sampled region, outperformed other schemes in terms of nMSE and SSIM. For PS case nMSE of 0.5% and a SSIM of 0.99 (Table 3) were found.
Qualitative analysis

In the reconstructed images, the signal inhomogeneity artifacts, which increase with Acc, are visible for example in tibia only in four times up-scaled absolute difference images for Acc ≤ 16. For Acc ≥ 24, inhomogeneity artifacts were visible in vastus lateralis muscle as well as femur, tibia, and fibula. Intra-articular structures including the anterior and posterior cruciate ligaments, the menisci, and cartilage appeared homogeneous. In Figure 6, blurring artifacts were found at the image periphery, for example, at the edge of the posterior knee extending partially toward the gastrocnemius muscles. The blurring at the periphery can be attributed to unwanted slice selection; hence, it was independent of Acc.

Despite the undersampling, individual joint structures, such as cartilage, menisci, and ligaments, could be identified with homogeneous signal. Descriptively, all soft tissue structures showed intermediate MR signal similar to muscle signal intensity. Cortical and trabecular bone and fluid showed slightly hypointense signal and was, therefore, well delineated in comparison to all soft tissue structures (menisci, cartilage, ligaments, fat).

Semi-quantitative analysis

In the semi-quantitative evaluation, inter-reader agreement was substantial according to the definition of Landis and Koch between the two readers (Kappa with
Quadratic Weighting $0.69 \pm 0.07$, 95% confidence interval [0.55, 0.84]). Mean values of both readers were used for statistical analyses. Although overall image quality visually remained good, paired $t$-tests showed a stepwise increase of artifact severity and a reduction in diagnostic quality ($P < .05$; Table 4). In the parameter-based analyses, significant differences were found for the artifacts and image quality in muscle and bone with $Acc = 12$. At $Acc \geq 24$, scores of 3 or below were given for bone and muscle artifacts, indicating limited or insufficient diagnostic quality, whereas artifacts and image quality for intra-articular structures was only reduced slightly. The contrast and the delineation decreased to diagnostic quality below the diagnostic quality of $Acc = 1$ only at 32 for the delineation of cartilage. For the differentiation of osseous trabeculae and fatty bone marrow the diagnostic quality was below that for $Acc = 1$ at $Acc \geq 16$ (score < 4) and the reduced delineation of osseous trabeculae and fatty bone marrow the diagnostic quality was already visible at the first acceleration step ($Acc = 8$), indicating this parameter to be the limiting factor. Although the overall differences were statistically significant for each acceleration step, for the total score the diagnostic quality was reduced only of $Acc \geq 24$. The total acquisition time was reduced from 7.8 for $Acc = 1$ down to 5.2 min for $Acc = 16$.

4 | DISCUSSION

In this work, CS was used in PETRA sequence to reduce the number of SPI samples. The performance of the proposed csPETRA method was demonstrated with different
acceleration rates and compared with the original PETRA technique on in vivo knee measurements. With csPETRA, 0.45-mm MRI of the knee was possible in clinically acceptable measurement times of about 5 min.

Compared with SPI with full sampling of Cartesian k-space, csPETRA can reduce the SPI acquisition time at the center of the k-space by 16-fold. In Equation (1), $N_{SPI}$ is cubically related to inverse of dwell time. Increasing the base resolution results in using smaller dwell times; thus, a higher number of SPI points is required. The effect of the proposed undersampling scheme on the total acquisition time is prominent especially in high-resolution protocols. In both phantom images and in vivo, knee imaging details were preserved even at high acceleration factors of 16, and the nMSE was less than 1% up to a 16-fold acceleration. Also, the overall structure of the images was preserved as can be seen by the high SSIM index values (>0.97) and SSIM maps.

The performance of the CS method is highly dependent on sparsity of the data and randomness of the undersampling scheme. In 3D radial imaging, undersampling of radial spokes results in a pseudo-random sampling scheme, where all the samples along a radial line are omitted. In this study, the end points of straight radial spokes form a 3D helix that covers k-space homogeneously to minimize acoustic noises and eddy-current artifacts. Regular radial undersampling was applied to shorten the scan time, and random undersampling was not used because it introduces coherent artifacts on the images. To further improve the scan efficiency, different trajectories with additional undersampling can be used, such as cones and rosette samplings.

Long ring-down time of coils and/or system switching times can prolong minimum TE up to 150 µs (Supporting Information Figure S3). Numerous studies in the literature have included PETRA sequence with TE = 70 µs.12,20,50,51 PETRA acquisitions are extremely long for TE values above a millisecond, since the time required for SPI acquisition increases with the third power of TE. The proposed approach allows PETRA technique to be used in quantitative high-resolution MRI applications, such as $T_2^*$ mapping, focusing on short-$T_2^*$ applications such as myelin, tendon, and cartilage. The csPETRA technique allowed visualization of tissues with short $T_2^*$ values in high resolution (<0.5 mm) keeping the measurement time almost the same as the ZTE technique, which has no single point acquisition to retrieve the missing data.

Compared to clinical $T_1$-weighted spin echo sequences, PETRA yields less contrast between cartilage and connective tissues, and intrastructural signal is more homogeneous for all joint structures. Assessment of structural pathologies, such as ruptures and degeneration, and the additional clinical value of PETRA sequences need to be assessed in larger cohorts in future studies and was not scope of this technical development study. It was shown recently that contrast of PETRA can be improved using long-$T_2$ and/or fat suppression techniques; such magnetization preparation techniques can also be included in csPETRA.

At a subjective score of 4, the diagnostic quality of csPETRA with $Acc = 16$ was rated identical to $Acc = 1$. Thus, despite the presence of inhomogeneity artifacts, $Acc = 16$ may be implemented for csPETRA sequences in clinical practice. Cartilage was well differentiated from...
subchondral lamina with csPETRA for all acceleration factors. In common TSE sequences, the delineation of the deep cartilage layers versus the subchondral lamina is challenging. This observation may suggest a useful clinical benefit of csPETRA but needs to be investigated further in a larger clinical study.

Despite the high rate of undersampling of low-frequency information, aliasing artifacts were not observed in csPETRA images. The structure of the difference images (Figures 5 and 6) shows that the performance of the method at the knee joint is only limited to SNR. nMSE mainly originated from the regions where SNR < 40; however, visual inspection and SSIM maps suggest that structural information even from the low SNR regions is preserved. The proposed iterative reconstruction technique using $\epsilon_1$-norm and TV as regularization operators significantly reduces the error compared to FFT-based reconstruction (Supporting Information Figure S4).

In PETRA, k-space data are acquired directly after a non-selective RF pulse. However, because the frequency-encoding gradients are already switched on before the RF excitation to increase the acquisition bandwidth, the excitation is not truly non-selective and a sinc-shaped slice centered at the isocenter is selected. The bandwidth of this slice profile is inversely proportional to the duration of the RF pulse and amplitude of the readout gradient. Hence, to avoid the unwanted slice selectivity artifacts very short RF pulses must be used and the center of the region of interest must be positioned as close as possible to the isocenter. In this work, we used a hard pulse with 8 µs duration, which was the shortest pulse that could create 6° flip angle, which resulted in highest SNR. Moreover, we positioned the knee joint at the isocenter. Thus, blurring artifacts from unwanted slice selectivity were only present at the image periphery in the posterior section of the knee, which is positioned farthest away from the isocenter.

Currently, PETRA provides a hybrid contrast of $T_1$ and spin density weighting. To improve contrast, preparation pulses can be included; however, this would result in longer TRs. The total time saved by using magnetization prepared csPETRA would be proportional to TR.

In this work, data from each channel were reconstructed independently due to high memory requirements. However, iteratively reconstructing all channels using parallel imaging techniques might increase the overall image quality, which can be highly valuable in clinical practice. On the other hand, the csPETRA images have already very high SSIM (≥0.97) and low nMSE (≤1%) for Acc ≤ 16, so that the possible improvement in terms of SSIM and nMSE will only be minor when parallel imaging is used.

In general, CS reconstructions can be time-consuming if data from many coil elements need to be processed. The overall reconstruction time depends on the available number of physical central processing unit (CPU) or graphical processing unit (GPU) cores and memory, because channels can be reconstructed in parallel. Moreover, ADMM is highly suitable for distributed computing. The computation time can theoretically be reduced to 5 min using state-of-the-art GPUs.

5 | CONCLUSION

csPETRA enables 3D imaging with isotropic submillimeter resolution within only a few minutes. As demonstrated in 3D csPETRA imaging of the human knee, acceleration factors up to 16 can be used, which reduce the overall scan time by 33%.

ORCID
Serhat Ilbey https://orcid.org/0000-0002-3574-4320
Pia M. Jungmann https://orcid.org/0000-0003-2198-3678
Johannes Fischer https://orcid.org/0000-0003-1499-6163
Matthias Jung https://orcid.org/0000-0002-1124-4284
Michael Bock https://orcid.org/0000-0001-9720-3506
Ali Caglar Özen https://orcid.org/0000-0003-3536-0826

REFERENCES
1. Du J, Bydder GM. Qualitative and quantitative ultrashort-TE MRI of cortical bone. NMR Biomed. 2013;26:489-506.
2. Gatehouse PD, Bydder GM. Magnetic resonance imaging of short T2 components in tissue. Clin Radiol. 2003;58:1-19.
3. Robson MD, Benjamin M, Gishen P, Bydder GM. Magnetic resonance imaging of the Achilles tendon using ultrashort TE (UTE) pulse sequences. Clin Radiol. 2004;59:727-735.
4. Weiger M, Pruessmann KP, Bracher A-K, et al. High-resolution ZTE imaging of human teeth. NMR Biomed. 2012;25:1144-1151.
5. Robson MD, Bydder GM. Clinical ultrashort echo time imaging of bone and other connective tissues. NMR Biomed. 2006;19:765-780.
6. Shapiro L, Harish M, Hargreaves B, Staroswiecki E, Gold G. Advances in musculoskeletal MRI: technical considerations. J Magn Reson Imaging. 2012;36:775-787.
7. Robson MD, Gatehouse PD, Bydder M, Bydder GM. Magnetic resonance: an introduction to ultrashort TE (UTE) imaging. J Comput Assist Tomogr. 2003;27:825-846.
8. Hafner S. Fast imaging in liquids and solids with the back-projection low angle ShoT (BLAST) technique. Magn Reson Imaging. 1994;12:1047-1051.
9. Lauterbur PC. Image formation by induced local interactions: examples employing nuclear magnetic resonance. Nature. 1973;242:190-191.
10. Madio DP, Howe JI. Ultra-fast imaging using low flip angles and FIDs. Magn Reson Med. 1995;34:525-529.
11. Weiger M, Pruessmann KP. MRI with zero echo time. Encycl Magn Reson. 2012;1:311-322.
12. Grodzki DM, Jakob PM, Heismann B. Ultrashort echo time imaging using Pointwise Encoding Time Reduction with Radial Acquisition (PETRA). Magn Reson Med. 2012;67:530-518.

13. Li C, Magland JF, Zhao X, Seifert AC, Wehrli FW. Selective in vivo bone imaging with long-T2 suppressed PETRA MRI. Magn Reson Med. 2017;77:989-997.

14. Wu Y, Ackerman JL, Ackerman JL, Graham L, Wang Y, Glimcher MJ. Density of organic matrix of native mineralized bone measured by water- and fat-suppressed proton projection MRI. Magn Reson Med. 2003;50:59-68.

15. Froidevaux R, Weiger M, Rössler MB, Brunner DO, Pruessmann KP. HYFI: hybrid filling of the dead-time gap for faster zero echo time imaging. NMR Biomed. 2021;34:e4493. doi:10.1002/nbm.4493

16. Idiyatullin D, Corum C, Park JY, Garwood M. Fast and quiet MRI using a swept radiofrequency. J Magn Reson. 2006;181:342-349.

17. Özen AC, Atalar E, Korvink JG, Bock M. In vivo MRI with concurrent excitation and acquisition using automated active analog cancellation. Sci Rep. 2018;8:10.1038/s41598-018-28894-w

18. Froidevaux R, Weiger M, Brunner DO, Dietrich BE, Wilm BJ, Pruessmann KP. Filling the dead-time gap in zero echo time MRI: principles compared. Magn Reson Med. 2018;79:2036-2045.

19. Wu Y, Dai G, Ackerman JL, et al. Water- and fat-suppressed proton projection MRI (WASPI) of rat femur bone. Magn Reson Med. 2007;57:554-567.

20. Lee YH, Suh JS, Grodzki D. Ultrashort echo (UTE) versus Pointwise Encoding Time Reduction with Radial Acquisition (PETRA) sequences at 3 Tesla for knee meniscus: a comparative study. Magn Reson Imaging. 2016;34:75-80.

21. Van Dyck P, Vanhevel F, Vanhoenacker FM, et al. Morphological MR imaging of the articular cartilage of the knee at 3 T—comparison of standard and novel 3D sequences. Insights Imaging. 2015;6:285-293.

22. Kobayashi N, Goerke U, Wang L, Ellermann J, Metzger GJ, Garwood M. Gradient-modulated PETRA MRI. Tomography. 2015;1:85-90.

23. Candès EJ, Romberg J, Tao T. Robust uncertainty principles: exact signal reconstruction from highly incomplete frequency information. IEEE Trans Inf Theory. 2006;52:489-509.

24. Donoho DL. Compressed sensing. IEEE Trans Inf Theory. 2006;52:1289-1306.

25. Lustig M, Donoho D, Pauly JM. Sparse MRI: the application of compressed sensing for rapid MR imaging. Magn Reson Med. 2007;58:1182-1195.

26. Yang J, Zhang Y, Yin W. A fast alternating direction method for TVL1-L2 signal reconstruction from partial Fourier data. IEEE J Sel Top Signal Process. 2010;4:288-297.

27. Huang J, Zhang S, Metaxas D. Efficient MR image reconstruction for compressed MR imaging. Med Image Anal. 2011;15:670-679.

28. Block KT, Uecker M, Frahm J. Undersampled radial MRI with multiple coils. Iterative image reconstruction using a total variation constraint. Magn Reson Med. 2007;57:1086-1098.

29. Fritz J, Ahlawat S, Dehmehr S, et al. Compressed sensing SEMAC: 8-fold accelerated high resolution metal artifact reduction MRI of cobalt-chromium knee arthroplasty implants. Invest Radiol. 2016;51:666-676.

30. Jungmann PM, Bensler S, Zingg P, Fritz B, Pfirrmann CW, Sutter R. Improved visualization of juxtaprosthetic tissue using metal artifact reduction magnetic resonance imaging: experimental and clinical optimization of compressed sensing. Magn Reson Imaging. 2019;54:23-31.

31. Zibetti MV, Sharafi A, Otazo R, Regatte RR. Accelerated mono- and biexponential 3D–T1p relaxation mapping of knee cartilage using golden angle radial acquisitions and compressed sensing. Magn Reson Med. 2020;83:1291-1309.

32. Madelin G, Chang G, Otazo R, Jerschow A, Regatte RR. Compressed sensing sodium MRI of cartilage at 7T: preliminary study. J Magn Reson. 2012;214:360-365.

33. Otazo R, Kim D, Axel L, Sodickson DK. Combination of compressed sensing and parallel imaging for highly accelerated first-pass cardiac perfusion MRI. Magn Reson Med. 2010;64:767-776.

34. Emid S, Creyghton JHN. High resolution NMR imaging in solids. Physica B+C. 1985;128:81-83.

35. Emid S. Ultra high resolution multiple quantum spectroscopy in solids. Physica B+C. 1985;128:79-80.

36. Rioux JA, Beyea SD, Bowen CV. 3D single point imaging with compressed sensing provides high temporal resolution R 2* mapping for in vivo preclinical applications. Magn Reson Mater Phys. 2017;30:41-55.

37. Speidel T, Paul J, Wundrak S, Rasche V. Quasi-random single-point imaging using low-discrepancy $k$-space sampling. IEEE Trans Med Imaging. 2018;37:473-479.

38. Cook RL. Stochastic sampling in computer graphics. ACM Trans Graph TOG. 1986;5:51-72.

39. Yuksel C. Sample elimination for generating Poisson disk sample sets. Comput Graph Forum. 2015;34:25-32.

40. Jackson JJ, Meyer CH, Nishimura DG, Macovski A. Selection of a convolution function for Fourier inversion using gridding (computerised tomography application). IEEE Trans Med Imaging. 1991;10:473-478.

41. Beatty PJ, Nishimura DG, Pauly JM. Rapid gridding reconstruction with a minimal oversampling ratio. IEEE Trans Med Imaging. 2005;24:799-808.

42. Boyd S, Parikh N, Chu E, Peleato B, Eckstein J. Distributed optimization and statistical learning via the alternating direction method of multipliers. Found Trends Mach Learn. 2010;3:1-122.

43. BART. https://mrirecon.github.io/bart/ Published 2015 doi:10.5281/zenodo.31907

44. Wang Z, Bovik AC, Sheikh HR, Simoncelli EP. Image quality assessment: from error visibility to structural similarity. IEEE Trans Image Process. 2004;13:600-612.

45. Jamieson S. Likert scales: how to (ab)use them. Med Educ. 2004;38:1217-1218.

46. Landis JR, Koch GG. The measurement of observer agreement for categorical data. Biometrics. 1977;33:159-174.

47. Gurney PT, Hargreaves BA, Nishimura DG. Design and analysis of a practical 3D cones trajectory. Magn Reson Med. 2006;55:575-582.

48. Noll DC. Multishot rosette trajectories for spectrally selective MR imaging. IEEE Trans Med Imaging. 1997;16:372-377.

49. Li Y, Yang R, Zhang C, Zhang J, Jia S, Zhou Z. Analysis of generalized rosette trajectory for compressed sensing MRI. Med Phys. 2015;42:5530-5544.

50. Niwa T, Nozawa K, Aida N. Visualization of the airway in infants with MRI using Pointwise Encoding Time Reduction with Radial Acquisition (PETRA). J Magn Reson Imaging. 2017;45:839-844.
SUPPORTING INFORMATION

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

TABLE S1 MR imaging parameters

FIGURE S1 Total number of acquisitions for the complete inner k-space for WASPI, PETRA, HYFI, and csPETRA methods. WASPI method is faster than PETRA and HYFI as it radially acquires data, whereas PETRA is the slowest among all other techniques, as each k-space point is sampled with SPI. The total number of acquisitions in HYFI method depends on the parameter $A \in [0,1]$ (or targeted $T_2^*$ value) and it leads to PETRA and WASPI for $A = 0$ and 1 (at limiting cases), respectively. The proposed csPETRA method with $Acc = 16$, has the shortest acquisition time up to $k_{gap} = 48 k_{dw}$.

FIGURE S2 Acoustic noise levels. Acoustic noise next to the magnet was $58 \pm 1$ dBA, whereas the acoustic noise during the radial and SPI part of the PETRA ($Acc = 1$) measurements were $59 \pm 1$ dBA, and $60 \pm 1$ dBA, respectively. The noise level has increased about 2 dB, 5 dB, 7 dB, 9 dB, and 11 dB for the $Acc = 8, 12, 16, 24$, and 32, respectively, compared to $Acc = 1$. Please note that, acoustic noise level of csPETRA with $Acc = 16$ was still approximately 23 dB lower than the typical acoustic noise levels of conventional techniques (e.g., TSE) and UTE.

FIGURE S3 FID measurements from a Tx/Rx knee coil (QED, Cleveland, Ohio) (left) and wrist coil (Siemens, Erlangen) (right). Signal oscillations only vanish at $t = 60 \mu s$ and 150 $\mu s$ when knee and wrist coils are used, respectively. Thus, even though in principle the SPI part could be made very small, in practice hardware limitations lead to an extended SPI section, and, thus, to longer measurement times.

FIGURE S4 Coronal view of high-resolution knee PETRA ($Acc = 1$) and csPETRA ($Acc = 16$) images of a volunteer reconstructed with Fourier Transform after zero-filling and proposed iterative reconstruction (CS) together with the twice up-scaled absolute difference images and SSIM maps. The proposed iterative reconstruction technique using $\ell_1$-norm and TV as regularization operators significantly reduces error.

How to cite this article: Ilbey S, Jungmann PM, Fischer J, Jung M, Bock M, Özen AC. Single point imaging with radial acquisition and compressed sensing. Magn Reson Med. 2022;87:2685–2696. doi:10.1002/mrm.29156