Combined Compressed Sensing and SENSE to Enhance Radiation Therapy Magnetic Resonance Imaging Simulation

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Abstract
Purpose: To assess the effect of a combination of compressed sensing and SENSitivity Encoding (SENSE) acceleration techniques on radiation therapy magnetic resonance imaging (MRI) simulation workflows.
Methods and Materials: Thirty-seven acquisitions were performed with both SENSE-only (SENSE) and combined compressed sensing and SENSE (CS) techniques in 24 patients receiving radiation therapy MRI simulation for a wide range of disease sites. The anatomic field of view prescription and image resolution were identical for both SENSE and CS acquisitions to ensure fair comparison. The acquisition time of all images was recorded to assess time savings. For each image pair, image quality, and ability to contour were assessed by 2 radiation oncologists. Aside from direct image pair comparisons, the feasibility of using CS to improve MRI simulation protocols by increasing image resolution, field of view, and reducing motion artifacts was also evaluated.
Results: CS resulted in an average reduction of 27% in scan time with negligible changes in image quality and the ability to contour structures for RT treatment planning compared with SENSE. Physician scoring of image quality and ability to contour shows that while SENSE still has slightly better image quality compared with CS, this observed difference in image quality did not affect the ability to contour. In addition, the higher acceleration capability of CS enabled use of superior-inferior direction phase encoding in a sagittal 3-dimensional T2-weighted scan for substantially improved visibility of the prostatic urethra, which eliminated the need for a Foley catheter in most patients.
Conclusions: The combination of compressed sensing and parallel imaging resulted in marked improvements in the MRI Simulation workflow. The scan time was reduced without significantly affecting image quality in the context of ability to contour. The acceleration capabilities allowed for increased image resolution under similar scanning times as well as significantly improved urethra visualization in prostate simulations.

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Introduction

It is well recognized that one of the major drawbacks of magnetic resonance imaging (MRI) is prolonged scan times, and substantial improvements in both patient comfort and health care economics can be achieved if examination times are reduced. When MRI is used for radiation therapy (RT) simulation, patients are often positioned in immobilization devices, and depending on the disease site, they may need to be simulated under breath-hold, arms above head, full bladder, or empty rectum conditions to ensure reproducible setup and minimize toxicity to normal organs. In addition to discomfort of such setup conditions, consistency between bladder and rectal filling is also desired between different MRI sequences to ensure adequate image registration and correlation for contouring, particularly in the context of MR-only prostate treatment planning. These constraints further motivate efforts to reduce scan times in the context of MRI simulation of radiation therapy.

Modern rapid MRI techniques, such as parallel imaging and compressed sensing, exploit redundancies to reduce the number of acquired k-space points and thus increase imaging speed. Parallel imaging techniques, such as SENSitivity Encoding (SENSE) and GeneRalized Autocalibrating Partial Parallel Acquisition developed in the late 1990s and early 2000s, take advantage of redundant information from multiple receiver coils to reduce the number of acquired phase-encoding steps. Compressed sensing exploits the compressibility of medical images to also reduce the number of phase-encoding steps, but in a different way than parallel imaging. In compressed sensing, k-space samples are taken on a nonuniform pattern to produce incoherent aliasing artifacts that can be removed using a nonlinear iterative reconstruction algorithm that promotes sparsity in the solution. Even though parallel imaging requires uniform k-space sampling and compressed sensing requires nonuniform k-space sampling, both techniques can be combined using the concept of joint multicoil sparsity to achieve higher acceleration factors than each technique alone. Compressed SENSE (CS) is a recent clinical implementation of the combined compressed sensing and parallel imaging approach, which uses wavelet sparsity as compression transform and the SENSE model for parallel imaging. Compressed SENSE has been demonstrated to accelerate the acquisition of 2-dimensional (2D) and 3-dimensional (3D) images in a diagnostic setting.

This work investigates the use of Compressed SENSE to enhance the workflow in MRI simulation for radiation therapy. The effect of Compressed SENSE on scan time, image quality, and ability to contour target or organ-at-risk volumes is evaluated with respect to SENSE on 37 different acquisitions performed on 24 patients.

Methods

Compressed SENSE

CS uses a variable density random k-space undersampling along the phase-encoding dimensions (ky for 2D imaging and ky-kz for 3D imaging). CS reconstruction algorithms combines SENSE parallel imaging in the imaging domain and compressed sensing with a 2D or 3D wavelet sparsity constraint. The iterative algorithm aims to solve the following optimization problem:

\[
I = \min \left( \sum_{i=1}^{\text{coils}} \| m_i - OS_i I \|_2^2 + \lambda_1 \| R^{-1/2} I \|_2^2 + \lambda_2 \| \psi I \|_1 \right).
\]

The left-hand side term enforces data consistency, the central term controls the power of the solution to limit noise amplification in SENSE reconstruction (Tikhonov regularization) and the right-hand side term enforces sparsity in the wavelet domain. \( I \) represents the image to be reconstructed, \( m_i \) is the measured k-space data for coil \( i \), \( O \) is the Fourier operator according to the sampling pattern, \( S_i \) is coil sensitivity for coil \( i \), \( R \) is a low-resolution SENSE reference image to give low-weights to pixels without signal (air), \( \psi \) is the wavelet transform, and \( \lambda_1 \) and \( \lambda_2 \) represent the regularization factors that balance data consistency with prior knowledge of image content and wavelet sparsity, respectively. The practical clinical implementation of CS software involves user input of a CS acceleration factor and selection of a preferred denoising level. The CS acceleration factor represents the ratio between the number of k-space lines of a fully acquired image and an image acquired with CS, with allowed range from 1 to 32. The denoising level parameter (options: no, weak, medium, strong, and system default) controls the sparsity regularization factor \( \lambda_2 \). Based on user input of the aforementioned 2 CS related parameters, along with the coil arrangement and geometric parameters of the planned imaging sequence at question, the optimal sampling strategy and regularization factors are automatically determined by the CS software.

Data acquisition and analysis

All scans were performed on a 3T Philips Ingenia (Philips Health care, Best, The Netherlands) with software version 5.4 or 5.6 and equipped with a flat tabletop for MRI-based simulation. The MRI examinations of 24 patients performed between September 2019 and June 2020 were analyzed retrospectively using a retrospective institutional review board protocol. A wide variety of anatomic sites, including brain, head and neck, spine, chest wall, and prostate were included in the study. Each
of these patients had undergone an MRI simulation procedure where one imaging series had been acquired twice: once using only SENSE and a second time using CS. The anatomic field of view prescription was identical for the 2 series to ensure fair comparison. The scan time was recorded for both series to evaluate overall time-savings that CS was able to achieve. A total of 37 image pairs were acquired from 24 patients, with disease site and tumor histology included in Table 1.

Table 2 contains the anatomic sites, series descriptions, and acceleration factors used for each SENSE and CS acquisitions. The selection of CS acceleration factors added to existing sequences with routinely acceptable image quality was based on the following vendor recommendations. For standard 2D sequences, recommended CS factor was 2 to 3, or 30% higher than the existing SENSE acceleration factor. For 2D fat-saturated or inversion-recovery (IR) sequences, the guideline range was 1.5 to 1.8, or 20% higher than existing SENSE factor. For all 3D sequences, the recommended CS factor range was 6 to 8, or 40% higher than current SENSE factor. A relative signal-to-noise ratio (SNR) factor generated by vendor CS software relative to the initial SENSE-only implementation was used to evaluate the adequacy of the CS acceleration factor. In our implementations, we kept the relative SNR factor above 0.85 while maintaining the same voxel resolution. The regularization parameter which generated the least denoising (option: weak) was used in all acquisitions to avoid image blurring. For each SENSE and CS pair, image quality and the ability to contour lesions were assessed by 2 radiation oncologists who routinely use MR images to delineate and contour tumor target volumes and surrounding organs-at-risk. For each image pair and evaluation category, the readers provided their opinion on which they thought was superior: CS, SENSE, or neither (in which case they stated “equivalent”). Image quality encompassed perceived SNR, sharpness, and presence or absence of artifacts.

The series used for imaging gold fiducial markers in the prostate were further assessed by contouring the markers and comparing the contoured volumes using a paired Student’s t test. Additional series using CS are discussed in Results to illustrate alternative image quality improvement strategies. Details of an optimized MRI-only prostate simulation protocol to incorporate CS acceleration and the clinical effect of resultant acceleration is shown in Table 3.

To assess the geometric integrity of scans under CS image acceleration, geometric distortion analysis was performed through experimental data acquisition on the QUASAR™ MRID3D geometric distortion phantom (Modus QA, London, ON, Canada). The phantom contains 1502 precisely machined mineral oil-filled fiducial markers uniformly distributed on the cylindrical boundary of the phantom. The fiducial-containing imaging field of view is 34.3 cm in diameter and 29.4 cm in length.

### Table 1

| Subject | Disease site         | Tumor histology                                      |
|---------|----------------------|------------------------------------------------------|
| 1       | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 2       | Prostate             | Prostate adenocarcinoma                              |
| 3       | Prostate             | Prostate adenocarcinoma                              |
| 4       | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 5       | Brain                | Endometrial adenocarcinoma                           |
| 6       | Brain                | Astrocytoma                                          |
| 7       | Prostate             | Prostate adenocarcinoma                              |
| 8       | Prostate             | Prostate adenocarcinoma                              |
| 9       | Prostate             | Prostate adenocarcinoma                              |
| 10      | Prostate             | Prostate adenocarcinoma                              |
| 11      | Spine: cervical      | Melanoma                                             |
| 12      | Spine: thoracic      | Rectal adenocarcinoma                                |
| 13      | Chest wall           | Lung adenocarcinoma                                  |
| 14      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 15      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 16      | Head and neck        | Squamous cell carcinoma of the oral cavity           |
| 17      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 18      | Spine: cervical and thoracic |                     |
| 19      | Head and neck        | Renal cell carcinoma                                 |
| 20      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 21      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| 22      | Spine: thoracic and lumbar |                       |
| 23      | Head and neck        | Liver adenocarcinoma                                 |
| 24      | Head and neck        | Squamous cell carcinoma of the oropharynx            |
| Patient no. | Image no. | Series                          | Sim examination         | SENSE | CS | Time change, % |
|------------|-----------|--------------------------------|-------------------------|-------|----|----------------|
| 1          | 1         | 2D T2-weighted fat-saturated FSE | Head and neck           | 5:21  | 2  | -33.33        |
| 2          | 2         | 3D balanced fast GRE            | Prostate                | 1:20  | 1.4| -23.75        |
| 3          | 3         | 3D balanced fast GRE            | Prostate                | 1:46  | 1.4| -24.53        |
| 4          | 4         | 3D balanced fast GRE            | Prostate                | 1:46  | 1.4| -24.53        |
| 5          | 5         | 2D T2-weighted fat-saturated FSE | Head and neck           | 4:11  | 2  | -25.11        |
| 6          | 6         | 2D T1-weighted post contrast FSE | Brain                   | 5:17  | 2  | -30.11        |
| 7          | 7         | 3D spoiled GRE                  | Brain                   | 4:01  | 2  | -24.48        |
| 8          | 8         | 3D balanced fast GRE            | Prostate                | 1:26  | 1.4| -25.58        |
| 9          | 9         | 3D balanced fast GRE            | Prostate                | 1:24  | 1.4| -26.39        |
| 10         | 10        | 3D balanced fast GRE            | Prostate                | 1:28  | 1.4| -31.82        |
| 11         | 11        | 3D balanced fast GRE            | Prostate                | 1:24  | 1.4| -30.55        |
| 12         | 12        | 2D T1-weighted FSE              | Spine                   | 2:19  | 1.7| -17.99        |
| 13         | 13        | 3D balanced fast GRE            | Spine                   | 3:31  | N/A| -31.28        |
| 14         | 14        | 2D T2-weighted FSE              | Spine                   | 2:39  | 1.2| -30.19        |
| 15         | 15        | 2D triggered T1-weighted FSE    | Chest wall              | 3:24  | 2  | -35.29        |
| 16         | 16        | 2D T2-weighted FSE              | Head and neck           | 3:43  | 2  | -25.11        |
| 17         | 17        | 2D T2-weighted FSE              | Head and neck           | 1:12  | 2  | -36.11        |
| 18         | 18        | 2D T2-weighted fat-saturated FSE | Head and neck           | 3:43  | 2  | -25.11        |
| 19         | 19        | 2D T2-weighted FSE              | Head and neck           | 1:12  | 2  | -26.39        |
| 20         | 20        | 2D T2-weighted fat-saturated FSE | Head and neck           | 4:11  | 2  | -33.47        |
| 21         | 21        | 2D T2-weighted FSE              | Head and neck           | 1:12  | 2  | -26.39        |
| 22         | 22        | 2D T2-weighted fat-saturated FSE | Head and neck           | 3:43  | 2  | -25.11        |
| 23         | 23        | 2D T2-weighted FSE              | Head and neck           | 1:09  | 2  | -25.74        |
| 24         | 24        | 2D T2-weighted fat-saturated FSE | Spine                   | 2:54  | 1.2| -29.89        |
| 25         | 25        | 2D T2-weighted fat-saturated FSE | Spine                   | 3:58  | 1.2| -29.83        |
| 26         | 26        | 2D T2-weighted fat-saturated FSE | Head and neck           | 3:43  | 2  | -25.11        |
| 27         | 27        | 2D T2-weighted fat-saturated FSE | Head and neck           | 4:11  | 2  | -33.47        |
| 28         | 28        | 2D T2-weighted FSE              | Head and neck           | 1:35  | 2  | -27.37        |
| 29         | 29        | 2D T2-weighted fat-saturated FSE | Head and neck           | 3:43  | 2  | -25.11        |
| 30         | 30        | 2D T2-weighted FSE              | Head and neck           | 1:10  | 2  | -24.29        |
| 31         | 31        | 2D T1-weighted post contrast FSE | Spine                   | 1:54  | 1.7| -17.54        |
| 32         | 32        | 2D T2-weighted fat-saturated FSE | Spine                   | 2:20  | 1.2| -20.00        |
| 33         | 33        | 2D T1-weighted post contrast FSE | Spine                   | 2:41  | 1.7| -18.63        |
| 34         | 34        | 2D T2-weighted fat-saturated FSE | Spine                   | 2:54  | 1.2| -20.11        |
| 35         | 35        | 2D T2-weighted fat-saturated FSE | Head and neck           | 4:11  | 2  | -33.47        |
| 36         | 36        | 2D T2-weighted fat-saturated FSE | Head and neck           | 3:43  | 2  | -25.11        |
| 37         | 37        | 2D T2-weighted FSE              | Head and neck           | 1:12  | 2  | -26.39        |

**Abbreviations:** 2D = 2-dimensional; 3D = 3-dimensional; CS = compressed sensing; FSE = multishot fast spin-echo; GRE = gradient echo.

Scan times are given in minute:second, and the percentage scan time change for CS relative to SENSE is reported. Two experienced radiation oncologists compared the series with regards to image quality and ability to contour tumor and organs at risk.
A 3D T1 fast-field-echo sequence with 1 mm³ isotropic voxels was used for all image acquisitions. The phase-encoding direction was set to right-left and frequency-encoding direction was anterior-posterior. The distortion field throughout the entire imaging volume was computed using the MRID3D geometric distortion analysis software. Routine large field of view distortion QA is typically performed with the quadrature body coil (QBC) alone. However, with the limited number of phased-array coil elements in the QBC, CS acceleration could not be properly assessed. To better understand the effect that CS acceleration has on geometric distortion, we acquired data with a 16-channel anterior torso coil with no CS acceleration, and with CS factors of 3, 5, and 7. The distortion field values for all acquisitions were extracted for diameter spherical volume (DSV) of 20 cm and 34 cm and compared.

**Results**

Example SENSE and CS images for head and neck, prostate, brain, and spine are shown in Figure 1. Table 2 contains a comparison of scan times for 37 SENSE and CS acquisitions, including imaging parameters, examination disease site and acceleration factors. The average CS scan time reduction with respect to SENSE for both 2D and 3D sequences was 27%. This translated to average time savings of 45 seconds among all sequences and up to 2-minute reduction for a longer sequence that was originally 5 and half minutes. Figure 2 shows the overall radiation oncologist assessment results when evaluating imaging quality and ability to contour when comparing SENSE and CS images. In Figure 2a and b, aside from the overall results from each reader, a consensus result between the 2 readers was determined for each pair of SENSE and CS images and results were summarized. Figure 2c and d demonstrate the image quality and ability to contour disease site-specific breakdown of the consensus reading results. In terms of image quality, Reader 1 found both image quality and ability to contour were overall equivalent between CS and SENSE, while Reader 2 assessed SENSE having superior image quality. The disease site breakdown shows that while head and neck and spine examinations had the majority of cases scored as equivalent, SENSE had slightly better image quality. The prostate image pairs consist of a small field of view 3D acquisition specifically designed for imaging gold seed fiducials, in which CS had superior image quality compared with SENSE. The average volume of all evaluated 18 prostate fiducial markers from 6 evaluated image pairs were 0.0283 mL and 0.0277 mL for the SENSE and CS group, respectively. Paired t test showed no significant difference for fiducial marker volumes for CS versus SENSE (P = .578).
Figure 1  Example images in the spine, prostate, brain, head and neck using compressed sensing (CS; top; combined SENSE) and SENSE (bottom; SENSE only). Imaging acceleration factors and scan times are specified in Table 2. Spine: image 12, prostate: image 8, brain: image 6, head and neck: image 5.

Figure 2  Physician image quality and ability to contour assessment results. (a) Image quality. (b) Ability to contour, of all images for each reader and consensus. (c) Image quality (d) ability to contour, by disease site. Combined compressed sensing (CS) and SENSE (CS) technique is superior, SEN: SENSE only is superior, EQ: 2 series are equivalent.
Not included in the table is a lower extremity MRI simulation, in which CS was applied to 7 2D fast spin echo series including axial upper/lower T1W and T2W, coronal T1W and T2W, and sagittal T2W. Compared with a previous lower extremity simulation, which used SENSE only, total scan time was reduced from 33 to 24 minutes using CS (27%).

Prostate simulations comprise the vast majority of our patient cohort (~80 cases/mo) and have therefore provided the most opportunity for optimization with CS. The 5 series comprising the prostate MR simulation (Table 3) demonstrate the variety of applications of CS aside from direct time savings. In series 2, 4, and 5, the acceleration provided by CS was traded for reduced scan time and increased in-plane spatial resolution. Series 1, a 3D scan which is used to contour the urethra, would normally be prescribed with phase encoding in RL and AP dimensions to limit scan time. However, phase encoding in the AP dimension results in respiratory motion artifacts which can obscure the urethra. Swapping phase-encoding to the superior-inferior dimension and oversampling to avoid aliasing resulted in a scan time of over 9 minutes with SENSE, which was unacceptable with patients. However, with a CS factor of 7.3, we were able to obtain the image in 5 minutes, 16 seconds. A demonstration of SENSE-only with AP phase encoding compared with CS for the high-resolution urethra sagittal T2 3D acquisition with SI phase encoding is shown in Figure 3. CS significantly improved urethra visualization and removed the need to use the Foley catheter for urethra visualization and contouring for most patients, reducing the overall workflow time and increasing patient comfort. Another example of using CS to increase image resolution in a head and neck mDixon examination while maintaining scan time is illustrated in Figure 4a and b, where marked improvement in clarity and resolution can be seen in the CS images (Fig. 4b). Another great example of the advantage of CS compared with SENSE was demonstrated when the acceleration factor was pushed to its limits in spine imaging with a custom flexible 32 channel array (8 in HF direction × 4 in LR direction). CS significantly outperformed SENSE for an acceleration factor of 8, presenting...
higher SNR and overall improved image quality (Fig. 4c and d). This is mainly because there were only 8 coil elements in the HF direction, which was insufficient to achieve an acceleration factor of 8 while maintaining reasonable image quality with coil-based acceleration (SENSE) alone. The CS algorithm, on the other hand, was able to automatically balance the optimal percentage of k-space sparse-sampling and coil-based acceleration to provide high quality images with an acceleration factor of 8 without problems.

Distortion phantom assessment results are shown in Figure 5, where the distortion vector distributions in x (left-right), y (anterior-posterior), z (superior-inferior), and overall magnitude are represented in boxplot format comparing standard QBC acquisition to anterior coil with no CS, and various CS acceleration factors up to 7. Figure 5a and b correspond to all distortion fields within the diameter spherical volume of 34 cm, and Figure 5c and d correspond to all distortion fields for DSV ≤ 20 cm. The results show little to no difference in distortion between all compared CS acceleration settings and standard QBC baseline values. The acquisition times (minutes:seconds) for anterior coil with no CS, CS = 3, CS = 5, and CS = 7 were 17:54, 5:35, 3:23, and 2:24, respectively.
respectively. For DSV ≤ 20 cm, the average distortion magnitude ranged from 0.38-0.43 mm between all acquisitions, with overall maximum of 0.84 mm. For DSV ≤ 34 cm, the average distortion magnitude ranged from 0.73 to 0.77 mm, with overall maximum value of 2.3 mm. A summary table containing the mean, standard deviation, and maximum distortion data for all acquisitions are included in Appendix E1, Table E1.

Discussion

CS is a productive MRI acceleration technique that leverages image compressibility, k-space undersampling, and redundancies from multiple receiver coils. Because the first demonstration of this method in cardiac perfusion MRI, it has also been shown to accelerate acquisition of 2D and 3D images, and garnered clinical adoption through vendor software implementations. In this work, we demonstrated the utility of CS in MRI simulation for radiation therapy treatment planning, an application in which an accelerated image acquisition could be even more crucial than it is in diagnostic applications, considering the added patient discomfort caused by the immobilization devices that are necessary to create a reproducible radiation therapy treatment position setup.

Our results have shown that CS permitted an average 27% reduction in MRI simulation scan time. Although the potential benefit of CS is greater in 3D series where there are 2 phase-encoding dimensions that can be exploited in acceleration, we were able to substantially improve scan times in 2D series as well. Qualitative evaluation on image quality and ability to contour performed by radiation oncologists on paired SENSE and CS images (except for the acceleration factor, which was higher in CS) demonstrated that while both techniques are...
equivalent in image quality for most of the cases, SENSE is slightly superior in terms of SNR, especially for head and neck and spine examinations. This result is expected given the higher acceleration factor in CS relative to SENSE. However, the overall qualitative evaluation on ability to contour demonstrated that the subtle image quality differences that could sometimes be noticeable between the 2 methods did not affect the ability to contour for treatment planning. CS had superior overall image quality in the evaluated prostate image series. This result suggests that the benefits of CS are better realized in 3D acquisitions (prostate) compared with 2D acquisitions (all other anatomic sites), which is expected because image compressibility is higher in 3D images compared with 2D images due to the greater sparsity of 3D k-space data (with 2 phase-encoding directions) compared with 2D k-space data (with only one phase-encoding direction). Aside from speed ups in image acquisition, we also used CS in other ways, including the increase of in-plane spatial resolution without affecting scan time and modification of phase encoding direction to reduce respiratory motion effects. The largest effect of enabling high-resolution sagittal T2w 3D without respiratory motion artifacts was the elimination of Foley catheter placement for urethra visualization and contouring in our routine workflow. This change in workflow not only indirectly reduced overall simulation workflow time significantly, but also increased overall patient comfort and reduced potential urethra displacement that could result from the Foley catheter.

The large field of view distortion phantom analysis performed with various CS acceleration factors demonstrated the geometric integrity of scans are well maintained with CS acquisition accelerations. Little to no change in the distribution of distortion values were seen, regardless of level of acceleration. This quantitative analysis result further strengthens our confidence in implementing CS in radiation therapy MR simulation, a setting in which geometric integrity is of utmost importance for tumor and organs-at-risk delineation in treatment planning.

The study has some limitations. First, while the range of disease sites and sequences evaluated were comprehensive, the patient cohort was still relatively small to establish statistical significance. In addition, quantitative image quality metrics such as SNR and contrast-to-noise ratio were not evaluated. SNR and contrast-to-noise ratio are very challenging to compute in compressed sensing MRI due to the nonlinear nature of the reconstruction algorithm. We deemed that, in the context of MRI simulation for radiation therapy, the qualitative impressions from radiation oncologists are far more important than quantitative metrics because, ultimately, the targets and surrounding organs-at-risk contours that they generate from their qualitative impressions determines the final treatment plan for each patient.

Future directions of applying CS in MRI simulation for radiation therapy include further exploiting the larger benefit in 3D imaging sequences by implementing more high-resolution CS-accelerated 3D sequences in standard simulation protocols, combining the acceleration potential with metal artifact reduction techniques, such as slice encoding for metal artifact correction18 combining this methodology with motion robust 3D radial image acquisitions19 and exploring faster quantitative MRI in clinically relevant time frames.

Conclusions

The application of a combined compressed sensing and SENSE using the Compressed SENSE (CS) implementation resulted in approximately 27% reduction of scan time compared with SENSE-only, while maintaining high image quality for MRI simulation of radiation therapy. Moreover, CS acceleration not only enabled increased image resolution in many simulation examinations without increasing scan time, but also allowed for significant improvements in high resolution urethra imaging with reduced breathing motion artifacts under reasonable scanning time frames.

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Supplementary materials

Supplementary material associated with this article can be found in the online version at doi:10.1016/j.adro.2021.100799.

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