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Hexagonal Undersampling for Faster MR Imaging near Metallic Implants

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Abstract

Purpose— Slice encoding for metal artifact correction (SEMAC) acquires a 3D image of each excited slice with view angle tilting to reduce slice and readout direction artifacts respectively, but requires additional imaging time. The purpose of this study is to provide a technique for faster imaging around metallic implants by undersampling k-space.

Methods— Assuming that areas of slice distortion are localized, hexagonal sampling can reduce imaging time by 50% compared with conventional scans. This work demonstrates this technique by comparisons of fully sampled images with undersampled images, either from simulations from fully acquired data or from data actually undersampled during acquisition, in patients and phantoms. Hexagonal sampling is also shown to be compatible with parallel imaging and partial Fourier acquisitions. Image quality was evaluated using a structural similarity index (SSIM).

Results— Images acquired with hexagonal undersampling had no visible difference in artifact suppression from fully sampled images. The SSIM index indicated high similarity to fully sampled images in all cases.

Conclusion— The study demonstrates the ability to reduce scan time by undersampling without compromising image quality.

Keywords

Metallic implants; SEMAC; Undersampling; Artifacts; Parallel Imaging

Introduction

Arthroplasty procedures are being performed at a growing rate in the United States and are projected to become more common over coming decades (1). Furthermore, such procedures often have a high revision burden, such as 8% and 17% for knee and hip replacements, respectively (2). This makes the ability to perform in vivo magnetic resonance imaging around metal implants highly desirable. However, the presence of metal can cause large distortions in the static magnetic field that lead to severe imaging artifacts in conventional

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MRI scans (3). In recent years, multi-spectral imaging (MSI) techniques have been developed for imaging near metal. These include MAVRIC (4), SEMAC (5), and related approaches MAVRIC-SL (6), and MSVAT-SPACE (7), all of which use interleaved excitation and 3D imaging. Although the clinical utility of these methods has been demonstrated in recent studies for various types of implants (8–12), the additional phase encoding leads to long imaging times.

Long scan times have led to the common use of k-space based acceleration methods in all MSI techniques. For example, partial Fourier acquisitions and parallel imaging are routinely used in SEMAC scans, often offering about a four-fold acceleration (13). Assuming a SEMAC scan uses 16 phase encodes for resolving each slice, it is still about 4 times longer than a non-phase encoded scan, often resulting in scan times of about 5–10 minutes with reduced spatial resolution as well. Techniques to further accelerate these scans are therefore desirable. Recently, compressed sensing has also been utilized in SEMAC (14) imaging, but reconstruction times remain a challenge for routine use. Phase encoding along all directions (single point imaging) has been shown to produce images of good quality under severe B_0 inhomogeneity (15,16), but suffers from very long scan times. A method to accelerate such scans for imaging near metal implants was recently published (17), but would still benefit from further acceleration.

In this work, we demonstrate a method for reducing SEMAC imaging time by up to 50% for many cases. This is achieved by undersampling k_{xy}-k_z space in a hexagonal (18,19) pattern, which is known to produce diagonally positioned image replicas in y-z image space. The shape and extent of the y-z signal in SEMAC imaging causes the aliased replicas to not interfere with the desired image. This allows for the replicas to be easily eliminated during post-processing. The method was tested on actually acquired data from a metallic implant phantom and by simulations from fully acquired data from *in vivo* scans where the field of view (FOV) extent is much larger than the extent of the implant induced distortions. The proposed undersampling scheme was also demonstrated to be compatible with partial Fourier and parallel imaging techniques.

Methods

Sampling of k-space and coverage of FOV in metal imaging

SEMAC resolves through-plane distortion caused by the presence of metal by acquiring a 3D phase encoded volume with standard ky-k_z sampling (Fig. 1a) for each distorted slice. In SEMAC, the volume is assumed to be large enough to contain the whole distorted slice profile. The phase encoding enables determination of the shape of the distorted slice. The contributions of each 3D volume to each slice are then combined using either a simple linear sum or a sum-of-squares, with the latter being used in order to reduce noise. An exemplary distorted slice is shown in Figure 1c. Under normal conditions, much of the y-z extent of the slice is undistorted, assuming the implant is small compared to the in-plane field of view. This means much of the y-z FOV of each of the 3D scan volumes is empty when ky-k_z space is fully sampled (Fig. 1a,1e), needlessly increasing scan time.

Hexagonal sampling has been shown to require fewer samples and thus shorter scan time when allowed by the imaged geometry. Using this method, k-space is sampled in a checkerboard pattern (Fig. 1b). This leads to aliasing, with the aliased replicas being packed in a similar checkerboard pattern (Fig. 1f).

The distorted slice shapes in SEMAC are ideally suited for hexagonal sampling (19). The sampling pattern gives a hexagonal region of support, with image replicas not overlapping as long as the image is limited in extent as shown in Figure 1 (20). When using hexagonal sampling, the FOV is assumed to have a slightly larger extent in z than is necessary for the fully sampled case, in order for the non-distorted segments of the replicas to not overlap with the distorted part of the desired signal. Empirical data has demonstrated this assumption to be valid. The aliases can then be zeroed out in the image domain during post-processing without affecting the desired image with a zeroing mask, as shown in Fig. 1g, before the volumes are finally combined together to form the final images. This leads to a field of view with 50% reduced area and can reduce the total scan time by up to 50%.

Clinical MRI scans usually employ acceleration methods such as partial Fourier or auto-calibrated parallel imaging in order to reduce scan time. Both techniques normally use a fully sampled central region of k-space as a reference to fill in missing data in the outer regions of k-space. In this work, the missing samples of the central region were treated as sampled by the reconstruction algorithm (such as homodyne reconstruction (21) for partial Fourier acquisitions and auto-calibrated techniques such as GRAPPA (22) or ARC (Autocalibrating Reconstruction for Cartesian Imaging) (23) for parallel imaging) (Fig 1c). The algorithm then treated the sample as having an actual value of approximately zero instead of being unsampled. The algorithm then assumes the k-space representation of the object to actually be hexagonally distributed, leading to the same aliasing pattern in image space as for a hexagonal scan without parallel imaging. The aliased replicas can then be removed as before.

The desired image need not be located at the center of y - z space as shown in Fig. 1e, as long as the reconstruction algorithm knows where to center the anti-aliasing mask. This can be based on a localizer scan or simply prior knowledge from the person operating the scanner, for example by requesting the operator to specify the approximate location (along y) of the implant at the user interface, with that location information subsequently fed to the reconstruction software.

Experiments

All studies were acquired on a 1.5 T MRI system (11Dx, GE Healthcare, Waukegan, Wisconsin, USA) except the phantom scans, which were acquired on a 3.0 T scanner (Discovery MR750, GE Healthcare). Informed consent was obtained from all subjects in accordance with the Institutional Review Board (IRB) protocol at our institution.

To demonstrate the feasibility of using hexagonal undersampling in actually acquired data, a SEMAC scan of a phantom was performed with and without undersampling with no other acceleration techniques used. The object scanned was a titanium shoulder prosthesis immersed in agar gel. The aliased copies were removed during post-processing by

multiplying with a mask such as the one shown in Fig. 1g, and the 3D scan volumes were then combined with a sum-of-squares operation to form the final images. The process was then repeated using 55% partial Fourier scanning in ky, and then using 2× parallel imaging in ky. The scan parameters are shown in Table 1.

Next, to show compatibility with partial ky scanning in vivo, the undersampling scheme was simulated from acquired data from spine scans of three subjects with metallic fusion hardware. The data was acquired with 55% partial ky sampling and no parallel imaging. Other imaging parameters are shown in Table 1. The undersampling scheme was applied to the entire ky-kz space, including the symmetrically sampled central region. Homodyne reconstruction was used to synthesize the missing outer regions of k-space, based on the undersampled central region as described. The aliased copies were then eliminated as described above and the final images created by adding up the contributions from each 3D scan volume.

Finally, we showed compatibility with partial-Fourier scanning and parallel imaging in vivo. The undersampling method was tested on three data sets with metallic knee pins and one data set with a metal hip replacement. All the scans had metal objects within the field of view and the undersampling scheme was simulated from acquired data. The data sets were acquired with 55% partial ky scanning and 2× parallel imaging acceleration in ky. Similar to the spine scans, the entire ky-kz space was undersampled for each data set and aliased copies removed after ARC and then homodyne reconstruction had been applied. The imaging parameters are shown in Table 1.

To determine the similarity of the images resulting from the undersampling scheme to the original images, a structural similarity index (SSIM) was used, which has been shown to give a good indication of image fidelity (24). The SSIM was computed using a 11×11 Gaussian kernel, while downsampling the images to 256×256. A region of interest (ROI) was drawn around the metal implant and the mean SSIM value computed for the ROI over all clinically relevant slices, to examine whether the area close to the implant was similar to the normally sampled case. The same was done for the entire anatomy, in order to see whether regions far from the metal were affected by the aliased replicas, excluding regions with inadequate signal-to-noise ratio (SNR), which was empirically estimated to be SNR less than 20. A mean SSIM value of 0.95 or above was considered to indicate the metal artifacts to be resolved equally well for practical purposes.

Results

The resulting images from undersampling the shoulder prosthesis acquisition had no visible difference in metal artifact suppression compared with the fully sampled images (Fig. 2). A subtle change in signal intensity can be seen as a vertical line one quarter of the FOV from the edges of the image. This artifact arises because the number of volumes included changes abruptly (Fig. 1g), which typically alters the SNR slightly.

The in vivo scans also have apparently identical image quality with and without the undersampling scheme simulated from the acquired data. The absolute difference images are low-amplitude and show no structure, indicating that artifacts are equally reduced in both

scans without and with undersampling (a representative spine scan and the hip scan are shown in Figs. 3–4). No aliasing effects can be seen in the undersampled images. The other 5 in vivo datasets similarly showed no difference in artifact correction from fully sampled images, again without any degradation in overall image quality.

For all experiments, we list the average, standard deviation, and range of mean SSIM values calculated over each relevant slice, as well as the average percentage of pixels that had SSIM greater than 0.95 over each slice. When calculations were done over an ROI containing the metal implant, the results were 0.994 ± 0.009 (ranging between 0.989–0.997) and 99.6% of pixels above 0.95 for the phantom scan without partial Fourier (PF) or parallel imaging (PI) applied; 0.991 ± 0.010 (0.984–0.994), 99.2% > 0.95 for the phantom scan with PF; 0.994 ± 0.008 (0.989–0.996), 99.7% > 0.95 for the phantom scan with PI; 0.988 ± 0.018 (0.952–0.999), 97.2% > 0.95 for the in vivo scans. When SSIM values were calculated over the whole phantom or anatomy, the results were 0.995 ± 0.009 (0.994–0.997), 99.3% > 0.95 for the phantom scan without PF or PI; 0.993 ± 0.008 (0.991–0.994), 99.4% > 0.95 for the phantom scan with PF; 0.994 ± 0.010 (0.993–0.996), 99.0% > 0.95 for the phantom scan with PI; 0.992 ± 0.011 (0.982–0.998), 99.1% > 0.95 for the in vivo scans. These results are also listed in Supporting Table 1.

The results demonstrate that the images obtained from hexagonal sampling are very similar to normal scans. Looking at the whole anatomy yields slightly better results than for a small ROI over the metal, demonstrating that areas far away from the metal are less affected by the metal. In vivo scans have slightly lower SSIM than phantom scans, likely due to lower SNR.

Discussion

Multi-Spectral Imaging (MSI) enables distortion-free MRI imaging near metal implants, which has been shown to be useful for diagnostic purposes (3–12). To date, most MSI resolves the distorted slice profile without making any assumptions about the spatial distribution of the distortion. Patients with metal hardware are often unable to tolerate long scan times, making rapid scanning essential. This study shows that by assuming a limited, non-rectangular region of support in the 3D volume, a significant reduction in scan time can be achieved by undersampling k-space and removing the resulting aliased copies during post-processing. The technique allows for undersampling of the central region of k-space when acceleration methods such as partial Fourier acquisition and parallel imaging are employed. Since the method effectively yields up to 50% reduction in scan time, the scanner operator has the option of maintaining the scan time, but doubling the number of phase encodes used for better resolution of the slice profile. This might prove useful when scanning at higher field strengths, such as 3T or higher. The operator also has the option of keeping the number of slice encodes and the scan time fixed, but increasing the in-plane resolution as SNR permits. The results indicate that hexagonal sampling of SEMAC scans cuts scan time without degrading artifact correction or overall image quality. Analyzing the whole anatomy instead of an ROI focused on the implant generally gave the same or better results in terms of SSIM, demonstrating that anatomy free of distortion is likely to have higher SSIM values. It should be noted that while the results of this study were all obtained

on titanium implants, this does not imply a fundamental limit of the method presented, which would work equally well on other implant types such as stainless steel or cobalt-chromium. Distortions close to and far from the metal are both proportional to the susceptibility of the metal (25). Therefore for metals with a larger susceptibility difference compared to water, such as stainless steel, the FOV would need to be increased when using SEMAC. However, the benefits of hexagonal sampling would be comparable to those when other types of metal are used.

The combination of hexagonally subsampled MSI with parallel imaging depends on numerous factors including the exact shape of the distortion as well as the number of coils and coil sensitivity geometry. Hexagonal subsampling may not offer further undersampling of the k-space regions outside the central calibration region in a 2×2 accelerated scan. Furthermore, when the method is combined with a 2×1 accelerated scan, these outer k-space regions will have the same sampling pattern as that of a 2×2 scan. The total scan time will still be decreased since the calibration region is undersampled. It should also be noted that sometimes the coil sensitivities do not allow 2×2 acceleration, while this will not affect the undersampling method presented here. Our experience has been that the combination of hexagonal undersampling with parallel imaging can offer greater flexibility for performance variations in different subjects and with different coil setups. A more thorough analysis of the g-factors that allow the undersampling method out of 2×2 parallel imaging could be done but is beyond the scope of this study.

When reconstructing the images from an undersampled calibration region, a different approach can be taken than the one used in this work (Fig. 1). The data from the calibration region can be used to reconstruct a low-resolution image, which will have aliases due to the undersampling. The aliases can be removed as for the full-resolution image. The resulting k-space data will then appear fully sampled, and can be used for reconstructing the full image from a partial Fourier or parallel imaging scan. This approach involves more computational steps and was therefore not used in this work. Another option is to leave the calibration region fully sampled and only undersample the outer k-space regions. This will improve image SNR but will also lengthen scan time.

The hexagonal subsampling method works with multiple coils and parallel imaging, and although this work focuses on its usefulness with SEMAC, it will also work with other methods such as MAVRIC-SL or MSVAT-SPACIR. The small amount of information contained in far off-resonance bins in the original MAVRIC method allows for more direct FOV reduction by tailoring the extent in the slice direction to each frequency bin (4) and has also been shown to benefit parallel imaging (26).

The method has an effect on the image SNR. Since scan time is reduced by half, the CNR of each scan volume is reduced by a factor of $\sqrt{2}$ before the volumes are combined. However, the volume areas that are removed by the masking operation would contain noise in a normal SEMAC acquisition, but now contain no signal, so the masking operation denoises parts of the image. This results in an abrupt change in SNR if a cross-shaped mask with a "hard threshold" is used, as in this paper (Fig 1g). This can appear as a vertical line in the final image, which might lead an observer unaware of this effect to mistake it for true signal.

intensity change. This artifact can be made less pronounced by using a softer threshold in the mask (or a trapezoidal mask slope in y/z , with a sloping boundary instead of a sharp one), but has a minor effect on the FOV shape. Testing this on the phantom in Figure 2 by applying the two different mask shapes and calculating the SSIM over the phantom yielded a value of 0.995 for the hard mask and 0.993 for the soft mask, with no vertical lines (supporting Fig. 1). The soft mask leads to a higher SSIM value since less noise is cut off. The resulting image will be noisier, however. In general, the SNR of SEMAC images greatly depends on the combination strategy used. To improve SNR, scan volumes are normally combined with a sum-of-squares operation, but more advanced techniques have also been applied (27).

This article has described a basic implementation of hexagonal sampling for MSI, but further improvements could be added. The location of the cross-shaped FOV could be centered in y based on signal detection (as is currently done in z). The location could also be determined by using a spectral scout scan, without readout gradients, to estimate the distortion as a function of y (28,29). The shape of the mask could similarly be determined automatically from such a prescan. This might also allow for the number of z -encodes to be set to an optimal value. The method bears similarities to other methods using controlled aliasing. The UNFOLD method (19,30) uses the same sampling scheme in k - t space. ANTHM (31) uses hexagonal sampling to better tailor the FOV to anatomical features. CALPIRNHA (32) uses a similar sampling strategy to decrease interference from aliasing and achieve a more robust reconstruction from parallel imaging.

In conclusion, we have shown that undersampling by $k_x k_z$ in a checkerboard pattern can improve the FOV/scan-time trade-off in SEMAC imaging, saving up to 50% in scan time with no additional hardware requirements and only a simple additional post-processing step. We have demonstrated compatibility with both partial k_y and parallel imaging.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1

A fully sampled k_y - k_z space (a) results in a single distorted excitation profile in the middle of the FOV (e). By undersampling the data in a checkerboard or hexagonal pattern (b), portions of the FOV, otherwise unused, contain aliased copies of the desired image (f). The aliased copies can be removed in post-processing with a masking operation (g). This can be thought of as using an effective FOV that is better tailored to the shape of the distorted slice. This can be combined with partial Fourier and parallel imaging, where the central k -space data is undersampled in a hexagonal pattern (c). The parallel imaging reconstruction algorithm (such as ARC) is made to treat the missing data points as sampled. The resulting k -space pattern will look as the central region (d), resulting in the same aliased image as before (f). If partial Fourier but no parallel imaging is used, the procedure would start at d.

Figure 2

A titanium shoulder prosthesis imaged with a basic SEMAC scan with no acceleration techniques applied (a), partial Fourier (b), and 2×1 parallel imaging (c), and their hexagonally undersampled counterparts (e-g). In each case, SSIM values were calculated over the region shown in figures e, f, and g. The values had a mean and standard deviation of 0.989 and 0.041 for figure e, 0.984 and 0.045 for figure f, and 0.989 and 0.044 for figure g. 97.6%, 96.8%, and 97.9% of pixels were above 0.95 in the ROIs in figures e, f, and g, respectively. Figure d shows a sample distorted slice from the acquisition in figure a. Figure h shows the corresponding slice when hexagonal sampling has been applied. Everything outside the dashed line is zeroed out, resulting in figure i, which looks the same as figure d.

Figure 3

An in vivo spinal image from a SEMAC scan with 55% partial Fourier sampling in k_y and elliptical sampling in k_y - k_z , resulting in a total 14% of k_y - k_z space sampled (a), and the corresponding scan with all of k_y - k_z hexagonally undersampled (b). The absolute difference is also shown (c). The arrows show spinal fixation hardware. The SSIM value over the region shown in Fig. b was calculated to be 0.995 with a standard deviation of 0.004, with 100% of the values being over 0.95. The coil geometry of such scans does not easily allow for parallel imaging, making them well-suited for hexagonal undersampling. After the removal of the aliases and the computation of the SSIM value, the images were intensity corrected to reduce signal drop-off in pixels further from the coil. Figure d shows a sample excited slice in the k_y - k_z plane, having aliased copies which do not interfere with the desired signal. Eliminating everything outside the dashed line results in figure e, which can then be used for image reconstruction. Figures f and g show axial views of the scans in figures a and b, respectively, demonstrating good similarity. The arrows again point to the spinal fixation hardware.

Figure 4

A hip scan acquired using SEMAC with 55% partial Fourier sampling in k_y , elliptical sampling in k_y - k_z , and APC 2x acceleration in k_y , resulting in a total sampling of 26% of k_y - k_z space, but otherwise conventionally sampled (a) and the corresponding scan with all of k_y - k_z undersampled (b). The absolute difference is also shown (c). The arrow shows a titanium hip implant. The SSIM value over the region shown in Fig. b was 0.997 with a standard deviation of 0.006, with 99.6% of the values being over 0.95. Figures d and e show a sample slice from hexagonal sampling, before and after zeroing, as in figure 3. Figures f and g show axial views of the scans in figures a and b, respectively, again showing good similarity. The arrows again point to the hip implant.

Table 1

Scan parameters for phantom and in vivo scans. The number of scans for each scan type is shown in parentheses in the top row. All data was acquired using sinc pulses with a bandwidth of 1.8 kHz. The column for the phantom combines the full/partial Fourier/parallel imaging scans.

	Phantom (3)	Spine (3)	Knee (3)	Hip (1)
TR [s]	3	4	1.6–3	3
TE [ms]	26	96–27	11–37	11
Pixel size [mm]	1	1.1–1.2	0.8	1.5
Field of View [cm]	26	28–30	20	38
Phase Field of View [%]	75	100	75–100	100
Number of Slices	32	20–22	24–28	30
Slice thickness [mm]	3	3	5	5
Echo Train Length	8	20	8	8
Readout bandwidth [Hz/pixel]	977	977	977	977
Resolution (frequency)	256	256	256	256
Resolution (phase)	256	224	192–256	256
Z encodes	16	16	16	16
Central region lines (in-plane phase direction)	24	24	24	24
Coil	8 channel receive head	8 channel receive linear array	8 channel T/R knee	8 channel receive cardiac array
Scan time (min:sec)	18:03/10:32/10:51	5:57–6:49	3:10–6:42	6:42
Scan time with hexagonal undersampling (min:sec)	5:39/5:27/5:27	2:59–3:25	1:50–2:21	3:21