Super Resolution of the Magnetic Resonance Imaging Reconstruction

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Abstract— **Medical images after reconstruction have some defects, and the enlargement the image to the desired magnification factor causes the defects to be appeared again and in dense forms and making the image is not suitable for transmission and displaying on the HDTV, or other high definition devices. We summarize the defects and removing it in denoising and debluring and remove aliasing and crosstalk using adaptive noise filter, also perform registration with decreasing aliasing effect, and removing the crosstalk by using 'intensity interpolation' of the MRI slices by simulation blur (Gaussian),noise(salt &pepper), aliasing of the medical image (type "shepp-logan"), and the MRI slices are originally reconstructed .the magnification MRI slices from 128 to 256, with time of reconstruction compared with time of super resolution in MRI slices.**

Keywords: MRI, super resolution, adaptive median filter, medical image, aliasing, salt and pepper.

I. INTRODUCTION

The reconstructed image has different in the model of reconstruction, medical image modalities like CT – scan and MRI scan .The MRI reconstruction steps are include variable controlling factors of the contrast MRI image, depending on the tissue proton density ,also the calibration of the dephasing process during the signals sequence produce different images like, spin echo, gradient echo image, which are used as the process of removing the decaying in the acquisition RF signals to achieve repetition as fast as possible and construct the image with fast scanning.

So the mechanics of acquisition of all the signals needed for reconstruction depends on the RF signals profiles, see Fig.1 the signal is repeated till complete signals from all the tissues of the plane will received. The MRI signal is mix of RF waves in the varieties of amplitudes, frequencies and phases and containing spatial information and the signal digitized, where a raw data written into a data matrix called k-space [1] converted to the image using Fourier Transform.The k-spaces are then sampled where some of the most popular trajectories types has used such as spiral and Cartesian trajectories. The output k-space is the raw data which is input to the Fourier transform and get the final output spatial image from frequency encoded image.The k-space is not complete and partial k-space is the input to the reconstruction algorithm to obtain the full k-space which has different calculations according to the algorithm of reconstruction used.
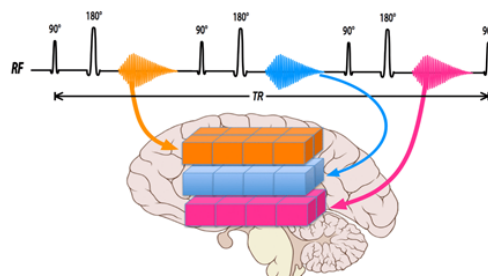

Figure 1. MRI pulses sequences

II. MRI SUPER RESOLUTION:

MRI reconstruction can be done with gridding method, where the gridding method applied to phase encode direction in the 2DFT, or in both directions in spiral imaging. The aliasing artifacts from the gridding method

avoided by putting the density of the grid higher than the highest sampling density of the actual trajectory, and the convolutional kernel must be made wide to cover the unsampled regions and to obtain an estimate of values of all grid points. But using variable density trajectories results in lower SNR in images.

The high frequency component contains little energy, under sampling them will not contribute sever aliasing artifacts, and the LF artifacts can be avoided by sampling the central k-space region sufficiently, and one can use a variable density trajectory, which sufficiently samples the central k-space region and samples the outer k-space region with lower density to achieve HR and minimal aliasing artifacts simultaneously without increasing the total scan time.[2]

The k-space origin is determined first by the actual object size to eliminate LF aliasing artifacts, the sampling density is then gradually decreased as the trajectory moves away from the origin, and if the sampling density decreases too quickly ,then the trajectory covers a greater extent in k-space and results in higher spatial resolution but the overall image is compromised by more severe aliasing artifacts[3].

III.   MEDICAL IMAGE DEFECTS:

The pre-processing of medical image reconstructed type CT – scan reconstructed image [4]and MRI image before enlargement to the desired value ,is to remove the noise and blurring of each low resolution image by treated the noise as external process out of super resolution process, we must achieve some processing of removing the noise and blurring from medical image , before any super resolution reconstruction

A.   Blurring effect

We began here with blurring on image and we have chosen the deconvolution using 'Wiener filter algorithm' with known point spread function(PSF) .We have tested that the effect of 'deconwnr' in the case of Pre-processing super resolution is not the same in the case of post- processing , comparing the result on Shepp-Logan (size 512)[5] ,which has used in medical images analyses, by down sampled to (256) ,with anti-aliased filter on the image before down sampling to generate the low resolution image ,and Gaussian blurring function are convolved with LR image , the size of PSF is (16) and variance of (2),where the results stated that the super resolution after debluring is more than in PSNR that the debluring of super resolution image as in the figures below:

Figure 2. Deblurring on image(a)blurring with Gaussian function(PSF)(b) deblurring then super resolution (c) super resolution then deblurring (worst case)

B.   Noise effect

The noise in the LR images which is the important side in image, must be treated before the interpolation which the interpolation generate more noisy sample in the output image if it is not cleared exactly before that ,so the median filter which has used as filter for remove noise with edge preserving ,the noise type   ('Salt &Pepper') noise ,and the median filter size range between the [3~9,3~9] in odd values but not robust to the high Pa and Pb (noise density of salt and pepper noise). Where the built in function in matlab 2017b, has used for 0.2 or less, so adaptive median filter[6] has tested it with the various noise density from 0.1 to 0.5 and it gives more fine result than the median filter size [3 3] which gives the closet result to the adaptive median filter and the noise density and denoising using adaptive median filter are listed in the figures below. If we repeat the adaptive median filter from 1~2 iteration, the output noise density will be improved and it is useful in a case of 0.5 noise density. The PSNR between 6~14, due to the noisy image has been taken as the reference image see Figures 3, 4, 5.
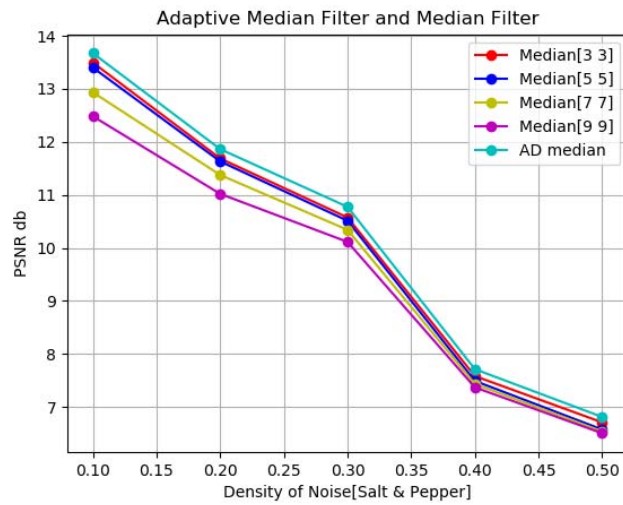
Figure 3. Comparison adaptive median (AD.M) filter with classical median filter in PSNR
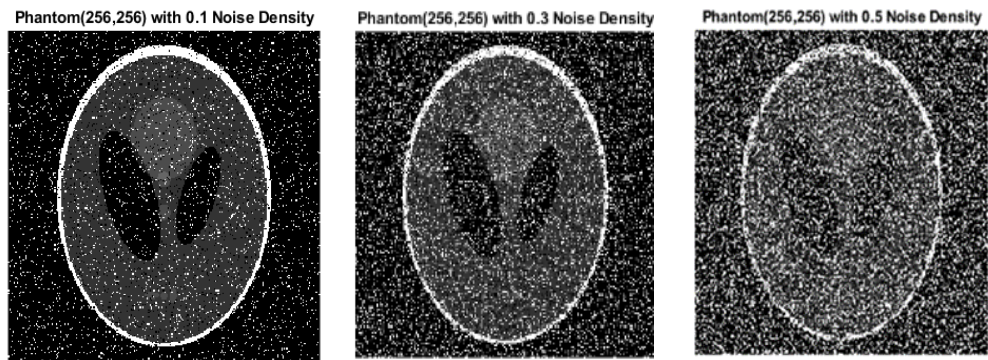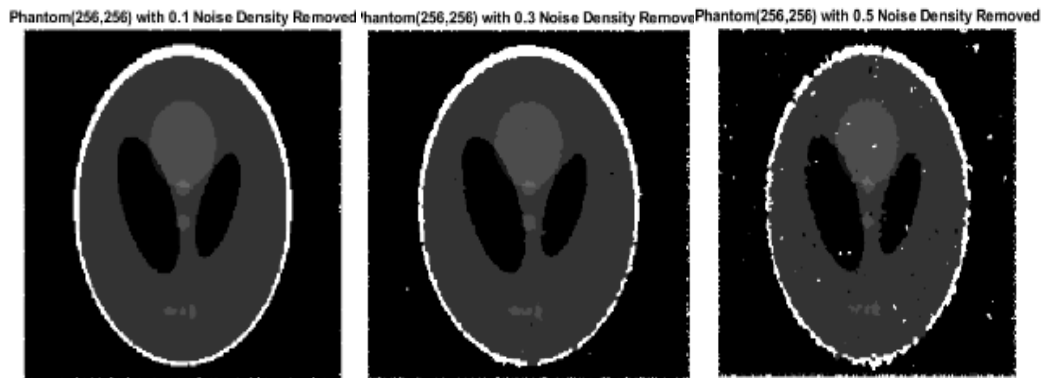

Figure 4. Noisy images with (a)0.1(b)0.3(c)0.5

Figure 5. Adaptive median filtering (a) 0.1(b) 0.3(c) 0.5

C. Aliasing effect:

The scanning process introduce different image in aliasing or shifting and multi modal image must be stacking to increase the signal to noise ratio for faint objects. The computer performs transformations on one image to make major features align with a second or multiple images The registration is process for converting multi low resolution sub pixel shifted images(or aliased image) to single image ,because it gives the information in all images into single image for analysis and diagnostics, with enhanced results, depending on the percentage

of edge pixels in the image, also the problem of fine registration depends on the aliased of the image, more aliasing means poor registration and the antialiasing filter to remove the aliasing, but causing increasing the blurred of the image which means fine registration with low resolution[7]
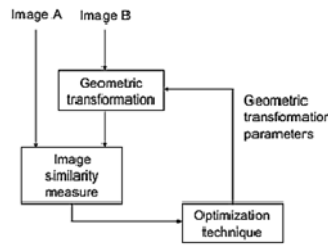

Figure 6. Image registration

Figure 6 clarify the general registration process, where are the transformation types has used to give the accurate aligning image with the other image comes from different times or different modality, we take here the transformation, as below:

1- Rigid transformations: the rigid transformations include (rotations, translations, reflections, or their combination). Any object will keep the same shape and size after a proper rigid transformation, and all rigid transformations are examples of affine

2- Affine transformations: the affine transformations in 2D can be represented as linear transformations in 3D.

And the following algorithms has used to perform the registration process here:

- Phase correlation method

- Intensity registration method

- Non - rigid registration method

Using the reconstructed medical image 'Shepp -Logan' reconstructed from the Filtered Back Projection algorithm[4],where LR images are generated by shifted it with subpixel shift ,We have dealt with three types of transformations and register them with the result in the following cases:

Case 1

 Aliasing free image: using 'anti-aliased filter' kernel = [1 4 16 4 1]/16, before down sampling process, in which the aliasing effect the fine registration for the following transformations types:

TABLE I.   CASE 1

Transformation type and other process	SSIM	Time	Registration method
anti-aliased ,rotation ,down sampling	0.984	1.16	phase correlation
anti-aliased ,translation , down sampling	0.988	0.23	phase correlation
anti-aliased ,non-rigid, down sampling	0.951	2.70	intensity registration

Case 2:

Aliasing image: we simulate the aliasing image using 'Bicubic interpolation' by down sampled the image only, and using the registration approaches above to fit the best alignment, there is no fine registration appear:

TABLE II.   CASE 2

Transformation type and other process	SSIM	Time	Registration method
Rotation , down sampling	0.842	1.20	Rigid registration
Translation , down Sampling	0.856	0.36	Intensity registration
Non-rigid , down sampling	0.462	0.69	Intensity registration

Case 3:

The filter kernel above has used, to remove the aliased signals after down sampling, which aliasing produce less quality registration.

TABLE III.　　CASE 3

Transformation type and other process	SSIM	Time	Registration method
Rotation, down sampling ,filter	0.996	1.11	Non-rigid
Translation ,down Sampling , filter	0.992	1.18	Non -rigid
Non rigid, down Sampling, filter	0.417	0.23	Intensity registration

We infer from results above the registration with the non-rigid transformation is lowest accuracy than other transformation ,the non-aliased images using antialiasing filter are registered in more quality than other aliased image, the aliasing has removed before the down sampled image using 'anti-aliased filter' , the result are more accurate in registration ,but it is more blurred in super resolution. Where the SSIM results stated that the registration was improved in the case 3, raw 1and 2, but the anti-aliased filter before down sampling will remove more information from the image rather than using it after the down sampling process.

A. Crosstalk effects:

The resolution of MRI slices is not homogenous, in which the high resolution of medical image reconstruction is in the in-plane and good or worse slices reconstruction .The number of slices mostly from 28 to128 slices for slice thickness approximately 5 mm to 0.5 mm, where the thickness alter the signal to noise ratio for each slice . The spaces between the slices called 'gap' which is percent value of the slice thickness.[8]
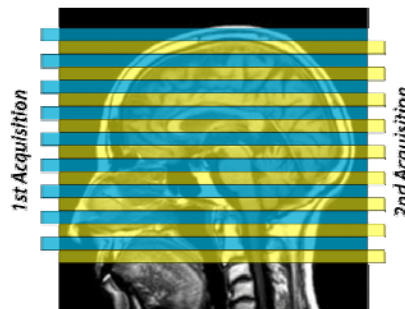

Figure 7. Slices reconstruction in two acquisitions to avoid crosstalk

To make the interpolation contribute in the reconstruction of the interslices, the time of reconstruction is depending on the interpolation type, and the slices are obtained from enter "load mri" in matlab 2017b in the command prompt, which is 28 slices in the slice select direction (z-direction).

 The Cross-Talk between slices affect the contrast of image slice reconstruction , and this avoided by setting the gap between slices [9] ,the gap is used to prevent the excitation of the contiguous slices during the excitation of one slice .The Cross-Talk increased in the case of small gab scanning and leads to more artifacts on the image, so the scanning must be done on the entire volume, and another slices reconstructed without additional scanning.

In the following table we have noted the super resolution does not construct different slices properties in the correlation values which is not altered less than 0.8 and the other result gives steady state values .The reconstructed slices with the original slices illustrated where the even slices interpolated from odd slices and the erre between the interpolated and reconstructed is as in the table IV:

TABLE IV.   SLICES RECONSTRUCTION

SLICES	SSIM	CORR	MSE	PSNR db	PSNR
2	0.76187	0.886338	0.003805	31.86159	24.1953
4	0.72836	0.843103	0.004361	31.61388	23.6034
6	0.75293	0.873989	0.003697	31.91333	24.3208
8	0.77235	0.890452	0.003116	32.21416	25.0636
10	0.81105	0.931331	0.001917	33.02215	27.1728
12	0.86434	0.960978	0.001106	33.8648	29.5617
14	0.88315	0.967343	0.00095	34.08536	30.221
16	0.86446	0.950929	0.001373	33.54118	28.6204
18	0.82377	0.923035	0.002236	32.77271	26.5034
20	0.8085	0.910674	0.002583	32.5337	25.8774
22	0.75511	0.875022	0.003512	32.00474	24.5442
24	0.75608	0.859996	0.00361	31.95569	24.4241
26	0.77635	0.841439	0.003843	31.84397	24.1527

IV. RECONSTRUCTION SLICES OF MRI

The scanner MRI image which has used to scan more than one region in the form of slices for each small region and the reconstructed slice is 2-Dimenstion image but the entire scan is 3-D image contain several discrete slices ,according to the scanning machine . The entire slices have low resolution through the plane of images but high resolution in the in-plane images. The in-plane slices have slice thickness in mm, large thickness meaning more SNR but low resolution through the plane, so to convert the entire discrete data to continuous data the interslice interpolation must be uses to accommodate un homogeneous scanning in the case of no Cross –Talk and more SNR of the original reconstruction.

The relationship between slice thickness p (inversely to the number of slices), SNR, and acquisition time, Tacq is given by[10]

$$SNR \propto p\sqrt{T_{acq}}$$ (1)

The time of reconstructed additional slices depends on the interpolation factor and the algorithm used, but the time of original slice reconstruction depend on the scanning image types which either T1-weighted image or T2 weighted image.

TABLE V.   MOST COMMON MRI SEQUENCES AND THEIR APPROXIMATE TR AND TE TIMES.

	TR(m sec)	TE(m sec)
T1-Weighted(short TR and TE)	500	14
T2-Weighted(long TR and TE)	4000	90

We can be noted the maximum time required to acquire high contrast image reach to 4sec+0.09 for one TR, TE sequence and this is repeated several times to complete one scan of slice .The gap between slices is percent value of the slice thickness, which is one of the scanning parameter used to decrease the slice Cross -Talk which affect on the image contrast. The slice thickness in MRI determined by what is termed the slice-selection pulse profile, which has in turn determined by hardware limitations coupled with pulse sequence timing considerations.

The challenge for SR in MRI is to increase the resolution in the slice-select dimension to achieve HR, isotropic, 3-D scanning images. A further challenge of scanning is to achieve the HR outcome without decreasing the SNR, which has achieved by interpolation process.

Each slice selection profile and gab between slices and thickness of slice must be calibrated in the scanning process to reduce the effect of the external signals on the reconstructed image, but we perform increasing in the slices also increasing in the pixels for using for the MRI slices of "load mri" command, and the slices dimension is 128*128*27, in plane resolution 128*128, and through plane equal of 27 slices.

The number of slices has increased by using interpolation kernels to 54 slices and interpolating it to 512*512 as in the figures below by where the larger image has compressed to display and the small image is the original slice view:
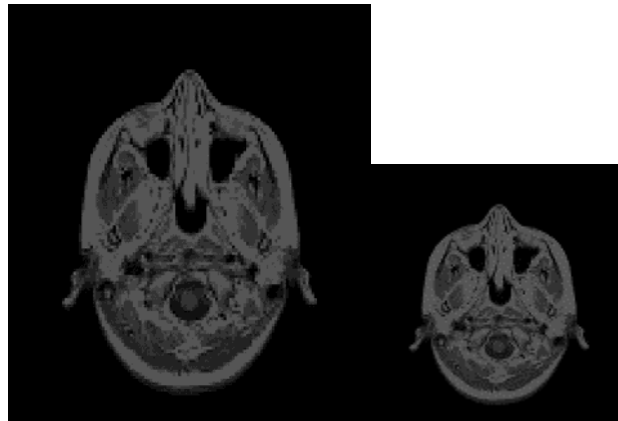
Figure 8. Interpolation MRI slices to from 128 to 512

The films must be prepared to accommodate the 54 slices rather than 27 slices, according to the displayed high definition image, the reconstructed slices and interpolated slices. The expanding time of the 27 slices to 54 slices reach to (98.45 sec) which is very important parameter for larger than 2 magnification factor, the magnification factor here is 4 (from 128 to 512), the time of interpolation using 'Bicubic' between slices achieved in (0.063418 sec.),while for 'nearest' (0.052631 sec.).Then the enlargement to the (512, 512, 54) require 0.38 sec using 'Bicubic', and 0.26 for nearest. The generated slices of size (512*512*54) has no reference image to record its quality of PSNR values, so that the (niqe) tool has used to give the all slices reconstruction quality measurement.
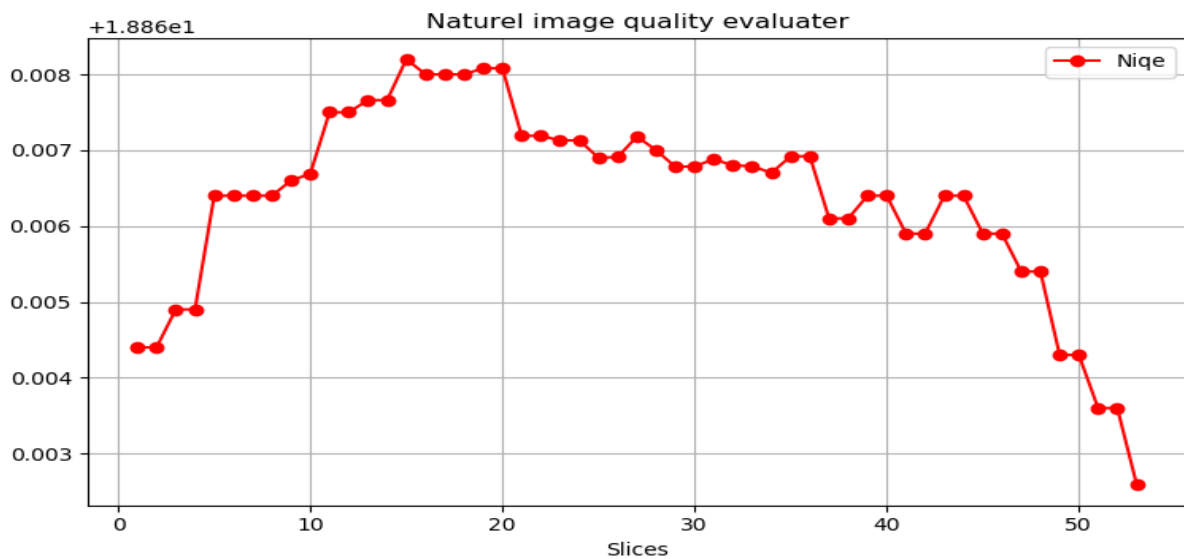

Figure 9. Natural image quality evaluator for 54 slices size (512*512)

The measurement values ranges from (18.868 to 18.863).

V. QUALITY MEASUREMENT:

1- The MSE(**M**ean **S**equare **E**rror)of the image is defined as:

$$MSE = \frac{1}{mn}\sum_{i=0}^{m-1}\sum_{j=0}^{n-1}[I(i,j) - J(i,j)]^2 \quad (2)$$

2- The PSNR(**P**eak **S**ignal to **N**oise **R**atio)of the image is also defined as:

$$PSNR = 10 * \log_{10}(\frac{MAX_I^2}{MSE}) \qquad (3)$$

Where MAX_I : is the maximum possible value of the image and for B=8-bit image the maximum

possible value is : $2^B - 1$,which is equal 255

3- The SSIM(**S**tructure **S**imilarity Measurement) of the image is calculated on various windows of x,y

size N*N image defined as:

$$SSIM(x, y) = \frac{(2\mu_x\mu_y + c_1)(2\sigma_{xy} + c_2)}{(\mu_x^2 + \mu_y^2 + c_1)(\sigma_x^2 + \sigma_y^2 + c_2)} \qquad (4)$$

Where

μ_x : the average of x, μ_y : the average of y , σ_x^2 :the variance of x , σ_y^2 the variance of y

$$\sigma_x^2 = E\left[(x - E(x))^2\right] = E(X^2) - E^2(X) ,$$

σ_{xy} the covariance between x and y , $c_1 = (k_1 L)^2, c_2 = (k_2 L)^2$

L :is the dynamic range of the pixel values, k_1 =0.01, k_2 =0.03

4- The CORR.(**Corr**elation between images)is defined in the following :

$$CORR(x, y) = \frac{cov(x,y)}{\sigma_x\sigma_y} = \frac{E[(x - \mu_x)(y - \mu_y)]}{\sigma_x\sigma_y} \qquad (5)$$

VI.   CONCLUSIONS:

The above defects (noise, blur, aliasing, crosstalk, subpixel shift) effect the super resolution of the image, the enlargement to the desired will distort the reconstructed image ,the reconstruction of MRI output with different calibration ,will generate these effects on the medical image, and any enlargement will decrease the super resolution values .The noise density with more than 0.5 will remove it using adaptive median filter, and the aliasing of the image and crosstalk which removed using filtering and reconstruction additional slices.

REFRENCES

[1] G. L. Zeng, *Medical image reconstruction: A conceptual tutorial*. 2010.
[2] D. W. McRobbie, E. A. Moore, M. J. Graves, and M. R. Prince, *MRI from picture to proton*. 2006.
[3] C. M. Tsai and D. G. Nishimura, "Reduced aliasing artifacts using variable-density k-space sampling trajectories," *Magn. Reson. Med.*, vol. 43, no. 3, pp. 452–458, 2000.
[4] M. T. Al Hussani, M. H. Ali Al-Hayani, "The Use of Filtered Back projection Algorithm for Reconstruction of tomographic Image," vol. 17, no. 2, pp. 151–156, 2014.
[5] L. A. Shepp and B. F. Logan, "The Fourier reconstruction of a head section," *IEEE Transactions on Nuclear Science*, vol. 21, no. 3. pp. 21–43, 1974.
[6] R. C. Gonzales and R. E. Wood, *Digital Image Processing*. 2005.
[7] D. L. Hill, P. G. Batchelor, M. Holden, and D. J. Hawkes, "Medical image registration.," *Phys. Med. Biol.*, vol. 46, no. 3, pp. R1–R45, 2001.
[8] D. Marinier-Sappey and A. Briguet, *Medical Imaging Based on Magnetic Fields and Ultrasounds*. 2014.
[9] A. Simmons, P. S. Tofts, G. J. Barker, and S. R. Arridge, "Sources of intensity nonuniformity in spin echo images at 1.5 T," *Magn. Reson. Med.*, vol. 32, no. 1, pp. 121–128, 1994.
[10] C. Tsai and D. G. Nishimura, "Reduced aliasing artifacts using variable-density k-space sampling trajectories," *Magn. Reson. Med.*, vol. 43, no. 3, pp. 452–458, 2000.