

Bioimaging: Principles of Spatial Encoding in MRI

(group 1)

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Contents

<i>Introduction.....</i>	<i>1</i>
General principle	1
Equations	1
<i>Experiments.....</i>	<i>2</i>
Introduction to experiment	2
Results.....	2
<i>Discussion.....</i>	<i>7</i>
<i>Appendix</i>	<i>8</i>
<i>References</i>	<i>9</i>

Introduction

General principle

Magnetic resonance imaging (MRI) is a non-invasive imaging technique that utilises a magnetic field and radiofrequency (RF) pulses to generate high-quality images. In contrast to other imaging modalities, such as computed tomography (CT) or positron emission tomography (PET), magnetic resonance imaging (MRI) does not involve the use of ionising radiation, thereby enhancing patient safety and facilitating repeated imaging. In contrast, MRI depends on the magnetic properties of nuclei within the body, particularly those of hydrogen atoms, due to their prevalence in water and fat [1]. The MRI system is comprised of a superconducting magnet, gradient coils, and RF coils [2].

The process of MRI imaging begins with the creation of a strong magnetic field by the magnet, which aligns the protons in hydrogen atoms either parallel or antiparallel to the magnetic field, B_0 . Subsequently, an RF pulse at the Larmor frequency is applied, resulting in a shift of the magnetization vector from the longitudinal axis (parallel to the magnetic field) into the transverse plane. This causes the protons to resonate. Once the RF pulse is terminated, the protons undergo a relaxation process. There are two principal types of relaxation: T1, which describes the recovery of the longitudinal magnetization vector as it returns to alignment with the z-axis (the direction of B_0), and T2, which describes the decay of transverse magnetization due to dephasing interactions. Both phenomena are described by the Bloch equations [3].

Equations

ImageJ was used for analysis of phantom images. Signal to noise ratio (SNR) is calculated as:

$$SNR = \frac{\mu_{foreground}}{\sigma_{background}} \quad (1)$$

where μ is the average signal of the object, and σ is the standard deviation of the background.

Measurement time, T, of one slice is described by an equation below

$$T = TR \times N_{phase} \quad (2)$$

where TR is the repetition time and N_{phase} is the number of phase encoding steps, i.e. a number of the rows in our matrix.

ITK-Snap was used for determination of the chemical shift length. We can calculate the number of pixel shift per bandwidth as:

$$\text{Pixel shift} = \frac{\text{Chemical water - fat shift (Hz)}}{\text{Bandwidth (Hz/px)}} \quad (3)$$

The size of the pixel in the image is determined by the equation:

$$\text{pixel size} = \frac{FOV}{\text{matrix size}} \quad (4)$$

The volume of the tissue, V, in the image is calculated as:

$$V = N_x \times N_y \times ST \quad (5)$$

where N_x is the pixel width, N_y is the pixel length and ST is the slice thickness.

Experiments

Introduction to experiment

3T Siemens MRI machine was used to obtain all the images. The practical consisted of two steps: in-vivo and in-vitro imaging.

In-vitro imaging (phantom)

A head coil had been placed in the MRI machine patient table and a phantom filled with water was placed in it to acquire the images. The height of the bed was adjusted and moved towards the iso-center of the magnet for image acquisition.

In MR console the new patient had been created. Since it was phantom the parameters had been randomly chosen as height to be 155 cm, weight to be 30 kg, and gender as "Other". We set the parameters of the setting to be as described in the protocol, and for each acquisition we changed one parameter to see the effect of each of parameters on the quality of the image.

In-vivo imaging (abdomen)

A volunteer was lied in the MRI machine's patient table and steps described in the manual were followed. The hold of breath and free breathing images were acquired. The goal was to see possible artifacts in abdominal MRI, such as motion artifact or chemical shift.

Results

Task 1.1:

The SNR for each image was calculated using equation 1 and shown in the table below

File	Mean	Std of background	SNR
Exp1_RefAcq	371.543	6.543	56.836
Exp1_SliceThickness_4mm	372.025	3.930	94.663
Exp1_SliceThickness_8mm	371.594	2.000	185.797
Exp1_MatrixSize_64x64	373.700	3.101	120.510
Exp1_MatrixSize_320x320	372.851	13.753	27.111
Exp1_NumberAve_4	372.561	3.365	110.716
Exp1_NumberAve_8	372.619	2.414	154.357
Exp1_BodyCoil	605.679	68.108	8.893

Table 1. Mean, standard deviation of the background and signal to noise ratio calculated for each observation.

Task 1.2:

Based on the results presented in Table 1, we can conclude the following effects of the parameters on SNR:

- Slice Thickness: increasing the slice thickness leads to a decrease in the standard deviation of the background noise while the signal remains relatively constant, resulting in an increase in SNR. This occurs because a thicker slice averages a

larger area, leading to more signal being captured due to the presence of more protons in that volume. As a result, the signal becomes more prominent relative to the noise.

- **Matrix Size:** decreasing the matrix size results in a less noisy image and an increase in SNR. Conversely, increasing the matrix size decreases SNR. When the field of view (FOV) is kept constant, changing the matrix size modifies the voxel size. Smaller voxels capture less proton signal, reducing the detected signal per voxel and thereby lowering the SNR.
- **Number of averages:** increasing the number of averages enhances the SNR of the phantom image. With each acquisition, the number of excitations increases, averaging both signal and noise. While the signal from the phantom remains consistent, the random noise decreases due to averaging, resulting in an increase in SNR.
- **Body coil:** Using body coil to detect signal from phantom decreased the SNR tremendously compared to the acquisition with the brain coil. This is because body coils generally have lower sensitivity and capture more background noise than specialized coils like the brain coil.

Task 1.3:

The phantom appears noisier in the images acquired with the FOV of both 140x140 and 100x100 matrix compared to RefAcq image. The reason is that pixel size is defined as in equation 2. Keeping the matrix size same as before and decreasing the FOV, we are increasing the size of the pixels and decreasing their number that were used to represent an image. Thus, we are adding more signal into same pixel than we did before. This decrease leads to partial volume effect, which is visible as distortions, such as lines appearing more prominent in the 140x140 FOV compared to the RefAcq image. Additionally, in the image acquired with FOV of 100x100, we weren't able to capture the whole object as the edges were outside of selected FOV (Fig. 1).

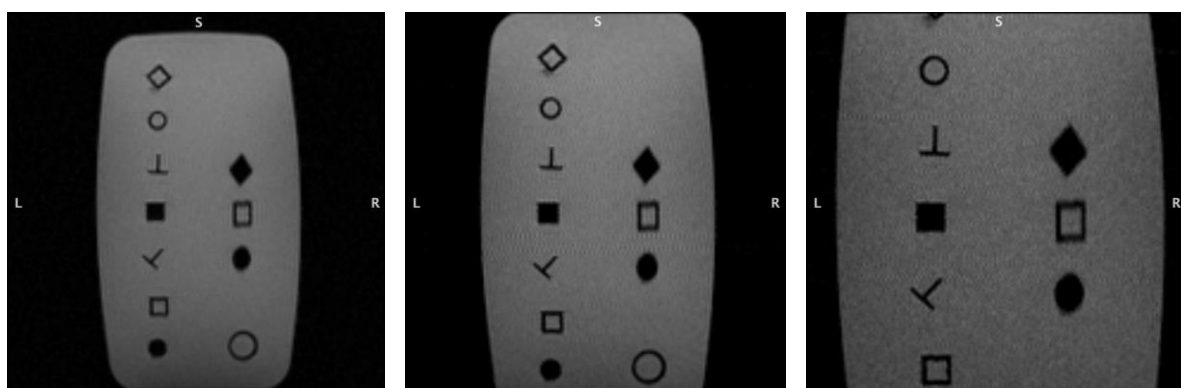


Figure 1. *Left:* The original reference image that we acquired. An acquisition with FOV of 140x140 (*middle*), and 100x100 (*right*).

Task 1.4:

The image acquired using the built-in body coil of the MRI system appears noisier than the RefAcq image acquired with the head coil. This reason is that body coil absorbs and transmits most of the generated RF energy, but the specific absorption rate (SAR) of

the body coil is limited to a few watts per kilogram (W/kg). In contrast, the head coil typically operates at a higher SAR, resulting in a less noisy image. Additionally, because the distance between the phantom and the body coil is relatively large, signals from surrounding air and nearby sources are also detected, which increases noise. Consequently, the ratio of the MR signal from the phantom to thermal noise is relatively low. Furthermore, the head coil has a greater number of coils and channels compared to the body coil, which contributes to better SNR.

Task 1.5:

In “Exp1_PerpExcitation_2/5/10/20mm”, we increased the number of slabs from 1 to 2, resulting in the acquisition being obtained in two separate sections. However, in all four different slice thicknesses, an artifact is detected. This artifact is likely due to incomplete magnetization recovery at the border of each slab, as each slab is excited separately. This can lead to a signal void at the slab boundaries. The larger the slice thickness, the larger the area of the signal void.

To determine the slice thickness, we drew a straight line through the image, plotted a light profile, and extracted the data table containing all values (Table 2). To determine the lengths of segments a and b (Fig. 2), we set the upper limit to 260 and the lower limit to 120. Based on this, we calculated a = 1 mm and b = 17 mm. The slice thickness is then calculated as follows:

$$\text{slice thickness} = \frac{17 + 1}{2} = 9 \text{ mm}$$

Calculated slice thickness is 1 mm lower than the expected thickness. This could be due to imprecise determination of the length or the partial volume effect.

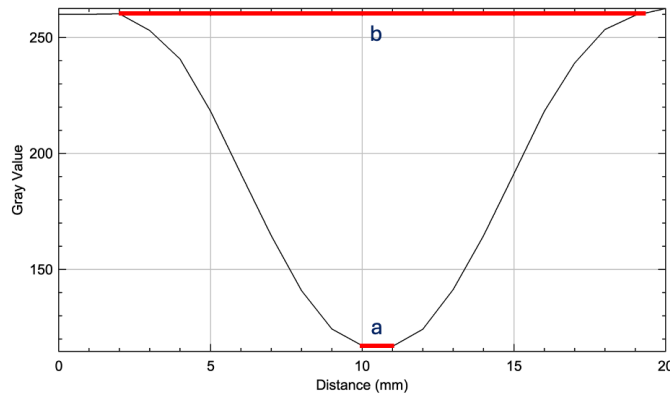


Figure 2. The light profile of the drawn line over the artefact.

0	260	11	116.839
1	260.044	12	124.219
2	260.212	13	141.349
3	253.037	14	164.428
4	240.754	15	191.348
5	218.221	16	218.217
6	191.231	17	239.03
7	164.605	18	253.444
8	140.87	19	259.568
9	124.35	20	262.497
10	117.12		

Table 2. Values of the light profile

Task 1.6:

Given that the TR time is 20 ms and that N_{phase} corresponds to the number of rows in the matrix (160), the measurement time (T) for one slice can be calculated using equation 2:

$$T = 20 \text{ ms} \times 160 = 3200 \text{ ms} = 3.2 \text{ s}$$

The measurement time depends only on the echo time and the number of phase encoding steps, and therefore independent of the slice thickness. However, the slice thickness affects the SNR of the image, and to improve this, one may increase the TR and the N_{phase} , which will affect the measurement time.

Task 2.1:

In “Exp2_FB_AP”, the motion artifact is more prominent than in “Exp2_GRE_2D” (Fig. 3). The artifact is more noticeable in the anterior-posterior (A-P) direction due to respiratory motion.

In “Exp2_FB_LR,” the imaging artifact is more prominent than in “Exp2_FB_AP” (Fig. 3). While the artifact observed in “Exp2_FB_AP” is primarily caused by respiratory motion, “Exp2_FB_LR” presents an additional issue related to phase encoding. Specifically, the field of view (FOV) did not completely cover the subject, resulting in several additional artifacts apart from the respiratory motion artifact:

- wrap-around artifact (aliasing), where we can see part of the arms are projected into the area between body and the arm in the FOV, causing incorrect representation of the anatomical structure of the arm.
- Loss of signal and resolution, due to the doubling of the anatomical area toward the edges of the FOV.

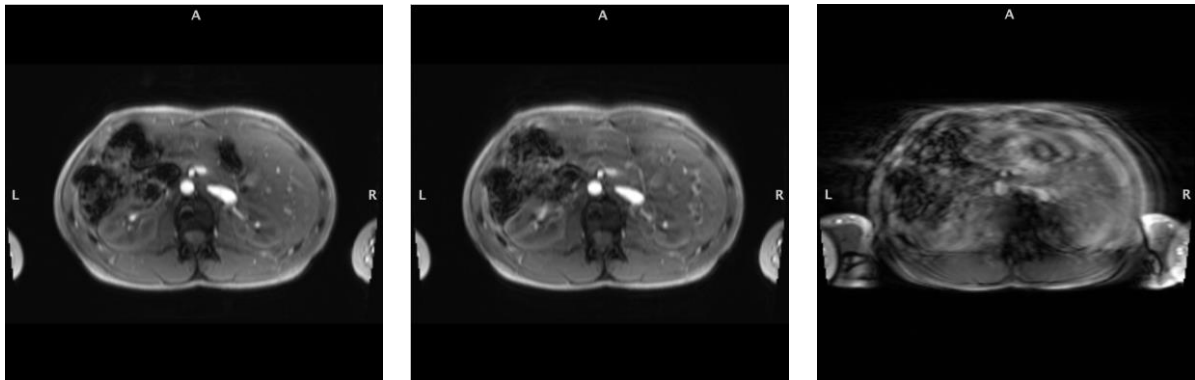


Figure 3. *Left:* an image of Exp2_RefAcq. *Middle:* an image of Exp2_FB_AP. *Right:* an image of Exp2_FB_LR.

Task 2.2:

In the image acquired with low bandwidth, we can observe the chemical shift artifact of adipose tissue, which appears on the left and shifts towards the right side. However, this artifact is not seen in the image acquired with high bandwidth (Fig. 4).

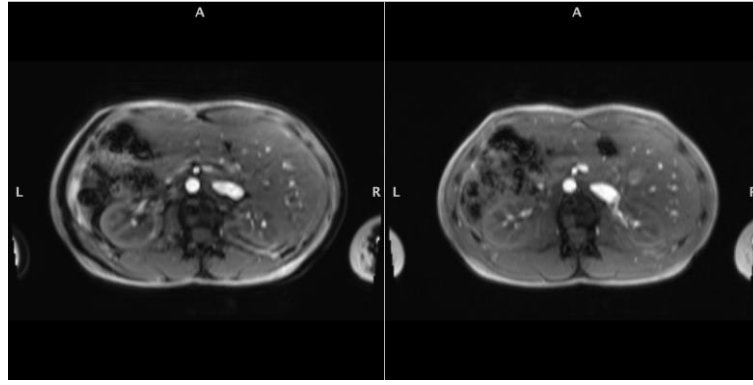


Figure 4. An MRI image of an abdominal area acquired with *left*: low bandwidth, and *right*: high bandwidth

Task 2.3:

The expected chemical water-fat shift at 3 T is:

$$150 \frac{Hz}{T} \times 3 T = 450 Hz$$

For each bandwidth we calculate the number of pixel shift using equation 3 as shown in the table below.

Bandwidth	Pixel shift
60 Hz/px	$\frac{450 Hz}{60 Hz/px} = 7.5 px$
545 Hz/px	$\frac{450 Hz}{545 Hz/px} = 0.83 px$
1950 Hz/px	$\frac{450 Hz}{1950 Hz/px} = 0.23 px$

The water-fat shift in the “Exp2_BH_LowBW” image is 7.677 mm towards the left (Fig. 5). Given that the pixel size is 1.367 mm, the shift corresponds to 5.62 pixels, which is much higher than the expected 0.83 pixels. This discrepancy could be due to motion artifacts.

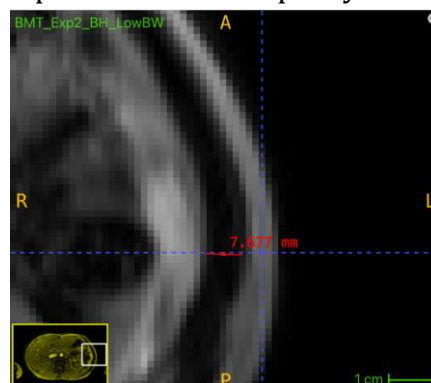


Figure 5. The chemical shift of fat-water.

Task 3.1:

To calculate the volume of tissue contributing to the image, we need to first calculate the size of the pixel. This can be done by equation 4:

$$\begin{aligned} \text{x-direction} \quad & \frac{400}{256} = 1.56 \text{ mm} \\ \text{y-direction} \quad & \frac{300}{128} = 2.34 \text{ mm} \end{aligned}$$

Thus, the pixel size is 1.56 x 2.34 mm. Since the slice thickness is 4 mm, we can calculate volume using equation 5:

$$V = 1.56 \times 2.34 \times 4 = 14.60 \text{ mm}^3$$

Task 3.2:

The rang of T1 time for the below mentioned areas at 3T is [3]:

Adipose tissue: 253 – 450 ms

Water containing tissue: ~ 898 – 1815 ms

Gray matter: 968 ± 85 – 1815 ± 170 ms

Muscle: 898 ± 33 – 1420 ± 38 ms

Free water in body:

CSF: 3817 ± 424 – 6873* ms

*values are displayed as mode of the population.

The T1 time depends on the imaging method and sequence used [3].

Task 3.3:

The order to achieve slice selective excitation of magnetization is as follows:

1. Slice selective gradient is turned on and a truncated *sinc* shaped RF pulse is applied to excite the H atoms within the selected slice. The slice thickness is determined by the RF pulse bandwidth. This creates different Larmor frequencies across different regions within the selected slice.
2. The gradient is then decreased and inverted for some time to allow the rephasing of the transverse magnetization. This allows maximization of the coherence of the transverse magnetization, enhancing the signal.

Task 3.4:

The center of k-space typically contains low spatial frequencies and appears brightest in the k-space image. These low frequencies are responsible for the contrast in the final image.

The peripheral regions of the k-space contain higher frequencies and appear darker than the center of k-space image. These higher frequencies contribute to the sharpness, edges, and definition of the image, helping to avoid smearing or partial volume effects in the final MRI image after Fourier transformation.

Discussion

Using a phantom—a box filled with objects containing water—we tested the signal-to-noise ratios (SNRs) under different scanning parameters. For images with the same matrix size, we kept the region of interest (ROI) in the phantom and the background consistent (Fig. A1).

From Table 1, we observe that while using the head coil for signal acquisition, the mean signal of the phantom remained relatively constant (mean = 372.41 ± 0.76). However, the standard deviation of the background noise varied based on the chosen scan parameters. Notably, when the slice thickness was doubled, the background noise decreased by half. However, we observed a smudging effect on the right side of the image. This is because a thicker slice averages a larger area, leading to increased signal detection, but smaller details can be lost or transferred to adjacent slices (Fig. A2). Furthermore, increasing the matrix size while keeping all other parameters constant resulted in an increase in background noise, while the signal remained largely unchanged. Since the field of view (FOV) was kept constant, increasing the matrix size reduced the voxel size, which decreased the amount of signal acquired per voxel. As a result, the noise became more prominent, and the SNR decreased by over 4 times. Conversely, decreasing the matrix size improved the standard deviation of the background by a factor of two, effectively doubling the SNR. However, this came at the cost of significant resolution loss in the phantom image, which improved with a larger matrix (Fig. A3). This is because smaller voxel sizes capture more detail, despite increasing background noise.

Increasing the number of averages (excitations) also improved the SNR. Since the intensity of the source is high, the signal won't change much, however, in each our observation we have a random noise that changes at each observation. Thus, averaging the noise across multiple observations reduces its impact, improving the SNR of the image. Mathematically, doubling the number of averages should increase the SNR by a factor of $\sqrt{2}$ [4], which closely matches our results. Some inconsistencies may be attributed to human error in defining the ROI. Additionally, as expected, images acquired with the body coil were noisier compared to those obtained with the head coil, due to the physical position and capabilities of the body coil.

When increasing the number of slabs, we observed a signal void, which allowed us to calculate the slice thickness based on this void. However, the calculated slice thickness was approximately 1 mm smaller than the actual size, possibly due to partial volume effects or human error in defining the upper and lower slice boundaries (Fig. 2, left).

Additionally, we successfully acquired the images of the abdomen with motion artifact and without, as well as the images without the motion effect with chemical shift. Also, we have been able to calculate the chemical shift that appeared to be covering more pixels than theoretically expected. This difference could be attributed to incomplete motion exclusion during imaging.

Overall, comparing our observations with the goals, we can conclude that most of the data we acquired met expectations, though some discrepancies were observed, such as the slice thickness calculation using the signal void with two slabs and the number of pixels covering the fat-water chemical shift.

Appendix

ROI of the object for the "Exp1_RefAcq"

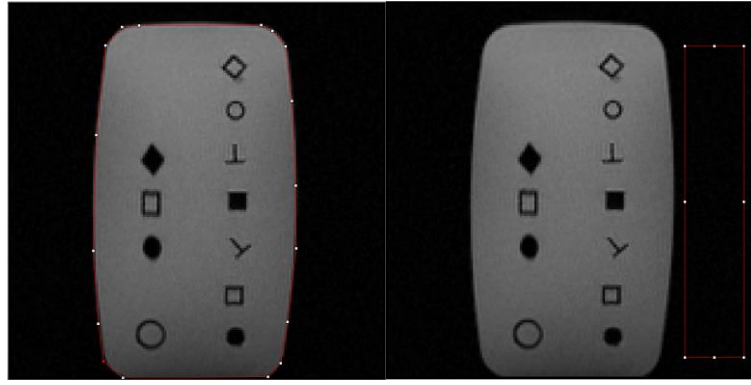


Figure A1. An ROI of *left*: phantom, and *right*: background

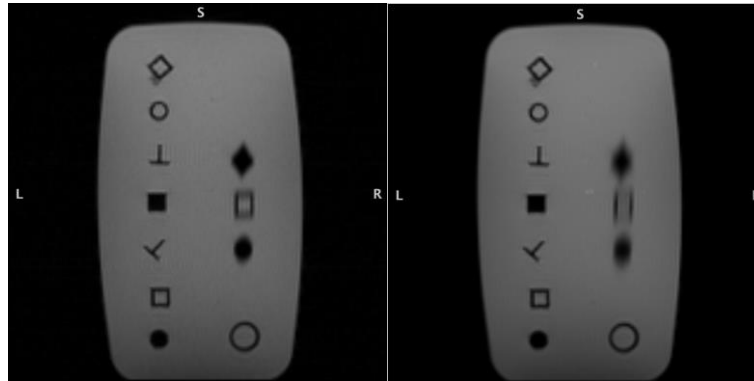


Figure A2. Image obtained with the slice thickness of *left*: 4 mm, and *right*: 8 mm

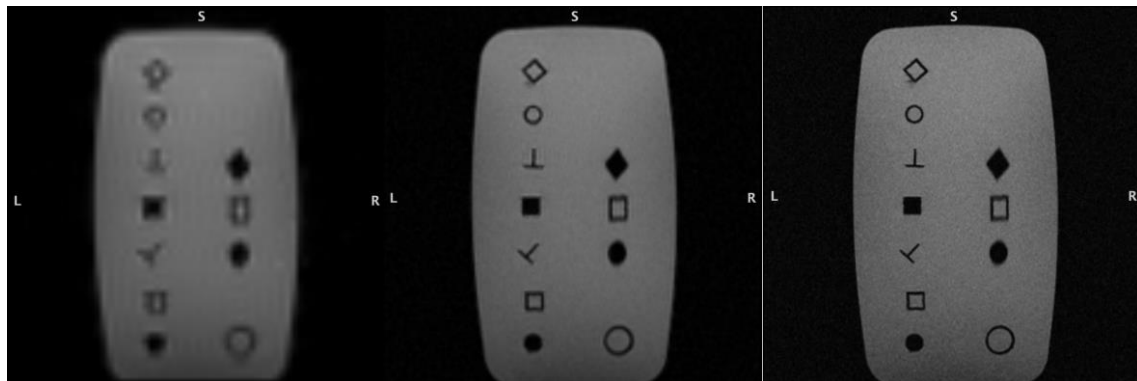


Figure A3. Images obtained with matrix size of *left*: 64x64, *middle*: 160x160 and *right*: 320x320, while keeping all other parameters constant.

References

- [1] S. C. Bushong and G. Clarke, *Magnetic Resonance Imaging: Physical and Biological Principles*. Mosby, 2003.
- [2] W. E. Kwok, "Basic Principles of and Practical Guide to Clinical MRI Radiofrequency Coils," *Radiographics*, vol. 42, no. 3, pp. 898-918, May-Jun 2022, doi: 10.1148/rg.210110.
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- [4] E. M. Haacke, Brown, R. W., Thompson, M. R., Venkatesan, R., "Signal, Contrast, and Noise," in *Magnetic Resonance Imaging*, 2014, pp. 325-373.

