

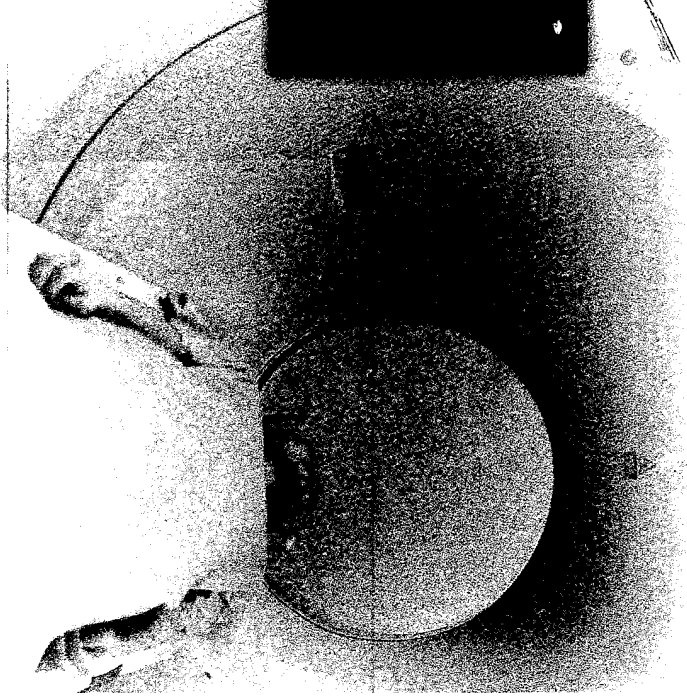
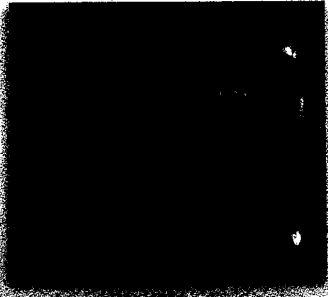
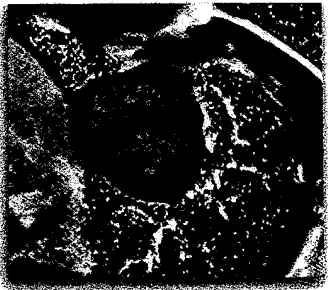
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The Basics



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Scan Parameters and Image Optimization

Introduction

In this chapter, we will discuss all the important parameters in MR imaging that the operator can control and adjust. We will then see how these changes influence the image quality. Every radiologist is comfortable with a particular set of techniques; therefore, "custom-made" techniques can be achieved only if the radiologist is aware of the parameters and trade-offs involved.

Primary and Secondary Parameters

Primary parameters are those that are set directly:

TR
TE
TI
FA (flip angle) } contribute to image contrast

Δz = slice thickness }
Interslice gap } contribute to coverage

FOV_x }
FOV_y } } contribute to resolution:
 N_x : number of frequency-encoding steps } Δx : spacing in x direction
 N_y : number of phase-encoding steps } Δy : spacing in y direction } contribute to S/N ratio
NEX }
Bandwidth }

From the primary parameters above, we can get the secondary parameters (which are also used to describe the image):

1. S/N ratio (SNR)
2. Scan time
3. Coverage
4. Resolution
5. Image contrast

Unfortunately, optimization of these parameters may involve some trade-offs. To gain some advantage with one parameter, we might have to sacrifice another parameter. Let's start out with the concept of signal-to-noise ratio (SNR).

Signal-to-Noise Ratio. What we want is signal. What we don't want is noise. Although we can't completely eliminate noise, there are ways to maximize the SNR. SNR is given by

$$SNR \propto (\text{voxel volume}) \sqrt{(N_y) (NEX) T_s} \quad (\text{Eqn. 17-4})$$

which makes sense because $N_y \times NEX \times T_s$ is the total time the machine is "listening" to the spin echoes.

Since $T_s = N_x \cdot \Delta T_s$ and $\Delta T_s = 1/BW$, then $T_s = N_x/BW$.

$$SNR \propto (\text{voxel volume}) \frac{\sqrt{(N_y) (N_x) (NEX)/BW}}{\quad} \quad (\text{Eqn. 17-5})$$

Therefore, SNR depends on

1. Voxel volume = $\Delta x \cdot \Delta y \cdot \Delta z$
2. Number of excitations (NEX)
3. Number of phase-encoding steps (N_y)
4. Number of frequency-encoding steps (N_x)
5. Full bandwidth (BW)

Let's go through SNR is affected

Voxel Volume

If we increase the number of pixels, before, increase (Fig. 17-1). T

Vox

where $\Delta x = p$ size in the y

NEX (Number of Acquisitions) times the scan signals (S_1 and associated with add up the sig

However, if v

$$N_1 + 1$$

This formula Why do we go to do with a concept and motion theory density of the

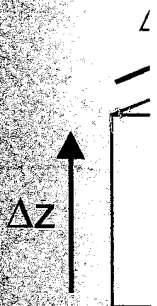


Figure 17-1. element with spins in a voxel having voxel size i

Let's go through each parameter and see how SNR is affected.

Voxel Volume

If we increase the voxel size, we increase the number of proton spins in the voxel and, therefore, increase the signal coming out of the voxel (Fig. 17-1). The voxel volume is given by

$$\text{Voxel volume} = \Delta x \cdot \Delta y \cdot \Delta z$$

where Δx = pixel size in the x direction, Δy = pixel size in the y direction, and Δz = slice thickness.

NEX (Number of Excitations or Acquisitions). NEX stands for the number of times the scan is repeated. Let's say we have two signals (S_1 and S_2), corresponding to the same slice (with the same G_y). There is constant noise (N) associated with each signal ($N_1 = N_2 = N$). If we add up the signals (assuming $S_1 = S_2 = S$), we get

$$S_1 + S_2 = 2S$$

However, if we add up the noise, we get

$$N_1 + N_2 = (\sqrt{2})N, \text{ where } \sqrt{2} \approx 1.41$$

This formula does not make sense at first glance. Why do we get $\sqrt{2}N$ and not $2N$? The answer has to do with a somewhat complicated statistical concept and the so-called random **Brownian motion** theory, which deals with the *spectral density* of the noise.

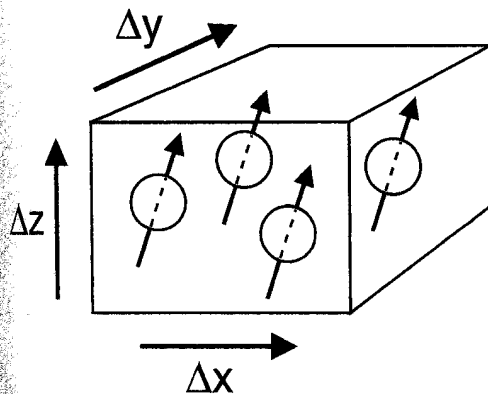


Figure 17-1. A voxel is a three-dimensional volume element with dimensions Δx , Δy , and Δz . The more spins in a voxel, the more signal. Therefore, increasing voxel size increases SNR.

In a simplistic approach, think of the noise as the **variance** (σ^2) of a Gaussian distribution (σ = **standard deviation**). Then, for the sum of the two noise distributions, the variance is additive and given by

$$\sigma_1^2 + \sigma_2^2 = \sigma^2 + \sigma^2 = 2\sigma^2$$

from which the standard deviation is calculated to be

$$\sqrt{(2\sigma^2)} = (\sqrt{2})\sigma$$

This is where the $\sqrt{2}$ factor comes from. However, you don't need to know the underlying math—you just need to understand the concept.

In summary:

$$\frac{S_1 + S_2}{N_1 + N_2} = \frac{2S}{\sqrt{2}N}$$

The resulting signal will be twice the original signal. The resulting noise, however, will be less—it will be the square root of 2 multiplied by the noise $\sqrt{2}N$.

In other words, if we increase the number of acquisitions by a factor of 2, the signal doubles and noise increases by $\sqrt{2}$, for a net $2/\sqrt{2} = \sqrt{2}$; thus, SNR increases by a factor of $\sqrt{2}$.

Therefore, \uparrow NEX by a factor of 2 $\rightarrow \uparrow$ SNR by a factor of $\sqrt{2}$.

Think of the NEX as an *averaging* operation that causes "smoothing" and improvement in the image quality by increasing the signal to a greater degree (e.g., factor 2) relative to the increase in the noise (e.g., factor $\sqrt{2}$). As another example, increasing NEX by a factor of 4 results in an increase of signal by 4 and an increase of noise by $\sqrt{4}$ or 2. Thus, SNR increases by 4/2 or twofold.

N_y (Number of Phase-Encoding Steps).

The same concept holds for N_y . That is, similar to NEX, there is a 41% ($\sqrt{2}$) increase in SNR when N_y is doubled. As with NEX, when the number of phase-encode steps doubles, signal doubles and noise increases (randomly) by $\sqrt{2}$ (for a net $\sqrt{2}$ increase in SNR).

Bandwidth. An inverse relationship exists between BW and SNR. If we go to a wider bandwidth, we include more noise, and the SNR decreases. If we decrease the bandwidth, we

Image Optimization

As above, we can get parameters which are also used

of these parameters affects. To gain some insight, we might have to consider: Let's start out with the signal-to-noise ratio (SNR).

What we want is signal-to-noise. Although we know there are ways to improve SNR, it is given by

$$\frac{S}{N} = \frac{NEX \cdot T_s}{\sqrt{NEX}} \quad (\text{Eqn. 17-1})$$

where $N_y \times NEX \times T_s$ is the "listening" to the signal

and $\Delta T_s = 1/BW$, then

$$\frac{S}{N} = \frac{NEX \cdot \Delta T_s}{\sqrt{NEX}} \quad (\text{Eqn. 17-2})$$

on the other hand, $\Delta T_s = 1/BW$, so $\Delta T_s \propto 1/BW$. Therefore, SNR $\propto \frac{NEX}{\sqrt{NEX} \cdot BW}$.

allow less noise to come through, and the SNR increases.

$$\downarrow \text{BW} \Rightarrow \uparrow \text{SNR}$$

To be exact, decreasing the BW by a factor of 2 causes the SNR to improve by a factor of $\sqrt{2}$.

In general, decreased bandwidth causes the following:

1. Increased SNR
2. Increased chemical shift artifact (more on this later)
3. Longer minimum TE (which means less signal due to more T2 decay). Remember that

$$\text{Bandwidth} = 1/\Delta T_s = N_x/T_s$$

Therefore, a longer sampling time (T_s), which is necessary for a decreased bandwidth, results in a longer minimum TE. With a long TE, increased T2 dephasing results in decreased signal. However, the contribution from reduced noise due to a lower bandwidth outweighs the deleterious effect of reduced signal due to greater T2 decay from increased TE.

4. Decreased number of slices. This decrease is caused by the longer TE. Remember,

$$\text{number of slices} = \text{TR}/(\text{TE} + T_s/2 + T_o)$$

where T_s is the total sampling (readout) time and T_o is the "overhead" time. A narrower bandwidth is usually used on the second echo of a T2-weighted dual-echo image because, with the second echo, we have a longer TE and we are able to afford the longer sampling time. On the first echo, however, we can't afford to use a narrower bandwidth because we can't afford to lengthen the TE. However, we probably don't need the smaller bandwidth anyway because we already have enough SNR on the proton density-weighted first echo of a long T2, double-echo image. A typical T_s for a 1.5-T scanner is 8 msec, resulting in a BW (for a 256 matrix) of

$$\begin{aligned} \text{BW} &= N_x/T_s = 256/8 = 32 \text{ kHz} \\ &= \pm 16 \text{ kHz} = 125 \text{ Hz/pixel} \end{aligned}$$

Note that BW can be described as "full bandwidth" (32 kHz in example above), \pm the

Nyquist frequency (which is ± 16 kHz above and defines the FOV) or "bandwidth per pixel" (which is unambiguous if you forget the \pm).

A typical "variable bandwidth" option includes

1. A wide bandwidth (± 16 kHz) on the first echo, and
2. A narrow bandwidth (± 4 kHz) on the second echo, thus increasing SNR and counteracting T2 decay effects.

Question: How does the gradient affect the BW?

Answer: Recall from Chapter 15 that the field of view (FOV) is given by

$$\text{FOV} = \text{BW}/\gamma G_x \quad \text{or} \quad G_x = \text{BW}/\gamma \text{FOV}$$

For a given FOV, increasing the gradient G_x causes increased BW and, therefore, decreased SNR.

SNR in 3D Imaging

In 3D imaging, we have the same factors contributing to SNR, plus an additional phase encoding step in the z direction (N_z):

$$\text{3D SNR} \propto (\text{voxel volume}) \sqrt{(N_y)(N_x)(N_z)(\text{NEX})/\text{BW}} \quad (\text{Eqn. 17-3})$$

From this equation, you can see why SNR in 3D imaging is higher than that in 2D imaging. Specifically,

$$\text{SNR}(3\text{D}) = \sqrt{N_z} \cdot \text{SNR}(2\text{D})$$

Another way to look at SNR is to say that SNR depends on only two factors:

1. Voxel size
2. Total sampling time

Sampling time (T_s) is the time that we sample the signal. Therefore, it makes sense that the more time we spend sampling the signal, the higher the SNR will be. Let's look again at the formula for SNR (in 2D imaging):

$$\text{SNR} \propto (\text{voxel volume}) \sqrt{(N_y)(N_x)(\text{NEX})/\text{BW}}$$

Recall that

$$T_s = N_x/\text{BW}$$

or

$$1/\text{BW} = T_s/N_x$$

so

$$\text{SNR} \propto (\text{voxel}$$

We know that N_y is steps, which is the the echo corresponds to the number of times we repeat the encoding gradient. In essence, the fact

$$T =$$

is the total sample received for a part

$$\text{SNR} \propto (\text{voxel volume}) \sqrt{\text{total s}}$$

In summary, SNR follows the following:

1. Increasing T
2. Decreasing T
3. Using a lower
4. Using volume
5. Increasing N
6. Increasing N
7. Increasing N
8. Increasing T

Resolution. Spatial resolution is the minimum distance between two points on an ir

$$\begin{aligned} \text{Pixel size} &= \\ &\uparrow N_y \end{aligned}$$

If we increase the number of steps, what happens to resolution? Usually, resolution usually improves if we look at Eqn. 17-2, the catch? The catch is that the constant while in

$$\text{Pixel size also}$$

By increasing N_y , the pixel size becomes smaller. Now, recall

$$\text{Voxel volume} =$$

Incorporating this into Eqn. 17-2 gives us another

so

$$\text{SNR} \propto (\text{voxel volume}) \sqrt{(N_y)(\text{NEX})(T_s)}$$

We know that N_y is the number of phase-encoding steps, which is the number of times we sample the echo corresponding to a particular phase-encoding gradient G_y , and that NEX is the number of times we repeat each phase-encoding step. In essence, the factor

$$T = T_s \cdot N_y \cdot \text{NEX}$$

is the *total sampling time* of all the echoes received for a particular slice. Thus,

$$\text{SNR} \propto (\text{voxel volume}) \frac{\sqrt{\text{NEX}}}{\sqrt{\text{total sampling time of all signals}}}$$

In summary, SNR can be increased by doing the following:

1. Increasing TR
2. Decreasing TE
3. Using a lower BW
4. Using volume (i.e., 3D) imaging
5. Increasing NEX
6. Increasing N_y
7. Increasing N_x
8. Increasing the voxel size

Resolution. Spatial resolution (or pixel size) is the minimum distance that we can distinguish between two points on an image. It is determined by

$$\text{Pixel size} = \text{FOV}/\text{number of pixels}$$

$$\uparrow N_y \rightarrow \text{better resolution}$$

If we increase the number of phase-encoding steps, what happens to SNR? Obviously, *better resolution usually means poorer SNR*. However, if we look at Equation 17-2, it appears that by increasing N_y , the SNR should increase! What's the catch? The catch is, we are keeping the FOV constant while increasing N_y . Take, for example,

$$\text{Pixel size along } y\text{-axis} = \Delta_y = \text{FOV}_y/N_y$$

By increasing N_y , we are making the pixel size smaller. Now, recall that

$$\text{Voxel volume} = \Delta x \cdot \Delta y \cdot \Delta z$$

$$= \text{FOV}_x \cdot \text{FOV}_y \cdot \Delta z / (N_x \cdot N_y)$$

Incorporating this information into Equation 17-2 gives us another way of expressing SNR:

$$\text{SNR} \propto (\text{FOV}_x)$$

$$(\text{FOV}_y) \Delta z \sqrt{\frac{\text{NEX}}{(N_x)(N_y)(\text{BW})}} \quad (\text{Eqn. 17-4})$$

This formula allows us to better separate the factors affecting SNR. From this, we can conclude the following:

1. If we keep FOV constant and increase N_y , we will decrease SNR.

$$\uparrow N_y, \text{FOV constant} \rightarrow \downarrow \text{SNR}$$

2. If we increase N_y and increase FOV, thus keeping pixel size constant, then we will increase the SNR. Yet the resolution doesn't change. What is the trade-off here? The answer is the acquisition time, which is proportional to N_y .

$$\uparrow \text{FOV, pixels fixed} \rightarrow \uparrow \text{SNR, } \uparrow \text{time}$$

3. If we increase slice thickness Δz , we get not only more SNR, but also more partial volume artifact.
4. If we increase NEX, we get more SNR at the expense of longer acquisition time. For 3D imaging, Equation 17-4 is modified to

$$\text{SNR (3D)} = (\text{FOV}_x)(\text{FOV}_y)(\text{FOV}_z)$$

$$\sqrt{\frac{\text{NEX}}{(N_x)(N_y)(N_z)(\text{BW})}} \quad (\text{Eqn. 17-5})$$

Basically, if we want better spatial resolution in a *given* acquisition time, we have to sacrifice SNR. Let's consider a few examples.

1. What happens if we increase the number of pixels with the FOV constant?
 - (a) Increase resolution.
 - (b) Decrease SNR (refer to Equation 17-4). Therefore, as we decrease the pixel size, we increase the resolution and decrease the SNR.
 - (c) Increase scan time (number of pixels increases in phase-encode direction).
2. What happens if we decrease the FOV and keep the number of pixels constant?
 - (a) Increase the resolution.
 - (b) Decrease SNR.
 - (c) Potentially increase aliasing artifact.

3. How do we determine the pixel size (resolution)?

It is determined by dividing the FOV by the number of encoding steps.

Example

For FOV = 250 mm and a 256×256 matrix

$$\begin{aligned} N_x &= N_y = 256 \\ \text{Pixel size (x)} &= \text{FOV}_x / N_x = 250/256 \\ &\cong 1 \text{ mm in x direction.} \end{aligned}$$

$$\begin{aligned} \text{Pixel size (y)} &= \text{FOV}_y / N_y = 250/256 \\ &\cong 1 \text{ mm in y direction.} \end{aligned}$$

In the x direction, there are two ways of increasing resolution (for a given FOV):

1. Increase N_x by reducing the sampling interval ΔT_s (i.e., by increasing the BW) and keeping the sampling time T_s fixed (recall that $T_s = N_x \cdot \Delta T_s$). The advantage here is no increase in TE; the trade-off is a reduction in SNR (due to increased BW).
2. Increase N_x by lengthening T_s and keeping ΔT_s (and thus BW) fixed. Here, the SNR does not change, but the trade-off is an increased TE (due to a longer T_s) and less T1 weighting (this is only a concern in short echo delay time imaging).

Acquisition Time

The acquisition time or scan time, as we have seen previously, is given by

$$\text{Scan time} = \text{TR} \cdot N_y \cdot \text{NEX}$$

where N_y is the number of phase-encoding steps (in the y direction).

For fast spin-echo (FSE) imaging (discussed in detail in Chapter 19), the above is modified to

$$\text{FSE time} = \text{TR} \cdot N_y \cdot \text{NEX} / \text{ETL}$$

where ETL = echo train length (4, 8, 16, 32).

For 3D imaging, the scan time is given by

$$\text{Time (3D)} = \text{TR} \cdot N_y \cdot N_z \cdot \text{NEX}$$

where N_z is the number of phase-encoding steps (partitions) in the z direction. In other words,

$$\text{Time (3D)} = N_z \cdot \text{time(2D)}$$

Multiplication by such a large number (e.g., $N_z = 32$ to 64 or 128) might at first seem to result in

an excessively long scan time for 3D imaging, but the TR used in 3D gradient-echo imaging is approximately 100 times smaller (order of 30 msec) compared with the TR used in conventional spin-echo imaging; we can perform a 3D scan in a reasonable time. Recently, 3D FSE imaging (discussed in Chapter 19) has also become feasible.

Example

1. Calculate the acquisition time of an SE sequence with TR = 3000 msec, $N_y = 256$, and NEX = 1.

$$\begin{aligned} \text{Solution: Scan time} &= 3000 \times 256 \text{ msec} \\ &= 768 \text{ sec} = 12.8 \text{ min} \end{aligned}$$

2. Calculate the acquisition time of an FSE sequence with the above parameters and an ETL of 8

$$\text{Solution: Scan time} = \frac{12.8 \text{ min}}{8} = 1.6 \text{ min}$$

3. (a) Calculate the acquisition time of a 3D gradient-echo sequence with TR = 30 msec, $N_y = 256$, NEX = 1, and $N_z = 60$.

$$\begin{aligned} \text{Solution: Scan time} &= 30 \times 256 \times 1 \times 60 \text{ msec} \\ &= 460.8 \text{ sec} = 7.68 \text{ min} \end{aligned}$$

- (b) If TR = 300 in the previous example then the scan time = 76.8 min = 1 hr and 16.8 min, which is, obviously, impractical. Hence, 3D techniques use gradient-echo sequences employing a very short TR.

TR. What happens if we increase or decrease TR?

1. Increasing TR:

- (a) increases SNR (according to the T1 recovery curve)
- (b) increases coverage (more slices)
- (c) decreases T1 weighting
- (d) increases proton density and T2 weighting
- (e) increases scan time

2. Decreasing TR:

- (a) decreases SNR
- (b) decreases coverage
- (c) increases T1 weighting

- (d) decreases weighting
- (e) decreases

Sometimes an M for a certain TR, though, can be achieved. Therefore he or she might increase T1 weighting, an undesirable effect.

Coverage

Coverage is the distance of acquisition. It depends on the slice thickness (Fig. 17-2). Because

number of slices

then

$$\text{Coverage} = \text{TR} / \text{thick}$$

where T_s is the "overhead" time, as chapters.

Let's summarize:

1. Coverage is in

- (a) increase slice thickness
- (b) increase TR
- (c) increase T1 weighting (i.e., increase TR)
- (d) decrease slice thickness (lower TE),

2. Coverage is de

- (a) increase T1 weighting
- (b) increase TR
- (c) increase slice thickness longer time

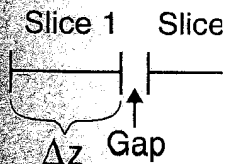


Figure 17-2. Coverage Δz and by the number of slices $\times (\Delta z +$

- (d) decreases proton density and T2 weighting
- (e) decreases scan time

Sometimes an MR technologist will find that, for a certain TR, the required coverage cannot be achieved. Therefore, to increase the coverage, he or she might increase the TR. However, in so doing, T1 weighting is decreased, which may be an undesirable effect.

Coverage

Coverage is the distance covered by a multislice acquisition. It depends on the number of slices and on the slice thickness and the interslice gap (Fig. 17-2). Because

$$\text{number of slices} = \text{TR} / (\text{TE} + T_s/2 + T_o)$$

then

$$\text{Coverage} = \text{TR} / (\text{TE} + T_s/2 + T_o) \times (\text{Slice thickness} + \text{gap})$$

where T_s is the sampling time and T_o is the "overhead" time, as we've discussed in previous chapters.

Let's summarize:

1. Coverage is increased if we:
 - (a) increase slice thickness
 - (b) increase interslice gap
 - (c) increase TR or decrease the last TE (i.e., increase TR/TE ratio)
 - (d) decrease sampling time T_s (resulting in a lower TE), that is, increase the bandwidth
2. Coverage is decreased if we:
 - (a) increase TE
 - (b) increase T_s
 - (c) increase ETL in FSE imaging (due to longer final TE)

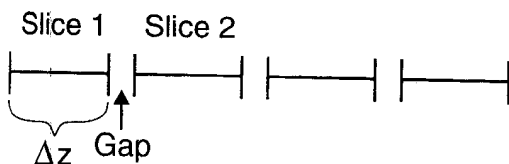


Figure 17-2. Coverage is determined by slice thickness Δz and by the interslice gap. Coverage = number of slices \times (Δz + gap).

3. Increasing interslice gap causes
 - (a) increased coverage
 - (b) decreased "cross-talk" artifact
 - (c) increased SNR (due to increasing effective TR by reducing cross-talk)
 - (d) decreased detection of small lesions (which may lie within the gap)

TE (Time to Echo)

Question: What happens if we increase or decrease TE?

Answer:

1. By increasing TE, we:
 - (a) increase T2 weighting
 - (b) increase dephasing and thus decrease SNR (according to the T2 decay curve)
 - (c) decrease number of possible slices (decrease coverage), because number of slices \approx TR/TE
 - (d) no change in scan time (unless, of course, the coverage is not adequate and either longer TR or extra acquisitions are required)
2. The reverse is true for decreasing TE:
 - (a) decrease T2 weighting and increase T1 or proton density weighting
 - (b) increase SNR (less dephasing). However, if TE is reduced by reducing T_s (i.e., increasing BW), SNR may be reduced!
 - (c) increase coverage
 - (d) no change in scan time

Question: What causes lengthening of the minimum TE?

Answer:

1. TE should be long enough so that the side lobes of the 180° pulse do not interfere with the side lobes of the FID or the echo (Fig. 17-3). Remember that we need a Fourier transform of the RF pulse with a square shape to be able to get ideal contiguous slices. To do this, the RF must be a sinc wave ($\text{sinc } t = \sin t/t$) with as many side lobes as possible. This, in turn, will lengthen the 90° and 180° RF pulse.
2. If TE is so short that it allows interference between the 180° RF pulse and the FID, an

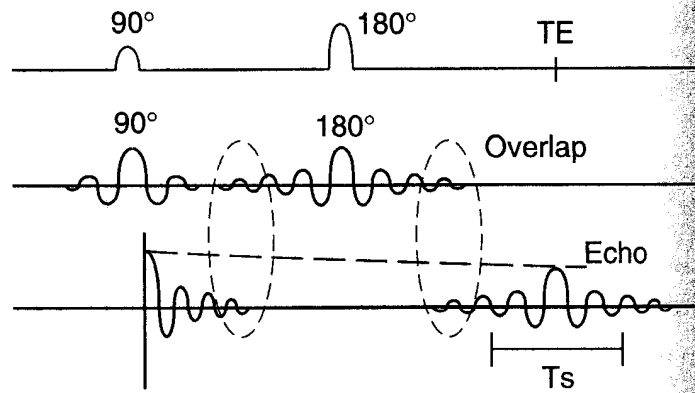


Figure 17-3. To avoid overlapping of the FID and the side lobes of the 180° pulse, you need to increase TE. This increase is one cause of lengthening the minimum TE.

FID artifact (or zipper artifact) will appear along the zero frequency line.

Question: How can TE be shortened?

Answer:

1. One way is to decrease the sampling time T_s . However, this results in a higher BW and therefore a lower SNR (Equation 17-2).
2. There is a limit as to how short TE could be. The factors limiting minimum TE include
 - (a) duration of RF pulse (especially the 180° pulse)
 - (b) duration of FID
 - (c) T_s or BW (which influence the SNR)
3. TE can also be shortened by switching to a gradient-echo sequence because a 180° refocusing pulse is no longer used.

Contrast on a spin-echo technique can be summarized (Table 17-1):

TI (Inversion Time)

As we saw in Chapter 7, inversion recovery sequences employ an additional 180° pulse before the 90° pulse.

Table 17-1

	TR	TE
T1W	Short	Short
PDW	Long	Short
T2W	Long	Long

Advantages

1. Can suppress various tissues by selecting the appropriate TI. More specifically, as we saw in Chapter 7, if

$$TI = 0.693 \times T1 \text{ (tissue } x \text{)}$$

- then tissue x is “nulled” or “suppressed.”
2. **STIR** (short TI inversion recovery) sequences suppress fat by selecting

$$TI = 0.693 \times T1 \text{ (fat)}$$

Since at 1.5 Tesla, $T1$ of fat is approximately 200 msec, then to null fat, we must select

$$TI = 0.693 \times 200 \cong 140 \text{ msec}$$

3. **FLAIR** (fluid-attenuated inversion recovery) sequences suppress fluid by selecting

$$TI = 0.693 \times T1 \text{ (fluid)}$$

This sequence is used, for example, in the brain to suppress cerebrospinal fluid (CSF) to increase the conspicuity of periventricular hyperintense lesions such as multiple sclerosis plaques. Since at 1.5 T, $T1$ of CSF is approximately 3600 msec, then to null CSF, we have to select

$$TI = 0.693 \times 3600 \cong 2500 \text{ msec}$$

Disadvantages

1. Decreased SNR
2. Decreased coverage (by a factor of about 2) due to the presence of the extra 180° pulse

Key Points

In this chapter, we discuss practical factors that influence imaging. To improve this, it is crucial to have a firm grasp of the concepts that, directly or indirectly, introduced the primary

Questions

- 17-1 For a TR = 1500 msec, a 128 matrix, calculate:
 - (a) a single slice
 - (b) 10 slices (p)
 - (c) 10 slices per slice (multi)
- 17-2 Calculate the achievable slice thickness if TE = 80 msec, TR = 20 msec, and “
- 17-3 The concept of “magic angle” is used to improve SNR. Calculate the angle for a selected. Suppose:
 - (a) How is the angle determined?
 - (b) What happens if the angle is not selected?
 - (c) How does the angle affect the number of slices?
- 17-4 The acquisition time for a gradient-echo sequence with TR = 30 msec, TE = 256 msec, and a matrix of 256 for acquisition:
 - (a) 15.36 sec
 - (b) 15.36 min
 - (c) 230.4 sec
 - (d) 230.4 min
 - (e) 230,400 sec
- 17-5 The SNR in 3D MRI is proportional to the SNR in 2D MRI. Calculate the SNR in 3D MRI if:
 - (a) N_z
 - (b) N_x
 - (c) N_y
- 17-6 Increasing TE affects the following:
 - (a) T2W
 - (b) T1W
 - (c) coverage
- 17-7 SNR can be improved by:
 - (a) increasing TE
 - (b) decreasing TE
 - (c) increasing TR

Key Points

In this chapter, we discussed the important and practical factors that influence the quality of MR imaging. To improve the quality of the images, it is crucial to have a firm grasp of the parameters that, directly or indirectly, affect the scan. We introduced the primary and secondary parameters

that are used to determine MR images (refer to the *Introduction* in this chapter). In a nutshell, the name of the game is "trade-offs." Often, one cannot gain advantage in one area without sacrificing another.

Questions

- 17-1** For a TR = 1500 msec, 2 NEX, and a 128×128 matrix, calculate the scan time for
- a single slice
 - 10 slices (performed one at a time)
 - 10 slices performed using a multi-slice (multiplanar) acquisition
- 17-2** Calculate the maximum number of achievable slices for a TR = 1000 msec, TE = 80 msec, sampling time $T_s = 20$ msec, and "overhead time" $T_o = 10$.
- 17-3** The concept of variable BW: in order to improve SNR, the lowest possible BW is selected. Suppose that the BW is halved:
- How is the SNR affected?
 - What happens to chemical shift artifacts?
 - How does this affect the maximum number of slices?
- 17-4** The acquisition time of a single acquisition gradient-echo sequence with TR = 30 msec, TE = 10 msec, NEX = 2, $N_y = 256$ for acquiring 15 slices is about
- 15.36 sec
 - 153.6 sec
 - 230.4 sec
 - 15,360 sec
 - 230,400 sec
- 17-5** The SNR in 3D imaging is equal to the SNR in 2D imaging times the factor:
- N_z
 - $\sqrt{N_z}$
 - N_y
 - $\sqrt{N_y}$
- 17-6** Increasing TE leads to a decrease in all of the following *except*
- T2W
 - signal
 - coverage
 - SNR
- 17-7** SNR can be increased by
- increasing NEX
 - decreasing BW
 - increasing N_y
 - increasing N_x
 - increasing voxel volume
 - increasing TR
 - decreasing TE
 - all of the above
 - only (a)–(e)
- 17-8** Increasing N_y leads to
- better resolution
 - increased SNR (fixed FOV)
 - increased SNR (fixed pixels)
 - increased scan time
 - all of the above
 - only (a), (c), (d)
 - only (a), (b), (d)
- 17-9** For a 128 square matrix and an FOV of 25 cm, the pixel size is about
- 0.5 mm
 - 1 mm
 - 1.5 mm
 - 2 mm
- 17-10** The SNR is proportional to the square root of
- $BW/N_x \cdot NEX$
 - $BW/N_y \cdot NEX$
 - $N_x \cdot N_y \cdot NEX/BW$
 - $N_y \cdot BW/NEX$
- 17-11** Increasing TR leads to an increase in all the following *except*
- scan time
 - SNR
 - T1W
 - T2W
 - coverage
- 17-12** Minimum TE can be reduced by
- reducing the duration of the RF pulses
 - reducing the sampling time T_s
 - increasing the bandwidth
 - using a sequence that doesn't use 180° pulses (as in gradient echo)
 - all of the above

- 17-13** The acquisition time in 3D imaging is equal to that in 2D imaging times the factor:
 (a) N_z (b) $\sqrt{N_z}$
 (c) N_y (d) $\sqrt{N_y}$
- 17-14** Coverage is increased by increasing all of the following *except*
 (a) slice thickness (b) interslice gap
 (c) TR (d) BW
 (e) TE
- 17-15** In STIR, T1 should be set to
 (a) 1.44 T1 (fat)
 (b) $(1/\sqrt{2})$ T1 (fat)
- (c) 2 T1 (fat)
 (d) 0.693 T1 (fat)
 (e) $(1/0.693)$ T1 (fat)
 (f) choices (b) or (d)
- 17-16** In FLAIR, T1 should be set to
 (a) 0.693 T1 (fluid)
 (b) $(\ln 2)$ T1 (fluid)
 (c) $(-1 \ln 0.5)$ T1 (fluid)
 (d) all of the above
- 17-17** Match (i) STIR; (ii) FLAIR with
 (a) dark fluid
 (b) dark fat

Introduction

MRI, as with any other share of artifacts.

It is important to and to have the tools to imize them. There are in MRI. These are sum

1. Image processing
 - (a) Aliasing
 - (b) Chemical shift
 - (c) Truncation
 - (d) Partial volume
2. Patient-related artifacts
 - (a) Motion artifacts
 - (b) Magic angle
3. Radio frequency (RF) artifacts
 - (a) Cross-talk
 - (b) Zipper artifacts
 - (c) RF feedthrough
 - (d) RF noise
4. External magnetic field artifacts
 - (a) Magnetic inhomogeneity
5. Magnetic susceptibility artifacts
 - (a) Diamagnetic, paramagnetic
 - (b) Metal
6. Gradient-related artifacts
 - (a) Eddy currents
 - (b) Nonlinearity

Introduction

MRI, as with any other imaging modality, has its share of artifacts.

It is important to recognize these artifacts and to have the tools to eliminate or at least minimize them. There are many sources of artifacts in MRI. These are summarized as follows:

1. Image processing artifact
 - (a) Aliasing
 - (b) Chemical shift
 - (c) Truncation
 - (d) Partial volume
2. Patient-related artifacts
 - (a) Motion artifacts
 - (b) Magic angle
3. Radio frequency (RF) related artifacts
 - (a) Cross-talk
 - (b) Zipper artifacts
 - (c) RF feedthrough
 - (d) RF noise
4. External magnetic field artifacts
 - (a) Magnetic inhomogeneity
5. Magnetic susceptibility artifacts
 - (a) Diamagnetic, paramagnetic, and ferromagnetic
 - (b) Metal
6. Gradient-related artifacts:
 - (a) Eddy currents
 - (b) Nonlinearity

- (c) Geometric distortion
7. Errors in the data
8. Flow-related artifacts
9. Dielectric effects

Let's discuss this list in more detail.

Image Processing Artifact

Aliasing (Wraparound). Refer to the discussion on *undersampling* in Chapter 12.

Spin-Echo Imaging. Let's say we're studying the abdomen (Fig. 18-1). If the field of view (FOV) only covers part of the body, we know that we may get **aliasing** (wraparound), but what causes the aliasing?

We have a gradient in the x direction (G_x), with a maximum frequency (f_{max}) at one end of the FOV, and a minimum frequency ($-f_{max}$) at the other end of the FOV. These are the **Nyquist** frequencies (discussed in Chapter 12). Any frequency higher than the maximum frequency allowed by the gradient cannot be detected correctly.

The gradient doesn't stop at the end of the FOV. The gradient is going to keep going because we still have magnetic fields outside the space designated by the FOV. The parts of the body outside the FOV (in this case, the arms) will be exposed to certain magnetic field gradients. One arm will receive a magnetic field that will generate a frequency higher than f_{max} for the FOV. It may be twice the frequency of f_{max} — twice the intended Nyquist frequency. The computer cannot recognize these frequencies above (f_{max}) or below ($-f_{max}$). They will be recognized

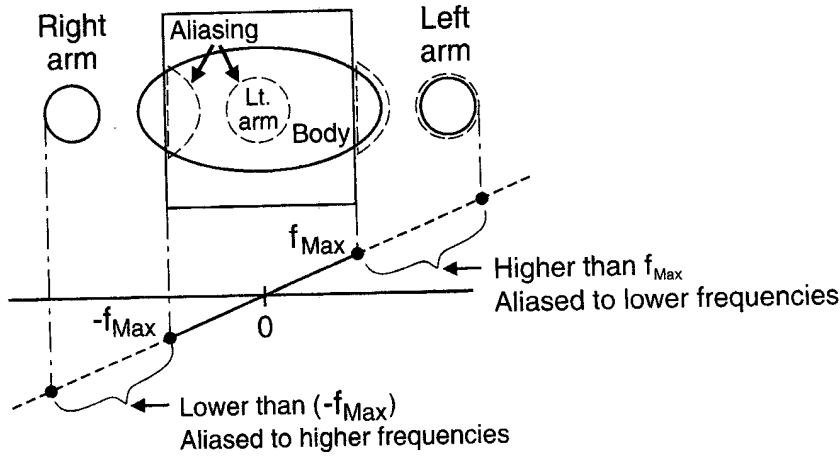


Figure 18-1. For a given FOV and gradient strength, the maximum frequency f_{max} corresponds to the edges of the FOV. Any part outside the FOV will experience a higher frequency. The higher frequencies outside the FOV may be aliased to a lower frequency inside the FOV. This will cause a wraparound artifact.

as a frequency within the bandwidth. The higher frequency will be recognized as a lower frequency within the accepted bandwidth.

For example, if the higher frequency were 2 kHz higher than (f_{max}), it would be recognized as 2 kHz higher than ($-f_{max}$), and therefore its information would be “aliased” to the opposite side of the image—the side of the FOV that corresponds to the lowest frequencies (Fig. 18-1).

The part of the body and arm on the left side of the patient that is outside the FOV and is exposed to a higher magnetic field will have spins oscillating at a frequency higher than (f_{max}). Thus, it will be identified as a structure on the right side of the patient—that side of the image associated with lower frequencies.

Likewise, the arm and body outside the FOV on the right side of the patient will experience spins oscillating at frequencies lower than ($-f_{max}$) and will also be incorrectly recognized by the computer. For example, if the lower frequency were 2 kHz lower than ($-f_{max}$), it would be recognized as 2 kHz lower than (f_{max}), and its information would be “aliased” to the opposite side of the image—the side of the FOV that corresponds to the higher frequencies. This process is also called **wraparound**—the patient’s arm gets “wrapped around” to the opposite side.

The computer can’t recognize frequencies outside the bandwidth (which determines the FOV). Any frequency outside of this frequency range is going to get “aliased” to a frequency that

exists within the bandwidth. The “perceived” frequency will be the actual frequency minus twice the Nyquist frequency.

$$f(\text{perceived}) = f(\text{true}) - 2f(\text{Nyquist})$$

Why then do we usually see wraparound in the phase-encoding direction? Remember that the number of phase-encoding steps is directly related to the length of the scan time. The phase-encoding steps can be lowered by shortening the FOV in the phase-encoding direction versus the frequency encoding direction also known as rectangular FOV (see Chapter 23). If the FOV is shortened too much in this direction versus the actual extent of the body then wraparound will occur. Figure 18-2 contains an example of wraparound.

3D Imaging. Wraparound artifact can also be seen in 3D imaging in all three directions.

1. It can be seen along the x and y directions as with spin-echo imaging.
2. It can also be seen along the slice-selected (phase-encoded) direction at each end of the slab (e.g., the last slice is overlapped on the first slice, as in Figs. 18-3 and 18-4).

Example

Suppose the frequency bandwidth is 32 kHz (± 16 kHz). This means that if we’re centered at zero frequency, the maximum frequency $f_{max} = +16$ kHz and minimum frequency ($-f_{max}$) = -16 kHz (Fig. 18-1). If we have a frequency in the arm (out



Figure 18-2. Axial MRI scan showing a patient's arm wrapped around the opposite side of the image. Arrows point to the wraparound artifact.

side the FOV) of +15 kHz, the frequency will be

$$f(\text{perceived}) = f(\text{true}) - 2f(\text{Nyquist})$$

Now, the arm, which has a frequency of -15 kHz, will be recognized as a low frequency—only 1 kHz above the negative end frequency.

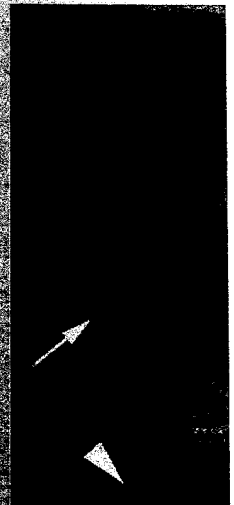


Figure 18-3. 3D gradient echo MRI scan showing gadolinium enhancement of the abdomen. Arrows indicate aliasing artifacts with the kidneys. Also note the phase-encoding direction in the anterior abdominal image. The kidney tissue “wrapping around” is visible.

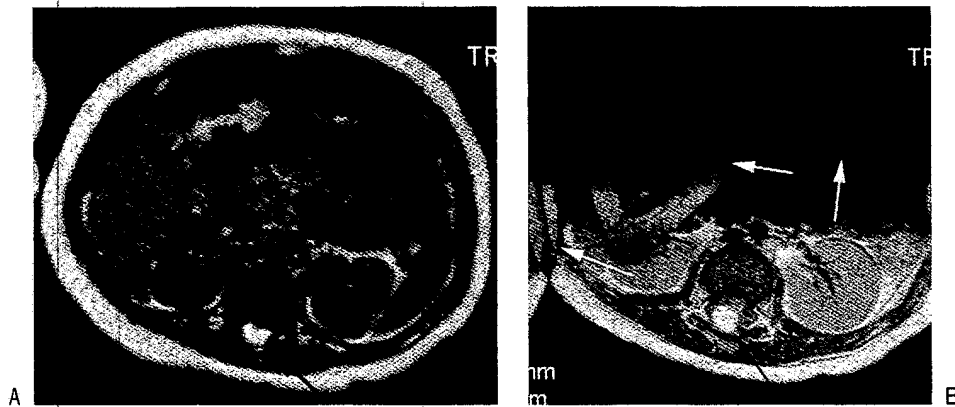


Figure 18-2. Axial T1 (A) and PD (B) images of the lumbar spine demonstrate aliasing of the arms in Figure B (arrows). Figure B was obtained with a smaller FOV resulting in aliasing artifact. Also note that the patient has a filum terminale lipoma (black arrows in A and B).

side the FOV) of +17 kHz, the perceived frequency will be

$$f(\text{perceived}) = +17 \text{ kHz} - 2(16 \text{ kHz}) \\ = -15 \text{ kHz}$$

Now, the arm, which is perceived as having a frequency of -15 kHz (rather than +17 kHz), will be recognized as a structure with a very low frequency—only 1 kHz faster than the negative end frequency of the bandwidth—and

will be identified on the opposite side of the image, the low-frequency side.

Remedy. How do we solve this problem?

1. *Surface coil:* The simplest way is to devise a method by which we don't get any signal from outside the FOV. With the patient in a large transmit/receive coil that covers the whole body, we will receive signal from all the body parts in that coil, and those parts outside the FOV will result in aliasing. But

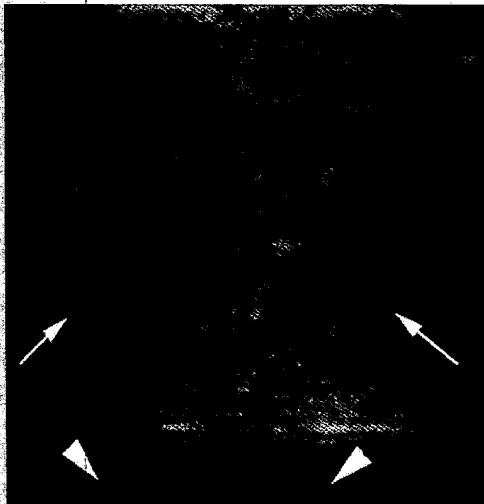


Figure 18-3. 3D gradient-echo T1 image without gadolinium of the abdomen shows slice direction aliasing with the kidneys appearing to be in the lungs (arrows). Also note that there is aliasing in the phase-encoding direction (anteroposterior) from the inferior abdominal image's anterior subcutaneous tissue "wrapping around" posteriorly (arrowheads).

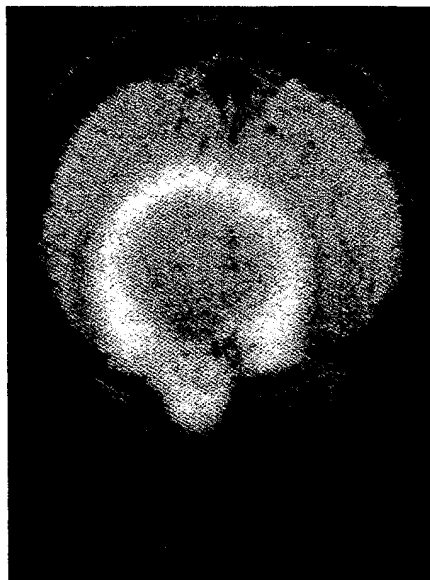


Figure 18-4. 3D coronal gradient-echo T1 image of the brain shows slice direction aliasing of the anterior skull on to the brain.

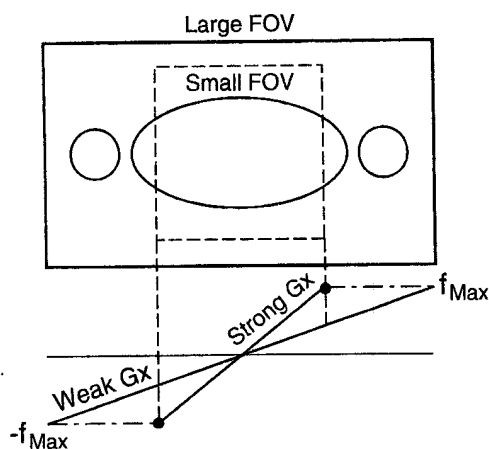


Figure 18-5. To avoid aliasing, increase the FOV.

if we use a **coil** that only covers the area within the FOV, we will only get signal from those body parts within the maximum frequency range, and no aliasing will result. This type of coil is called a **surface coil**. We also use a surface coil to increase the signal-to-noise ratio (SNR).

2. **Increase FOV:** If we double the FOV to include the entire area of study, we can eliminate aliasing. To do so, we have to use a weaker gradient. The maximum and minimum frequency range will cover a larger area, and all the body parts in the FOV will be included within the frequency bandwidth; therefore, no aliasing will result (Fig. 18-5). To maintain the resolution, double the matrix with a weaker gradient (G_x). The maximum and minimum frequency range will still be the same as the stronger gradient. They will just be spread out over a wider distance. Remember, to increase the FOV, we have to use a weaker gradient.
3. **Oversampling:** Two types are discussed:
 - (a) Frequency oversampling (no frequency wrap [NFW])
 - (b) Phase oversampling (no phase wrap [NPW])
 - (a) **Frequency oversampling (NFW):** Frequency oversampling eliminates aliasing caused by **undersampling** in the frequency-encoding direction (refer to the sampling theorem in Chapter 12). **Oversampling** can also be performed in the phase-encoding direction by increasing

the number of phase-encoding gradients.

- (b) **Phase oversampling (NPW):** We can double the FOV to avoid aliasing and, at the end, discard the unwanted parts when the image is displayed (Fig. 18-6). This is called **no phase wrap (NPW)** by some manufacturers. It is also called **phase oversampling** by other manufacturers. Because N_y is doubled, NEX is halved to maintain the same scan time. Thus, the SNR is unchanged. (The scan time might be increased slightly because **overscanning** performs with slightly more than $\frac{1}{2}$ NEX.) An example of this is seen in Figure 18-7.

4. **Saturation pulses:** If we saturate the signals coming from outside the FOV, we can reduce aliasing.
5. **3D imaging:** In 3D imaging, if we see this artifact along the slice-select axis, we can simply discard the first and last few slices.

Chemical Shift Artifact. The principle behind the chemical shift artifact is that the protons from different molecules precess at slightly different frequencies. For example, look at fat and H_2O . A slight difference exists between the precessional frequencies of the hydrogen protons in fat and

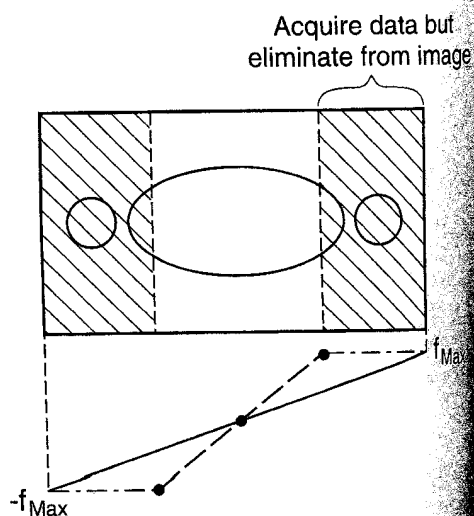


Figure 18-6. In no phase wrap, aliasing is avoided by doubling the FOV in the y direction and, at the end, discarding the unwanted part of the image.

A

H_2O . Actually, faster than the ppm. Let's see

Example
Consider a 1.5-T magnet. The precessional frequency is as follows:

1. Frequency of fat is 63.8 MHz
2. 3.5 ppm
3. $(3.5 \times 10^6) \times 63.8 \text{ MHz} = 223.3 \text{ kHz}$

In other words, the precessional frequency of fat is 223.3 kHz and in H_2O is 226.8 kHz.

Example
We now have a 1.5-T magnet. The precessional frequency of fat is 63.8 MHz and in H_2O is 67.3 MHz.

Therefore, at 0.1 ppm, the precessional frequency of the H_2O is only 73 Hz higher than that of fat. How does this affect the chemical shift artifacts at the fat-water interfaces, and

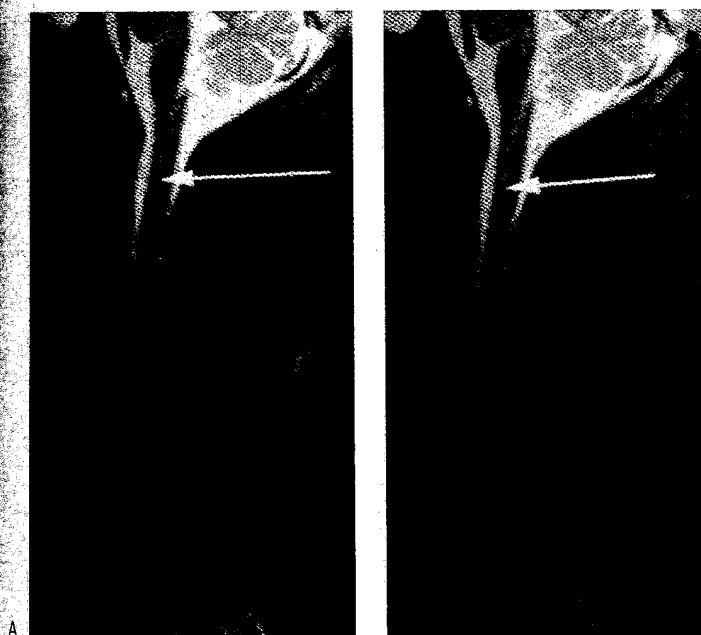


Figure 18-7. Sagittal STIR image (A) of the cervical spine with craniocaudal phase-encoding direction demonstrates aliasing of the brain onto the upper thoracic spine. (B) The same image after no phase wrap was applied. Truncation artifact is also seen (arrows).

H_2O . Actually, the protons in H_2O precess slightly faster than those in fat. This difference is only 3.5 ppm. Let's see what this means by an example.

Example

Consider a 1.5-T magnet. The precessional frequency is as follows:

1. Frequency = $\omega_0 = \gamma B_0$
 $= (42.6 \text{ MHz/T}) (1.5 \text{ T})$
 $\approx 64 \text{ MHz} = 64 \times 10^6 \text{ Hz}$
2. $3.5 \text{ ppm} = 3.5 \times 10^{-6}$
3. $(3.5 \times 10^{-6})(64 \times 10^6 \text{ Hz}) \approx 220 \text{ Hz}$

In other words, at 1.5 T, the difference in precessional frequency of the hydrogen protons in fat and in H_2O is 220 Hz.

Example

We now have a 0.5-T magnet. The precessional frequency of protons in a 0.5-T magnet is 1/3 of a 1.5-T magnet. The frequency difference is then

$$1/3(220 \text{ Hz}) = 73 \text{ Hz}$$

Therefore, at 0.5 T, the difference in precessional frequency of the hydrogen protons in fat and in H_2O is only 73 Hz. In other words, if we use a weaker magnet, we will get less chemical shift.

How does this affect the image? Chemical shift artifacts are seen in the orbits, along vertebral endplates, in the abdomen (at organ/fat interfaces), and anywhere else fatty structures

abut watery structures. In a 1.5-T magnet, the sampling time (T_s) is usually about 8 msec. Let's take 256 frequency points in the frequency-encoding direction.

$$\begin{aligned} \text{BW} &= N/T_s \\ &= 256/8 \text{ msec} \\ \text{BW} &= 32 \text{ kHz} \end{aligned}$$

These formulas show that the entire frequency range (i.e., bandwidth) of 32 kHz covers the whole length of the image in the x direction. Because we have the FOV of the image in the x direction divided into 256 pixels, each pixel is going to have a frequency range of its own, that is, each pixel has its own BW:

$$\begin{aligned} \text{BW/pixel} &= 32 \text{ kHz}/256 \\ \therefore \text{BW/pixel} &= 125 \text{ Hz} \end{aligned}$$

(This representation of BW on a "per pixel" basis is used by Siemens and Philips. It has the advantage of less ambiguity than the $\pm 16 \text{ kHz}$ designation, should the " \pm " be deleted.) Thus, each pixel contains 125 Hz of information (Fig. 18-8). Stated differently, the pixel bin contains 125 Hz of frequencies. Now, because fat and H_2O differ in the precessional frequency of hydrogen by 220 Hz at 1.5 T, how many pixels does this difference correspond to?

$$\begin{aligned} \text{Pixel difference} &= 220 \text{ Hz}/125 \text{ Hz/pixel} \\ &\approx 2 \text{ pixels} \end{aligned}$$

se-encoding gra-

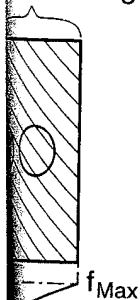
γ (NPW): We can
 o avoid aliasing
 d, discard the
 en the image is
 6). This is called
 NPW) by some
 is also called
 ng by other man-
 e N_y is doubled,
 o maintain the
 hus, the SNR is
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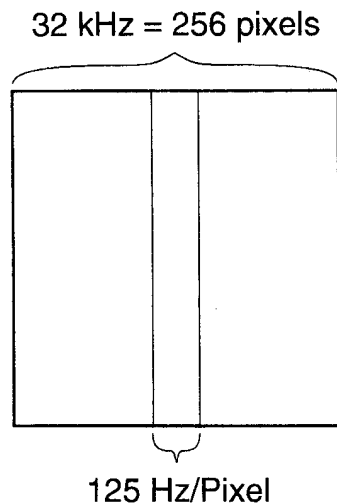


Figure 18-8. At 1.5 T with a BW of 32 kHz and 256 pixels, there will be about 125 Hz/pixel ($32 \text{ kHz}/256 = 125 \text{ Hz}$), that is, there is 125 Hz of information in each pixel. This may be a better way of describing the BW of a scanner because there is no \pm confusion.

This means that fat and H_2O protons are going to be **misregistered** from one another by about 2 pixels (in a 1.5-T magnet using a standard $\pm 16 \text{ kHz}$ bandwidth). (Actually, it is *fat* that is misregistered because position is determined by assuming the resonance property of water.) If pixel size $\Delta x = 1 \text{ mm}$, this then translates into 2 mm misregistration of fat.

MATH: For the mathematically interested reader, it can be shown that

Chemical shift

$$\begin{aligned}
 &= \frac{3.5 \times 10^{-6} \gamma \mathbf{B}}{\text{BW}/N_x} \text{ (in pixels)} \\
 &= \frac{3.5 \times 10^{-6} \gamma \mathbf{B}}{\text{BW}/N_x} \times \frac{\text{FOV}}{N_x} \\
 &= \frac{3.5 \times 10^{-6} \gamma \mathbf{B} \times \text{FOV}}{\text{BW}/N_x} \text{ (in mm)}
 \end{aligned}$$

where $\gamma = 42.6 \text{ MHz/T}$, \mathbf{B} is the field strength (in T), BW is the bandwidth (in Hz), and FOV is the field of view (in cm).

Let's now consider chemical shift artifact visually (Fig. 18-9). Remember that H_2O protons resonate at a higher frequency compared with the hydrogen protons in fat. With the polarity of the frequency-encoding gradient in the x direction set such that higher frequencies

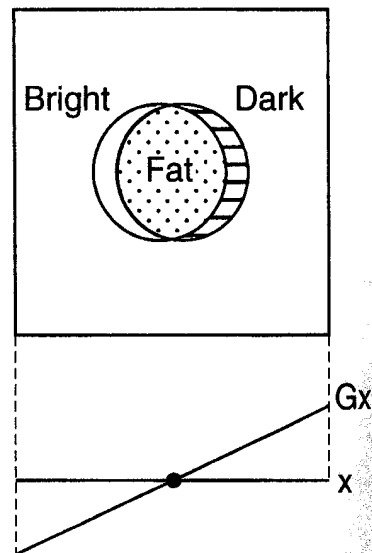


Figure 18-9. Chemical shift effect between fat and water causes a bright band toward the lower frequencies (due to overlap of fat and water at lower frequencies) and a dark band toward the higher frequencies (due to lack of fat and water signals).

are toward the right, H_2O protons are relatively shifted to the right (toward the higher frequencies), and fat protons are relatively shifted to the left (toward the lower frequencies). This will result in overlap at lower frequency and a signal void at higher frequency. This in turn results in a *bright band* toward the lower frequencies and a *dark band* toward the higher frequencies on a T1-weighted (T1W) or proton density (PD)-weighted conventional spin-echo image. (On a T2W CSE, fat is dark, so the chemical shift artifact is reduced. Unfortunately, on T2W FSE [fast spin echo—see Chapter 17], fat is bright, and the chemical shift artifact is more pronounced. We will see this misregistration artifact in more detail later where that we have a fat/ H_2O interface. Remember that this fat/ H_2O chemical shift artifact only occurs in the frequency-encoding direction (in a conventional spin-echo image) and not in gradient-echo [GRE] imaging).

Example—Vertebral Bodies

With frequency-encoding direction—fat is bright—going up and down (and “up” having higher frequency)—the fat in the vertebral bodies is misregistered down, making the lower

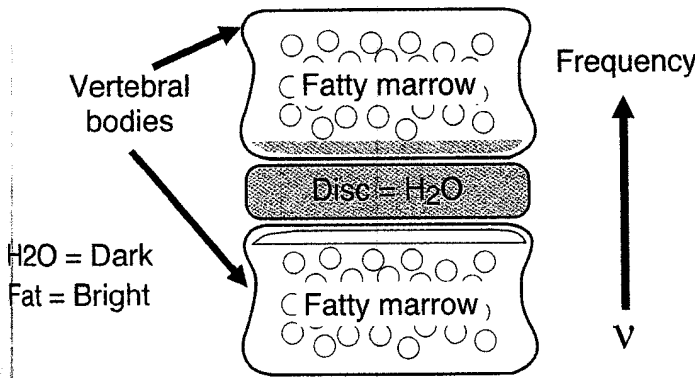


Figure 18-10. Chemical shift artifact in the vertebral endplates produces a dark band in the inferior endplates and a bright band in the superior endplates (assuming the frequency-encoding direction to be upward).

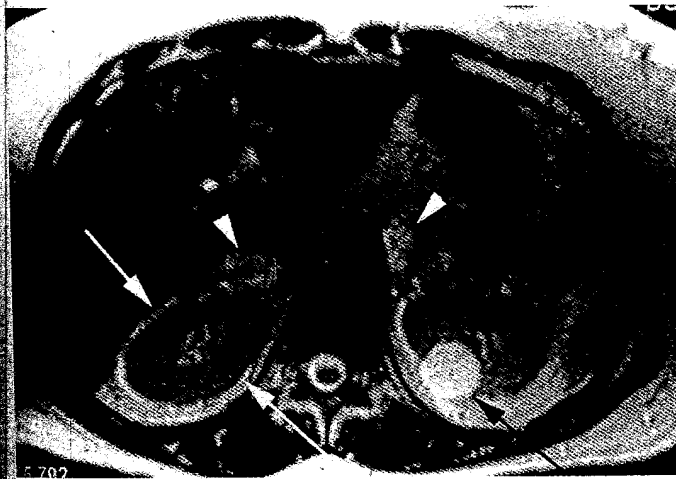


Figure 18-11. Axial T2 FSE image shows alternating bright and dark signal around the kidneys along the frequency-encoding (transverse) direction (*white arrows*). Patient also has bilateral pheochromocytomas (*white arrowheads*), a nonfunctioning islet cell tumor in the tail of pancreas (*wide black arrow*) and a simple cyst in the left kidney (*black arrow*). Patient had von Hippel-Lindau syndrome.

bright due to overlap of water and fat and the top endplate dark due to water alone (Fig. 18-10). If we increase the pixel size, the misregistration artifact will increase.

Figures 18-11 through 18-15 are examples of chemical shift artifact.

Question: What factors increase chemical shift artifact?

Answer:

1. A stronger magnetic field strength
2. A lower BW:

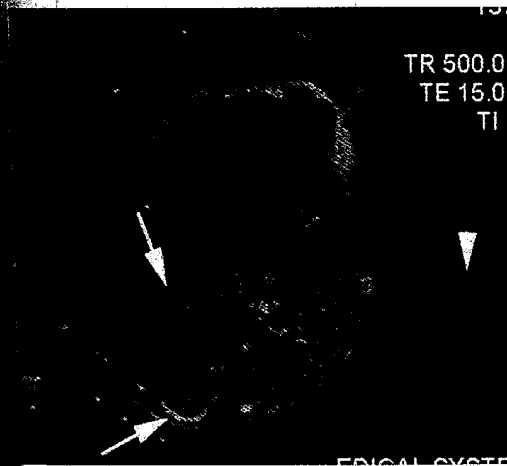


Figure 18-12. Axial T2* gradient-echo image of the knee shows first-order chemical shift (*arrows*) along the frequency direction (anteroposterior). Note that phase-encoding (transverse) direction "ghosting" artifact is also seen (*arrowhead*). A knee effusion is also demonstrated.

between fat and lower frequencies

are relatively higher frequencies

This shifting frequency and sign in turn leads to frequencies

artifact any interface. Also chemical shift artifact

n--in this case higher frequency body would be lower endplate

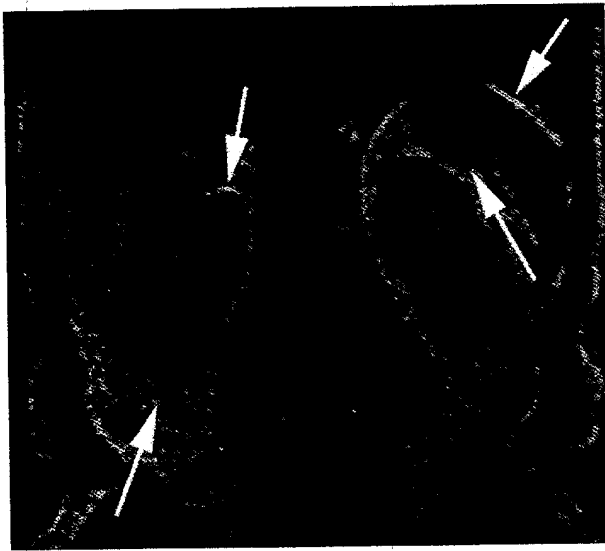


Figure 18-13. Coronal spoiled gradient-echo T1 (TR 93/TE 1.8 msec) image of the abdomen shows the typical alternating bands of dark and bright signal at the fat/water interfaces from chemical shift artifact along the frequency-encoding direction (cranio-caudal).

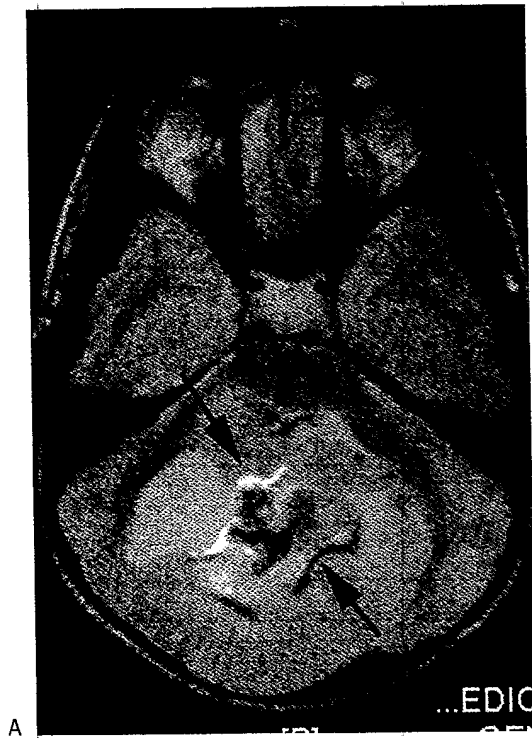


Figure 18-14. PD (A) and T2 (B) images of the posterior fossa show alternating bright and dark bands along the frequency-encoding direction (anteroposterior; *arrows*). Notice that the thickness of the artifact is wider in the T2 image secondary to a bandwidth (BW) of ± 4 kHz versus the PD image's BW of ± 16 kHz. Also note that the T2 image shows only the dark band well since the fat has low signal in this CSE T2 sequence, which minimizes the amount of bright signal. Patient had mature teratoma.

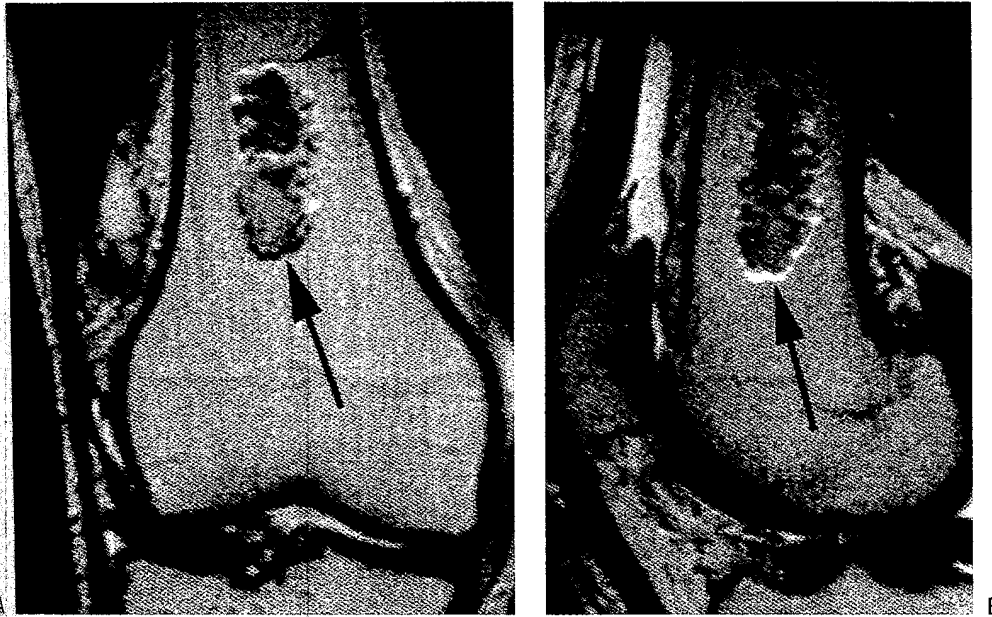


Figure 18-15. Coronal T2 (A) of the knee demonstrates chemical shift artifact in the frequency-encoding direction (craniocaudal) secondary to the fatty marrow juxtaposed to an enchondroma. Additional sagittal T2 (B) image again shows the artifact; however, now the bright and dark bands have swapped directions due to the frequency-encoding direction swapping (e.g., lower frequencies are superior in the coronal and inferior in the sagittal image). Finally, sagittal T2 image with chemical (spectral) fat saturation (C) shows no chemical shift effect.



If we decrease the BW, we have a lower BW per pixel and fewer frequencies/pixel. As an example, if instead of 32 kHz, we have a 16 kHz BW, then

$$\begin{aligned} BW/pixel &= 160 \text{ kHz}/256 \\ &= 62.5 \text{ Hz/pixel} \end{aligned}$$

Now, each pixel covers 62.5 Hz, but the chemical shift is still 220 Hz. Consequently,

$$220 \text{ Hz}/62.5 \text{ Hz/pixel} \cong 4 \text{ pixel misregistration}$$

Decreasing bandwidth results in increased chemical shift artifact.

This is one of the side effects of selecting a lower BW on your scanner. Unfortunately, the chemical shift due to field strength and that

coiled gradient-ec) image of the l alternating bands t the fat/water hift artifact along irection (cranio-

nd dark bands along artifact is wider in the Also note that the T2 which minimizes the

...ED

due to BW are independent and additive; thus, higher field/low BW techniques have the worst chemical shift artifact (Fig. 18-14).

3. Smaller pixels:

If we keep the BW of 32 kHz and the FOV the same but increase the number of frequency-encoding steps to 512 (instead of 256), the pixel bin will have half as many frequencies:

$$\begin{aligned} \text{Pixel bin} &= 32 \text{ kHz}/512 \\ &= 62.5 \text{ Hz/pixel} \end{aligned}$$

again leading to a greater chemical shift, as above (i.e., 4 pixels instead of 2).

Solution. How can you fix chemical shift artifacts?

1. Get rid of fat using fat suppression. If there is no signal from fat, there can be no chemical shift. This can be done with a spectroscopic "fat sat" pulse or a STIR sequence (Fig. 18-15).
2. Increase pixel size by keeping FOV the same and decreasing N_x (trade-off: deteriorates resolution).
3. Lower the magnet's field strength (not practical!).

4. Increase bandwidth (trade-off: lower SNR).
5. Switch phase and frequency direction. This will just change the direction of the chemical shift.
6. Use a long TE (causes more dephasing and less signal from fat).

Chemical Shift of the "Second Kind" phenomenon applies to GRE techniques (Chapters 20 and 21). As discussed previously, fat and water protons precess at slightly different frequencies in the transverse plane (200 Hz at 1.5 T). Because water precesses faster, it is 360° ahead of fat after a short period. Thus, there will be times (TE) when water spins will be totally in phase and when they will be 180° out of phase. And water and fat are in phase every 4.5 msec. This number is derived by the following:

$$\begin{aligned} \text{Frequency difference between fat and water} &= 220 \text{ Hz} \\ \text{Period} &= 1/\text{frequency} = 1/(220 \text{ Hz}) \\ &= 0.0045 \text{ sec} = 4.5 \text{ msec} \end{aligned}$$

In Figure 18-16, fat and water are initially in phase at TE = 0 msec, go out of phase at TE = 2.25 msec, and are back in phase at TE = 4.5 msec. In general, at 1.5 T, fat and H_2O

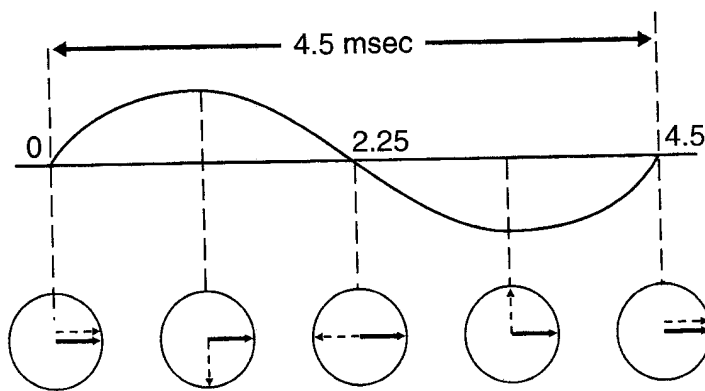
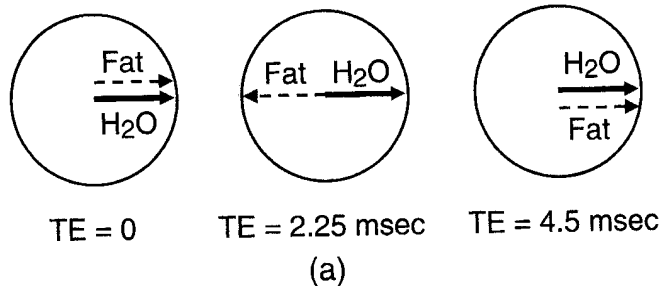


Figure 18-16. Chemical shift effect of the second kind. Fat and water protons get in and out of phase at various values of TE. Specifically, they are in phase at TE of 0, 4.5, and 9 msec. They are 180° out of phase at TE of 2.25, 6.75, and so on. This effect is represented graphically by the sine wave function.

out of phase every 2.25 msec. This is called a **chemical shift effect of the second kind.**

Boundary Effect. If the selected TE is 2.25, 6.75, 11.25, 15.75 msec, and so on, fat and water protons will be out of phase and a dark boundary will be seen around organs that are surrounded by fat (such as the kidneys and muscles). This result is called the **boundary effect, bounce point artifact, or India Ink Etching**, which is caused by chemical shift of the second kind. This type of imaging is referred to as **out-of-phase scanning**, referring to the fact that at these TEs, fat and water spins will be 180° out of phase. This phenomenon does not just occur along the frequency-encoding axis (as with the chemical shift artifact of the first kind) because it is a result of fat and water protons phase cancellation in all directions (Fig. 18-17). (Boundary effect does not occur in conventional SE techniques because

of the presence of the 180° refocusing pulse, which is absent in GRE techniques.)

Remedy

1. Make fat and H₂O in phase by picking appropriate TE.
2. Increase the BW (trade-off: decreases SNR).
3. Use fat suppression.

Truncation Artifact (Gibbs Phenomenon).

This artifact occurs at high contrast interfaces (e.g., skull/brain, cord/cerebrospinal fluid (CSF), meniscus/fluid in the knee) and causes alternating bright and dark bands that may be mistaken for lesions (e.g., pseudo syrinx of the spinal cord or pseudo tear of the knee meniscus).

The cause is inability to approximate exactly a steplike change in the signal intensity due to a limited number of samples or sampling time. The ripples in Figure 18-18 are responsible for

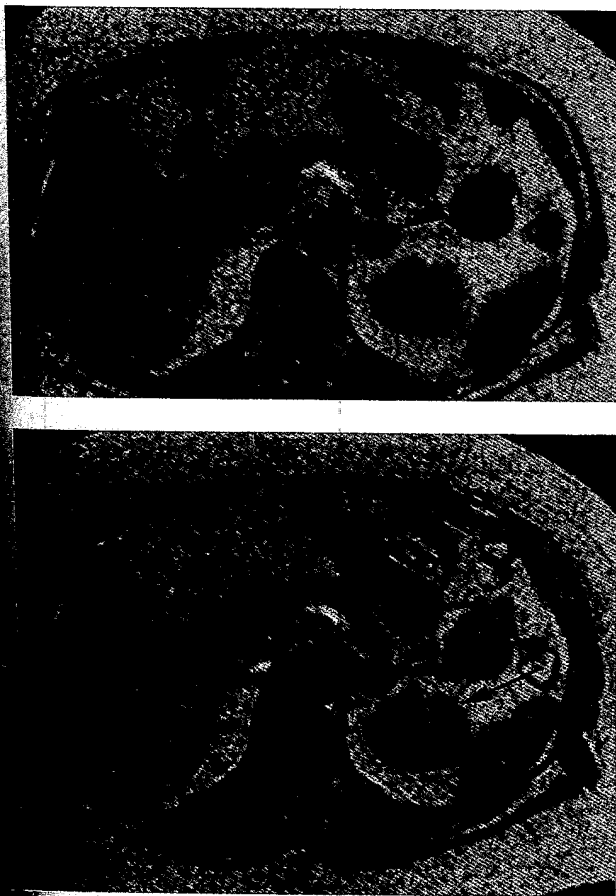


Figure 18-17. In-phase (A) and out-of-phase (B) spoiled gradient T1 images show the “boundary effect” on the out-of-phase images circumferentially at every fat/water interface (arrows in B). Also note that a left adrenal adenoma loses signal substantially on the out-of-phase image (arrowhead).

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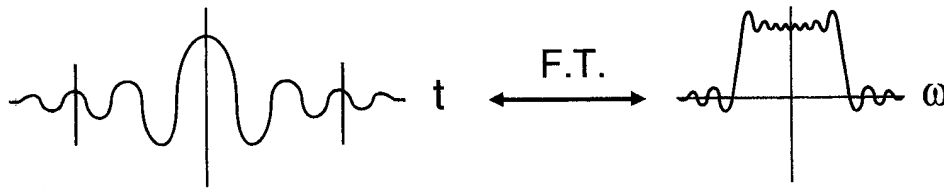


Figure 18-18. Truncation artifact causes a ring-down effect because the FT of a truncated sinc function has ripples at its edges.

the parallel bands seen at such sharp interfaces. This artifact is seen mostly in the phase direction (because we typically have few pixels and lower resolution in phase compared with frequency). Incidentally, the correct term is “truncation artifact.” “Gibbs phenomenon” refers to the infinitely thin discontinuity that still persists with an infinite number of pixel elements. Figures 18-7, 18-19, and 18-20 contain examples of truncation artifact.

Remedy

1. Increase sampling time (\downarrow BW) to reduce the ripples. (Remember, a wider signal in time domain means a narrower one in frequency domain.)

2. Decrease pixel size:

- (a) increasing the number of phase encodes, and
- (b) decreasing the FOV

Partial Volume Artifact. This artifact has the same concept as computerized tomography (CT). To reduce it, decrease the slice thickness (Δz). Figure 18-21 contains an example of partial volume artifact.

Patient-Related Artifact

This artifact is caused by voluntary or involuntary patient motion and by the patient's anatomy. Pulsating motion in vessels is also

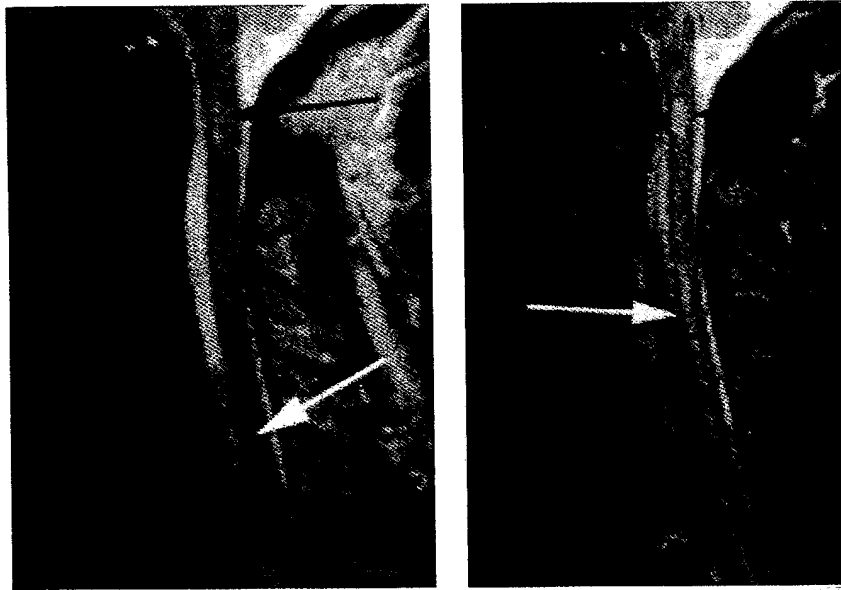


Figure 18-19. Sagittal fat-saturated T2 image (A) shows minimal truncation artifact (*white arrow*); a sagittal STIR image (B) shows wider truncation artifact (*white arrow*). The T2 image was obtained with 224 phase-encoding steps, and the STIR used 192 steps. Patient also had a small nonhemorrhagic cord contusion at C1/2 (*black arrows in A and B*).



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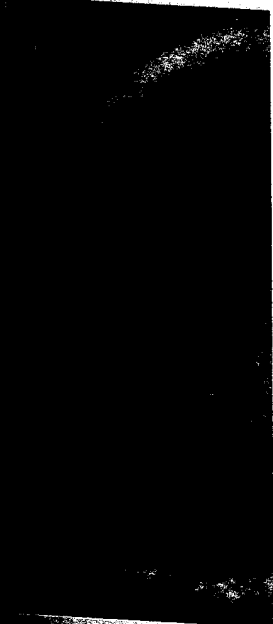
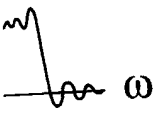


Figure 18-21. Axial FLAIR is
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the lesion (*arrows*). The
was flow-related enhance



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This artifact has the same appearance as the slice thickness artifact. It is an example of partial volume artifact.

Artifact

Partial volume artifact is caused by the patient's voluntary or involuntary movements and by the patient's motion in vessels is also an example of partial volume artifact.

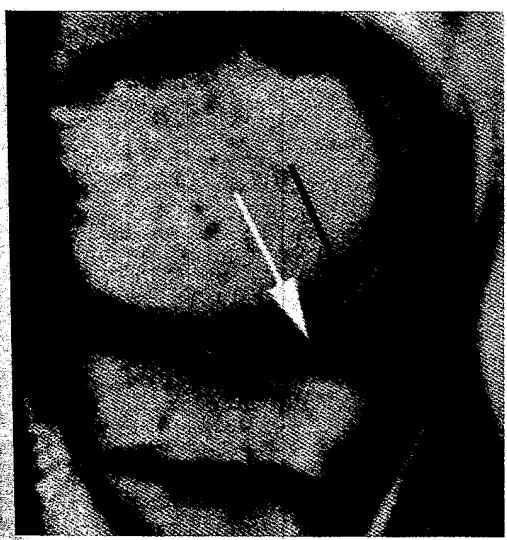


Figure 18-20. Proton density sagittal image of the knee shows truncation artifact mimicking posterior medial meniscus tear (*white arrow*). Note extension of high signal beyond the meniscus (*black arrow*).

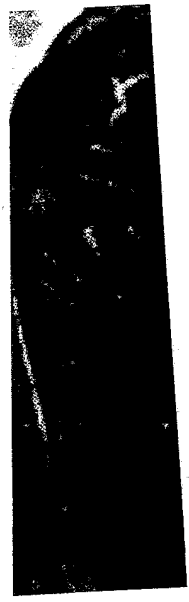
Partial volume artifact is an interesting source of motion-related artifacts. (More on this in later chapters.)

Motion Artifact. Motion artifact is caused by the patient's (voluntary or involuntary) movements (**random**) or by pulsating flow in vessels (**periodic**). We only get motion artifacts in the *phase-encoding* direction.

Question: Why is motion artifact only seen in the phase-encoding direction?

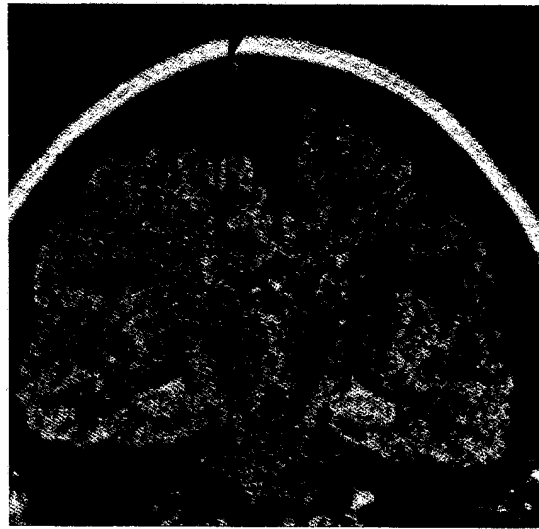
Answer: The reason is twofold:

1. First of all, motion along any magnetic field gradient results in abnormal phase accumulation, which mismaps the signal along the phase-encoding gradient.



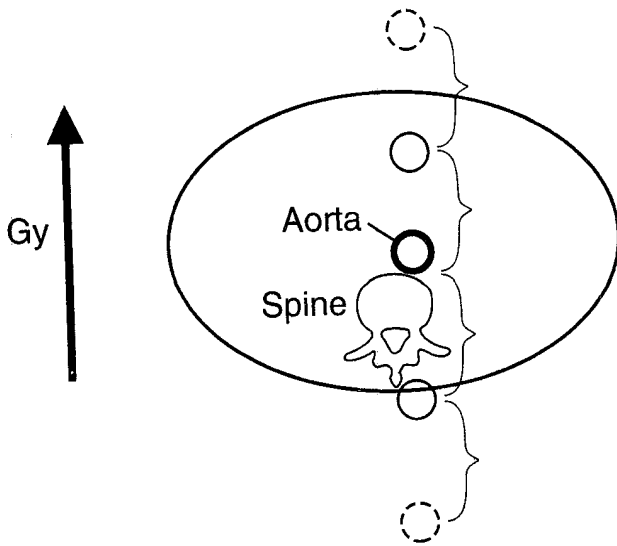
B

Partial volume artifact (*white arrow*). The T2 image of the patient also had a partial volume artifact.



B

Figure 18-21. Axial FLAIR image (A) shows some signal within a right convexity lesion that would not be consistent with a simple arachnoid cyst. Additional coronal FLAIR image (B) shows the signal to be around, but not within the lesion (*arrows*). The high signal on the axial image was due to partial volume averaging. The high signal on the coronal image was flow-related enhancement of CSF flowing around the cyst.



2. Also, there is a significant asymmetry in the data space (see Chapter 13) so that it takes much less time to sample the signal via frequency encoding (on the order of milliseconds) than to do a single phase-encoding step (on the order of seconds). Thus, most motions experienced during clinical MRI are much slower than the rapid sampling process along the frequency-encoding axis. This disparity between frequency- and phase-encoding periods allows motion artifacts to be propagated mainly along the phase-encoding axis. Motion artifacts along the frequency-encoding axis may occur, but they are insignificant (at best, they may cause minimal blurring).

Periodic Motion. Periodic motion is caused by pulsating or periodic motion of vessels, heart, or CSF. In the example in Figure 18-22 (also Fig. 18-23), with a cross section of the body through the aorta, and with the phase-encoding in the AP direction, we will get "ghost" artifacts of the aorta equally separated. The artifacts become fainter with increasing distance from the original structure. The separation (SEP) between the "ghosts" is given by

$$SEP = \frac{(TR)(N_y)(NEX)}{T(\text{motion})}$$

Figure 18-22. Ghost artifacts are equidistant replica of a pulsating structure, such as the aorta, along the phase direction.

Another way of expressing this is

$$SEP = \text{acquisition time} / T(\text{motion})$$

where $T(\text{motion})$ is the period of motion of the object (in this case, the aorta).

Example

The aorta pulsates according to the heart rate. If the heart rate is

$$HR = 60 \text{ beats/min} = 60 \text{ bpm} = 1 \text{ beat/sec}$$

then the period of motion = $T(\text{motion}) = 1 \text{ sec}$



Figure 18-23. Axial STIR image of the aorta. The image shows both the original structure (arrow) and ghost artifacts (arrowheads) in the phase-encoding direction (anteroposterior).

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This means that we have a pulsation every 1 sec. For example, if we have a TR = 500 msec = 0.5 sec, NEX = 1, $N_x = 256$, then

$$SEP = 0.5 \times 256/1 = 128/1 = 128 \text{ pixels}$$

Therefore, we get two ghosts in the image. If the heart rate is 120 bpm, then we get

$$SEP = 128/0.5 = 256 \text{ pixels}$$

and only one ghost.

$$SEP = \frac{(TR)(N_y)(NEX)}{T(\text{motion})}$$

= Separation between ghosts (in pixels)

If we multiply this by pixel size, we get the distance between the "ghosts." Therefore, if we increase TR, number of phase-encoding steps, or NEX, we can increase the separation of ghosts so that they won't be so numerous within the body part we're studying. More rapidly pulsating flow (i.e., shorter period) also causes more separation. If the FOV is too small, the "ghost" images outside the FOV might get "aliased" into the FOV. The ghosts may be dark or bright depending on the phase of the pulsating structure with respect to the phase of the background. If they are in phase, they'll be bright, and if they are out of phase, they'll be dark—Figures 18-24 and 18-25.

Remedy

1. Use spatial presaturation pulses to saturate inflowing protons and reduce the artifacts.

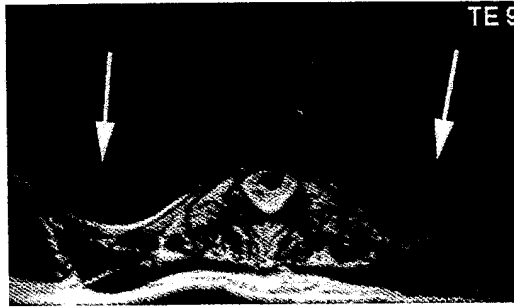


Figure 18-25. Axial T2 image of the thoracic spine shows positive "ghost" artifacts from the CSF that simulate lung nodules.

2. Increase separation between ghosts by increasing TR, N_y , or NEX (which is tantamount to increasing scan time).
3. Swap phase and frequency: although this only changes the direction of the artifacts, it does allow differentiation between a true lesion and an artifact.
4. Use cardiac gating.
5. Use flow compensation.

Random Motion. Random motion is caused by the patient's voluntary or involuntary movements (e.g., breathing, changing position, swallowing, tremors, and coughing). It causes blurring of the image. We may get parallel bands in the phase-encoding direction as well (Fig. 18-26). Although

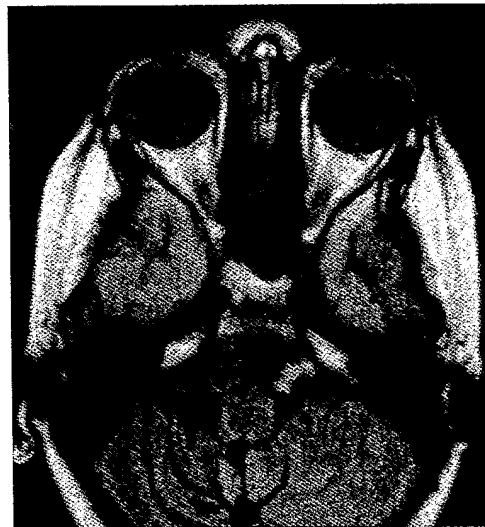


Figure 18-26. Axial FLAIR image shows marked motion artifact in the phase-encoding direction (transverse) from eye movement.

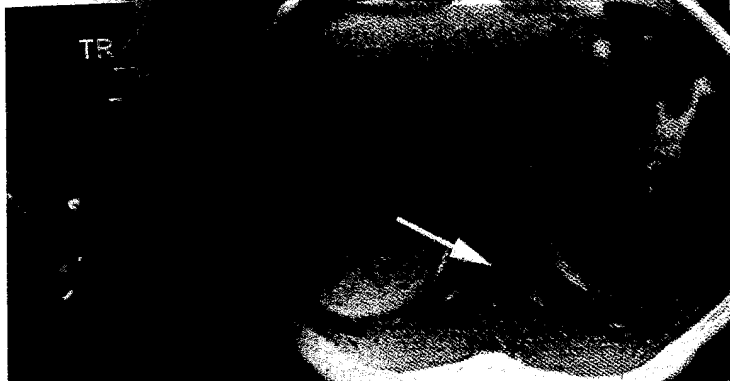


Figure 18-24. Axial spoiled gradient image of the abdomen shows both positive (*black arrow*) and negative (*white arrow*) "ghost" artifacts from the aorta. The negative "ghost" mimics a vertebral body lesion.

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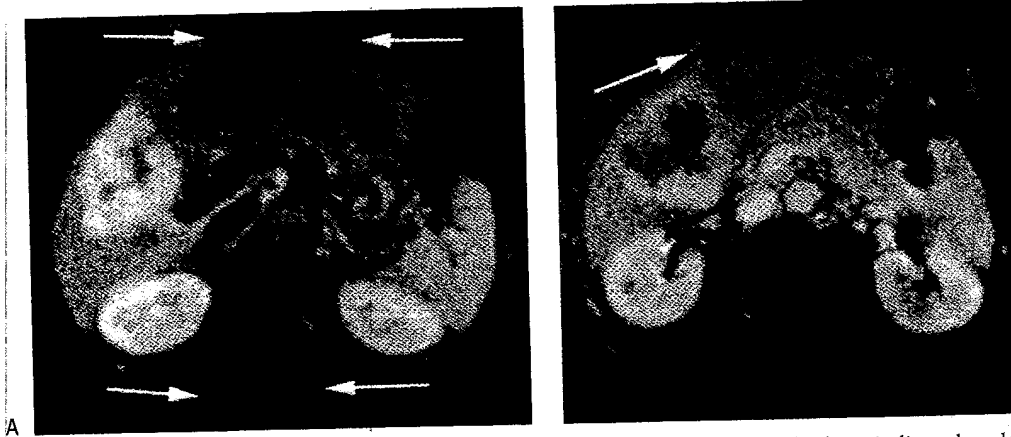


Figure 18-27. Axial CSE T1 image with fat saturation post gadolinium (A) shows both periodic and random phase direction (anteroposterior) artifact (arrows). A breath-hold spoiled gradient T1 image with fat saturation post gadolinium (B) shows near complete resolution of motion artifact. Note that there is increased magnetic susceptibility artifact (arrow in B) in the gradient-echo image. Patient had a liver hemangioma.

this may simulate truncation artifacts, it is different in that truncation causes *fading* parallel bands.

Remedy

1. Patient instruction: Don't move! (probably the most useful remedy)
2. Respiratory compensation (RC) (uses chest wall motion pattern to reorder scan and minimize motion)
3. Use of glucagon in the abdomen to reduce artifacts due to bowel peristalsis
4. Sedation
5. Pain killers
6. Faster scanning (FSE, GRE, EPI, etc.); sequential 2D rather than 3D scanning (see Fig. 18-27 for an example).

CSF Flow Effects. Dephasing of protons due to CSF motion may sometimes simulate a lesion. Flow compensation techniques can reduce this effect. Examples include the following:

1. Pseudo aneurysm of basilar artery due to pulsatile radial motion of CSF around it (Fig. 18-28).
2. Pseudo MS plaques in the brainstem due to CSF flow in the basal cisterns.
3. Pseudo disc herniation, again secondary to CSF flow.

Remedy

1. Be certain that "lesions" are seen on all pulse sequences (artifacts tend to only be seen on one image).
2. Use cardiac gating.
3. Use flow compensation.



Figure 18-28. Axial T2 image shows marked void around the basilar artery mimicking an aneurysm in this 3-year-old patient.



Figure 18-29. Signal void by increased signal on the tendon (dark signal on the tendon).

Magic Angle

points, if a tendon is oriented at a magic angle (55°) relative to the magnetic field, then the tendon will appear bright on PD-weighted images. This artifact is not seen on T2-weighted images because of the high density of the tendon.

Collagen, which is the main component of tendon composition, has anisotropic properties that vary with the orientation of the tendon. This anisotropy is responsible for the signal void seen on their orientation. (The signal void is not a true void, however, have proper orientation.)

the magic angle is increased. If the TE is long. If the TE or PD-weighted signal is increased on this T2 proton density image (mathematical derivation of the signal void to zero at $\theta = 55^\circ$ for examples).



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Figure 18-29. Sagittal PD (A) and T2 (B) fat-saturated images of the knee show magic angle artifact as seen by increased signal on the short TE PD image (arrow in A), whereas the tendon itself is not thickened and has dark signal on the T2 image (B). Joint effusion is also seen. (Courtesy of D. Beall, MD, San Antonio, Texas.)

Magic Angle Artifacts. In imaging the joints, if a tendon is oriented at a certain angle (55°) relative to the main magnetic field, then the tendon appears brighter on T1- and PD-weighted images, but normal on T2-weighted images. This artifactual increased intensity might potentially be confused with pathology.

Collagen, which is responsible for the majority of tendon composition, has an **anisotropic** structure. This anisotropic structure has properties that vary with the direction of measurement and is responsible for dependence of T2 of tendons on their orientation. (**Isotropic** structures, however, have properties independent of their orientation.)

At the magic angle, the T2 of the tendon is slightly increased. This increase is negligible when TE is long. However, when TE is short (as in T1- or PD-weighted images), the result is increased signal intensity. The mathematics behind this T2 prolongation has to do with some of the mathematical terms in the Hamiltonian going to zero at $\theta = 55^\circ$ (see Figs. 18-29 and 18-30 for examples).

MATH: This **magic angle** effect is the solution to the equation

$$3(\cos \theta)^2 - 1 = 0 \rightarrow (\cos \theta)^2 = 1/3$$

or

$$\cos \theta = \sqrt{1/3}$$

which is calculated to be $\theta \approx 55^\circ$. The above equation comes from a complicated mathematical theory dealing with the so-called **dipolar Hamiltonian**.

RF-Related Artifacts

Cross-talk. We have already discussed this issue in previous chapters. The problem arises from the fact that the Fourier transform (FT) of the RF pulse is not a perfect rectangle but rather has side lobes (Fig. 18-31). We shall use a simpler version of the RF profile, as in Figure 18-32. If we consider two adjacent slices, there will be an overlap in the FT of their RF pulses (Fig. 18-32). Cross-talk causes the effective TR per slice to decrease (due to saturation of protons by the RF signals for adjacent slices). Thus, more

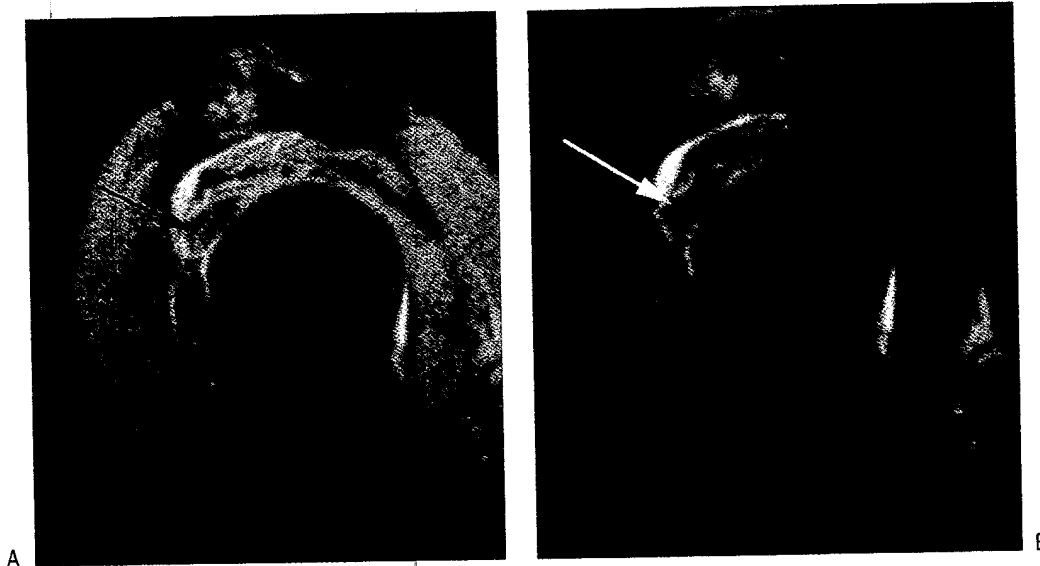


Figure 18-30. Angled sagittal PD (A) and T2 (B) fat-saturated images of the shoulder demonstrate magic angle artifact of the intra-articular biceps tendon. There is increased signal on the PD image (A), whereas the tendon has dark signal and overall normal appearance on the T2 image. Acromioclavicular joint high signal is from osteoarthritis changes. (Courtesy of D. Beall, MD, San Antonio, Texas.)

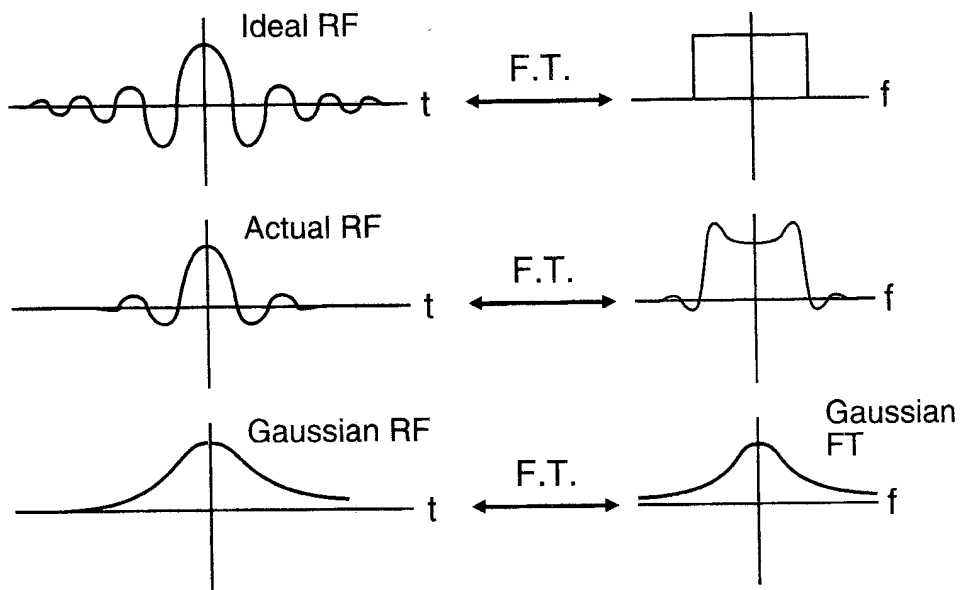


Figure 18-31. The actual RF has a finite time span, yielding side lobes or rings. A Gaussian RF pulse has a Gaussian FT.

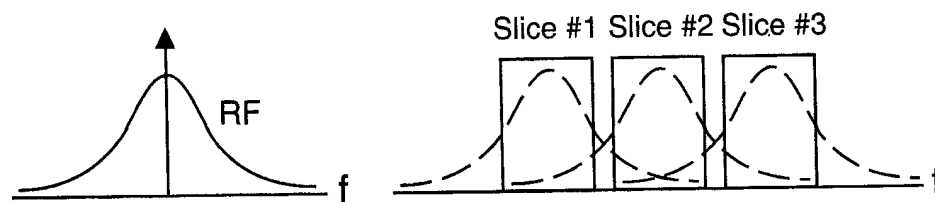
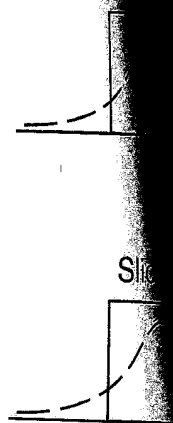


Figure 18-32. Side lobes of the FT of RF pulses (such as in the case of Gaussian curves) may overlap, causing cross-talk.

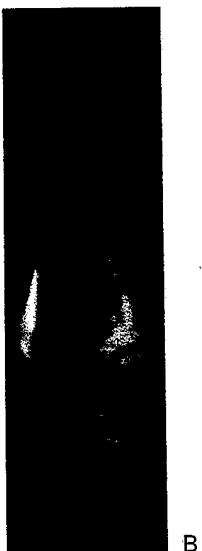


T1 weighting will be problematic for PD. Also, due to reduced SNR, the signal will decrease.

In short, cross-talk and decreased SNR.

Remedy

1. Gaps can be used between slices (Fig. 18-33).
2. Two acquisitions can be interleaved.
3. The RF pulse can be made more rectangular. Let's discuss this further.
 1. If we increase the slice thickness, we reduce cross-talk. However, this is an increase in the slice thickness and the increase in the slice thickness will cause small lesions to be missed.
 2. It doesn't matter how many slices (the slices) are used. The slice thickness will be the same range and way to eliminate cross-talk.



B

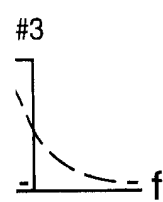
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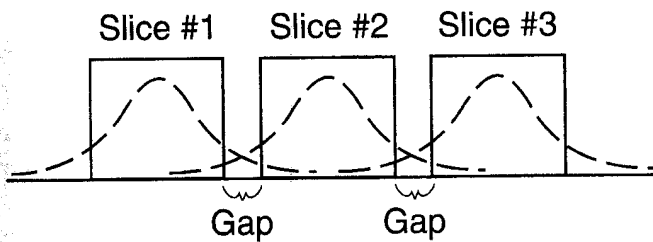


Figure 18-33. To reduce cross-talk, gaps are introduced between slices.

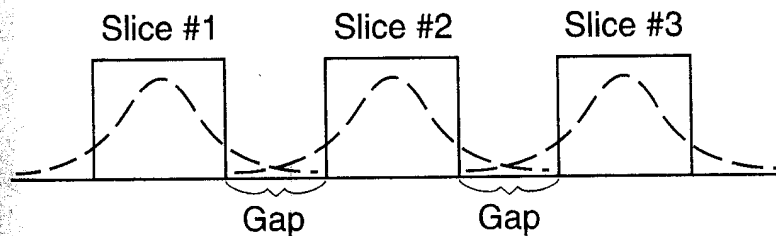


Figure 18-34. The larger the interslice gap is, the less cross-talk is observed.

T1 weighting will result (this is particularly problematic for PD- and T2-weighted images). Also, due to reduced effective TR, the SNR will decrease.

In short, cross-talk causes increased T1 weighting and decreased SNR.

Remedy

1. Gaps can be introduced between adjacent slices (Fig. 18-33).
2. Two acquisitions with 100% gaps can be interleaved.
3. The RF pulse can be lengthened to achieve a more rectangular pulse profile.

Let's discuss these in more detail:

1. If we increase the gap between slices, we reduce cross-talk (Fig. 18-34). The trade-off is an increase in the unsampled volume and the increased potential for missing a small lesion located within the gap.
2. It doesn't matter which way we order the slices (we can do slice 1, then slice 3, then slice 2, etc.). Adjacent slices still will be sharing a certain frequency range and cause cross-talk. The only way to eliminate cross-talk is to do two

separate sequences each with a 100% gap, such as:

First sequence: odd slices 1, 3, 5, 7, ...

Next sequence: even slices 2, 4, 6, 8, ...

This is the technique of "true" interleaving. Interleaving within a single sequence will not totally eliminate cross-talk, although it might reduce it somewhat. The interslice gap in this case is usually 25% to 50% of the slice thickness and a simple sequence is performed. Interleaving in the true sense, however, will double the scan time because it employs two separate sequences.

Contiguous Slices. The RF pulse on newer scanners more closely approximates a rectangular wave (Fig. 18-35). With this feature, we may have a 10% to 20% interslice gap without significant cross-talk. However, with reduced interslice gap, we reduce coverage and need more slices. Remember, we are talking trade-offs again.

RF Zipper Artifact. This artifact is one form of central artifacts (the other form is RF feedthrough,



Figure 18-35. The closer the profile of the RF pulse (actually its FT) is to a rectangle, the better we can achieve contiguous slices without encountering cross-talk.

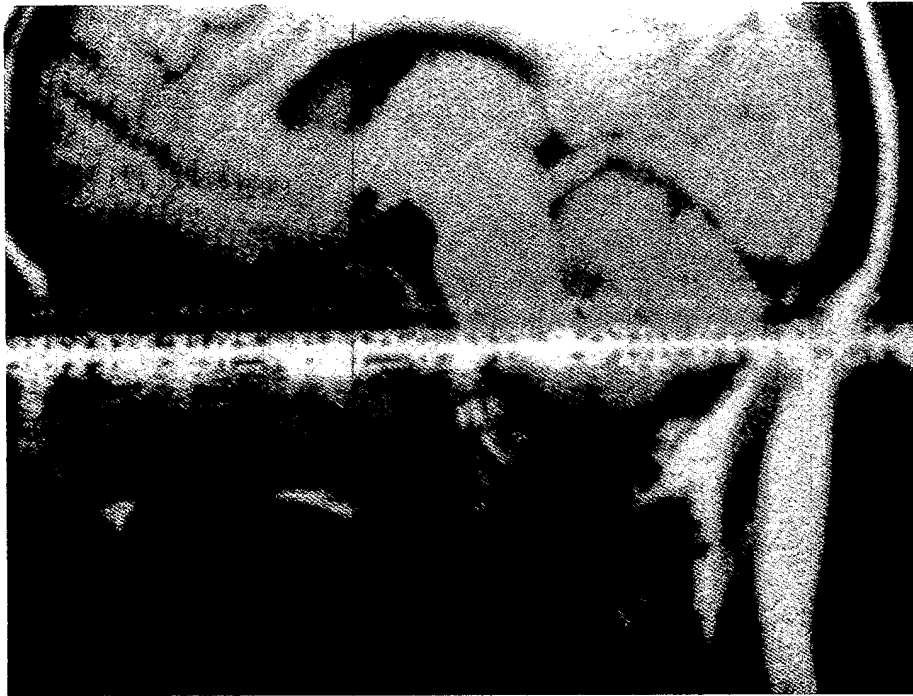


Figure 18-36. Zipper artifact at zero phase.

discussed later). They are referred to as **zippers** due to the formation of a central stripe of alternating bright and dark spots along the *frequency-encoding* axis (at zero phase), as in Figure 18-36. Two sources of zipper artifacts are discussed here.

FID Artifacts. Free induction decay (FID) artifacts occur due to overlapping of side lobes of the 180° pulse with the FID, before it has had a chance to completely decay (Fig. 18-37). This overlapping causes a “zipper” artifact along the *frequency-encoding* direction.

Remedy

1. Increase the TE (increases the separation between the FID and the 180° RF pulse).
2. Increase slice thickness (Δz). This in effect results from selecting a wide RF BW, which narrows the RF signal in the time domain, thus lowering chances for overlap.

Stimulated Echo. This artifact also appears as a narrow- or wide-band noise in the center along the *frequency-encoding* axis. The mechanism is similar to FID artifacts. In this case, imperfect RF

pulses of adjacent slices or imperfect 90° – 180° – 180° pulses of a dual-echo sequence form a stimulated echo that may not be phase-encoded, thus appearing in the central line along the *frequency-encoding* axis.

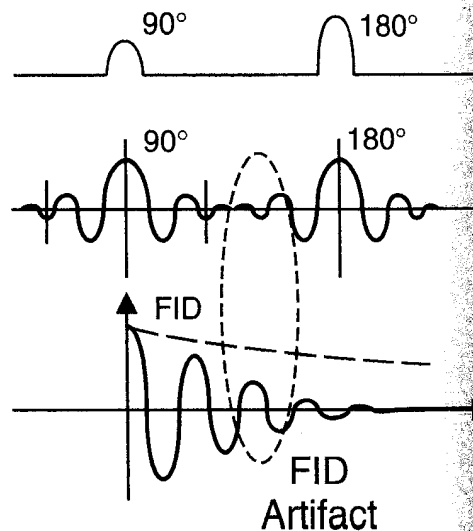


Figure 18-37. FID artifact. The side lobes of the 180° and the FID may overlap, causing a zipper artifact at zero frequency along the phase direction.



Remedy

1. Use spoiler gradient
2. Adjust the trans
3. Call the service

RF Feedthrough artifact occurs when not completely gated and “feeds” through 1 as a “zipper” stripe axis at zero frequency

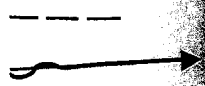
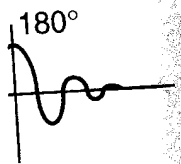
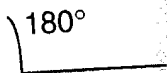
Remedy. Alternate the RF pulses by 180° on averaged phase-alternally eliminate RF fee

RF Noise. RF noise external RF noise (station, a flickering



Figure 18-38. RF feedthrough causes a zipper artifact at zero frequency along the phase direction.

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causing a zipper arti
phase direction.

Remedy

1. Use spoiler gradients.
2. Adjust the transmitter.
3. Call the service engineer.

RF Feedthrough Zipper Artifact. This artifact occurs when the *excitation* RF pulse is not completely gated off during data acquisition and "feeds" through the receiver coil. It appears as a "zipper" stripe along the *phase-encoding* axis at zero frequency (Fig. 18-38).

Remedy. Alternate the phase of the excitation pulses by 180° on successive acquisitions; the averaged phase-alternated excitations will essentially eliminate RF feedthrough.

RF Noise. RF noise is caused by unwanted external RF noise (e.g., TV channel, a radio station, a flickering fluorescent light, patient

electronic monitoring equipment). It is similar to RF feedthrough except that it occurs at the specific frequency (or frequencies) of the unwanted RF pulse(s) rather than at zero frequency (Fig. 18-39).

Remedy

1. Improve RF shielding.
2. Remove monitoring devices if possible.
3. Shut the door of the magnet room!

External Magnetic Field Artifacts

Artifacts related to B_0 are usually caused by magnetic inhomogeneities. These nonuniformities are usually due to improper shimming, environmental factors, or the far extremes of newer short-bore magnets. This can lead to

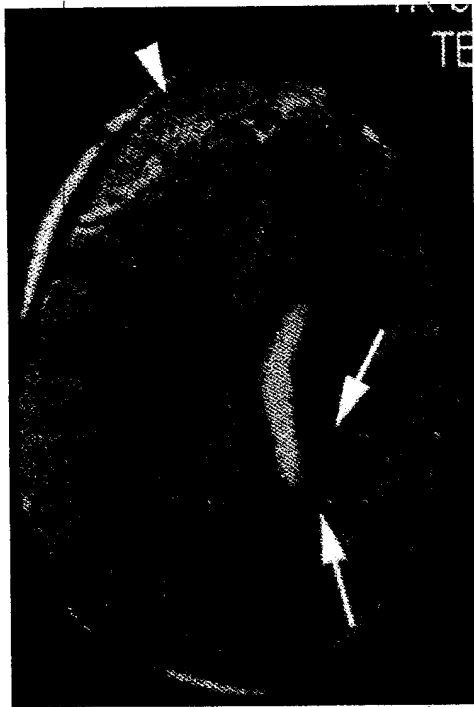


Figure 18-39. Axial T2 image shows RF noise (arrows) from monitoring devices in this recent post-operative patient. There is also an epidural hematoma (arrowhead).

image distortion (Fig. 18-40). They can be reduced in SE and FSE imaging by using 180° refocusing pulses. They can be a source of image inhomogeneity when a fat suppression technique is used.

In GRE imaging, small spatial nonuniformities cause **moiré fringes** (zebra pattern) due to the overlay of the primary image and aliased overlay (Fig. 18-41).

Remedy. Appropriate **shimming coils** (auto shimming) can minimize the problem.

Magnetic Susceptibility Artifacts

As discussed in Chapter 2, all substances get magnetized to a degree when placed in a magnetic field, and their **magnetic susceptibility** (denoted by the Greek symbol χ) is a measure of how magnetized they get.

There are three types of substances—each with a different magnetic susceptibility—commonly dealt with in MRI: paramagnetic, diamagnetic, and ferromagnetic. These substances were described in Chapter 2 and are briefly reviewed here:

1. **Diamagnetic** substances with no unpaired electrons have negative magnetic susceptibility χ (i.e., $\chi < 0$ and $\mu = 1 + \chi < 1$). They are basically nonmagnetic. The vast majority of tissues in the body have this property.
2. **Paramagnetic** substances contain unpaired electrons, have a small positive χ (i.e., $\chi > 0$ and $\mu > 1$), and are weakly attracted by the external magnetic field. The rare earth element **gadolinium** (Gd) with seven unpaired electrons is a strong paramagnetic substance. Gd is a member of the **lanthanide** group in the periodic table. The rare-earth element **dysprosium** (Dy) is another strong paramagnetic substance that belongs to this group. Certain breakdown products of hemoglobin are paramagnetic: deoxyhemoglobin has four unpaired electrons and methemoglobin has five. Hemosiderin, the end stage of hemorrhage, contains, in comparison, more than 10,000 unpaired electrons. It is in a group of substances referred to as **superparamagnetic**, with magnetic susceptibilities 100 to 1000 times stronger than paramagnetic substances.
3. **Ferromagnetic** substances are strongly attracted by a magnetic field and have a large positive χ , even larger than that of superparamagnetic substances. Three types of ferromagnets are known: iron (Fe), cobalt (Co), and nickel (Ni). Susceptibility artifacts in MRI occur at interfaces of differing magnetic susceptibilities, such as at tissue-air and tissue-fat interfaces (examples include paranasal sinuses, skull base, and sella). These differences in susceptibilities lead to a distortion in the local magnetic environment, causing dephasing of spins, with signal loss, mismatching (artifacts), and poor chemical fat saturation

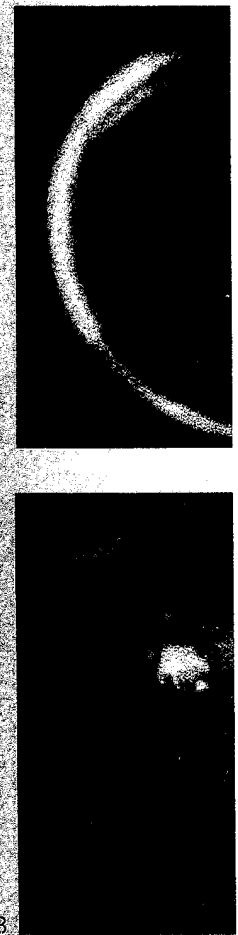


Figure 18-40. Axial T2 fat saturation secondary slice more inferiorly (B) in patient (C), a sagittal T1 image on a short-bore magnet. Patient

(Figs. 18-42 through 18-44) substances (such as foreign bodies), with artifacts, lead to susceptibility artifacts (Figs. 18-

Question: Which MRI sequences are sensitive to magnetic susceptibility artifacts?
Answer: In decreased signal-to-noise ratio imaging (EPI), gradient echo imaging (GRE), and conventional spin echo imaging (FSE). FSE is less sensitive to susceptibility effects. Multiple refocusing

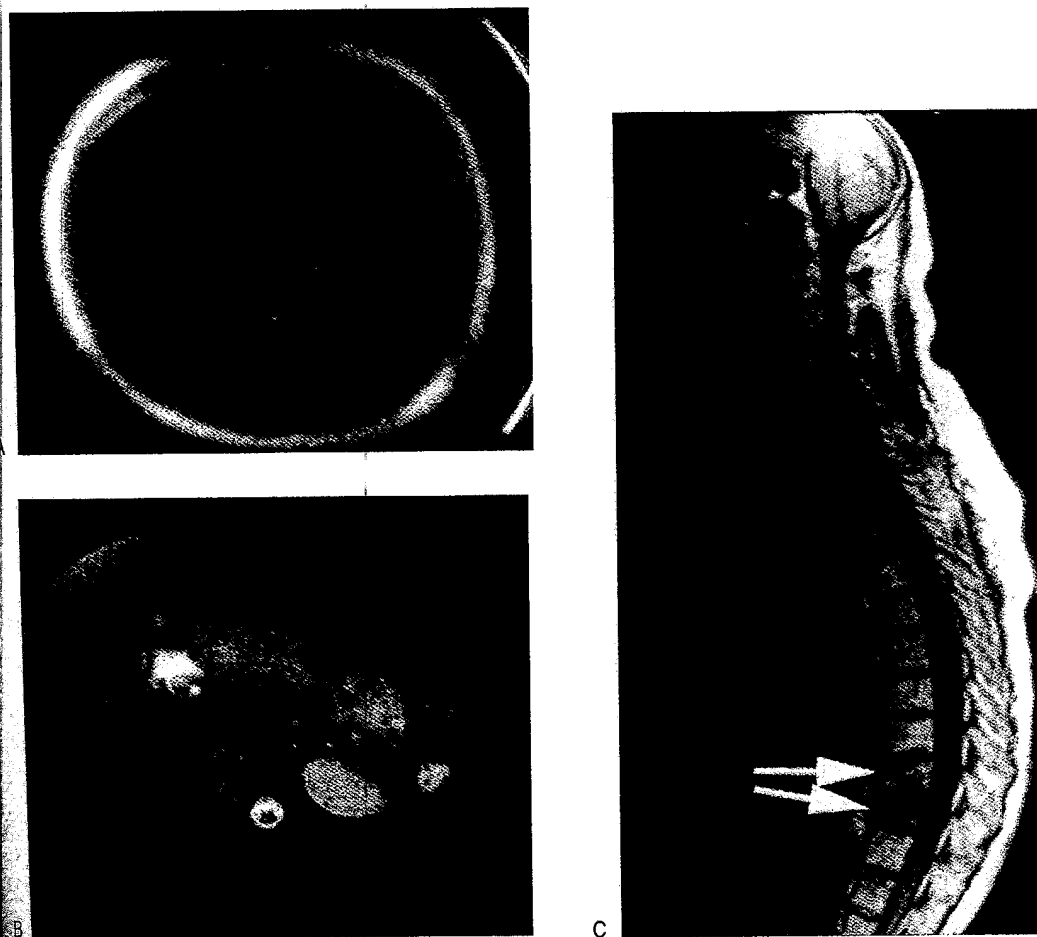


Figure 18-40. Axial T2 fat-suppressed image (A) shows distortion of the upper abdomen and lack of effective fat saturation secondary to magnetic field inhomogeneities at the fringe of a short-bore magnet. Comparative slice more inferiorly (B) has expected fat saturation and normal appearance without distortion. In another patient (C), a sagittal T1 image of the spine shows distortion of the extreme cranial and caudal features in a short-bore magnet. Patient also had two lower thoracic spine compression fractures (*arrows*).

(Figs. 18-42 through 18-44). Ferromagnetic substances (such as metallic clips and foreign bodies), with their large susceptibilities, lead to substantial field distortion and artifacts (Figs. 18-45 through 18-47).

Question: Which MR technique is least sensitive to magnetic susceptibility effects?

Answer: In decreasing order, echo planar imaging (EPI), gradient-echo (GRE) imaging, conventional spin echo (CSE), and fast spin echo (FSE). FSE is least sensitive to magnetic susceptibility effects due to the presence of multiple refocusing 180° gradients.

Gradient-Related Artifacts

Eddy Currents. Eddy currents are small electric currents that are generated when the gradients are rapidly switched on and off (i.e., the resulting sudden rises and falls in the magnetic field produce electric currents). These currents will result in a distortion in the gradient profile (Fig. 18-48) and in turn cause artifacts in the image.

Nonlinearities. Ideal gradients are linear. However, as in other aspects of life, there is no such thing as an ideal gradient. These nonlinearities cause local magnetic distortions and image



Figure 18-41. Coronal postgadolinium spoiled gradient T1 image with chemical (spectral) fat saturation (A) shows moiré fringes (*black arrows*). Comparison with single-shot FSE (SSFSE) T2 (B) image demonstrates decreased artifact. Also note ghosting on image A from the heart and aorta (*white arrow*) and increased magnetic susceptibility from an inferior vena cava (IVC) filter (*white arrowhead*).

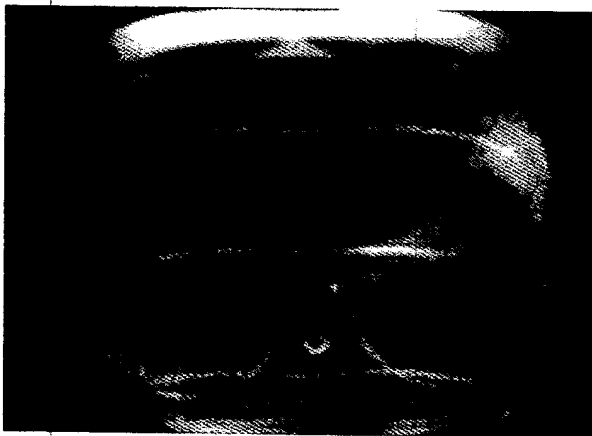


Figure 18-42. Axial T2 FSE with inhomogeneous fat saturation shows "ghosting" artifact from the unsaturated anterior abdominal subcutaneous fat.

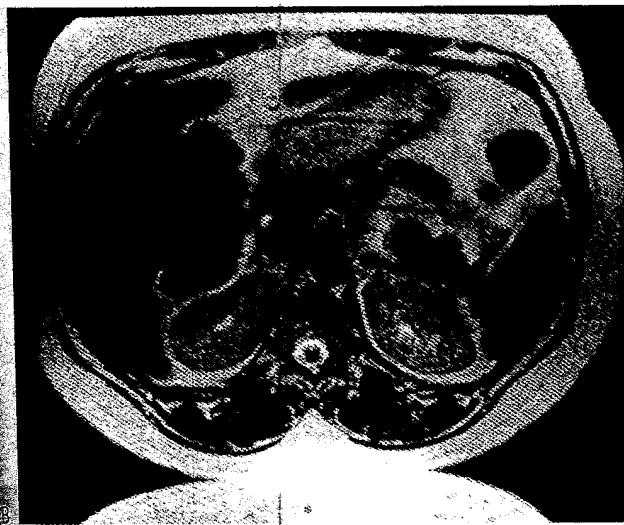


Figure 18-43. Axial postgadolinium fat-saturated gradient-echo T1 image of the abdomen (A) shows “blooming” artifact from the interface of the diamagnetic gas and the adjacent soft tissues (best at the splenic flexure). This effect is minimized on the T2 FSE (B) image. Note aliasing in the phase-encoding (anteroposterior) direction on both images. Image A also has inhomogeneous fat saturation at the diamagnetic interface.



Figure 18-44. Coronal postgadolinium gradient-echo T1 image with fat saturation shows magnetic susceptibility from the densely concentrated paramagnetic substance gadolinium, resulting in dark renal collecting systems with a fringe of bright signal. Also note mild moiré fringes artifact.

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ial T2 FSE with inhom
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Figure 18-45. Axial T2 FSE image shows metallic susceptibility artifact from an MRI compatible aneurysm clip in the area of the left internal carotid terminus.



Figure 18-46. Coronal T1 image shows metallic susceptibility from metallic foreign body at the base of the fifth digit.

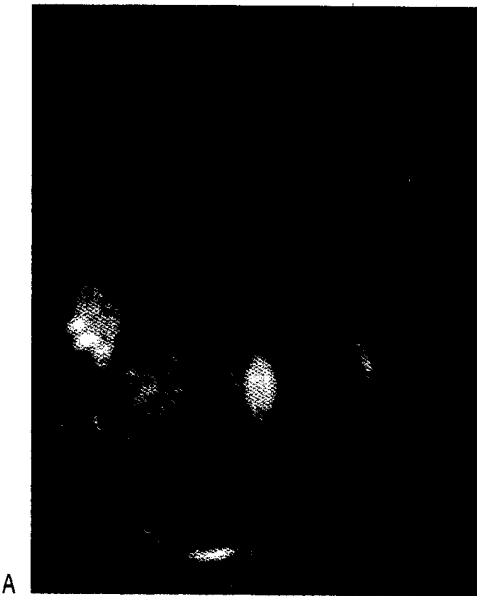


Figure 18-47. Axial EPI B₀ (A), CSE T2 (B), CSE PD (C), and FSE T2 (D) images show the varying effects of different pulse sequences on metallic susceptibility in a patient with dental braces. The EPI is the worst. The CSE T2 is worse than the PD due to a lower BW (± 4 kHz) for the T2 versus the higher BW (± 16 kHz) for the PD. Finally, the T2 FSE is the best (BW still ± 16 kHz), secondary to multiple 180° refocusing pulses.

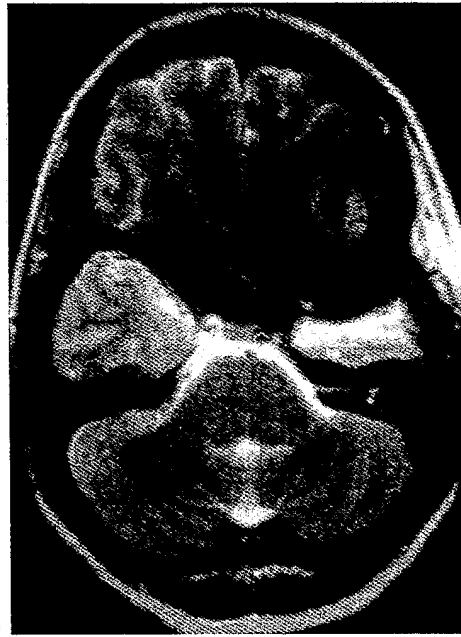


Figure 18-47. (Continued.)

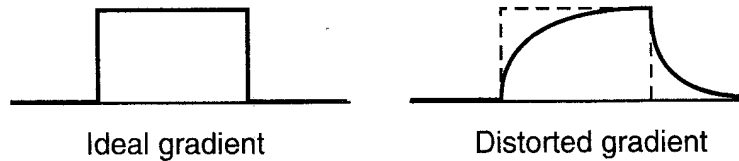


Figure 18-48. Eddy currents result from rapid on-and-off switching of the gradient and cause distortion in the gradient profile and thus the image.

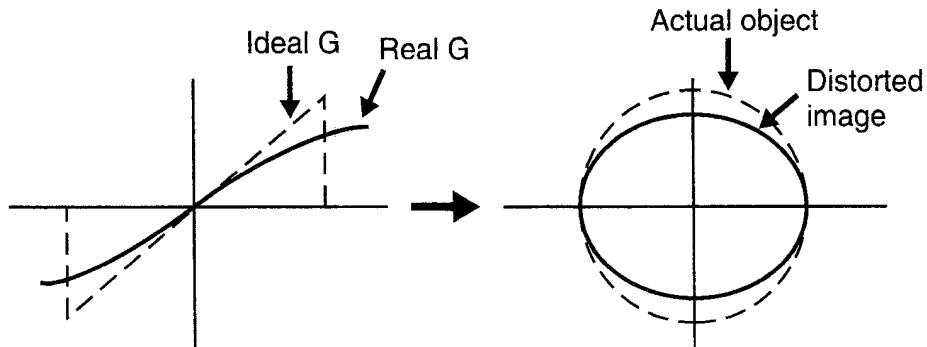


Figure 18-49. Nonlinearities in the gradient cause distortion in the image. For instance, a circle may appear elliptical.

artifacts. The effect is similar to artifacts related to B_0 inhomogeneities.

Geometric Distortion. Geometric distortion is a consequence of gradient nonlinearities or gradient power drop-off. Figure 18-49 illustrates this concept. The real gradient

has dampened peaks, causing image distortion (e.g., a circle may appear elliptical). (Figure 18-50 is an example due to gradient nonlinearities in the more demanding echo planar sequence.) If you find this is a problem, then you need to call your service engineer to fix it.

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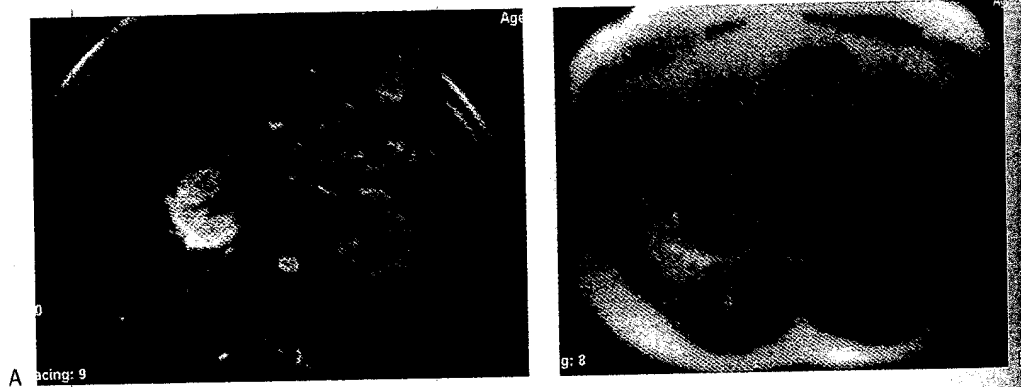


Figure 18-50. Axial T2-weighted EPI image (A) of the abdomen shows geometric distortion of the normal ovoid shape of the abdomen. An accompanying SSFSE T2 image (B) shows this patient's true shape.

Errors in the Data

Errors in the data are caused by a single calculation error in processing the data related to the k-space of a single slice. The result is a crisscross striation artifact that is present across a single image and not present on any other image. (See Figure 18-51.)

Remedy

1. Delete the discrete error and average out the neighboring data.
2. Simply repeating the sequence solves the problem.



Figure 18-51. Axial T2-weighted image of the cervical spine on a 0.23 T magnet shows diagonal lines coursing throughout the image related to a single data point error in k-space.

Flow-Related Artifacts

Motion artifacts were discussed previously, including periodic flow artifacts. Other flow-related phenomena are discussed in Chapters 26 and 27.

Dielectric Effects

As the wavelength of the radiowave approaches the dimensions of the body part being imaged, there can be areas of brightening and darkening due to standing waves. This is most pronounced at 3 T and above. Since the body is a conducting medium, the artifact is often called "dielectric effect" (Fig. 18-52). It seems to be worse in large body parts, that is, the abdomen, and seems to be quite common when ascites is present. The solution for dielectric effects is parallel transmission or "transmit SENSE."

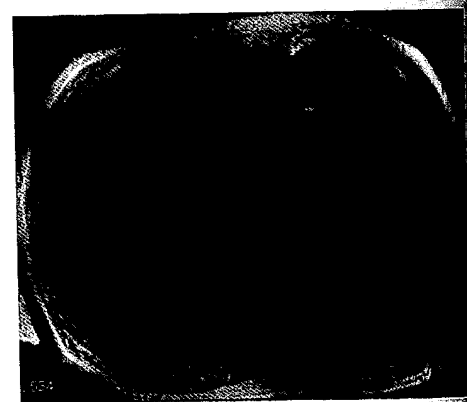


Figure 18-52. 3.0 T proton density-weighted image of the abdomen demonstrates cirrhosis and ascites with regional area of decreased signal intensity due to dielectric effect or "standing waves" (arrow).

Key Points

In this chapter, we discuss the most important causes of artifacts in MR imaging that every technologist should be aware of. For a list of artifacts, see the following sections.

Questions

- 18-1** Regarding chemical shift artifacts:
- (a) protons in fat are shifted higher than those in water
 - (b) at 1.5 T it is 3.5 pixels
 - (c) at 1.5 T and 256 × 256 matrix, it is 3.5 pixels
 - (d) all of the above
 - (e) only (b) and (c)

- 18-2** (a) Determine the relationship between the following parameters:

B_0	0.5 T
BW	50 kHz
Δf	10 kHz
Δx	4 kHz

- (b) Repeat this experiment with a slice thickness of 240 mm.

- (c) What is your conclusion?

- 18-3** Periodic motion artifacts along the phase-encoding direction. The number of pixels in the phase-encoding direction is given by:

$$N_y = \frac{TR}{T_m} \times \frac{1}{\Delta t}$$

where T_m = period of motion.

- (a) Calculate the number of pixels in the phase-encoding direction (HR = 60 bpm, TR = 200 ms, $N_x = 256$).

- (b) What is the relationship between the number of pixels in the phase-encoding direction and the number of pixels in the frequency-encoding direction along the x-axis?

- (c) What is the relationship between the number of pixels in the phase-encoding direction and the number of pixels in the frequency-encoding direction along the y-axis?

Key Points

In this chapter, we discussed the most common and important causes of potential artifacts in MR imaging that every MR radiologist should be aware of. For a list of these artifacts, refer to

the *Introduction* in this chapter. There are a few other less significant sources of artifacts that were not discussed in this chapter.

Questions

18-1 Regarding chemical shift artifact:

- (a) protons in fat resonate at 3.5 ppm higher than protons in water
- (b) at 1.5 T it is about 220 Hz
- (c) at 1.5 T and for a 32 kHz BW and 256×256 matrix, it is about 2 pixels
- (d) all of the above
- (e) only (b) and (c)

18-2 (a) Determine the chemical shifts (in terms of numbers of pixels) for the following situations (assume 256 frequency-encoding steps):

B_0	0.2 T	0.5 T	1.0 T	1.5 T
BW				
50 kHz				
10 kHz				
4 kHz				

- (b) Repeat this table in terms of millimeters, given an FOV = 24 cm = 240 mm.
- (c) What is your conclusion?

18-3 Periodic motion causes "ghost" artifacts along the phase-encoding direction. The number of *pixels* between two consecutive ghosts is given by (SEP = separation)

$$\text{SEP} = \text{TR} \cdot \text{NEX} \cdot N_y / T$$

$$= \text{acquisition time} / T$$

where T = period of the oscillating motion.

- (a) Calculate SEP for the aortic ghosts (HR = 60 bpm, i.e., $T = 1$ sec) when TR = 200 msec = 0.2 sec, NEX = 1, $N_y = 256$.
- (b) What is the maximum number of ghosts you could potentially see along the phase-encoding axis in example (a)?
- (c) What is the effect of increasing NEX?

18-4 Wraparound can be reduced by all of the following *except*:

- (a) using a surface coil
- (b) decreasing the FOV
- (c) using presaturation pulses
- (d) using a no phase wrap option
- (e) using a no frequency wrap option

18-5 Truncation artifacts can be reduced by all of the following *except*:

- (a) decreasing pixel size
- (b) increasing sampling time
- (c) increasing N_y
- (d) increasing FOV

18-6 T/F Chemical shift artifact causes a bright band toward the higher frequency and a dark band toward the lower frequency at a water/fat interface.

18-7 Chemical shift is decreased by all of the following *except*:

- (a) lowering the bandwidth
- (b) using a fat suppression technique
- (c) using a lower field magnet
- (d) using a longer TE

18-8 T/F Fat and water protons get out of phase at TE of odd multiples of 2.25 msec.

18-9 Chemical shift in general can be represented by

- (a) $3.5 \times 10^{-6} \gamma B \cdot N_x / \text{BW}$
- (b) $3.5 \times 10^{-6} \gamma B \cdot \text{FOV} / \text{BW}$
- (c) $3.5 \times 10^{-6} \gamma B / (\text{BW} \cdot N_x)$
- (d) both (a) and (b)

18-10 (a) Calculate the separation (in pixels and mm) between aortic ghosts for TR 500 msec, NEX 1, N_y 128, HR 80 bpm, and FOV 20 cm.

- (b) What is the maximum number of ghosts you could potentially see within the FOV?

- 18-11** Paramagnetic elements include all of the following *except*:
- (a) gadolinium
 - (b) dysprosium
 - (c) cobalt
 - (d) methemoglobin
 - (e) both (c) and (d)
- 18-12** Motion artifacts can be reduced by all of the following *except*:
- (a) fast scanning
 - (b) sedation
 - (c) 3D imaging
 - (d) flow compensation
- 18-13** CSF flow can lead to all of the following artifacts *except*:
- (a) pseudo MS plaques in the brainstem
 - (b) pseudo disc herniation
 - (c) pseudo basilar artery aneurysm
 - (d) pseudo syrinx
- 18-14** T/F Magic angle artifact demonstrates increased signal on proton density images in a tendon that is positioned perpendicular to the main magnetic field.
- 18-15** Cross-talk artifact can be reduced by all of the following *except*:
- (a) increasing the gradient strength
 - (b) increasing interslice gaps
 - (c) double acquisition with 100% gaps interleaved
 - (d) improving the RF profile
- 18-16** The number of ghost artifacts can be reduced by all of the following *except*:
- (a) flow compensation
 - (b) presaturation pulses
 - (c) decreasing N_y
 - (d) increasing TR
- 18-17** Truncation artifacts include
- (a) pseudo meniscal tear
 - (b) pseudo syrinx
 - (c) pseudo MS plaques
 - (d) all of the above
 - (e) only (a) and (b)
 - (f) only (a) and (c)
- 18-18** T/F Motion artifacts occur only along the phase-encoding direction.