磁振造影專業基礎課程

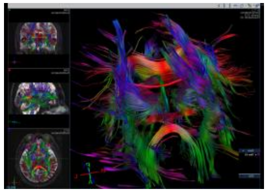
教育主題：磁振造影專業課程

主辦單位：中華民國醫事放射學會、國泰綜合醫院

上課時間：民國 105 年 6 月 19 日(星期日)

上課地點：台北國泰醫院

MRI 基本物理原理



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磁振造影的特性

- **無放射性**
- **高解析度**
- **高組織間對比**
- **多重切面(Sagittal view、Coronal view、Axial view)**

---- 未來百年內醫學影像的主流

何謂MRI？

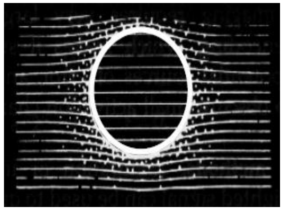
- M (magnetic)：訊號的來源，人體中小磁鐵的磁化。
- R (resonance)：共振，小磁鐵激發偵測的原理，小磁鐵和射頻脈衝間的交互作用。
- I (imaging) ：訊號轉為影像的方式。

磁化 (Magnetization)

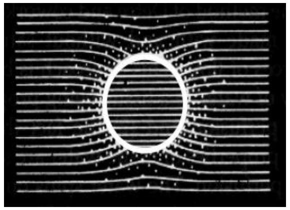
Magentic susceptibility

- 所有物質放在磁場中時，均有一定的磁化程度，而一個物質的感磁性是用來衡量他們被磁化了多少。
- MRI所使用的物質共可分為三種不同的感磁性-順磁性、反磁性和鐵磁性。

Magentic susceptibility



反磁性



順磁性

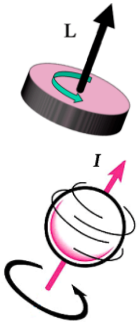
B=μ*H
B=磁通密度(magnetic flux density)
μ=磁導率(magnetic permeability)
H=磁場強度(magnetic field intensity)

物質的磁性與在MRI的用途

- **順磁性(paramagnetic)物質:**
- 導磁係數(μ)與空氣相近者，如:鋁、鉻等物質(釓、鏑、血液)
- **反磁性(diamagnetic)物質:**
- 導磁係數(μ)比空氣或真空小者，如:金、銀等物質(水及大部份組織)
- **鐵磁性(ferromagnetic)物質:**
- 導磁係數(μ)比空氣大者，如:鐵、鋼等物質(動脈瘤夾、榴霰彈片) 。

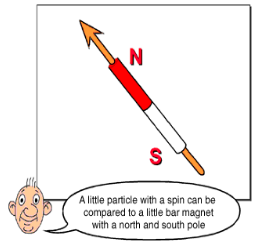
自旋 (Spin)

- 古典物理中，旋轉中的物體具有角動量(angular momentum)的特性，其大小和物體的外型、尺寸、質量以及向量的大小有關。
- 而在原子或次原子的領域中，則以自旋來表示，也就是說明粒子繞著本身軸位旋轉的特性。



人體內的小磁鐵

- 電子、質子帶有電荷且有自旋現象，故其行為類似於微小的電流迴路，因為移動的電荷會產生磁場，故電子、質子就好像是微小的磁鐵，有著北極和南極，故稱為磁偶極(magnetic dipole)。

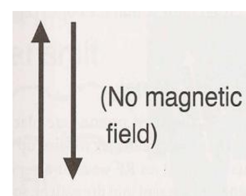


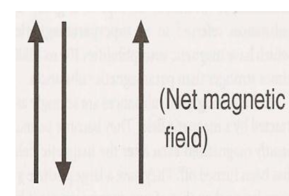
- 不同於古典力學的想法，自旋角動量只能有某些特定的值，即有量子化的現象，例如電子、質子等粒子的自旋量子數(S) 為1/2，其自旋角動量在某一軸向的分量只能有兩個值(能階數= 2S+1)，正負號表示方向，故有spin-up 與spin-down 兩種狀態之稱。

- 根據庖立不相容原理(Pauli exclusion principle)，兩個電子不能處在同一個量子狀態，故若其一為spin-up，另一必為spin-down。若一原子有偶數個電子，即有偶數個質子，則spin-up、spin-down兩種狀態互相抵消，所以無法有可被觀察到的自旋現象。然而，若一原子有奇數個電子(即不成對)，亦有奇數個質子，此時自旋現象才能被彰顯出來，氫原子就是其中一個例子。

NET MAGNETIC FIELD

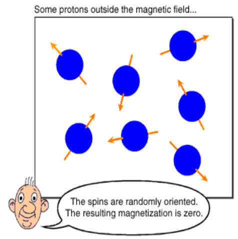
- 原子核中質子數目為**偶數(even)** * 她們的磁場會互相抵消，而使得淨磁場為0。
- 原子核中質子數目為**奇數(odd)** * 因而可以產生一個淨磁場或稱 (magnetic diopole moment, MDM)





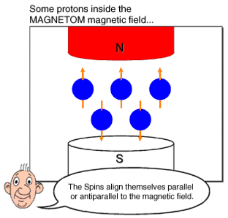
淨磁偶極的平衡

- 沒有外加磁場時，氫原子核(僅有一個質子)的磁偶極沒有特定的指向，淨磁化強度(所有磁偶極的加總，net magnetization)等於**0**。



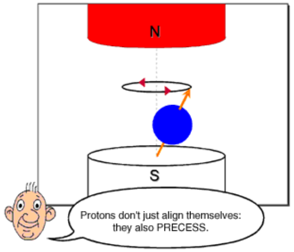
當外加磁場施加時....

- 場域內的氫原子核，受主磁場的強大作用力影響，產生順著主磁場方向(spin-up) 與逆著主磁場方向(spin-down) 的排列。
- 最終，在相互抵消的作用下，產生一個順著主磁場方向(spin-up)的**靜磁矩**。



原子的旋進(Precession)

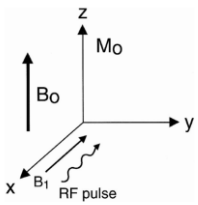
- 除了在自己的軸位上產生自旋，質子也會順著主磁場的方向，以特定的頻率產生繞進的現象，我們稱之為---**旋進**。

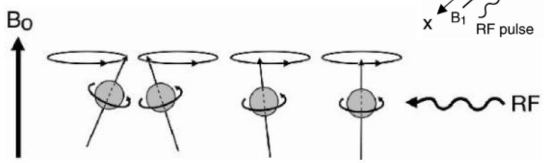


共振(resonance)

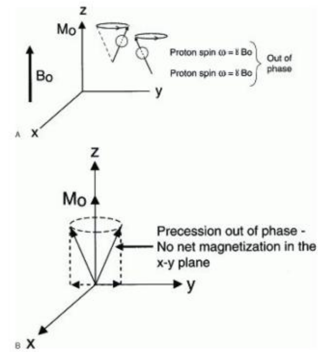
Radio Frequency Pulse

- 電磁波---傳遞能量
- 能量的傳遞須符合----拉莫頻率
- 施加的方向與B0垂直
- 將縱向磁矩偏轉到橫向平面

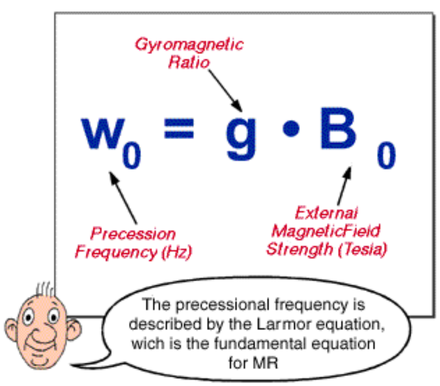




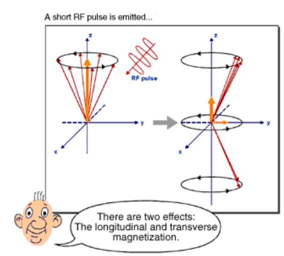
磁場的向量變化(RF施加前)



旋進頻率(拉莫頻率)



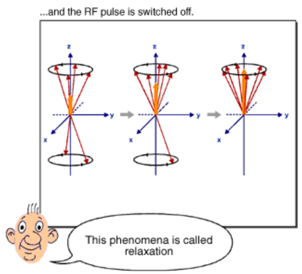
當RF施加時....



- RF頻率與系統之頻率相同
- 能量傳遞給系統，將spin-up之質子轉移至spin-down
- 縱向磁矩偏轉至XY軸平面，磁矩消減為0
- 橫向磁矩因質子同相，產生最大橫向磁矩

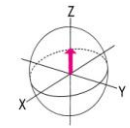
如果將RF關掉....

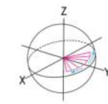
- 縱向磁矩回復至Z軸平面，磁矩回復為M0
- 橫向磁矩因質子產生失相，橫向磁矩消減為0

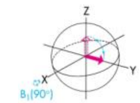


發生了什麼事???

- 縱向磁場的回復----
T1 recovery
- 橫向磁場的衰減----
T2 decay







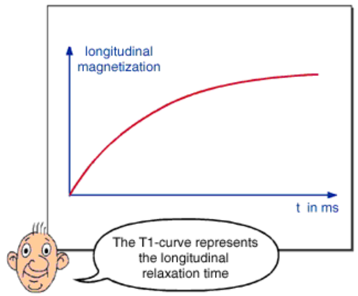
弛緩

- 在B1 關閉後，氫原子核要從激發狀態回到平衡狀態，與主磁場對齊，主要有兩個互相獨立的歷程，分別稱為自旋晶格弛緩(spin-lattice relaxation)和自旋自旋弛緩(spin-spin relaxation)，分別是Z 分量的回復和X-Y 分量的歸零，其弛緩的時間常數(time constant)分別稱為T1 和T2，故又稱為T1 弛緩和T2 弛緩。

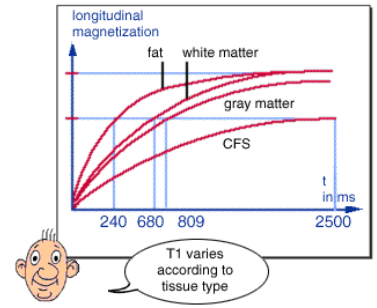
T1 弛緩

- 氫原子核將先前吸收的能量以熱能的方式釋放到鄰近的組織(lattice)中，使得氫原子核可和主磁場對齊。弛緩的時間常數，即Z 分量回復到原來M0 的63%所需的時間稱為T1，大約需要5 倍T1 的時間，Z 分量可完全恢復。

T1弛緩(relaxation)



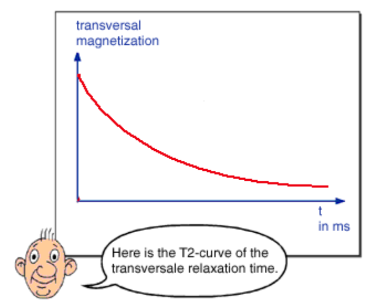
不同組織的T1曲線



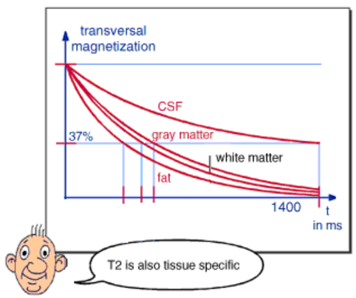
T2 弛緩

- 當有B1 磁場時，眾多氫原子核以同樣的相位(phase)自旋，當B1 關閉時，外力的協助消失，氫原子核間會有隨機的運動，彼此碰撞交換能量(所以稱為 spin-spin relaxation)，使相位一致性(phase coherence)消失，有的氫原子核進動較快，有的氫原子核進動較慢，使X-Y 分量互相抵消逐漸回復到零。X-Y 分量減少到M0 的37%(也就是衰減了63%)的時間稱為T2，T2 通常短於T1。

T2弛緩(relaxation)



不同組織的T2曲線



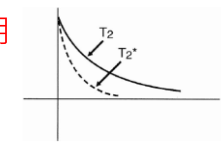
T2 vs T2*

- 造成橫向磁場衰減的原因:

--- Spin-Spin 間的交互作用(失相)

--- 主磁場的不均勻 (可以被修正)

- T2 ---造成橫向磁場衰減的原因僅為Spin-Spin 間的交互作用
- T2*--- Spin-Spin 間的交互作用+主磁場的不均勻



成像與造影(Imaging)

想,想,想.....

- 不同組織在T1、T2都有不同訊號的表現
- 在適當的時間擷取訊號
- 不同的組織就可以被區分出來了!
- 傅立葉轉換---- 時間函數 ⟹ 頻率函數
- 適當的空間編碼，填入K-space 中，MRI的影像就出來啦!!!

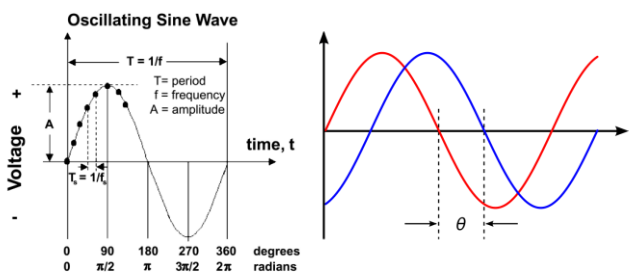
成像(空間編碼)

- MRI 量測到的是人體某一區塊中的所有氫原子核激發、弛緩的訊號，為了要了解人腦中不同位置的結構或功能性變化，必須在訊號中加入空間位置的訊息，簡單來說，這分為在Z 方向的切面選擇(slice selection)，X-Y 平面上任一軸例如X 方向的頻率編碼(frequency encoding)，和X-Y 平面上另一軸例如Y 方向的相位編碼(phase encoding)，要在那個方向做切面選擇、頻率、相位編碼視實際需求而定。所用到的技術是梯度磁場的概念以及二維傅利葉轉換(2-D Fourier transform)。

相位 vs. 頻率

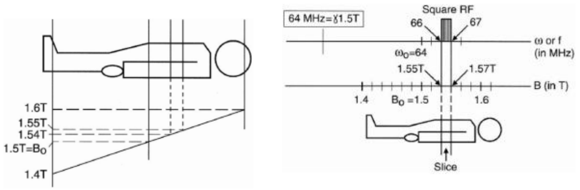
- MRI訊號:具週期性，以sin or cos 呈現，具有0~360 º 的相位變化
- 相位(Phase): 在特定的時間點中訊號波形(Waveform) 循環中的位置
- 頻率(Frequency):單位時間內，產生訊號週期的數目 (Cycle/Sec)

訊號 、頻率、相位與相位偏移

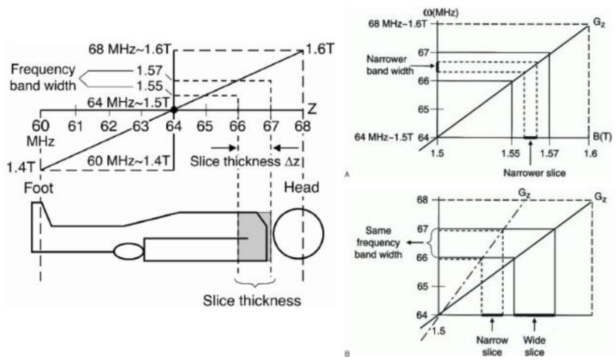


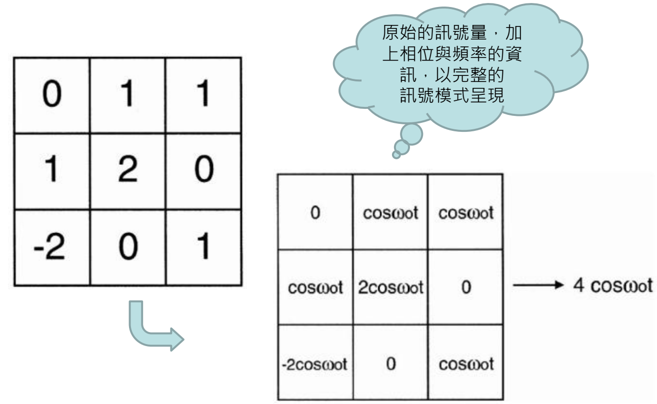
切面選擇(Slice Selection)

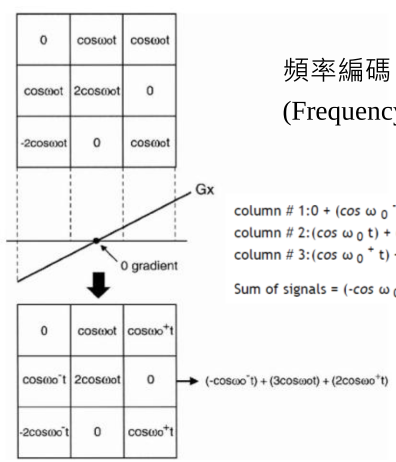
- 空間編碼的第一步
- 在切面方向施加一個梯度磁場，藉此在不同的位置產生相對不同的磁場強度 (B0+Gz)

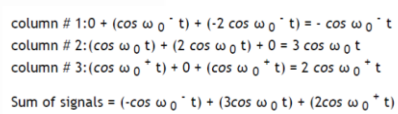


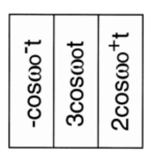
選擇切面的方式



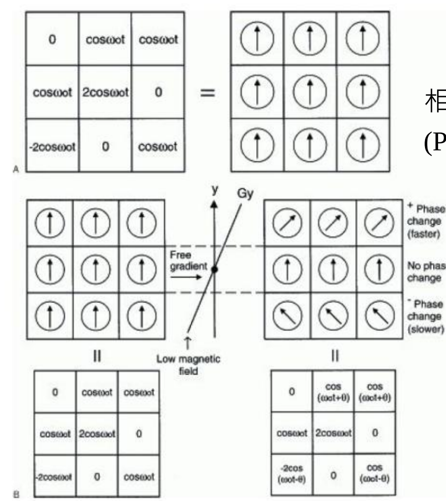






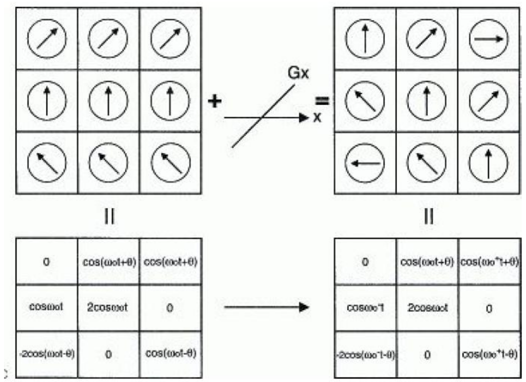


頻率編碼
(Frequency Encoding)

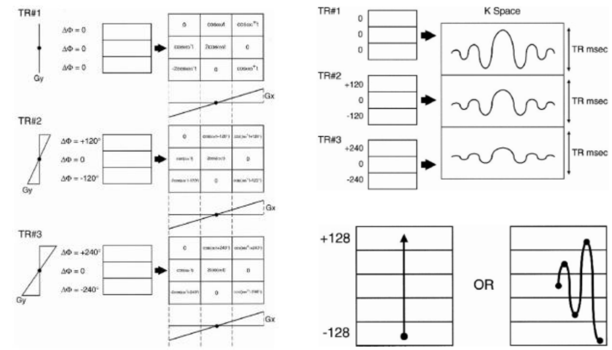


相位編碼
(Phase Encoding)

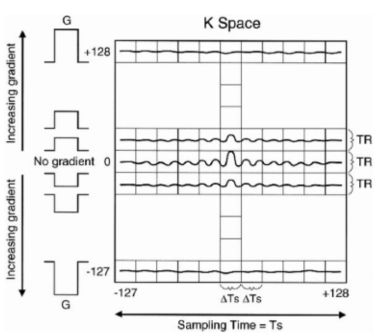
空間編碼的結果(相位+頻率)

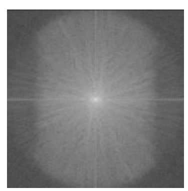


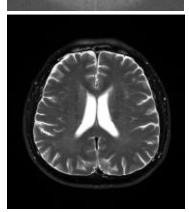
相位編碼與TR



Data Space to image







硬體上的需求

- 主磁場
  ---- 永久磁場
  ---- 電磁場
  ---- 超導磁場
- 線圈(coil)
  ---- Gradient coils
  ---- Shim coils
  ---- 射頻線圈
  ---- 接收線圈

磁場產生的方式

- 永久磁場
- 電磁場

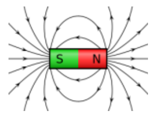
--- 電生磁，磁生電

- 超導磁場

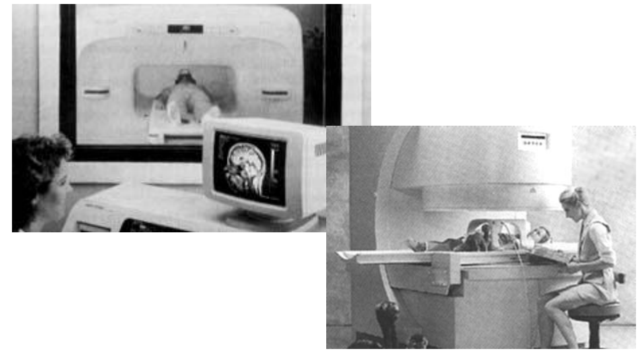
--- 導體的電阻為零

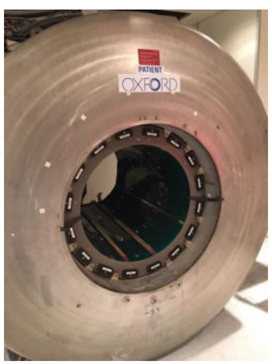
---指電流流通時無阻力的現象，也就是產生永久電流(persistent current）

--- 絕對低溫(4-6゜K)



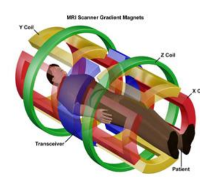
永久磁場

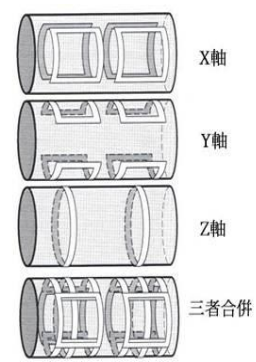




Gradient coils

- X軸、Y軸、Z軸都需要形成梯度磁場，　　就可以得到不同的斜切位。





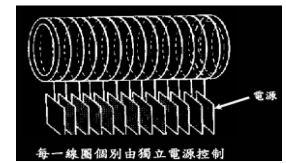
Gradient coils

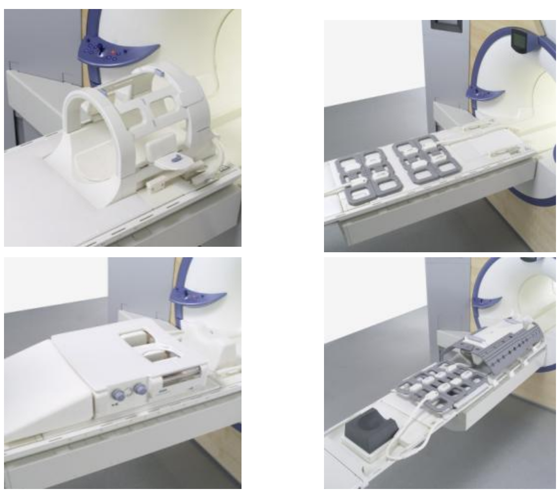
– Gz slice-selection切面選擇
– Gy phase-encoding相位編碼
– Gx frequency-encoding 頻率編碼

	slice-selection	phase-encoding	frequency-encoding
Axial	z	y	x
Sagittal	x	y	z
coronal	y	x	z

Shim coils

• 被動式
---維持主磁場的均勻性
• 主動式
• 維持局部磁場的均勻性，特別是在梯度回音或脂肪的化學位移消除技術，shimming可以使變動降低但不盡然完全消除。





MRI safety

• 影響裝置功能或MRI儀器本身產生之危害
--- 心臟節律器
--- 血管夾
--- 金屬物品(剪刀、髮夾、氧氣瓶…..)
• RF產生的熱效應
--- 紋眉、紋身
--- 精油、髮膠
---掃描過程中病人身體(皮膚)不要直接觸碰磁體內壁及各種導線

SAR (Specific Absorption Ratio)

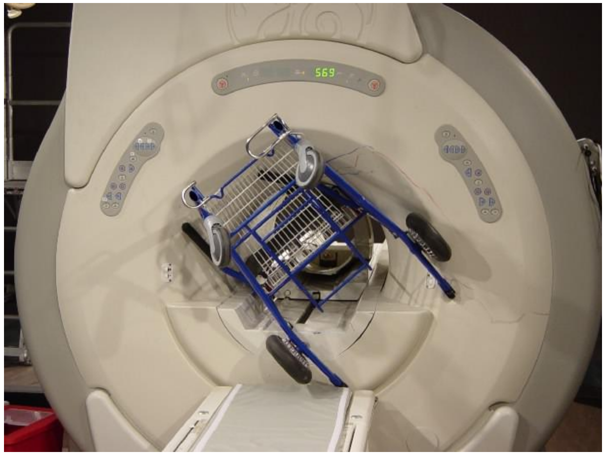
- 每單位質量的物體，因RF能量傳遞，造成物體能量吸收的比例 (單位: W/Kg)
- 依據FDA的規範， 不得超過 4 W/Kg
- 不同的部位，SAR值的吸收也有所不同
- SAR值產生的效應，以熱的方式來表現

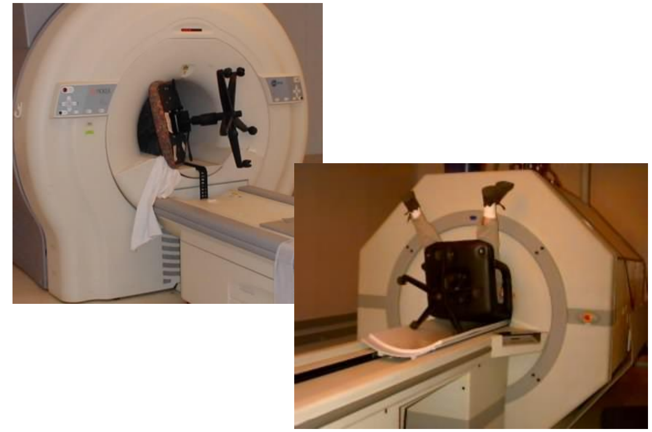
Table 10.1 RF temperature limits

Operating mode	Core temperature rise (°C)	Spatially localized temperature limits (°C)		
		Head	Torso	Extremities
Normal	0.7	38	39	40
First-level controlled	1	38	39	40
Second-level controlled	>1	>38	>39	>40

如何降低SAR值

- 使用 quadrature coil 進行RF的發送
- 在有適合的coil 可同時進行RF的發送與接下，盡量避免使用 body coil進行RF的發送
- 增加TR
- 減少掃描張數
- 減少ETL的長度
- 減少TSE中 refocusing pulse flip angle



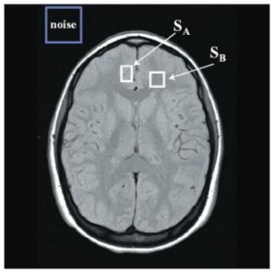


對比(Contrast)

對比的定義
影響對比的參數與臨床應用

Contrast

- Contrast was introduced in terms of the image appearance, or relative brightness of different tissues and pathology.
- Image contrast arises (or doesn't) when tissues generate MR signals which have different intensities because of their physical properties, i.e. T1 and T2 relaxation times and proton density.



Here's the maths bit

Mathematically we can define contrast as

$$C = \frac{S_A - S_B}{S_A + S_B}$$

where S_A and S_B are signal intensities for tissues A and B.

Signal-to-noise ratio (SNR) is defined as

$$\mathrm{SNR} = \frac{signal}{noise}$$

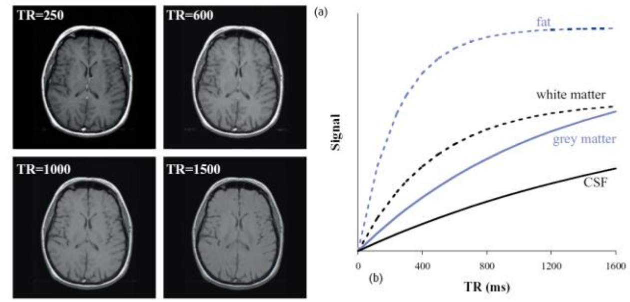
Contrast-to-noise ratio (CNR) is defined for tissues A and B as

$$CNR_{AB} = \frac{S_A - S_B}{noise}$$

In the simplest terms spatial resolution of the voxels is related to the field of view (FOV) and matrix thus

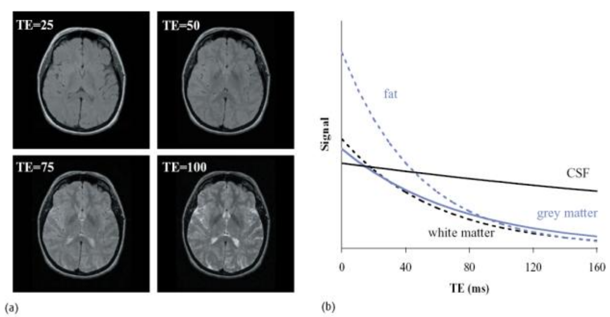
$$\Delta x = \frac{\mathrm{FOV}}{N_{\mathrm{FE}}} \qquad \Delta y = \frac{\mathrm{FOV}}{N_{\mathrm{PE}}} \qquad \Delta z = \text{slice width}$$

TR and tissue contrast

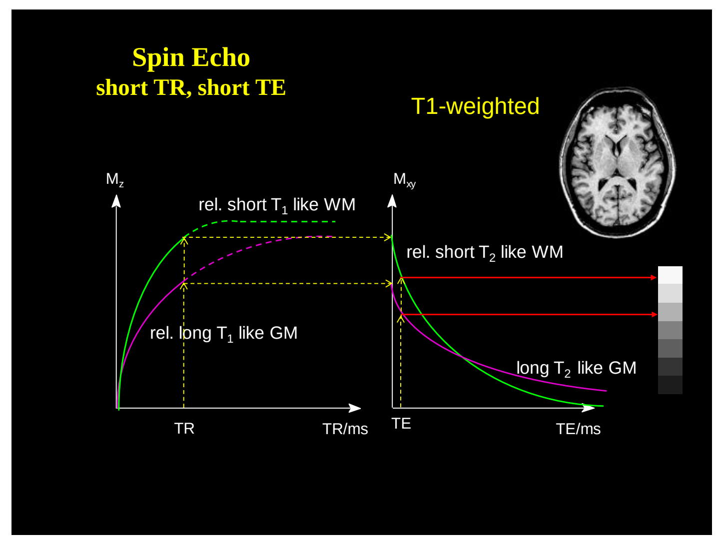


TE=10

TE and tissue contrast



TR=1500

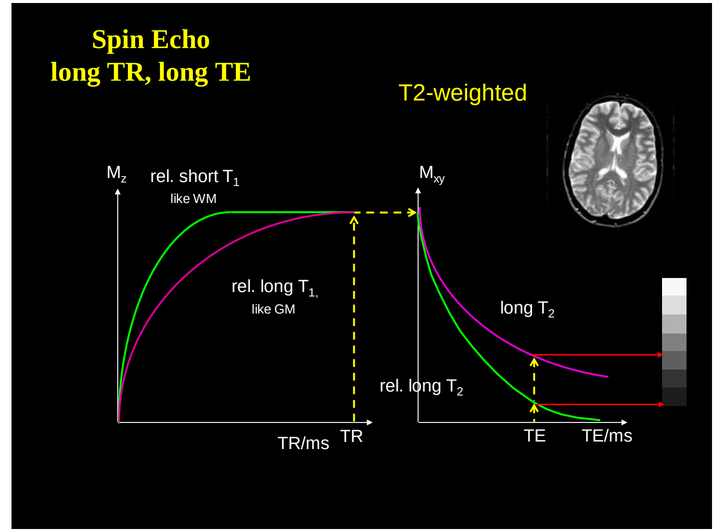


T1 effects and pathology

T1 effects on the image

short T1 - bright
- fat, fresh bleeding
- paramagnetic contrast agent (gadolinium)

long T1 - dark
- Neoplasm,
- edema, inflammation,
- pure fluid, CSF



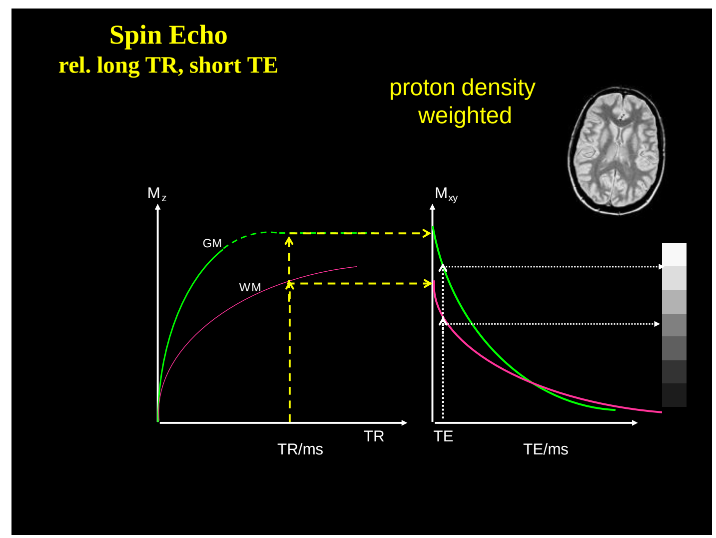
Clinical image appearance

T2 Effects on the image

short T2 - dark
- iron deposits in liver,
- magnetic suseptibility effects

long T2 - bright
- edema,
- inflammation, Gliosis,
- pure fluid, CSF

TR > 1800 ms TE > 80 ms



Clinical image appearance

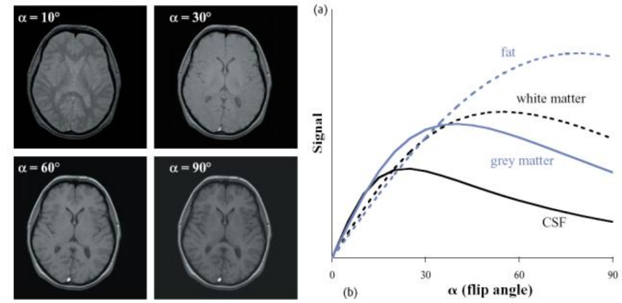
proton density effects

low proton density- dark
- calcium, air, cortical bone, ligamen

high proton density- bright
- fat, bone marrow

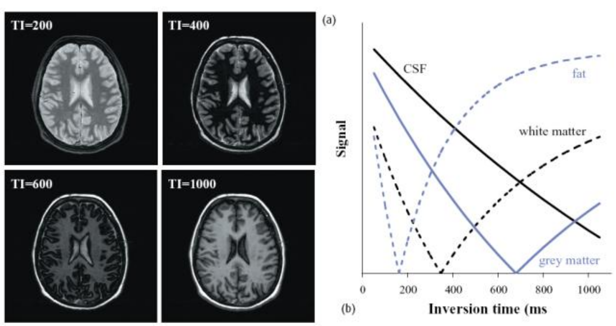
TR ~ 1200 ms   TE ~ 25 ms

Flip angles and contrast



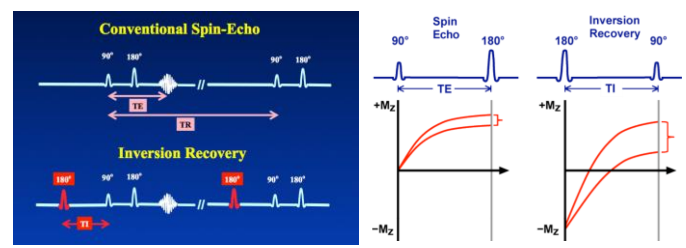
TR=150, TE=4.6

TI and contrast

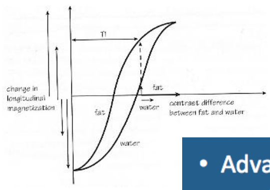


TR=4000, TE=19

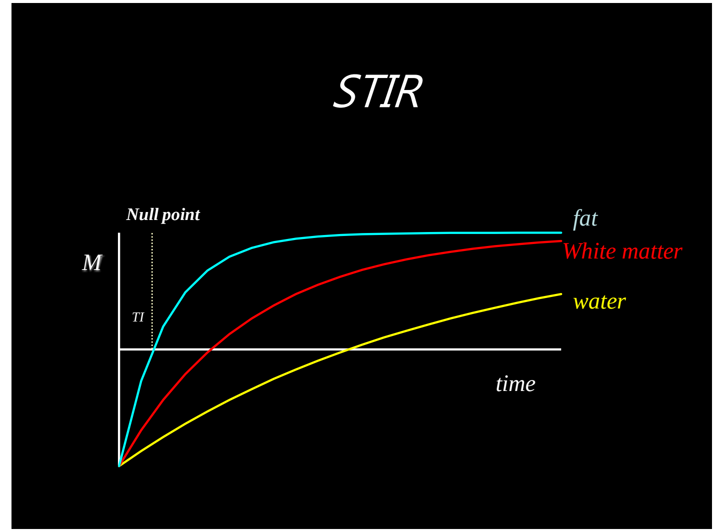
SE vs.IR



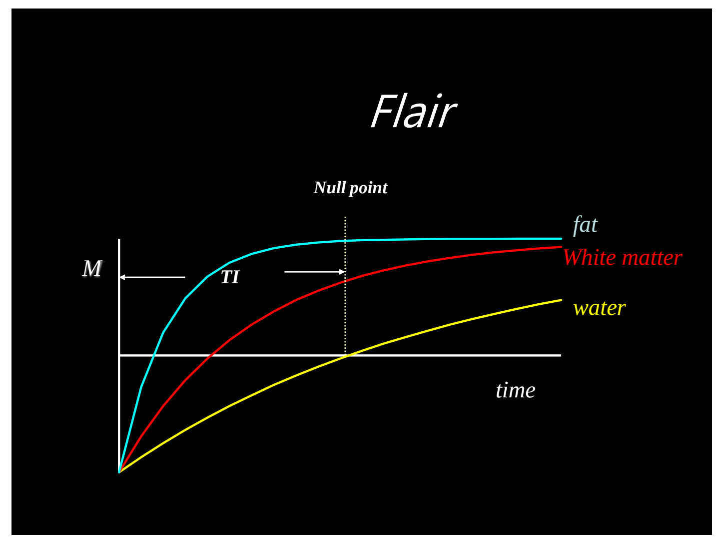
Invention recovery (IR)



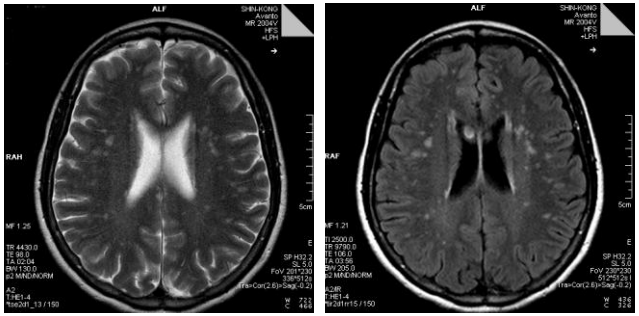
- Advantages of IR
  - Selective tissue suppression possible
  - Twice the sensitivity to T1 differences
  - Additive T1 and T2 contrast
- Disadvantages of IR
  - Longer imaging time
  - Higher energy deposition (SAR)

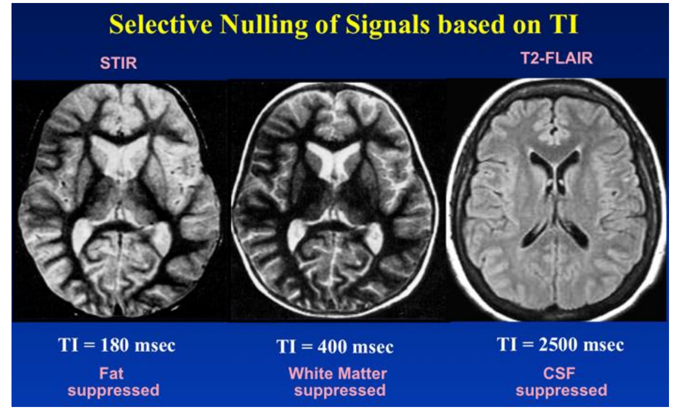


STIR image



FLAIR in Brain MRI





Disadvantages of IR

- Longer scan times
- Increase in flow-related artifacts
- Signal-to-noise can decrease as tissues are suppressed
- Higher specific absorption rate (SAR) due to additional 180° pulses

休息一下吧!!!

磁振造影專業基礎課程

MR影像建構與脈衝序列圖

□□□

MR Image Construction & Pulse Sequ

□□□

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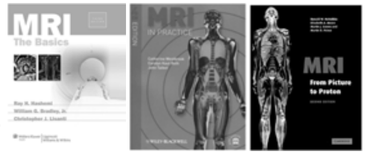
本次課程內容

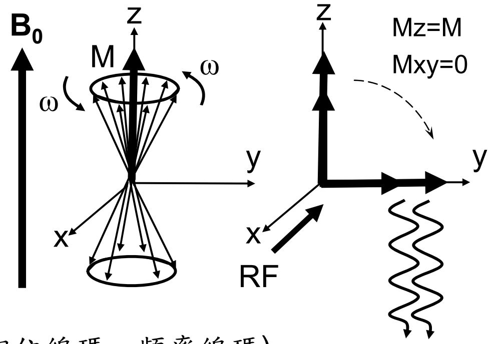
- MR基本原理與自旋回音(Spin echo)
- 脈衝序列圖(Pulse sequences diagram)
- 空間編碼(Spatial encoding)
  - Slice selection encoding(Gz)
  - Frequency encoding(Gx)
  - Phase encoding(Gy)
- K-space與訊號填入
- 快速自旋回音(Fast spin echo (FSE))

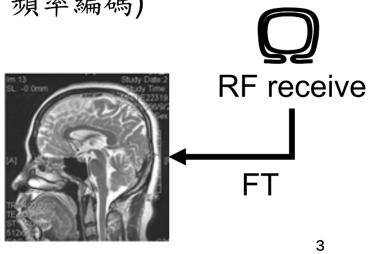
Reference:
1.MRI The Basics (3rd) (Chapter 7~14)
2.MRI IN PRACTICE(4td) (Chapter 3.5)
3.MRI From Picture to Proton(2nd)(Chapter 7)



MRI的成像過程

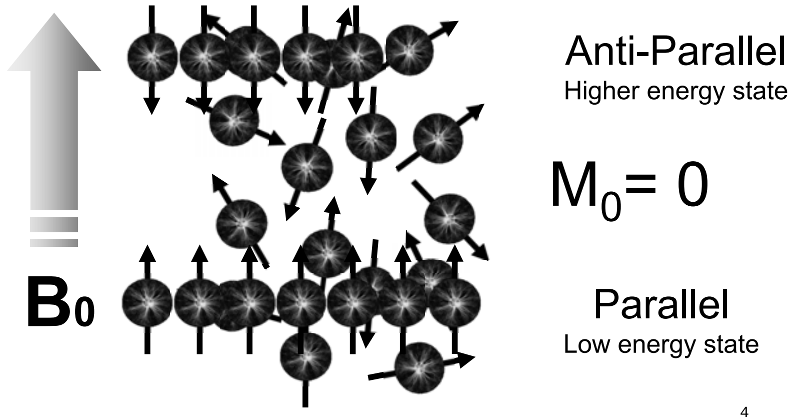
- 磁場 → 磁化現象
- RF脈衝 → 磁化量激發
- 梯度磁場 → 切面選擇
- 梯度磁場 → 空間編碼(相位編碼、頻率編碼)
- 信號取號 → 接收線圈取樣頻率
- 影像計算 → FOURIER轉換





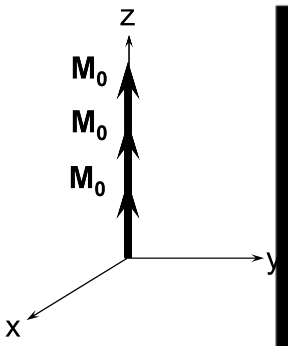
MR basic principle

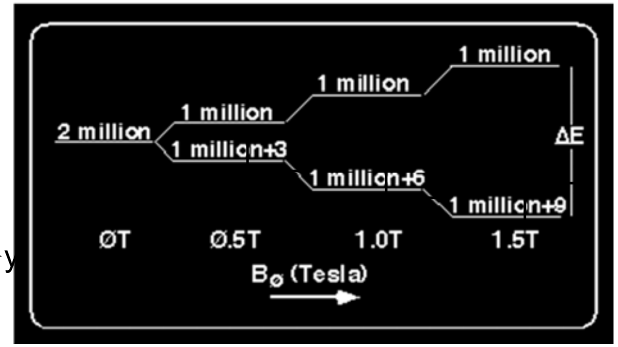
5/1000000(每一百萬個氫原子核中最多5~6個)



淨磁矩與外加磁場的關係

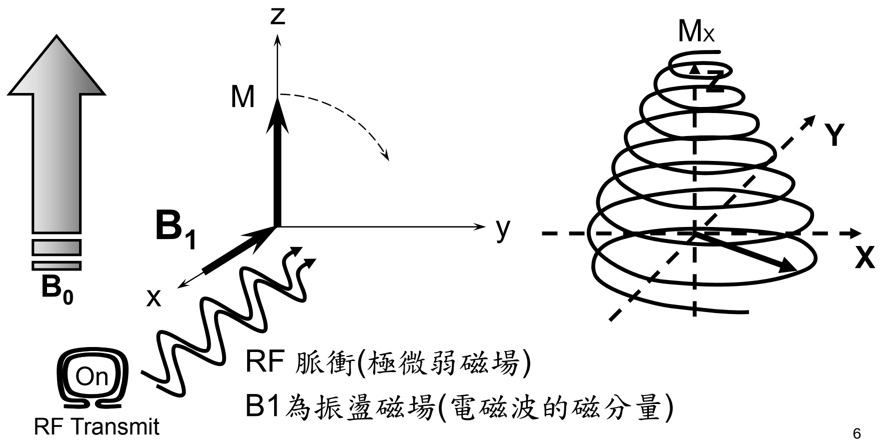
- 外加磁場愈大淨磁矩愈大





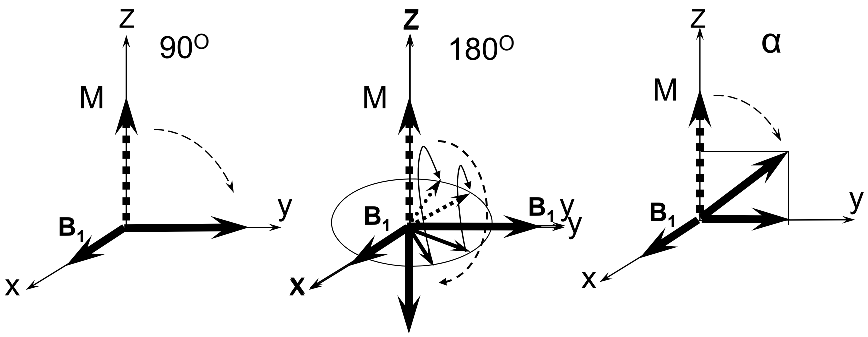
如何偏轉人體磁鐵？

- 電磁波產生一橫向磁矩(B_1)使人體磁鐵(M)偏轉
- RF的作用下M_0的運動軌跡為螺旋形



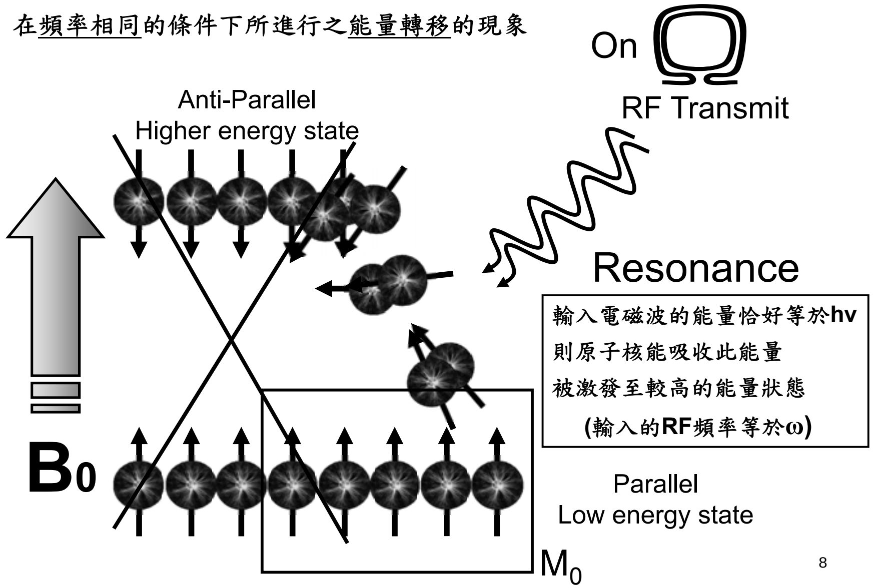
RF常用的脈衝角度

- Larmor frequency $\omega=\gamma B_0$
- 傾角(flip angle) $\Theta=\gamma B_1 t$ (B_1為RF脈衝磁場，t為時間)

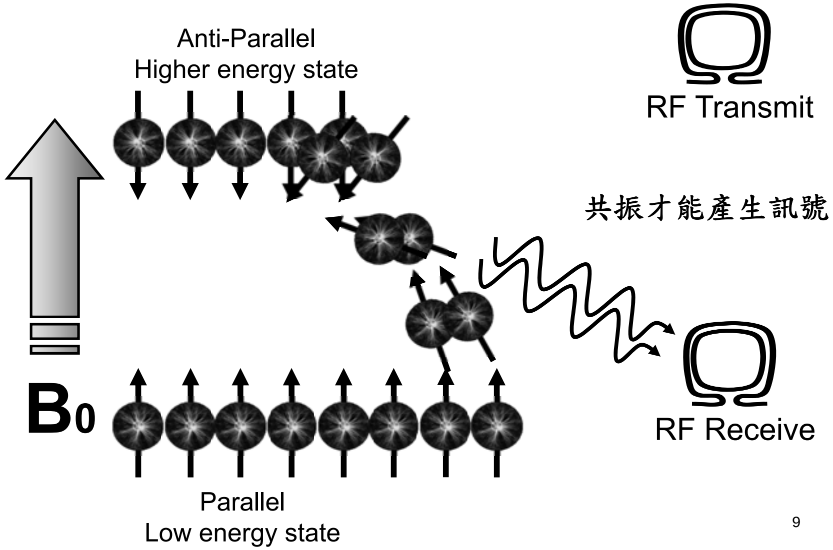


共振(resonance)

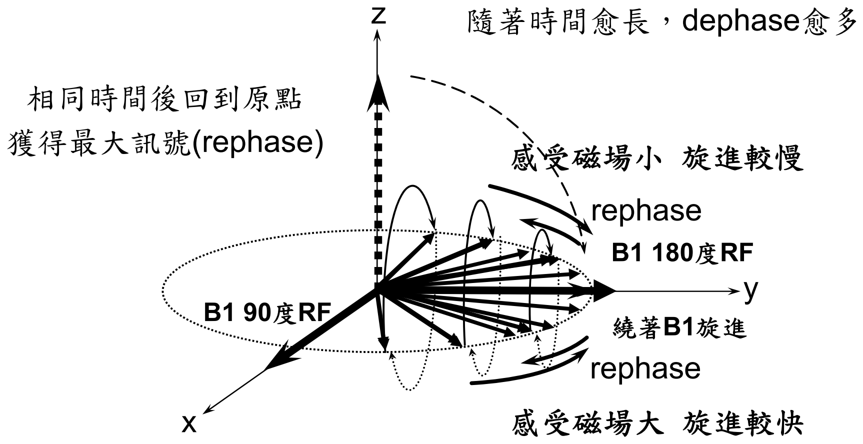
在頻率相同的條件下所進行之能量轉移的現象



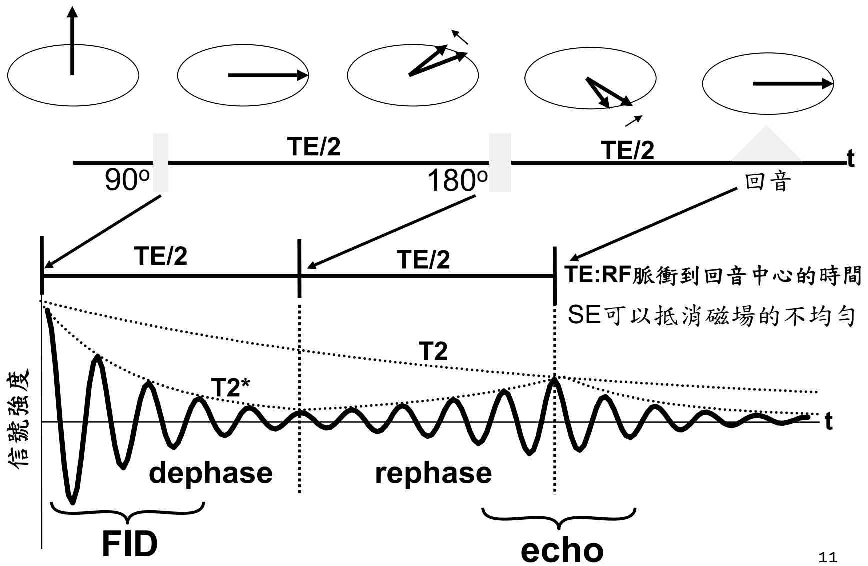
共振(resonance)



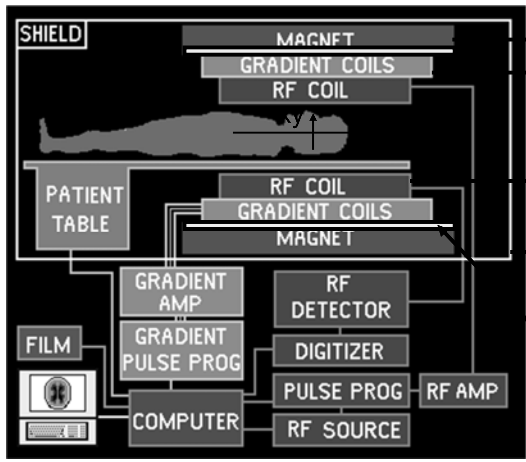
dephase 和 rephase的過程



自旋迴訊(spin echo)脈衝序列



磁振造影儀器基本硬體-線圈與磁場



產生淨磁向量

給予slice selection, phase encoding 和 frequency encoding 所需的磁場梯度

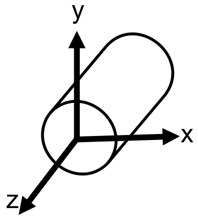
可給予 RF脈衝,
也可以接收信號

以銅牆,銅門,銅窗
阻隔外面的電磁波

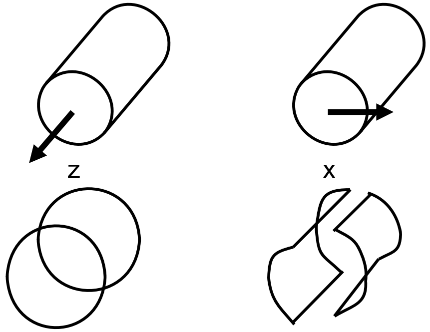
銅牆鐵壁(屏壁)

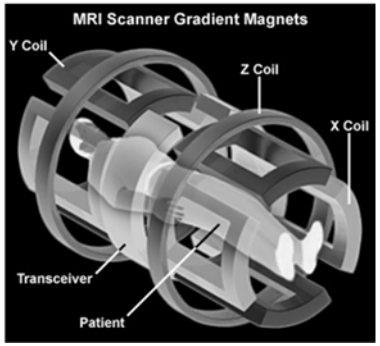
墊補磁鐵(shim coil)
修飾磁場的均勻度±1~5ppm
穩定度:10ppm/h (6~7 best)
分為被動式(小金屬片)
主動式(shim coil)

梯度線圈 gradient coils



- Alter the primary magnetic field
- Gradient coils in the in the x, y & z axis
- Responsible for loud noises of MRI





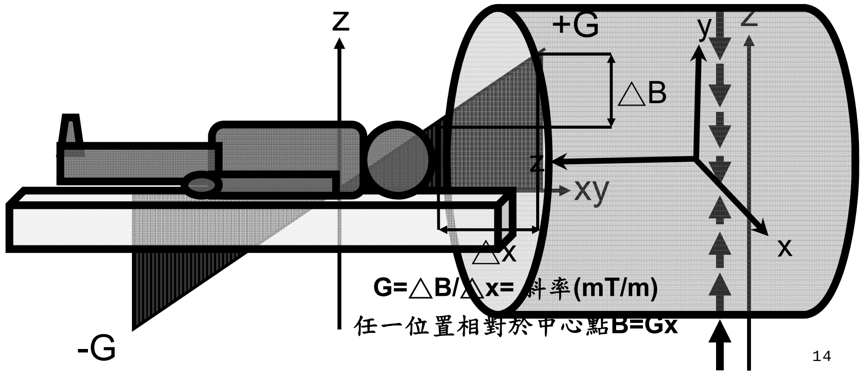
梯度磁場(gradient coil，Gx， Gy，Gz)

z(slice-select)、x (phase-encoding) 、y (readout or frequency-encoding)

- 磁場方向沿著**3**軸，強度隨著位置改變(**ex:**沿著**z**，隨**xy**改變)
- 故意製造磁場均勻度的擾動
- 改變氫原子核感受到的磁場大小

Z : no signal

X-Y : signal



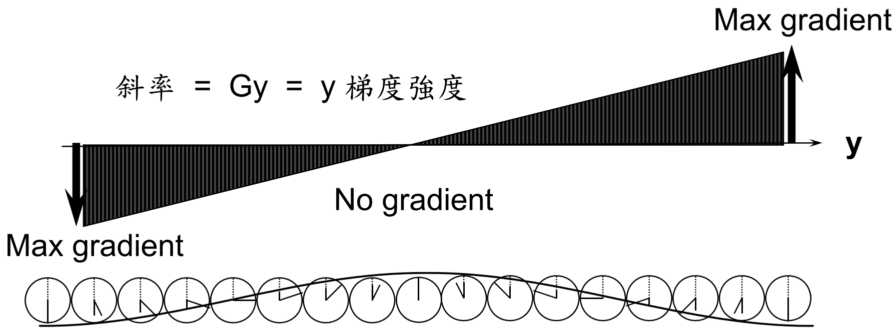
梯度磁場-相角隨位置而改變:

$\omega_0=\gamma B_0$

$\omega_0+\triangle\omega=\gamma(B_0+Gy)$

$\omega_0-\triangle\omega=\gamma(B_0-Gy)$

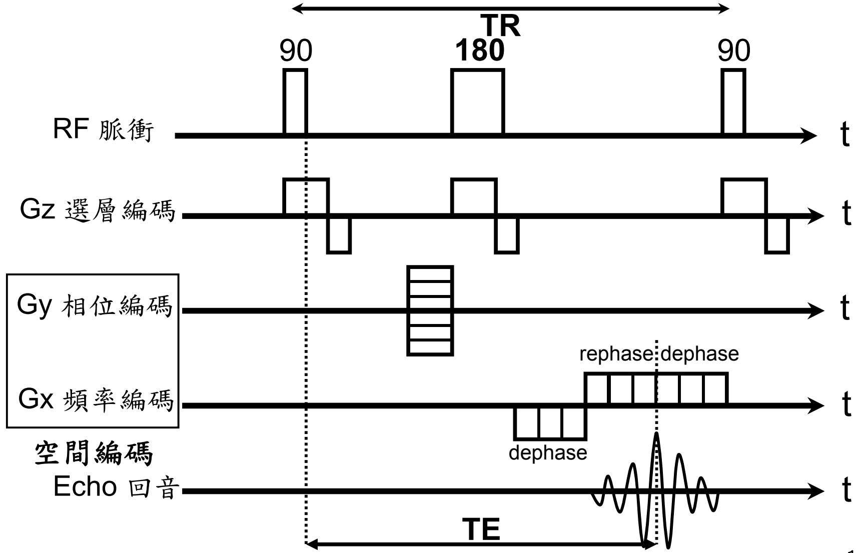
- 梯度是由一點到另一點的磁場改變-磁場的擾動通常是**linear**
- 三個軸暫時製造出線性磁場的不均勻，取得的位置資訊



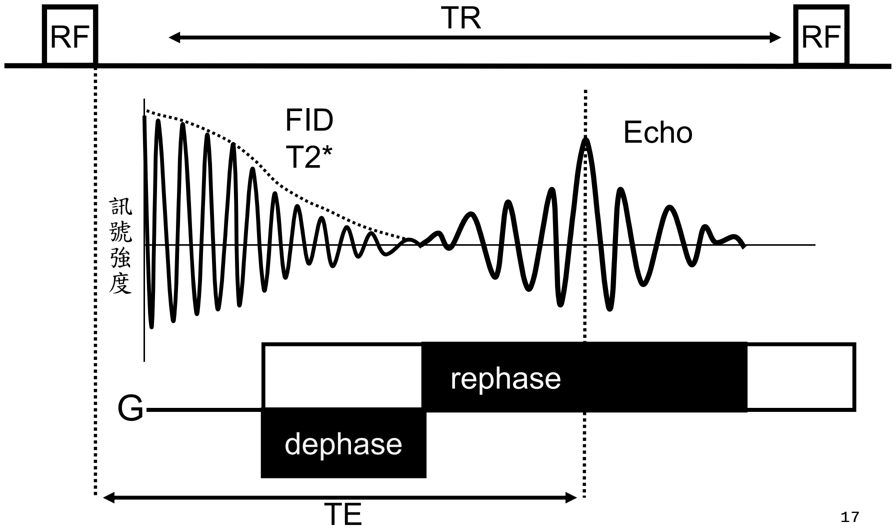
cosine 波形

from:鍾孝文教授ppt改編動畫

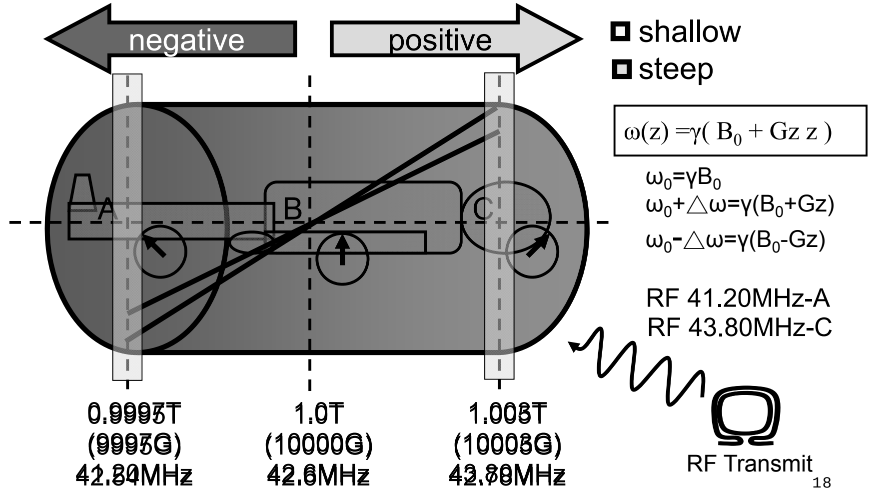
Spin Echo (SE) 波序圖



梯度磁場下的訊號衰減

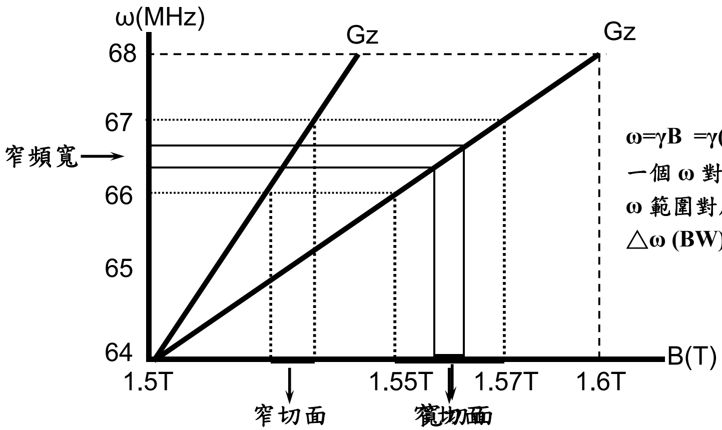


梯度磁場和變化磁場強度
(Gradients and changing field strength)
(How to select a slice?)



切面厚度slice thickness

- We can excite one slice by an RF pulse with a specific frequency range.
- This range of frequencies determines the slice thickness and is referred to as the bandwidth. (射頻脈衝頻寬和梯度強度的斜率有關)



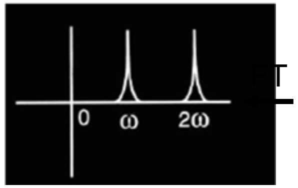
ω=γB =γ(Bo + Gz z)
一個 ω 對應到一個 z 位置
ω 範圍對應到 z 位置的範圍
△ω (BW) = γ Gz △ z

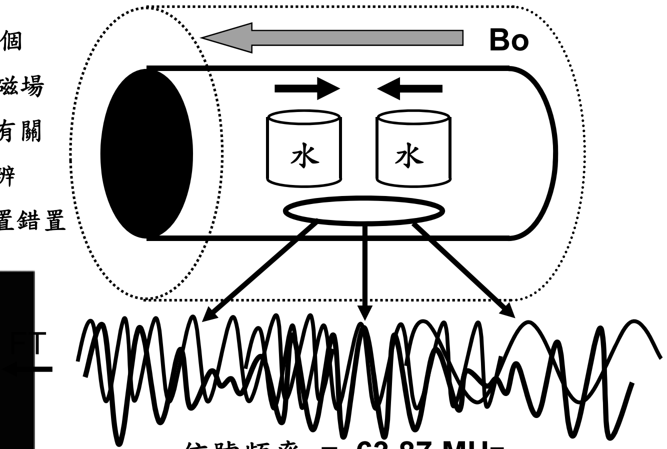
選層梯度磁場 (slice-select gradient, Gz)

- We transmit an RF pulse with a bandwidth that has the appropriate center frequency.
- This gradient is turned on only when we transmit the RF pulses.
- When we transmit the 180° pulse(rephasing pulse) for the same slice, we activate the same gradient.
- Two types of RF pulses
  - Slice-selective: only a certain slice of the body. used in 2D imaging.
  - Non-selective: excites every part of the body. used in 3D imaging.

選層梯度磁場 (slice-select gradient, Gz)

在 Bo 上另外加一個
“隨位置變化” 的磁場
如果信號頻率與位置有關
物體位置即可分辨
取樣錯誤就會造成位置錯置





信號頻率 = 63.87 MHz

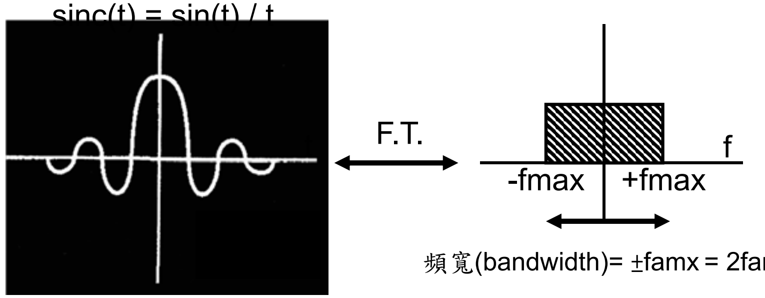
from:鍾孝文教授ppt改編動畫

Fourier Transform 轉換 (RF)

- **time domain**轉至**frequency domain**。提供訊號中的頻率範圍
- 有頻率的範圍及振幅就能把訊號重建回來
- 傳送一定範圍頻率的**RF**脈衝有一定頻寬的頻率

time domain **frequency domain**

sinc(t) = sin(t) / t

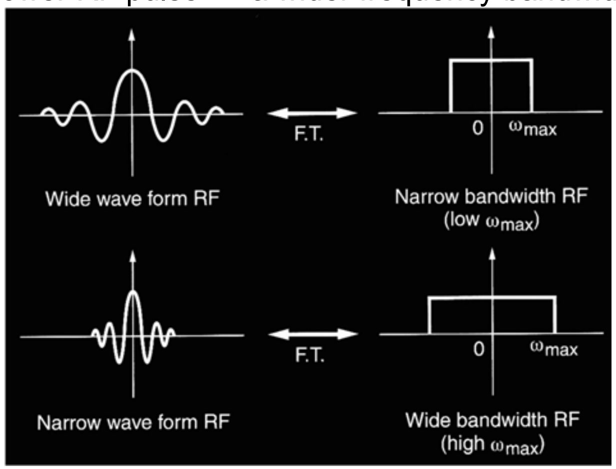


頻寬(bandwidth)= ±famx = 2famx

脈衝寬度與脈衝頻寬為反比

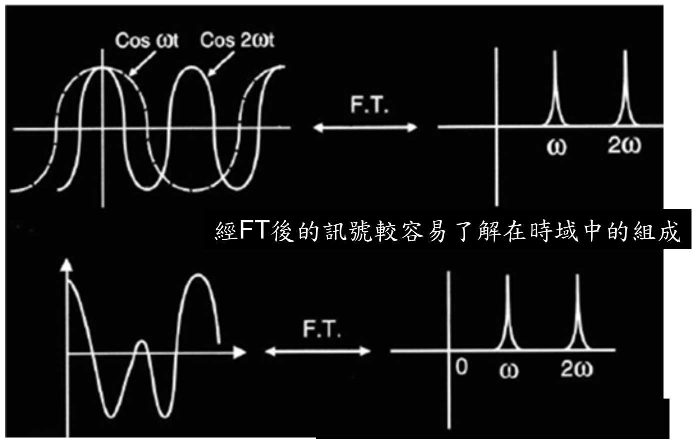
Fourier Transform 轉換 (RF)
waveform and bandwidth

- A narrower RF pulse ➞ a wider frequency bandwidth



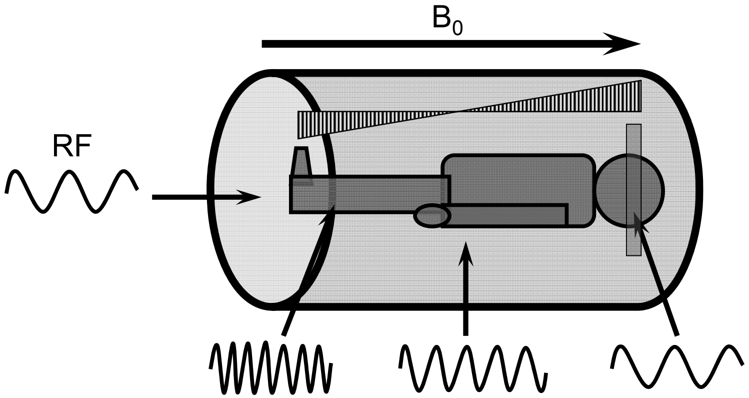
Fourier Transform 轉換 (Echo)

- **time domain**轉至**frequency domain** 。提供訊號中的頻率範圍
- 有頻率的範圍及振幅就能把訊號重建回來



切面選層梯度 (slice-select gradient, Gz)

- 信號只從一個平面出來，只激發該平面內的氫原子核
- 調整射頻脈衝頻率 選擇欲激發的切面



空間編碼(spatial encoding)

(phase encoding, Gy、frequency encoding, Gx)

- The spatial information regarding each slice

$4\cos\omega_0 t$ ↑

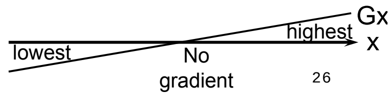
0	$\cos\omega_0 t$	$\cos\omega_0 t$
$\cos\omega_0 t$	$2\cos\omega_0 t$	0
$-2\cos\omega_0 t$	0	$\cos\omega_0 t$

→ frequency encoding 改變頻率差

$-1\cos\omega_0^- t + 3\cos\omega_0 t + 2\cos\omega_0^+ t$ ↑

0	$\cos\omega_0 t$	$\cos\omega_0^+ t$
$\cos\omega_0^- t$	$2\cos\omega_0 t$	0
$-2\cos\omega_0^- t$	0	$\cos\omega_0^+ t$

$\omega_0=\gamma B_0$
$\omega_0+\triangle\omega=\gamma(B_0+Gx)$
$\omega_0-\triangle\omega=\gamma(B_0-Gx)$

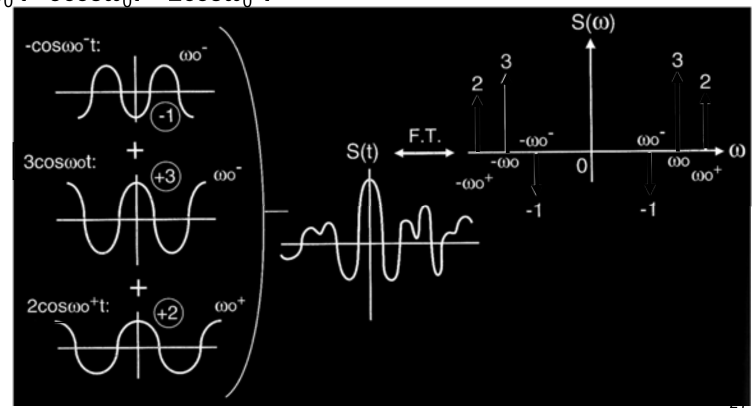


空間編碼(spatial encoding)

(phase encoding, Gy、frequency encoding, Gx)

- We can analyze the magnitude of each frequency component using FT.

$-\cos\omega_0^- t + 3\cos\omega_0 t + 2\cos\omega_0^+ t$



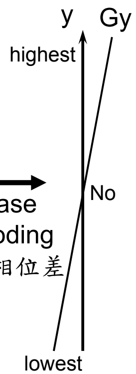
空間編碼(spatial encoding)

(phase encoding, Gy、frequency encoding, Gx)

- Turn on phase encoding (Gy) between 90º and 180º RF pulses or between 180º pulse and the echo.

0	$\cos\omega_0 t$	$\cos\omega_0 t$
$\cos\omega_0 t$	$2\cos\omega_0 t$	0
$-2\cos\omega_0 t$	0	$\cos\omega_0 t$

→ phase encoding 改變相位差

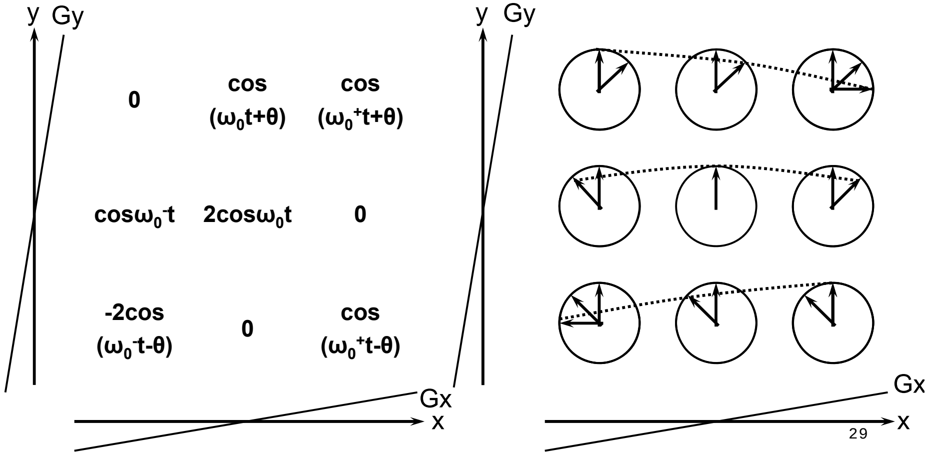


0	$\cos(\omega_0 t+\theta)$	$\cos(\omega_0 t+\theta)$
$\cos\omega_0 t$	$2\cos\omega_0 t$	0
$-2\cos(\omega_0 t-\theta)$	0	$\cos(\omega_0 t-\theta)$

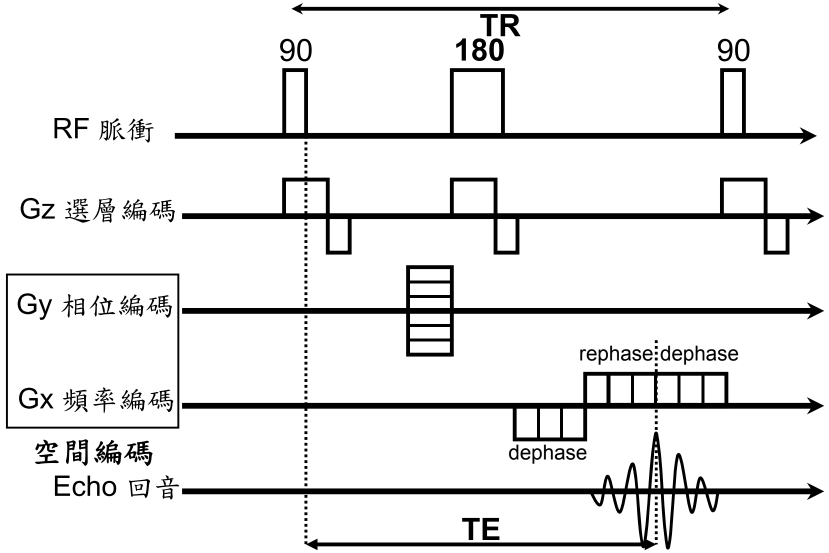
$\omega=\gamma Gy$
$\psi=\triangle\omega t=\gamma Gy\ t$

空間編碼(spatial encoding)
(phase encoding, Gy、frequency encoding, Gx)

- 將氫原子核座標位置的訊息帶入訊號中(需知道像素內有多少訊號)
- 使氫原子核旋進頻率隨位置改變。使磁向量旋進相位隨位置改變

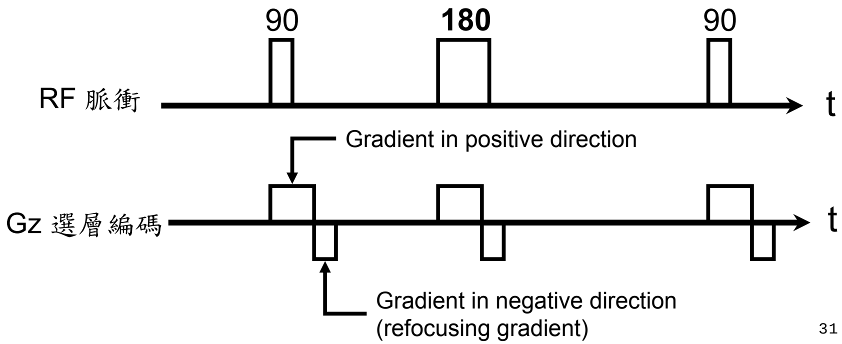


Spin Echo (SE) 波序圖



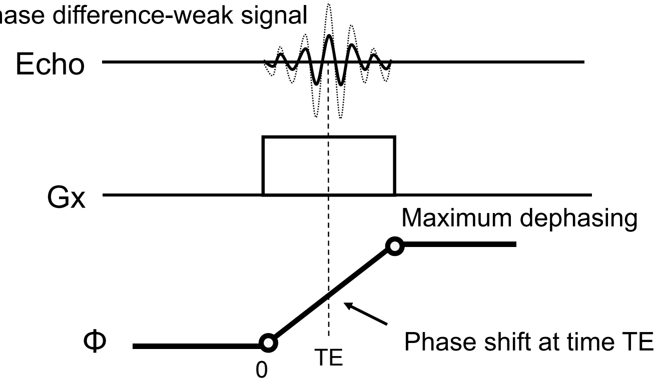
切面選層梯度 (slice-select gradient, Gz)

- Every time we apply a gradient. we dephase the spins.
- We apply a Gz gradient to select a slice. but after the slice is selected, we need to reverse the effect of dephasing (refocusing).



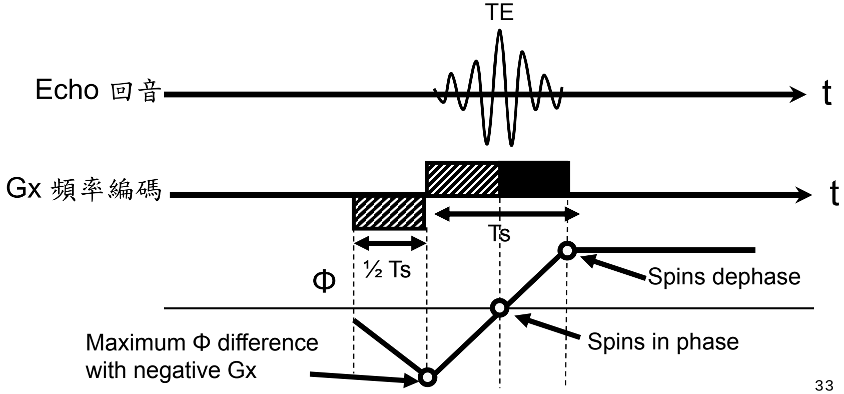
頻率編碼梯度
(frequency-encoding gradient, Gx)

- Without a proper compensation, the signal intensity will be….
  - Decrease at the time in the middle of echo
  - Maximum dephasing at the time in the end of echo
  - Phase difference-weak signal

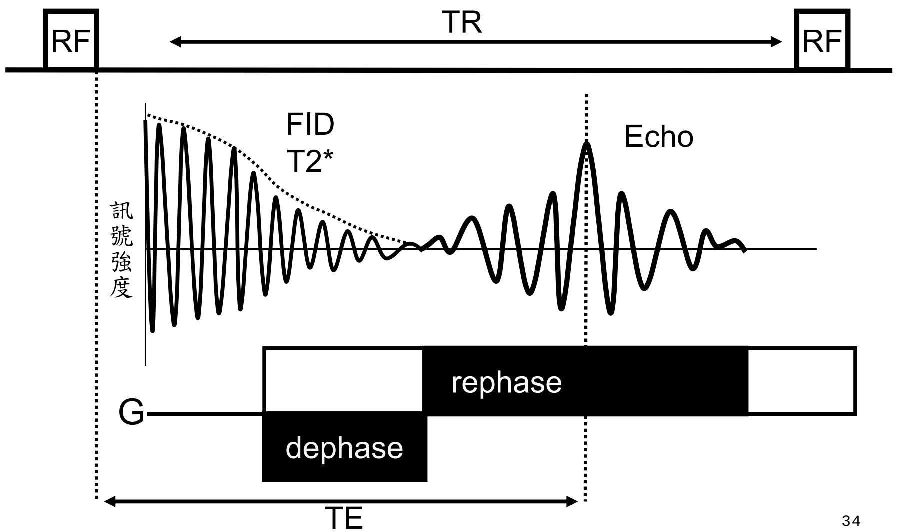


頻率編碼梯度 (frequency-encoding gradient, Gx)

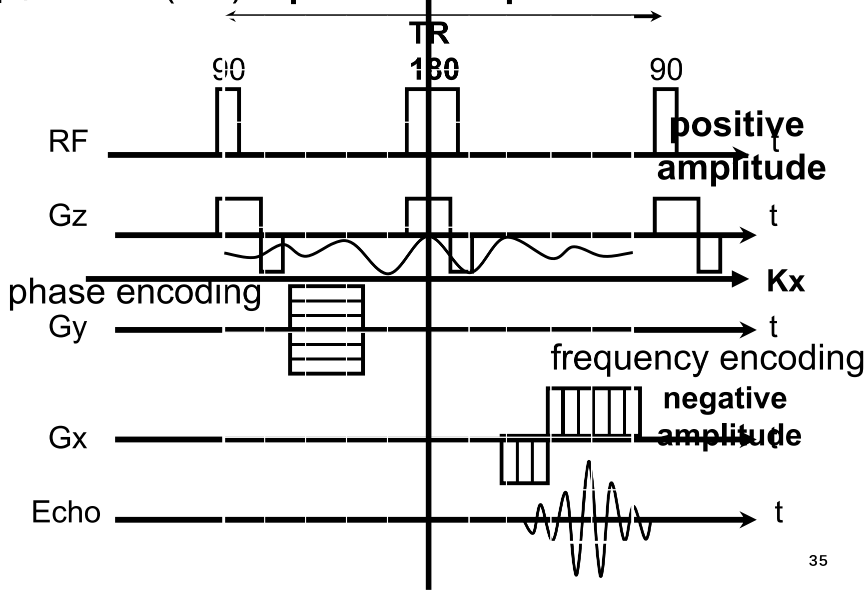
- We apply a gradient in the negative direction that has an area equal to ½ of that of the readout gradient.



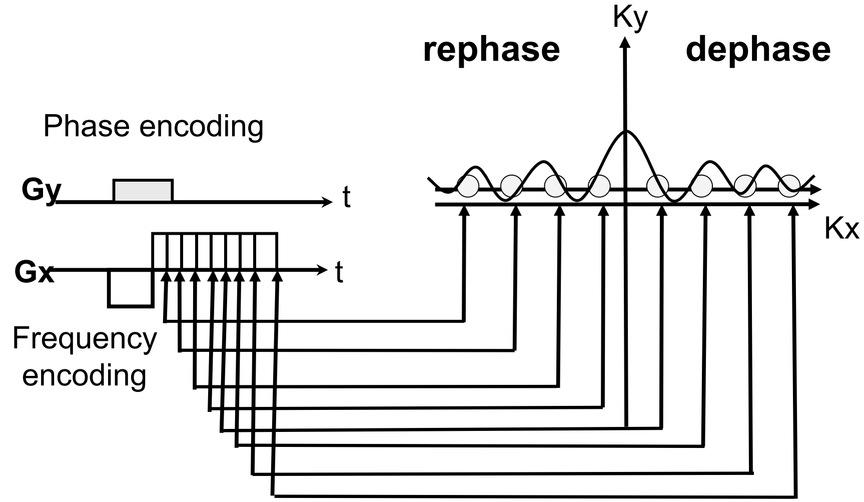
梯度磁場下的訊號衰減



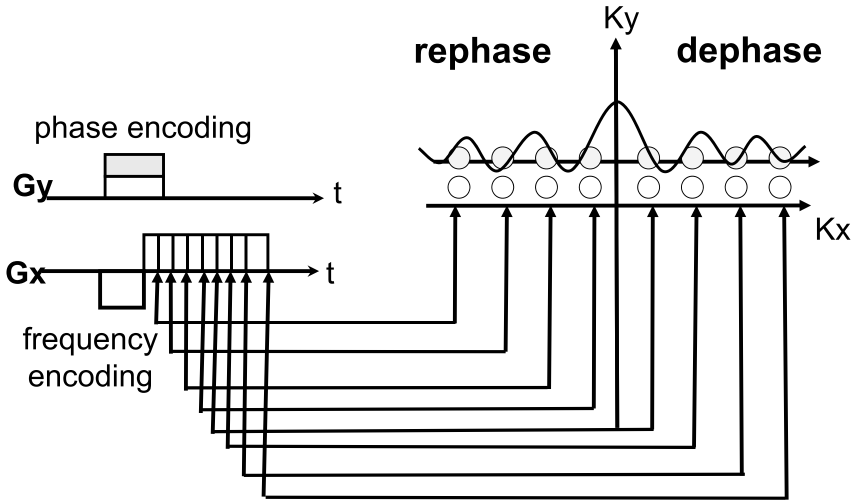
Spin echo (SE): K-space 簡圖(SE)



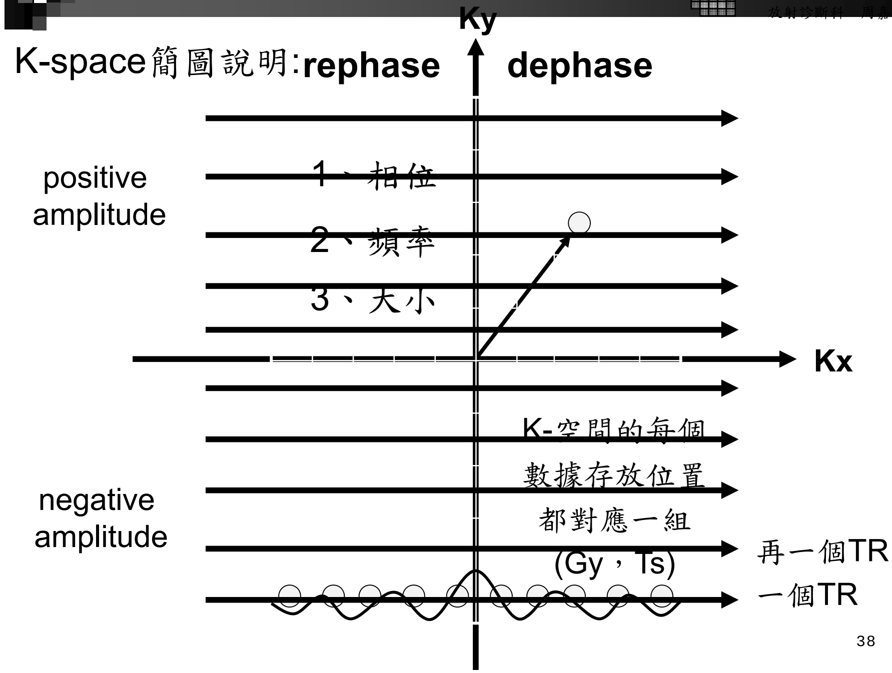
- Spin Echo K-space



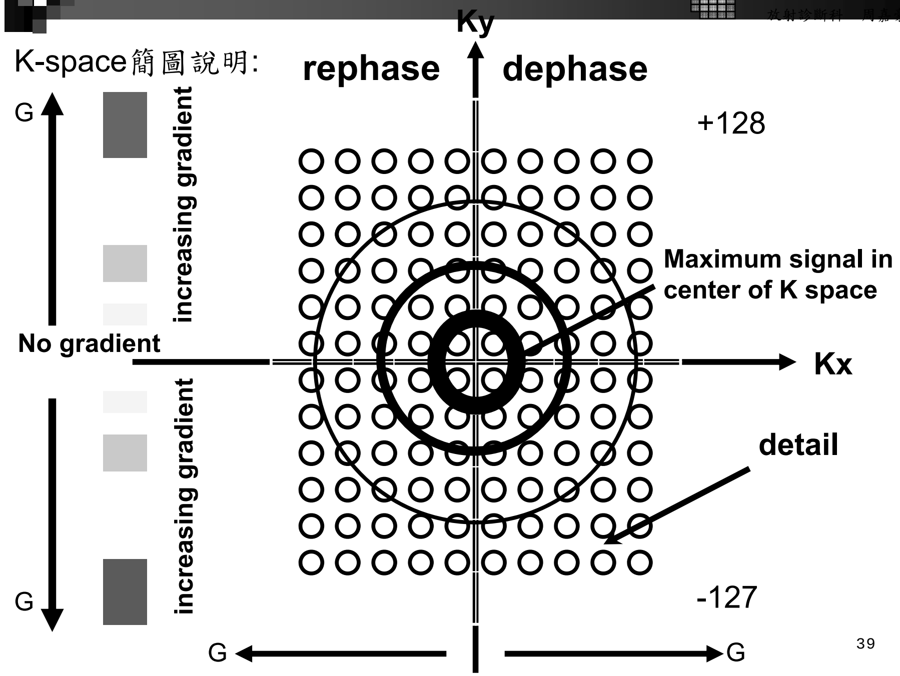
- Spin Echo K-space



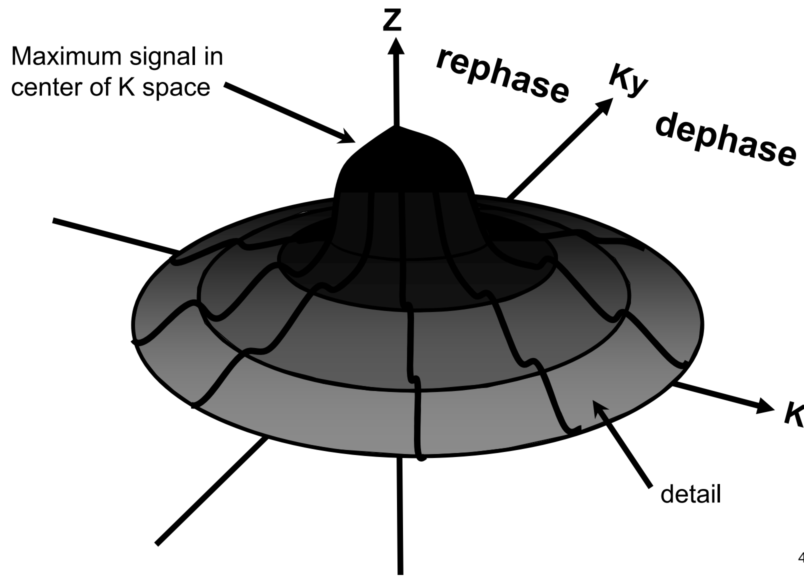
K-space簡圖說明:



K-space簡圖說明:

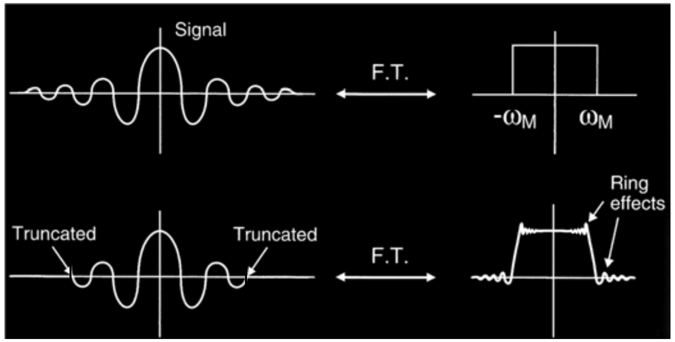


3維(3D)坐標方向上的K-space:



K-空間的邊緣(Edges of k-Space)

- The periphery of k-space provides information regarding “fineness” of the image and clarity at sharp interfaces.
- The FT of a truncated sinc function has ring down effects.
- The shades of gray are determined by the magnitude or amplitude of signal at each pixel.



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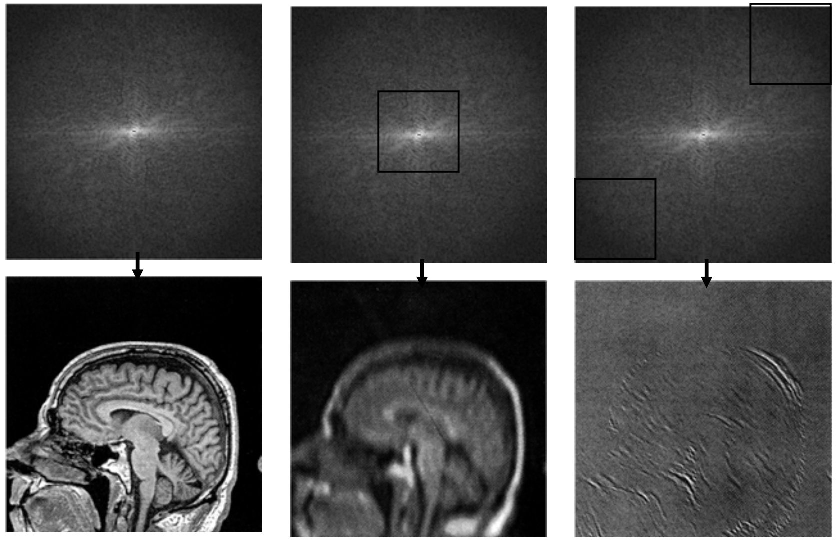
K-空間的邊緣(Edges of k-Space)



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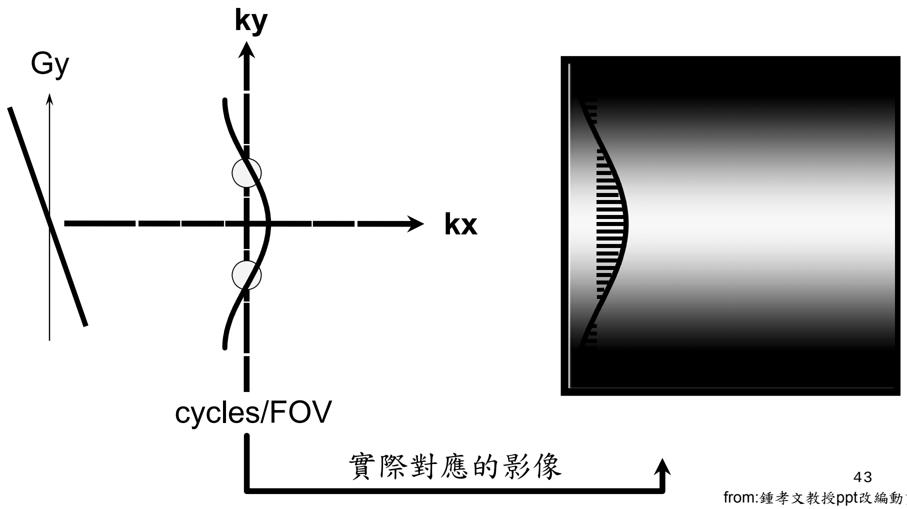
K-space位置和實際對應圖:

ky = +/- 1

kx = 0



from:鍾孝文教授ppt改編動畫

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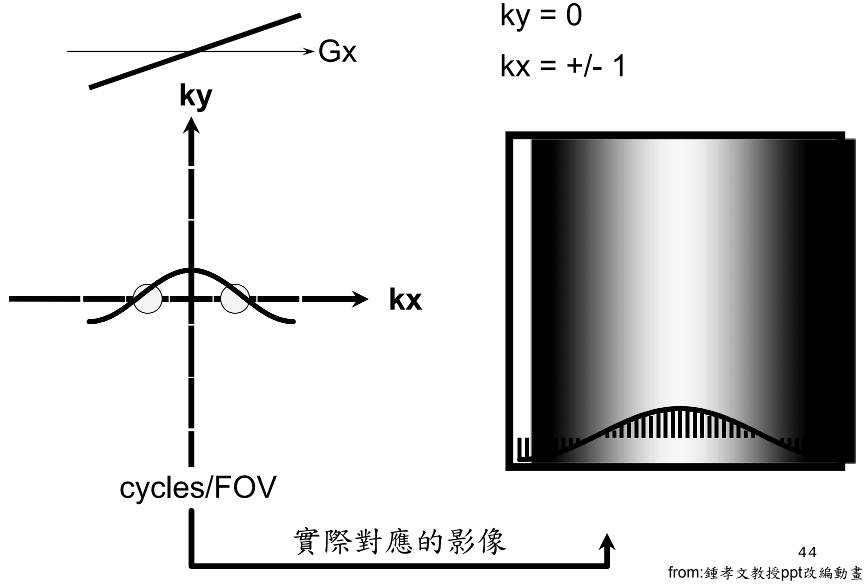
K-space位置和實際對應圖:

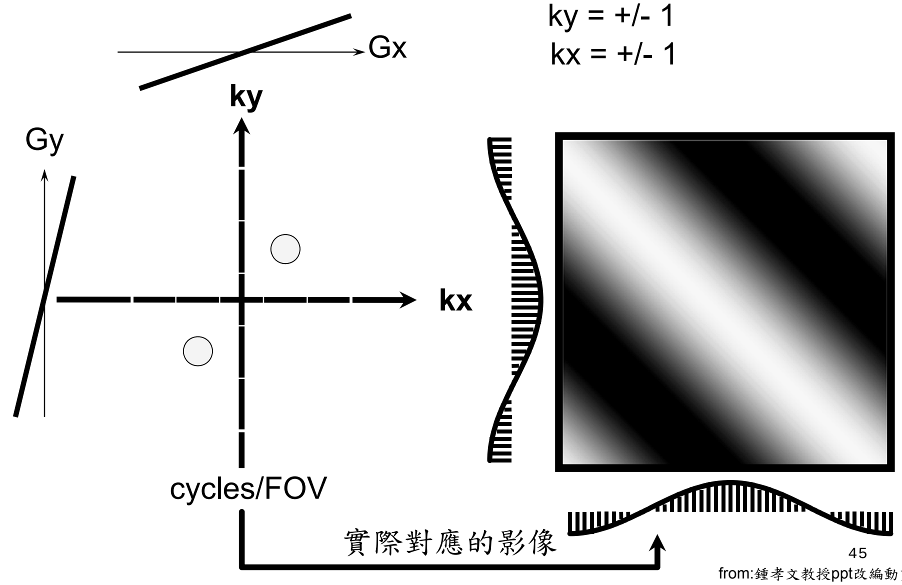
ky = 0

kx = +/- 1



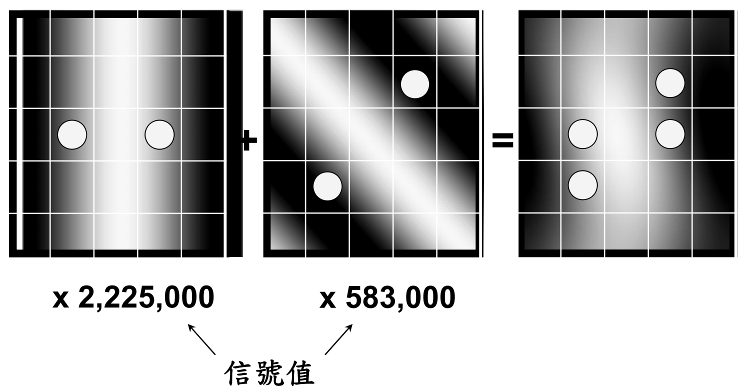
from:鍾孝文教授ppt改編動畫

K-space位置和實際對應圖:



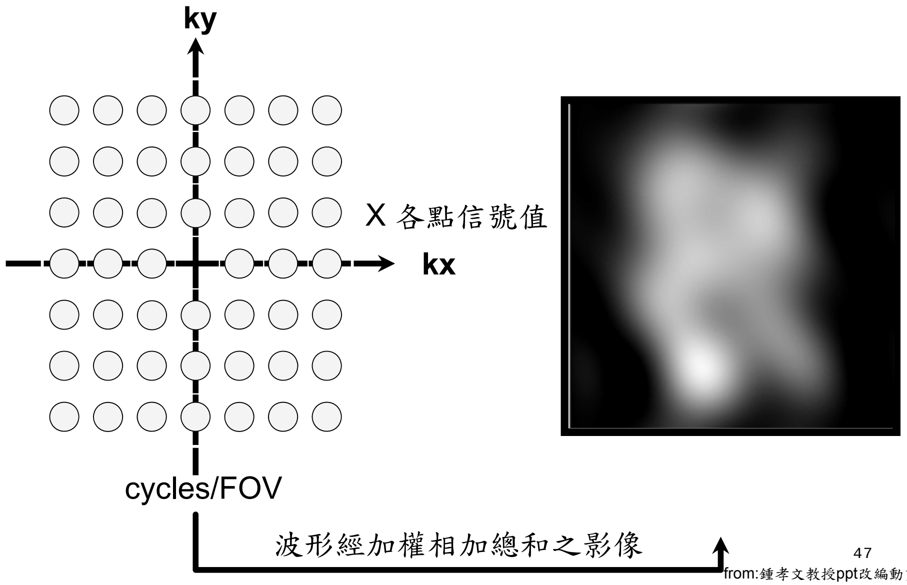
from:鍾孝文教授ppt改編動畫

影像 = 各波形成份的加權總和:



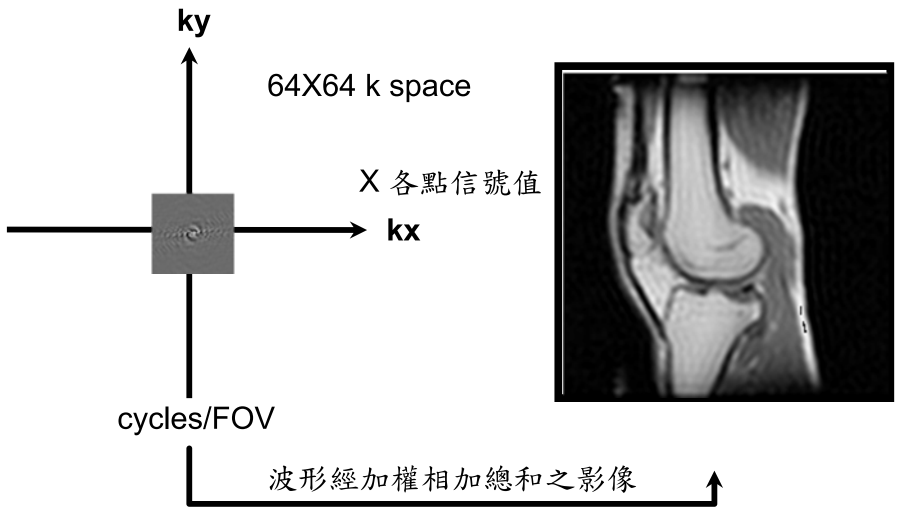
from:鍾孝文教授ppt改編動畫

影像 = 各波形成份的加權總和:



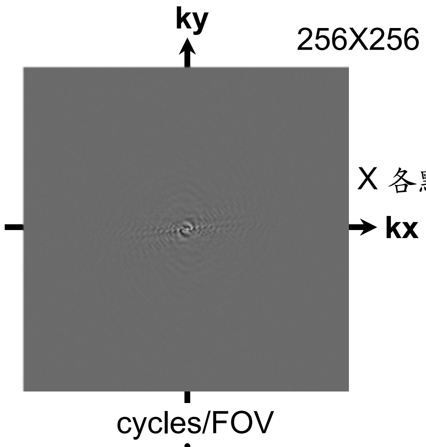
from:鍾孝文教授ppt改編動畫

影像 = 各波形成份的加權總和:



from:鍾孝文教授ppt改編動畫

影像 = 各波形成份的加權總和:

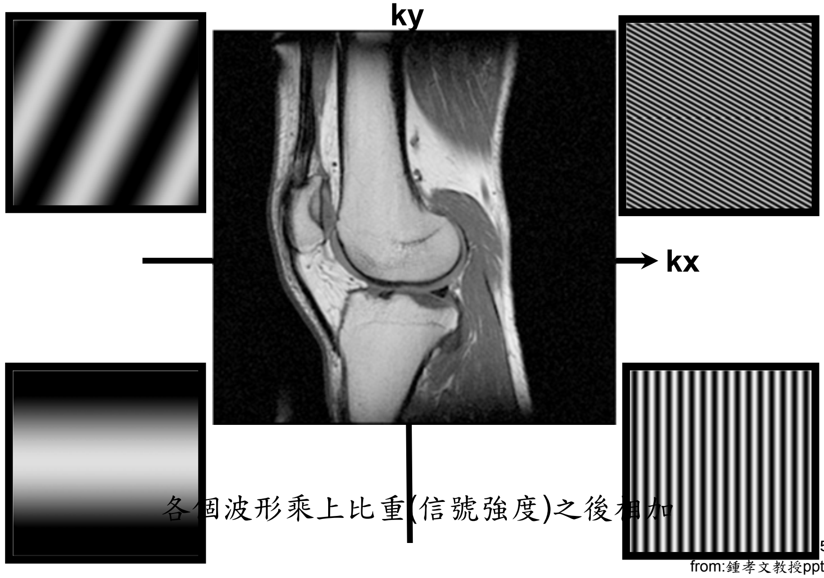


X 各點信號值

波形經加權相加總和之影像

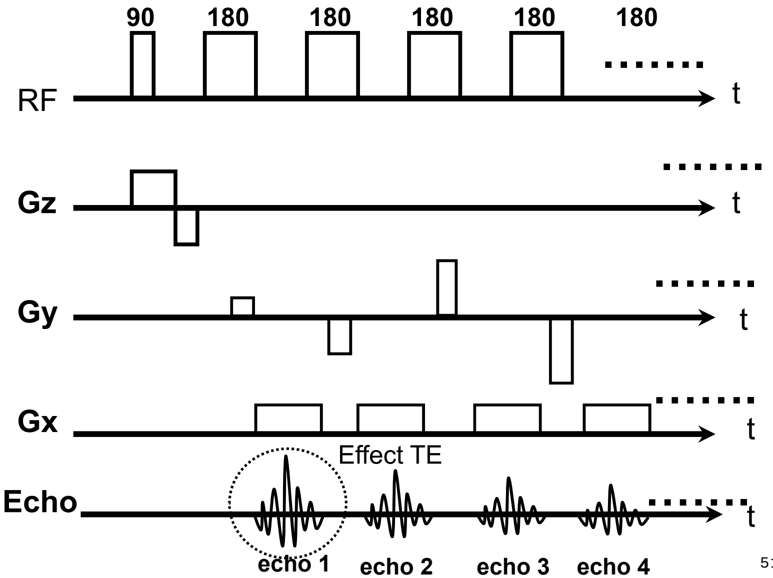
from:鍾孝文教授ppt改編動畫

k-space 數據 … 得到 MR 影像
數萬個不同的波形 … 種 … 波形:

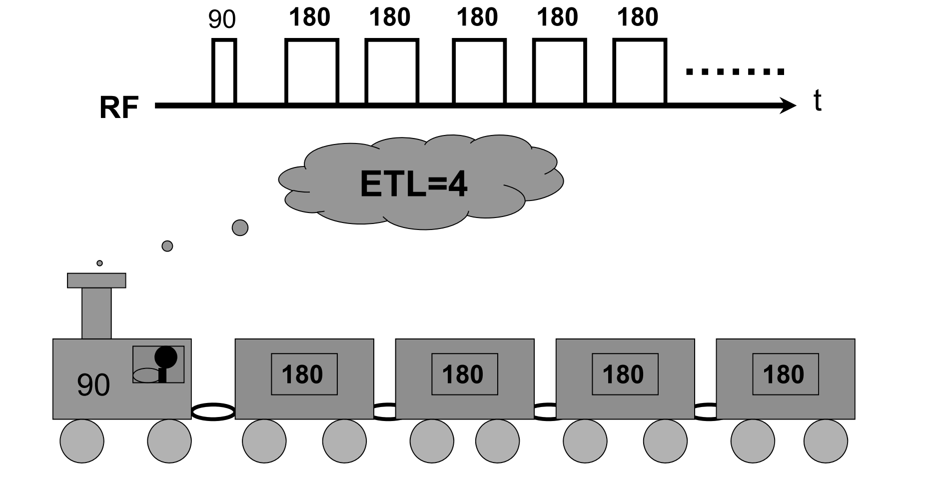


from:鍾孝文教授ppt改編動畫

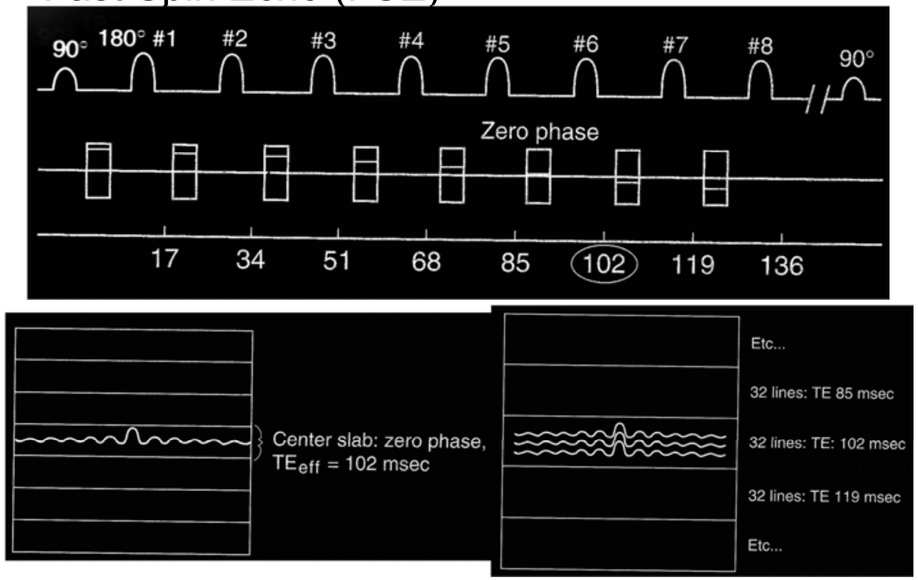
- Fast Spin Echo (FSE)



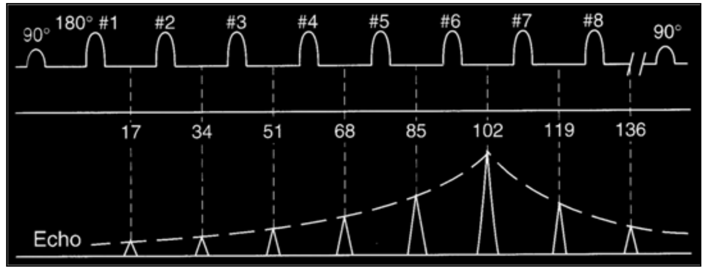
- Echo Train Length (ETL)



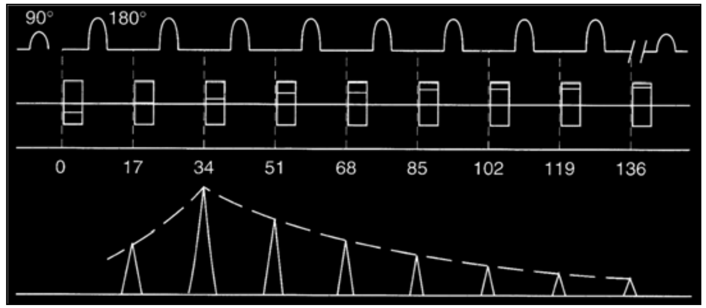
Fast Spin Echo (FSE)



Fast Spin Echo (FSE):對比平均



T2W為主



PDW為主

FSE優點

1.掃描時間降低(ETL)

2.SNR維持一樣(Ny相位不會改變)

3.合理的時間內作高解析度造影(512x512的內耳道造影)

4.運動假影比較不嚴重
(180^{o}等距離分開，偶數回音重聚相(even-echo rephasing))

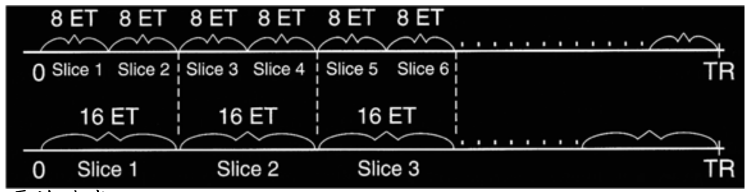
5.金屬物體在FSE影像中扭曲較小(含有許多rephase的180^{o})

6.FSE影像對不均勻磁場的磁鐵耐受度比CSE影像來得高(ETL)

FSE缺點

$$\text{切面數目} = \frac{TR}{TE+Ts/2+T_0}$$

1.切面數目減少，涵蓋範圍較低



2.對比平均造成:
•CSF在PDW影像中較亮
(極長TE的訊號，會有一些T2效應(CSF會較亮)，使用短的ETL和較高的BW)
•MS斑及其他在腦部的病灶可能被忽略(解決問題:較短的ETL和利用FLAIR)

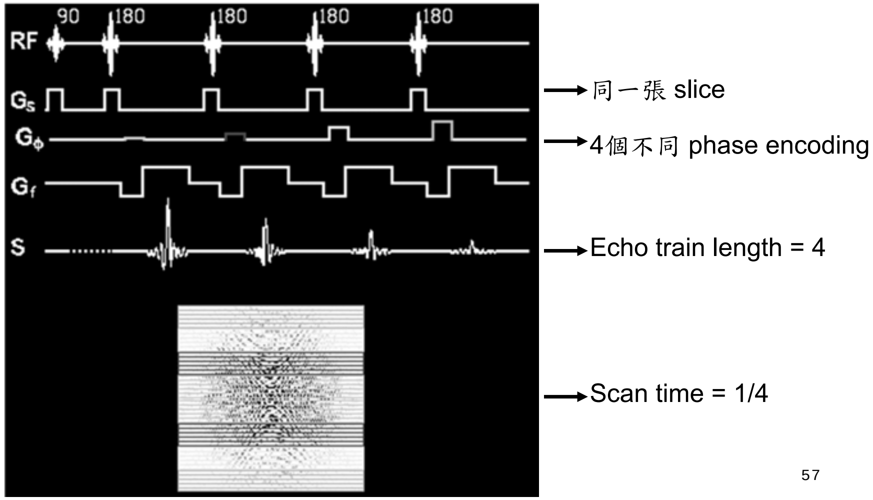
3.FSE中產生的MT轉移(許多快速的180^{o}脈衝，寬廣的BW造成)

4.正常椎間盤之T2W影像的亮度在FSE不如在CSE中(原因為MT造成)

5.感磁性效應比CSE低(出血不敏感)

6.脂肪在T2W的FSE像中較亮(解決問題: FAT SAT)

Fast SE:一個TR多次phase encoding



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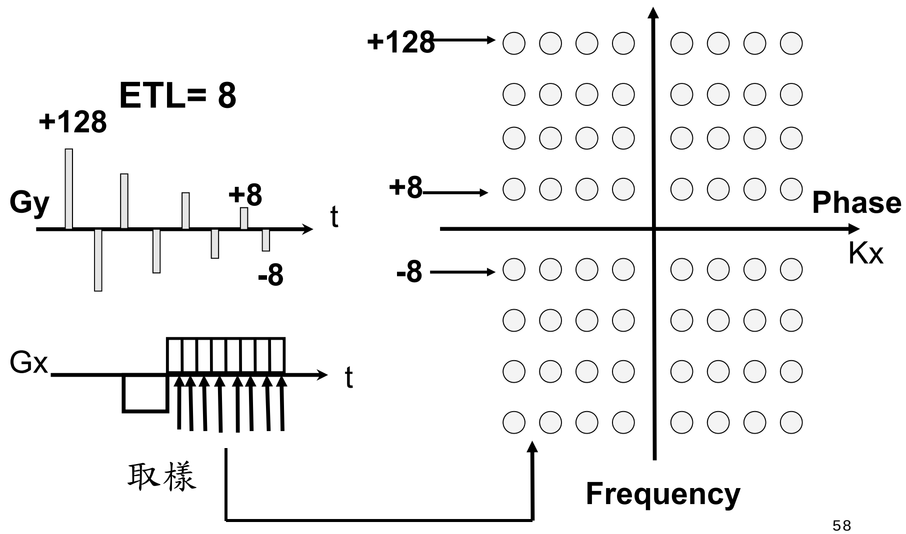
- FSE K-space



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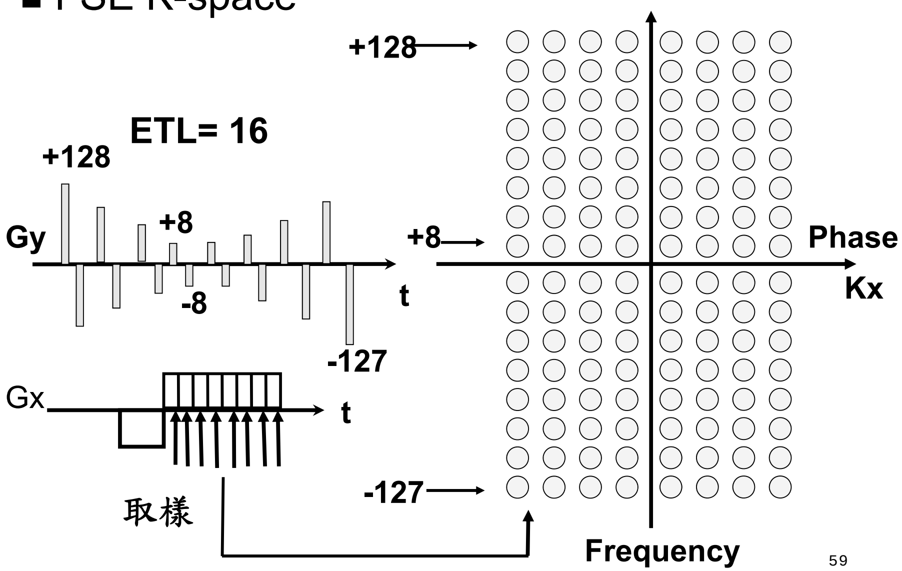
- FSE K-space



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Scan time (SE):

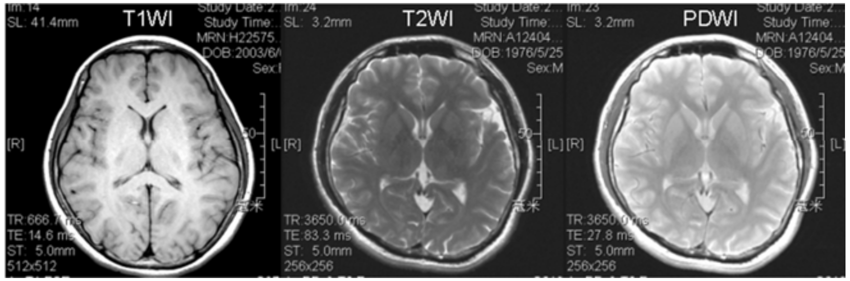
T = TR x Ny x NEX

- TR : Repetition time
- Ny : Phase encoding的次數
- Naq : Nex (Number of excitation)

- Scan time (FSE):

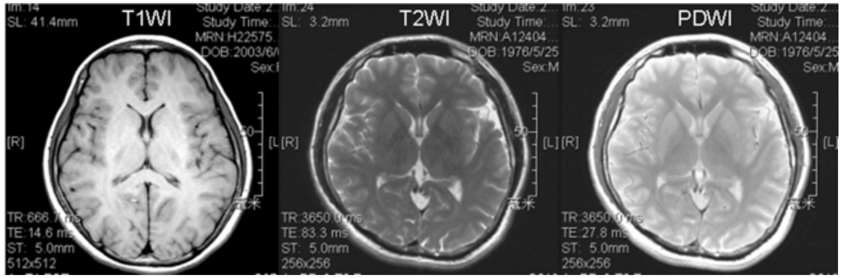
T = TR x (Ny/ETL) x NEX

Spin echo: acquisition time



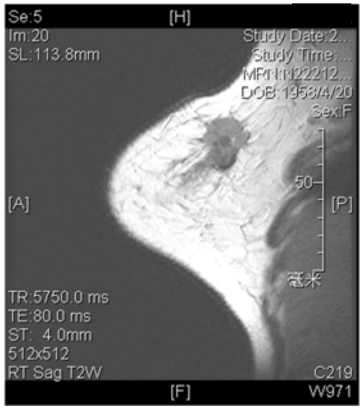
- T1WI: 666.7 msec (TR) x 512 (phase) x 1 (NEX)
  = 341350 msec = 341 sec = 5 min 41 sec
- T2WI: 3650 msec (TR) x 256 (phase) x 1 (NEX)
  = 934400 msec = 934 sec = 15 min 34 sec
- PDWI: 3650 msec (TR) x 256 (phase) x 1 (NEX)
  = 934400 msec = 934 sec = 15 min 34 sec

Fast SE: acquisition time

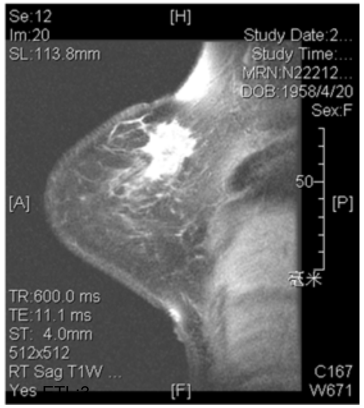


- T1WI: 666.7 msec (TR) x 512 (phase) x 1 (NEX) / 4 (echo train length)
  = 1 min 8 sec
- T2WI: 3650 msec (TR) x 256 (phase) x 1 (NEX) / 4 (echo train length)
  = 3 min 54 sec
- PDWI: 3650 msec (TR) x 256 (phase) x 1 (NEX) / 4 (echo train length)
  = 3 min 54 sec

Fast SE: 乳房磁振造影

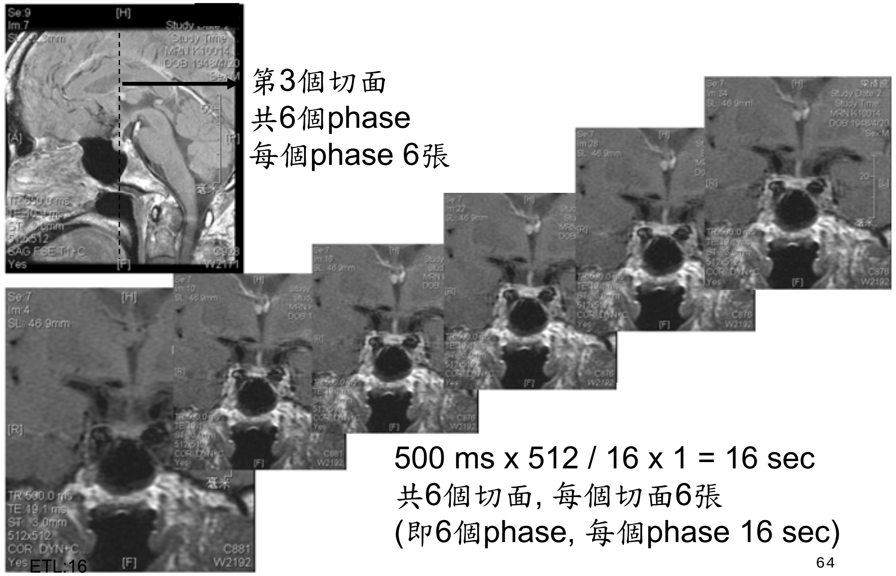


T2WI: 5750 ms x 512 x 1 / 14
= 210 sec
= 3 min 50 sec



T1WI+fat sat +C: 600 ms x 512 x 1 / 3
= 102 sec
= 1 min 42 sec

Fast SE: 動態腦下垂體磁振造影

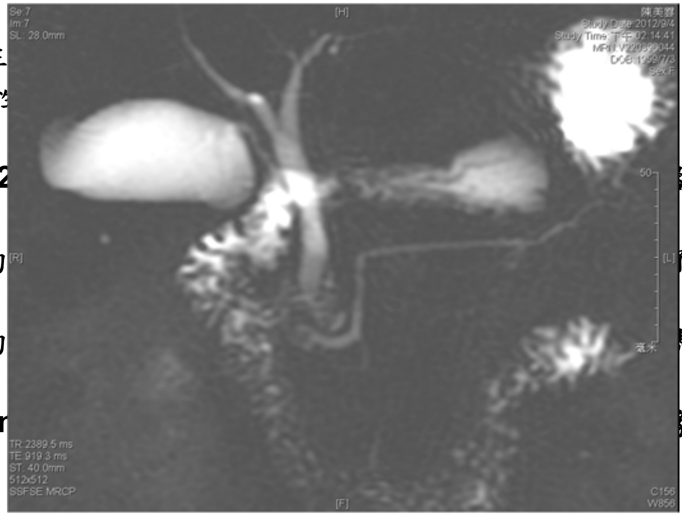


第3個切面
共6個phase
每個phase 6張

500 ms x 512 / 16 x 1 = 16 sec
共6個切面, 每個切面6張
(即6個phase, 每個phase 16 sec)

MR cholangiography (MRCP)(SSFSE)

- 不須注
  (阻塞性
- 利用T2 … 官顯影
- 静止的 … 像)
- 周圍的 … 號）
- MIP (r … 像

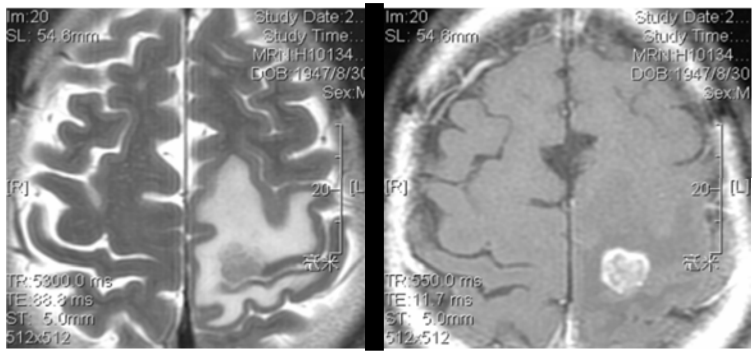


看圖說故事時間到了!!

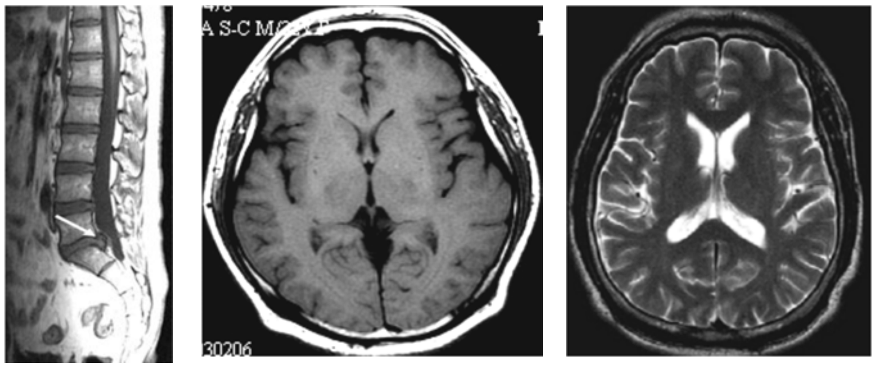
Sequence	TR Short	TR Long	TE Short	TE Long
SE	250–700	> 2000	10–25	> 60

Note.—Values are given in milliseconds.

- 各位放射師們請看影像，想想下列二個問題:
  1. 是檢查那個部位?是那種加權影像?
  2. 計算所需掃描時間?

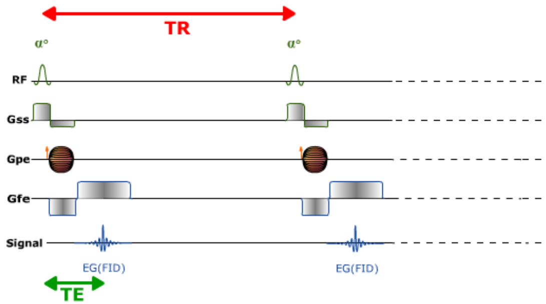


看圖說故事(2)



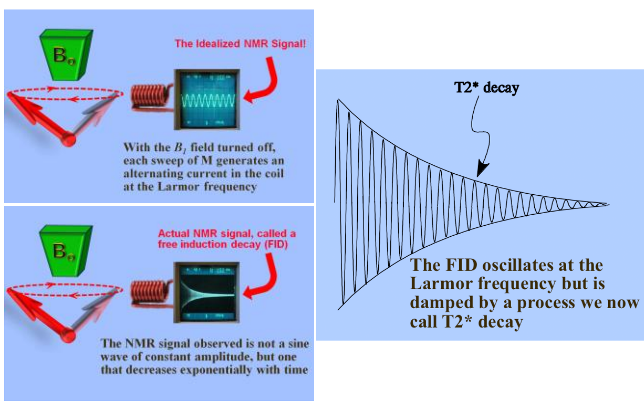
Thanks for your attention!

Gradient echo



What is a free induction decay (FID)?

- Pulsed methods :the main magnetic field is held constant while an RF-field at the Larmor frequency is pulsed on and off.
- Immediately after the RF pulse , transverse magnitude dephasing and induce signal called "nuclear induction decay" or "free induction," which today is commonly referred to as the ***free induction decay (FID)***.



Characteristics

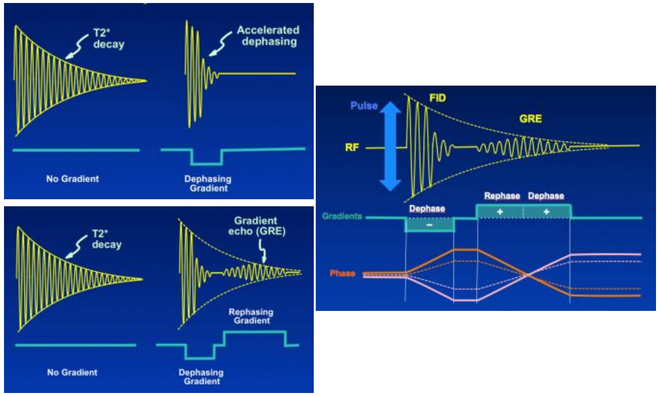
- The gradient echo sequence differs from the spin echo sequence in regard to:

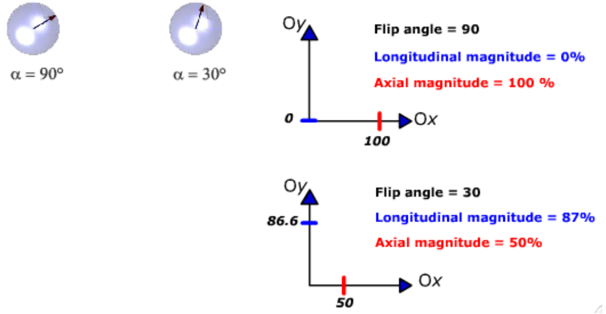
--- the flip angle usually below 90°

--- the absence of a 180° RF rephasing pulse

Type of MR Signal	Method of Formation
Free induction decay (FID)	1 RF pulse
Gradient echo (GRE)	1 RF pulse + gradient reversal
Spin echo (SE)	2 RF pulses
Stimulated echo	3 or more RF pulses

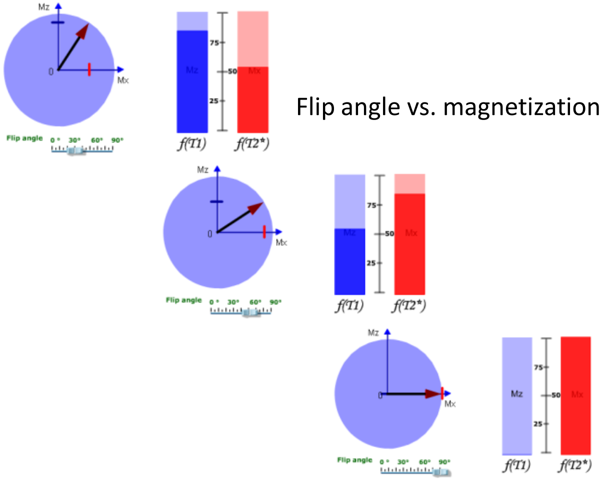
What is a gradient echo, and how does it differ from an FID?



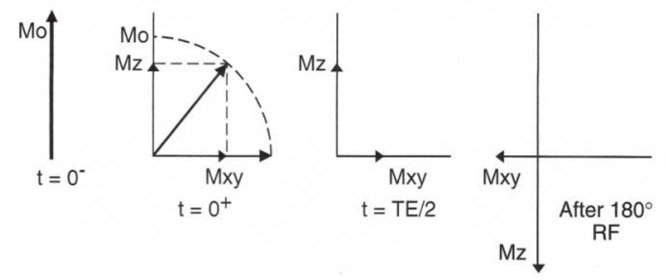


The advantages of low-flip angle excitations

- A flip angle lower than 90° (partial flip angle) decreases the amount of magnetization tipped into the transverse plane.
- The consequence of a low-flip angle excitation is a faster recovery of longitudinal magnetization that allows shorter TR/TE and decreases scan time.

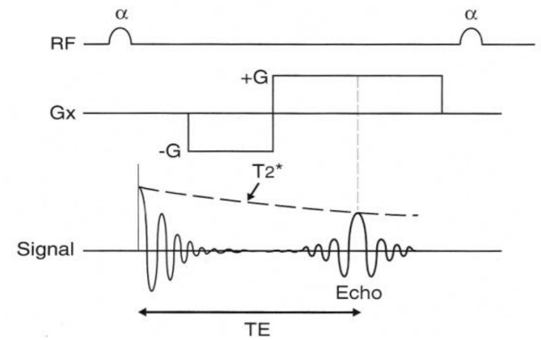


180° RF Pulse In GRE

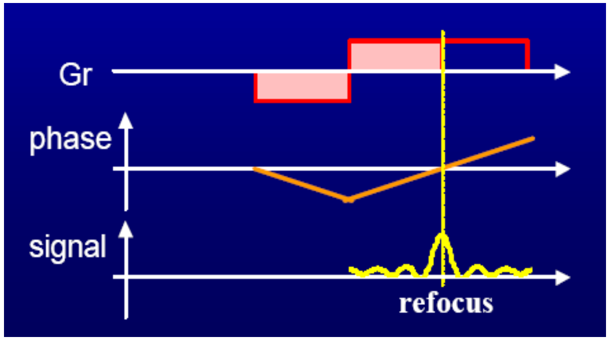


Echo time in GRE

Bi-lobed Gradient



Bi-lobed Gradient



Advantages of GRE

1. Increased speed
2. Increased sensitivity to magnetic susceptibility effects of hemorrhage
3. 3D imaging in reasonable time
4. Imaging of flowing blood (i.e MRA)

Disadvantages of GRE

1. Decreased SNR caused by small α, reducing the transverse magnetization, and very short TR
2. Increased magnetic susceptibility artifacts, most noticeable at air tissue interface such as in the region of paranasal sinuses or the absomen
3. T2* decay since there are no 180° rephasing pulses

The contrast of gradient echo

	Small	Large
α	↑ PDW	↑ T1W
TR	↑ T1W	↑ PDW
TE	↑ PDW	↑ T2*W

Short TRs

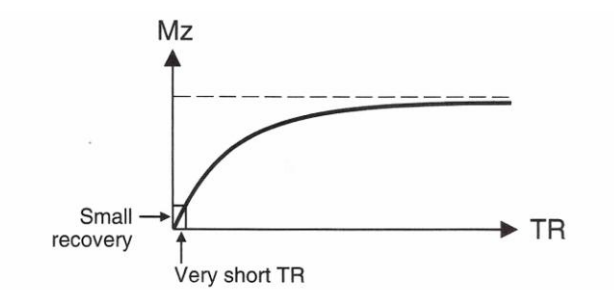


Figure 20-1 After a 90° pulse, longitudinal recovery after a short TR will be very small.

Short TRs

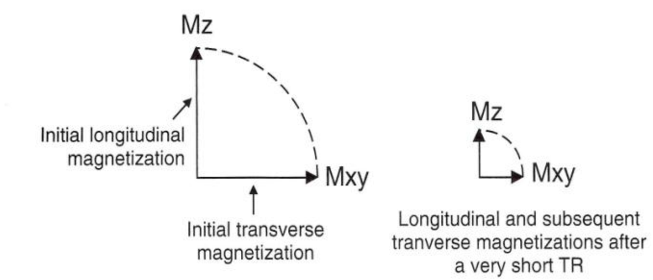
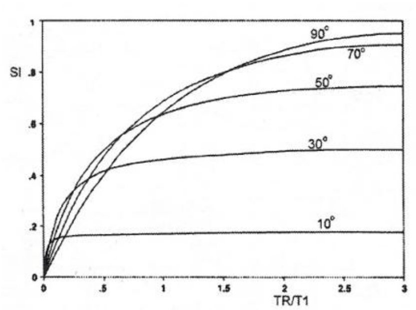


Figure 20-2 (A) The initial longitudinal and transverse magnetizations. (B) After a short TR, the subsequent longitudinal and transverse magnetizations will be smaller for a 90° RF pulse.

Small Flip Angle vs. short TR



1) α =90˚, TR=0.1 T1

→ Mz = 0.095 M_0

(2) α =25˚, TR=0.1 T1

→ Mz = 0.22 M_0

Tissue Contrast - FA

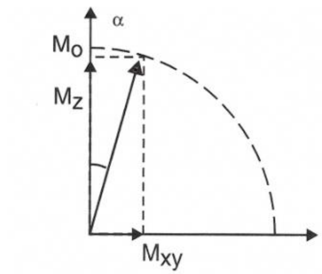


Figure 20-11 A small flip angle results in a large amount of longitudinal magnetization.

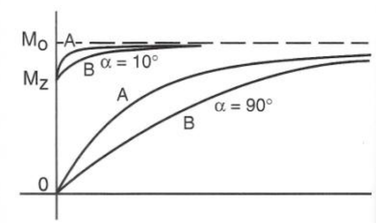
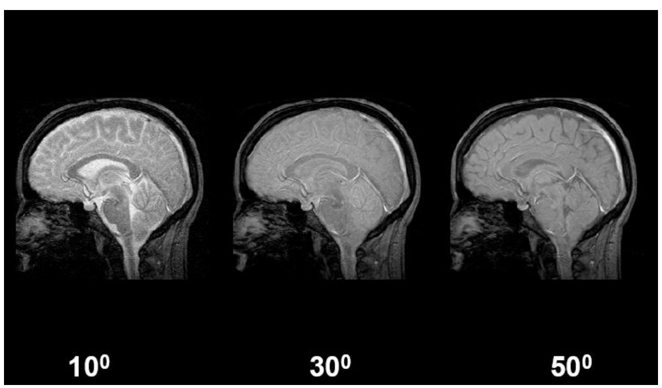
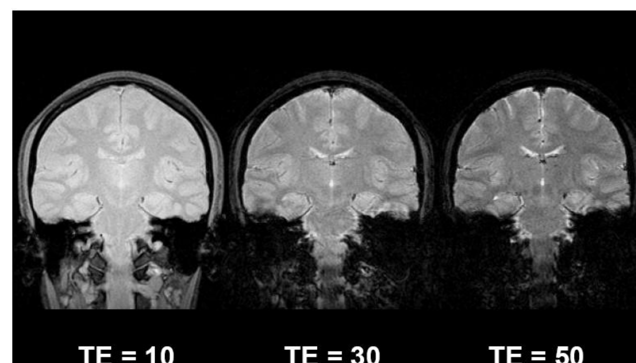


Figure 20-12 When the flip angle is small, it is difficult to discriminate the T1 contrast between two tissues. Thus, small α reduces T1 weighting.

Tissue Contrast - FA



Tissue Contrast - TE



Fast Scanning Techniques

GE	Siemens	Philips
GRASS	FISP	TFE
SPGR	FLASH	T1 FFE*
SSFP	PSIF	T2 FFE*
FSPGR	Turbo-FLASH	T1 TFE**

*TFE, turbo field echo

**FFE, fast field gradient echo

Steady state

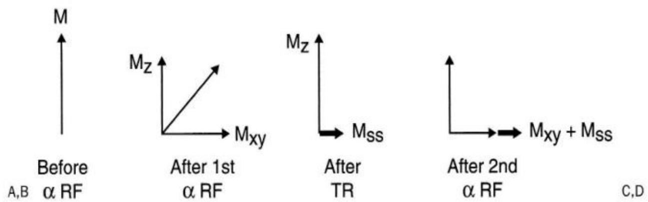


Figure 20-9 A–D: Because TR is short, a fraction of transverse magnetization remains at the end of the cycle, which eventually reaches a steady-state M_{ss}. This steady-state component is affected by the next RF pulse.

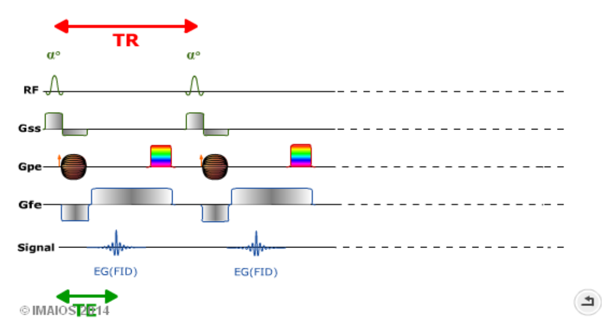
How can we do???

To mengent residual transverse magnetization is managed:

- gradient echo sequences with spoiled residual transverse magnetization
- steady state gradient echo sequences that conserve residual transverse magnetization and therefore participate in the signal.

Spoiled gradient echo sequences

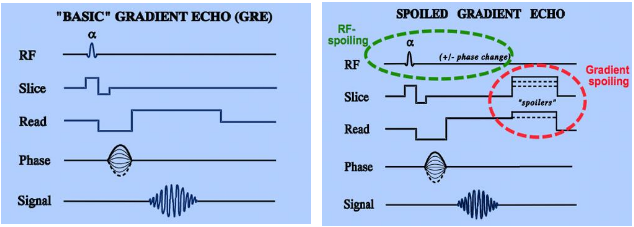
Type of sequence	Philips	Siemens	GE	Hitachi	Toshiba
Spoiled GE	T1-FFE	FLASH	SPGR MPSPGR	RSSG	RF-spoiled FE



Principles

- In certain cases, the steady state can be detrimental, namely for obtaining T1 weighted sequences.
- To resolve this problem, gradients and/or RF pulses (spoilers) are used to eliminate residual transverse magnetization.

Basic GRE vs. Spoiled GRE



SPGR (spoiled GRASS)

- The word “spoiling” refers to the elimination or spoiling of the steady-state transverse magnetization:

1. by lengthening TR
2. by applying variable gradient spoilers
3. by applying RF spoiling

Lengthening TR

- The method to achieve spoiling of Mss is by lengthening TR
- When TR is sufficiently large (generally over 200 msec), there is enough time to allow complete dephasing of the spins in the transverse plane

Variable Gradient Spoilers-2

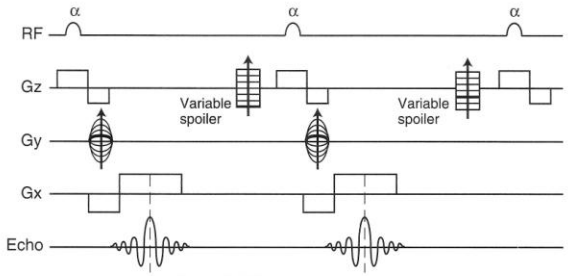
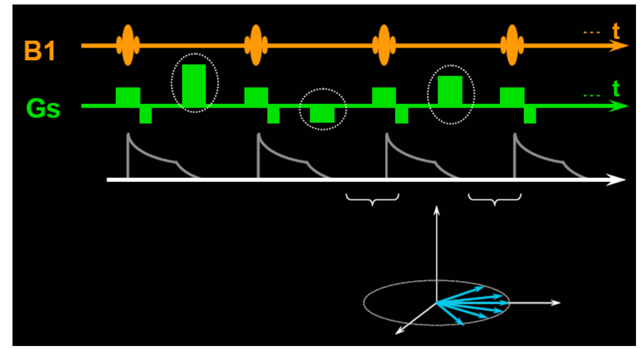
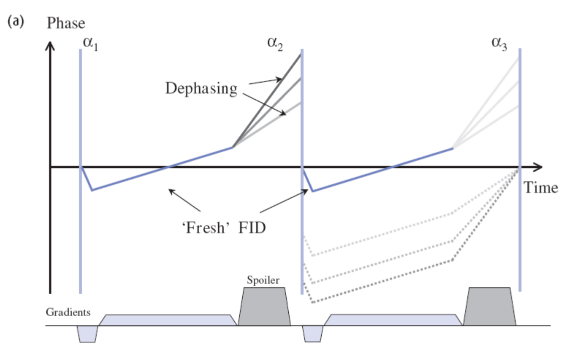


Figure 21-6 A PSD for spoiled GRE by using gradient spoilers.

Variable Gradient Spoilers





RF Spoiling (phase offset)

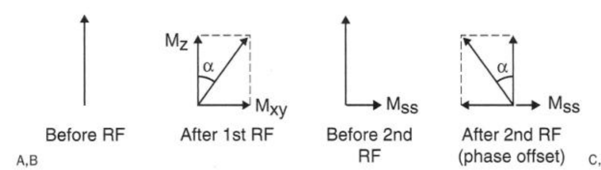
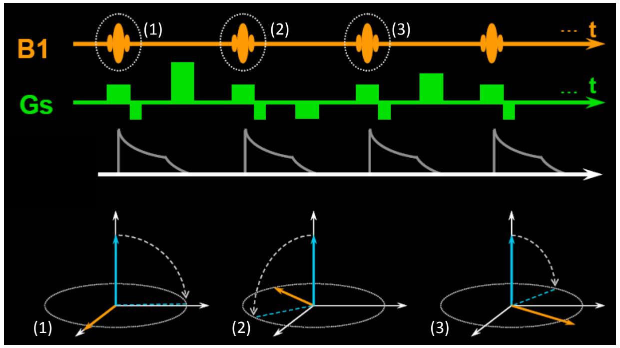
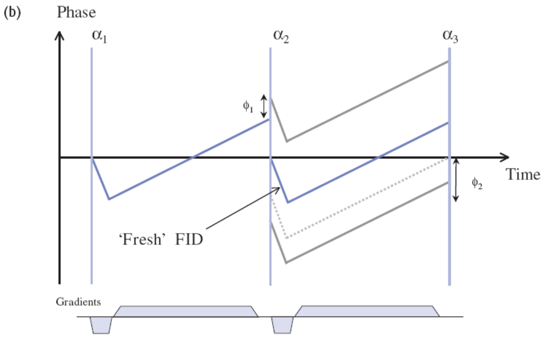


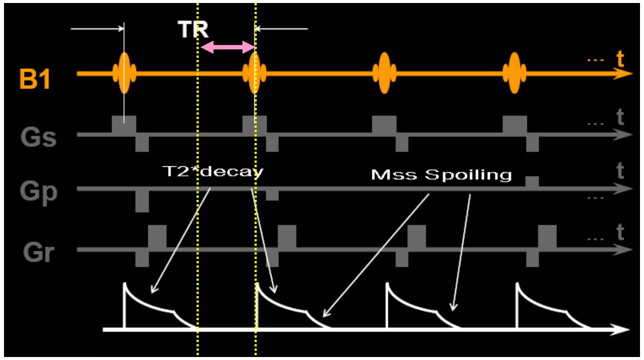
Figure 21-3A–D: Spoiling of the steady-state transverse magnetization can be done via RF spoilers (as in SPGR), in which a phase offset is added to each successive RF pulse.

RF Spoiling-2





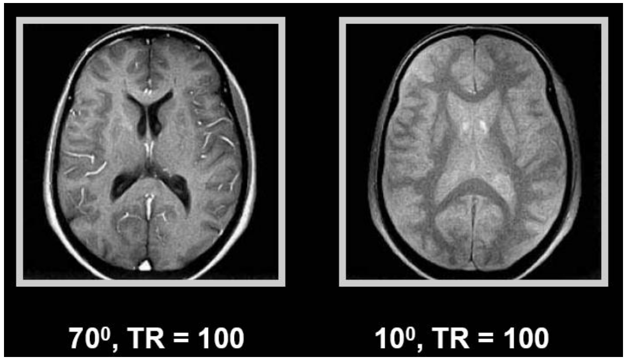
Mss Spoiled – TR can be shorten



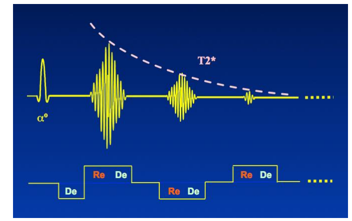
Disadvantage of SPGR(FLASH)

- Increased dephasing caused by inhomogeneities in B_0
- Increased magnetic susceptibility artifacts
- Increased chemical shift artifact

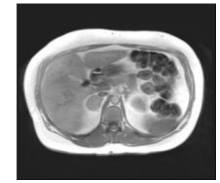
FLASH

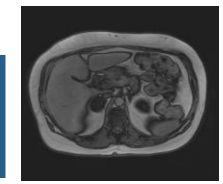


Dual/Multi-echo GRE



- Several echoes can be created after a single RF-pulse by sequential gradient reversals
- Permits "in-phase/out-of-phase" imaging
- Echoes may be combined (MERGE/MEDIC)

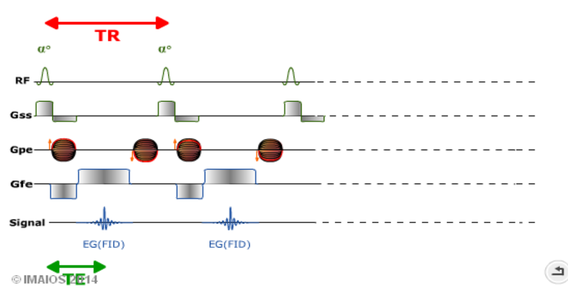




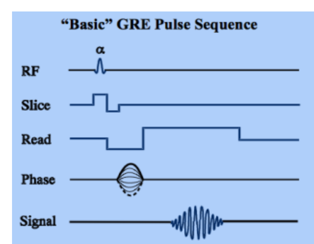
Steady-state gradient echo

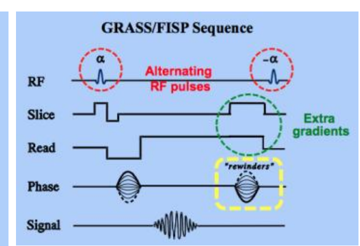
Type of sequence	Philips	Siemens	GE	Hitachi	Toshiba
Steady state GE	FFE	FISP	MPGR, GRE	TRSG	FE

Steady-state gradient echo



Basic GRE vs. GRASS/FISP





GRASS/FISP

- Residual Transverse Magnetization

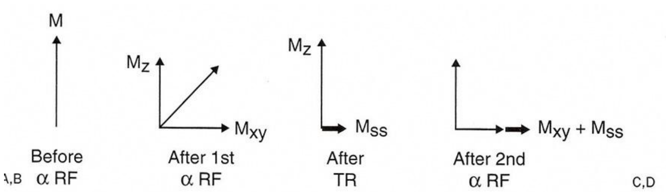


Figure 21-1A–D: A residual transverse magnetization that reaches a steady state M_{ss} remains after a short TR.

GRASS/FISP

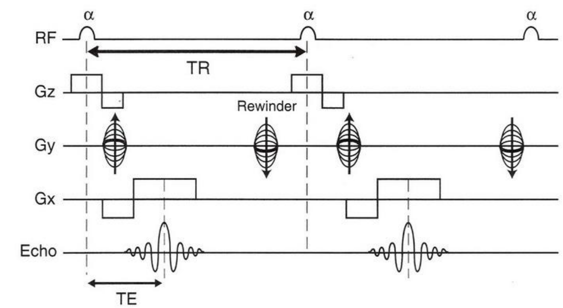
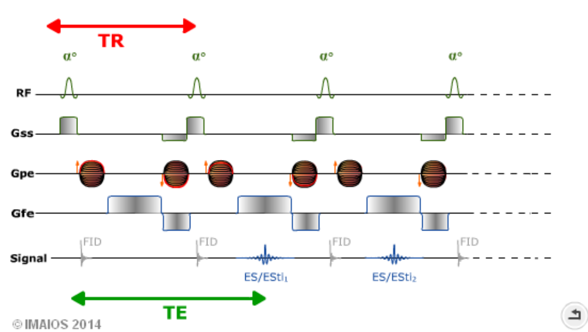


Figure 21-2 A PSD for GRASS/FISP. A "rewinder" gradient is applied along the y axis at the end of the cycle to reverse the effect of phase-encoding gradient.

T2-enhanced steady-state gradient echo

Type of sequence	Philips	Siemens	GE	Hitachi	Toshiba
T2-enhanced steady-state GE	T2-FFE T2	PSIF	SSFP		FE



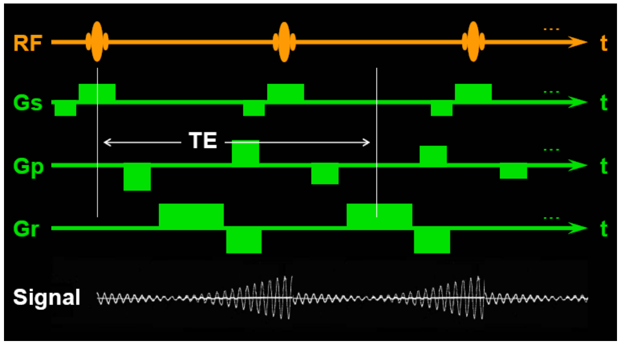
- In T2-enhanced steady-state gradient echo sequences:

--- residual transverse magnetization is conserved

-- the sequence is inverted in time, compared to the preceding sequences

--- only the echo corresponding to the Hahn echo, dependent on T2 but weaker than spin echo, is recorded

SSFP/PISF



TR<TE<2TR → T2W

SSFP/PISF

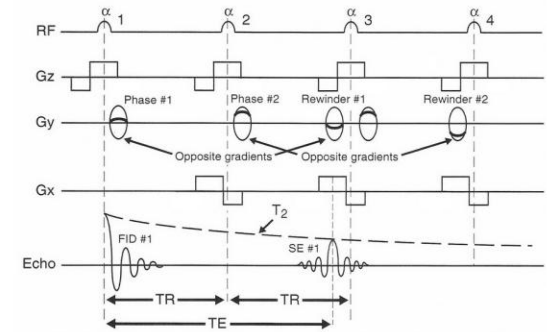


Figure 21-7 A PSD for SSFP/PSIF. Each α pulse contains some 180° pulse embedded in it that acts like a refocusing pulse. This in turn will result in a spin echo (SE) at the time of the next α pulse. Hence, contrast is determined by T2 (not T2*).

SSFP/PISF

Advantage of SSFP/PISF

- Decrease dephasing due to inhomogeneities in B_0 compare with GRASS and SPGR
- Decrease magnetic susceptibility artifacts compare with GRASS and SPGR
- Decrease chemical shift artifacts(dark band) compare with GRASS and SPGR

Disadvantage of SSFP/PISF

- Decrease SNR due to the use of longer TEs (TE>TR)
- Increase sensitivity to non-stationary tissue

The Characteristics of Various GRE

GRE Technique	SNR	CNR	Comments
GRASS/FISP	Highest	Best possible T2/T1	Preserves steady-state component
SPGR/FLASH	Intermediate	Best possible T1W	Spoiled steady-state component
SSFP/PSIF	Lower	Provides T2W	Gradient-recalled SE, TR<TE<2TR

The Characteristics of Various GRE

	GRASS/FISP	SPGR/FLASH
T1 contrast	-	Medium flip, Short TR/TE
Spin density contrast	Low flip, Medium TR, Short TE	Low flip, Medium TR, Short TE
T2/T1 contrast	Medium flip, Short TR/TE	-
T2* contrast	Long TE	Long TE

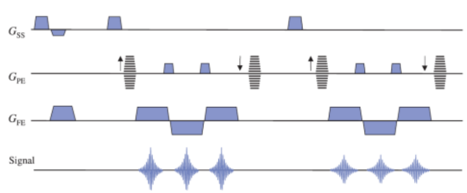
What is GRASE ?

- GRadient And Spin Echo (GRASE) or Turbo Gradient Spin Echo(TGSE)
- A fast segmented sequence that combines a multiple spin-echo train and intermediate gradient echoes

$$\text{Scan time} = \frac{\text{TR} \times N_{\text{PE}}}{N_{\text{spin echoes}} \times N_{\text{gradient echoes}}}$$

What is GRASE ?

- Some systems call the number of spin echoes the 'turbo factor' and the number of gradient echoes the 'EPI factor'
- Typically three gradient echoes will be used for each spin echo



Advantages of GRASE

- Much less RF power is used
- Higher 'turbo factors'
- More like T2-weighted spin echo than FSE

But-------

- substantial ringing artifacts in the phase encode direction

休息一下吧!!!

磁振造影專業基礎課程

MR組織壓抑技術

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MR Tissue Suppressi

□□□

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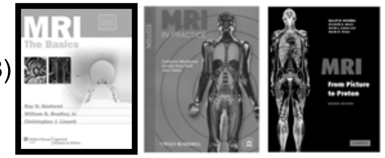
本次課程內容

- 瞭解並說明常用的組織壓抑技術, 包括下列幾項:
  - FLAIR, STIR, fast FLAIR , Double IR (Chapter 7,25,28)
  - Spatial presaturation (Chapter 23)
  - Chemical presaturation (Chapter 23,25)
  - Magnetization Transfer saturation (Chapter 25)

Reference:

1.MRI The Basics (3rd) (Chapter 7,23,25,28)
2.MRI IN PRACTICE(4td) (Chapter 5,6)
3.MRI From Picture to Proton(2nd)



組織壓抑技術
tissue suppression techniques

如何將組織信號抑制?

- 利用saturation的概念
- Tissue共振頻率不同
- T1 relaxation 的不同

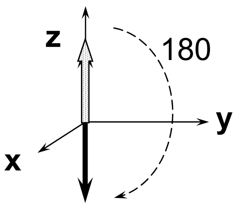
臨床上壓抑的主要對象:

- Suppress two common targets fat and water
- Suppress other silicone and blood

組織壓抑技術種類

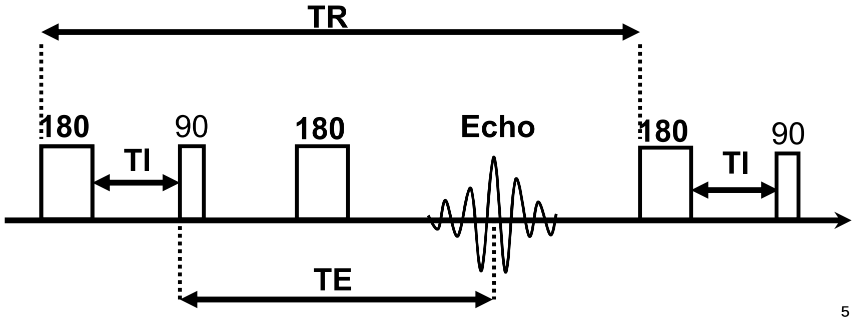
- 反轉回覆技術 (Inversion Recovery , IR)
  - STIR、FLAIR、fast FLAIR 、 DIR
- 化學(頻譜)預飽和 (chemical persaturation)
- 空間預飽和 (spatial persaturation)
- 磁量轉移 (Magnetization Transfer , MT)

MR時間-基本參數



磁化向量與T1 有關

- TR (repetition time)重複時間
- TE (echo time)回音時間
- TI (inversion time)反轉時間
- FA (flip angle)偏離角度

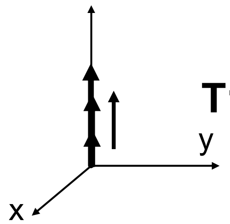


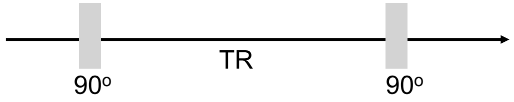
T1 弛緩現象

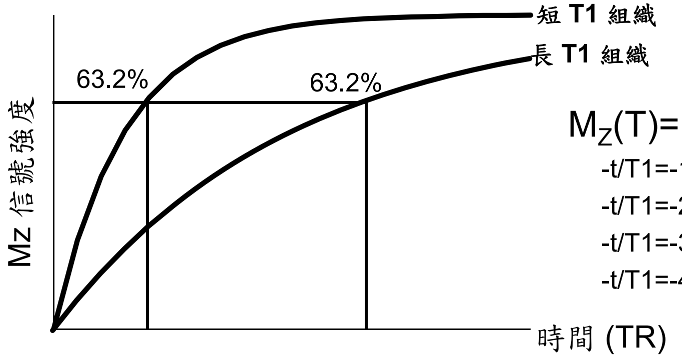
T1 弛緩 (T1 relaxation)

T1 value = Mz 恢復到 M_0 的 63.2% 所需之時間

T[illegible]衝的時間(ex:90°到90°的時間)







$$M_Z(T)=M_0(1-e^{-t/T_1})$$

-t/T1=-1, Mz=0.632

-t/T1=-2, Mz=0.865

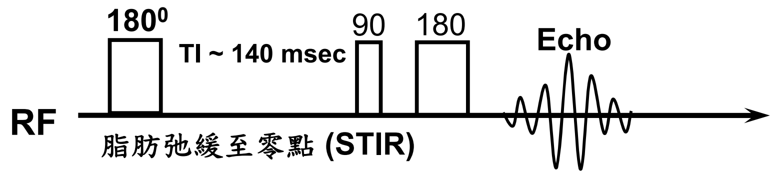
-t/T1=-3, Mz=0.950

-t/T1=-4, Mz=0.982

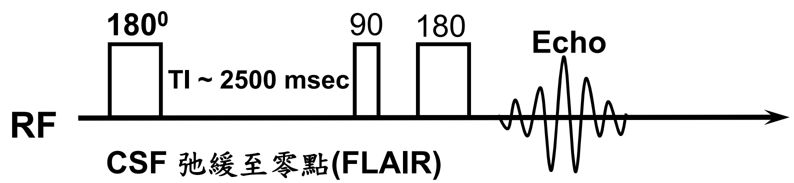
反轉回復技術 Inversion Recovery

- 180 度脈衝，所有磁向量均反向
- 等待 T1 弛緩 (TI： Inversion Time), 提供好的T1對比
- 到脂肪(或CSF)回復至零點後開始取像
- 1.5T 下 TI 通常約 140 msec (STIR-抑制fat)
  (Short Tau Inversion Recovery)
- 1.5T 下 TI 通常約 2500 msec (FLAIR-抑制CSF)
  (Fluid-Attenuated Inversion Recovery)

STIR、FLAIR (1.5T)

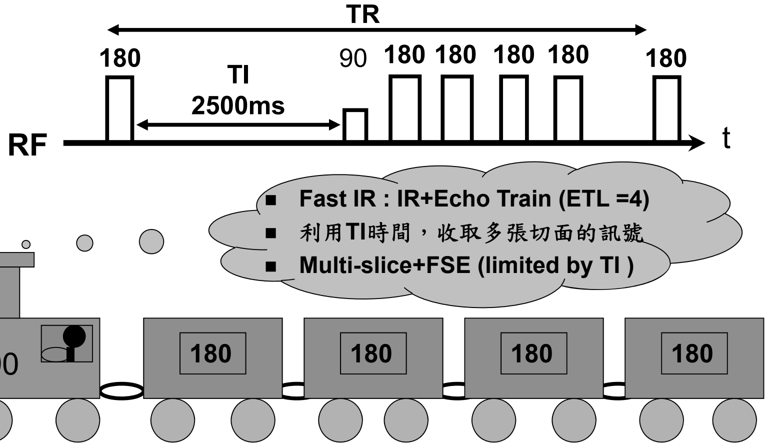


脂肪弛緩至零點 (STIR)

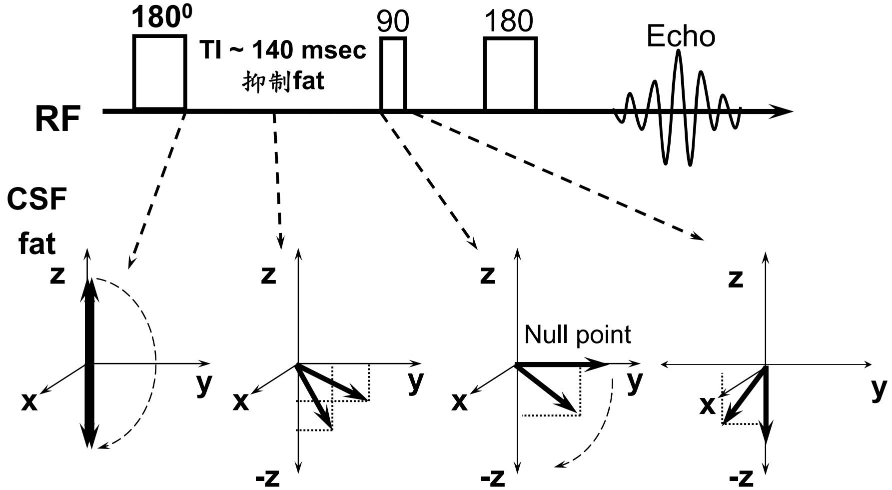


CSF 弛緩至零點(FLAIR)

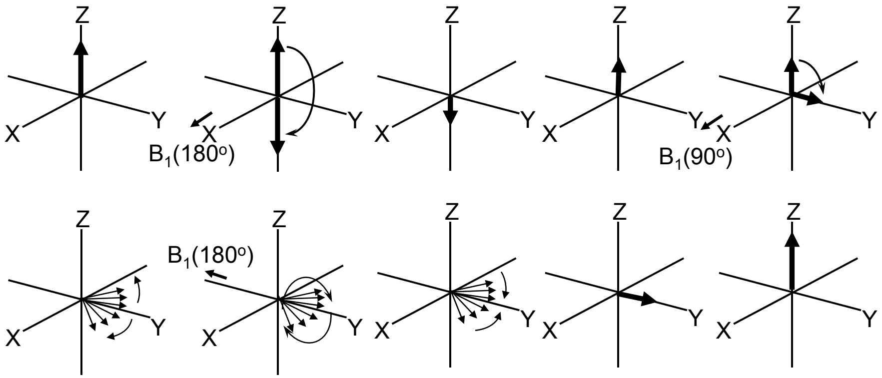
反轉回復技術 – fast FLAIR
IR + Echo Train (ETL 回音列車)



反轉回復原理示意圖



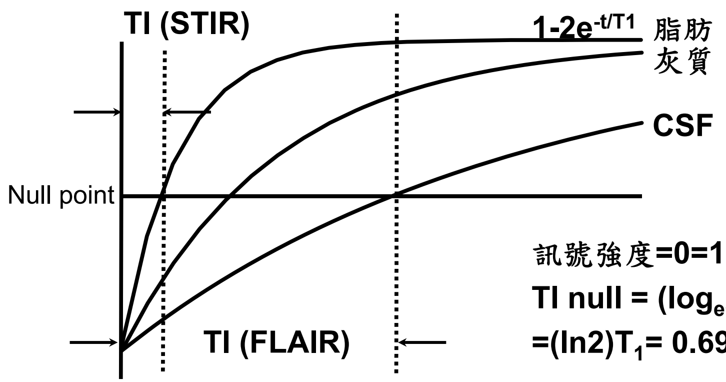
磁化量在IR波序上的變化



反轉回復曲線

$SI=M_0(1-2e^{-TI/T1})(1-e^{-TR/T1})$

90°



訊號強度=0=$1-2e^{-TI/T1}$

TI null = $(\log_e 2)T_1$

$=(\ln 2)T_1 = 0.693T_1$

TI 時間的長短可以決定抑制何種組織的信號

- STIR TI (fat) at 1.5 T = 0.693 x 200 msec = 140 msec
- FLAIR TI (CSF) at 1.5 T = 0.693 x 3600 msec = 2500 msec

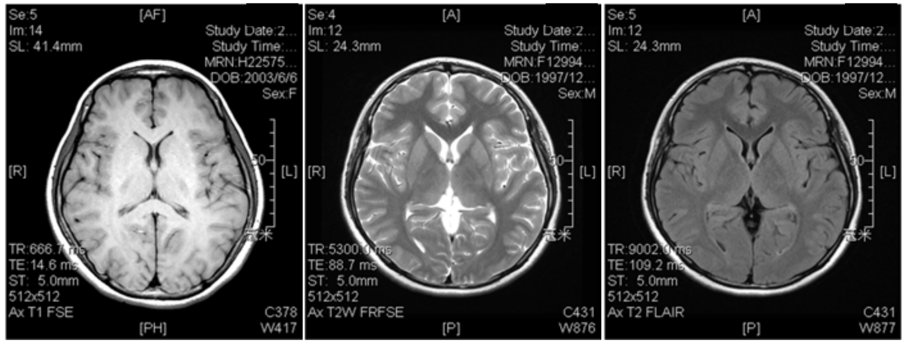
STIR 特性
IR Advantages & Disadvantages

- 不須特別均勻主磁場，影像含有 T1 成份
- Inverse T1 weighting
  - 長 T1 組織： 亮
  - 短 T1 組織： 暗
  - 與普通短 TR 之 T1 影像相反
  - CSF > edema > 灰質 > 白質

IR advantages	1.沒有明顯額外的RF加熱 2.不會有磁場不均度所造成的變動
IR disadvantages	1.相似T1值的組織通通被壓制掉而無法區別(ex: Gd effect) 2.長TR造成的長擷取時間 (3T的話TI就會更長) 3.較低的SNR(部分被飽和的關係)

Fluid-Attenuation Inversion Recovery
利用IR技術把水的信號壓掉

提供沒有CSF信號T2WI，降低灰白質對比，形成灰色背景顯現病灶

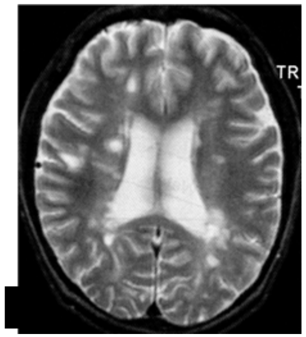


T1W　　T2W　　FLAIR

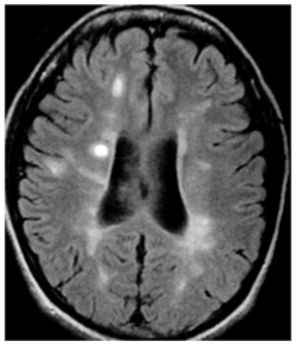
通常病灶有較長的T2時間(cortical lesion, peripheral subcortical, periventricular)
故利用T2W影像進行診斷，靠近CSF的病灶不好辨識(high CSF SI)

Fluid-Attenuation Inversion Recovery
利用IR技術把水的信號壓掉

多發性硬化病灶multiple sclerosis (MS)



T2W

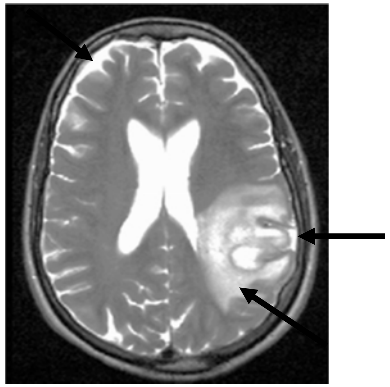


FLAIR

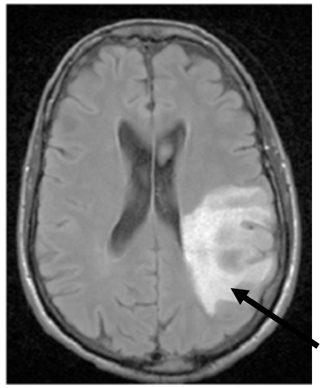
MS:神經傳導有關，發生在白質，會產生去髓鞘化的動作，會變白質產生病變

Fluid-Attenuation Inversion Recovery
利用IR技術把水的信號壓掉

Glioblastoma 神經膠母細胞瘤



T2W



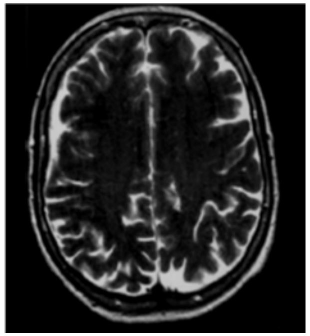
FLAIR

腦在病變(tumor)時產生的edema會和CSF接近，用FLAIR區分

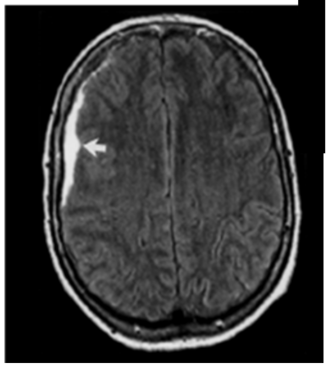
Andrea Hawkins-Daarud, Front. Oncol., 04 April 2013

Fluid-Attenuation Inversion Recovery
利用IR技術把水的信號壓掉

蜘蛛網膜下出血(SAH)及腦室內出血的診斷很有用

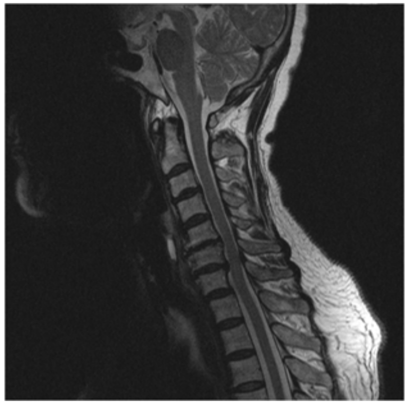


T2W



FLAIR

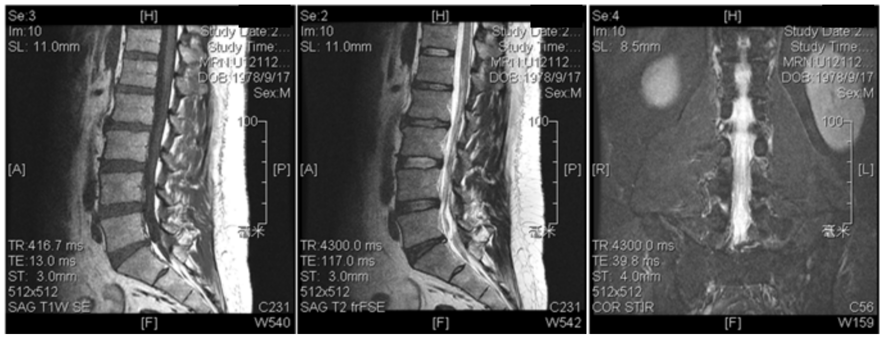
Short Tau Inversion Recovery
利用IR技術把脂肪的信號壓掉



T2W

STIR (SNR較差)

Short Tau Inversion Recovery
利用IR技術把脂肪的信號壓掉

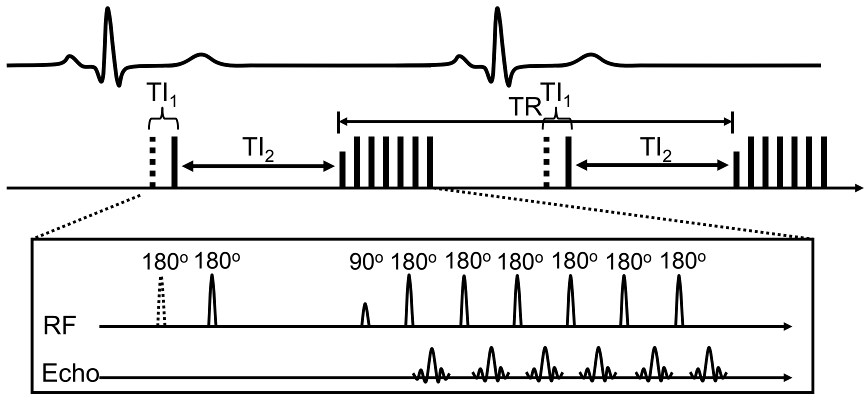


T1W　　T2W　　STIR

Double IR: black blood imaging

常使用於心臟血管磁振造影
可以清楚看到血管壁的構造

Double IR: black blood imaging原理

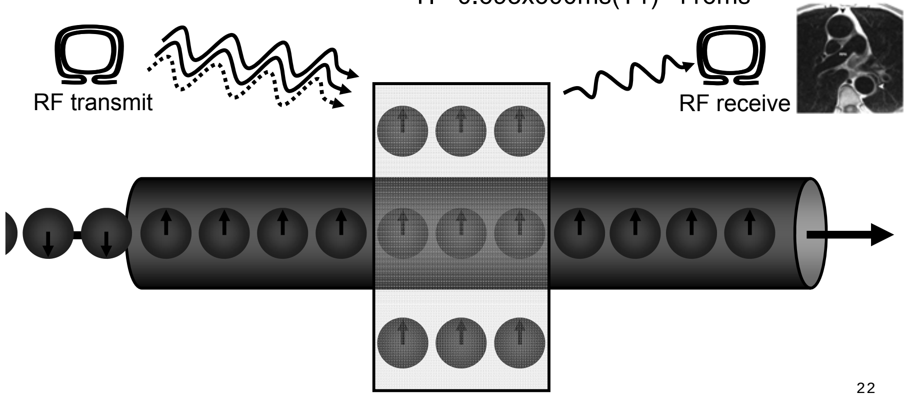


non-selective 180º → slice-selective 180º → TI flowing-in blood at nulling point → Fast Spin Echo T1WI

Double IR: black blood imaging的原理

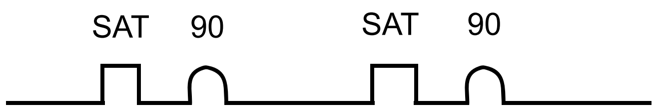
non-selective 180º → slice-selective 180º → TI flowing-in blood at nulling point → FSE T1WI 90-180-180…

TI =0.693x600ms(T1)=416ms



化學(頻譜的)預飽和
Chemical (Spectral) persaturation

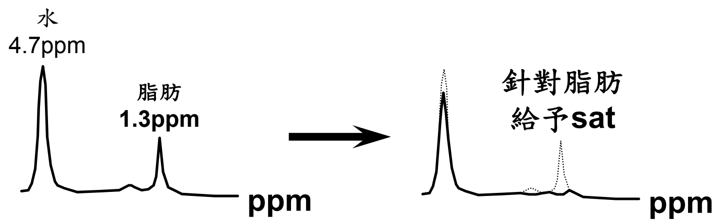
- RF脈衝施加之前加一個預飽和脈衝
- 消除脂肪和特殊的縱向磁量 CHESS: CHEmical Shift Selective
- SAT類似於90度(適當的選擇頻率進行飽合)
- 臨床上又叫fat sat ，主要是用來作脂肪的壓抑
- 受磁場不均勻度的影響
- Breast，Muscle，Bone，Spine常用
- Brain用的比較少(fat較多在皮下)



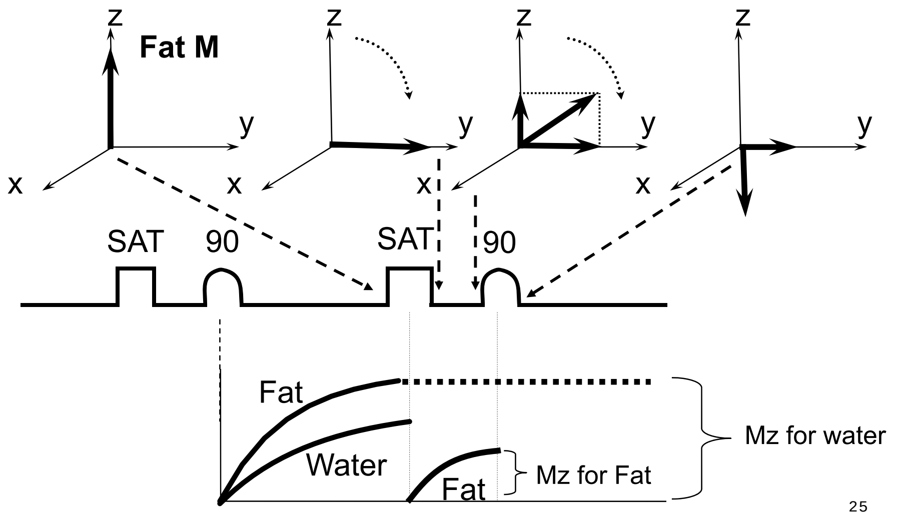
Chemical shift effect

- 水分子的質子的 Lamor frequency 比脂肪的質子的 Lamor frequency 快了3.5 ppm (parts per million)

  ω(頻率)= γ (磁旋比) Bo
- 3.0T相差3.5 x 10^{-6} x 42.6 MHz/T x 3.0T= 440 Hz
- 1.5T相差3.5 x 10^{-6} x 42.6 MHz/T x 1.5T= 220 Hz
- 0.5T相差3.5 x 10^{-6} x 42.6 MHz/T x 0.5T= 73 Hz



化學(頻譜的)預飽和
Chemical (Spectral) persaturation



Chemical (Spectral) persaturation Advantages & Disadvantages

Chemical (Spectral) persaturation advantages

1.可以區別T1相似的組織(ex: fat & Gd-enhanced tumors的)

2.除被壓抑的組織外，其他組織的訊號不會有影響
(IR affects the contrast of all tissues)

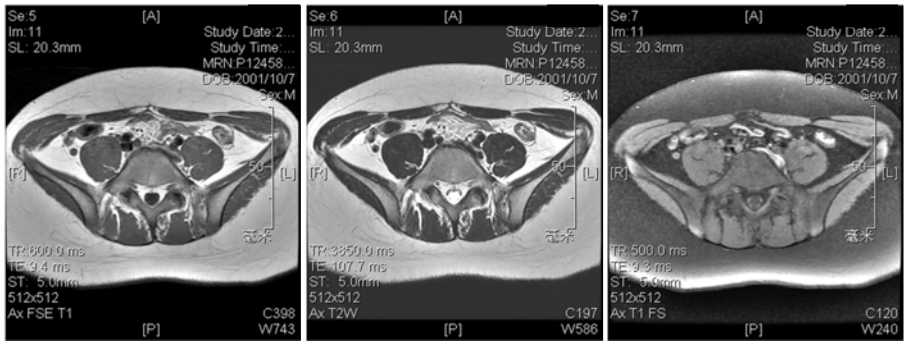
Chemical (Spectral) persaturation disadvantages

1.頻率技術的選擇對磁場不均度非常的敏感
(ex: metallic susceptibility artifacts/可用STIR改善)

2.需要額外的時間(TR increasing the scan time 5~8ms)

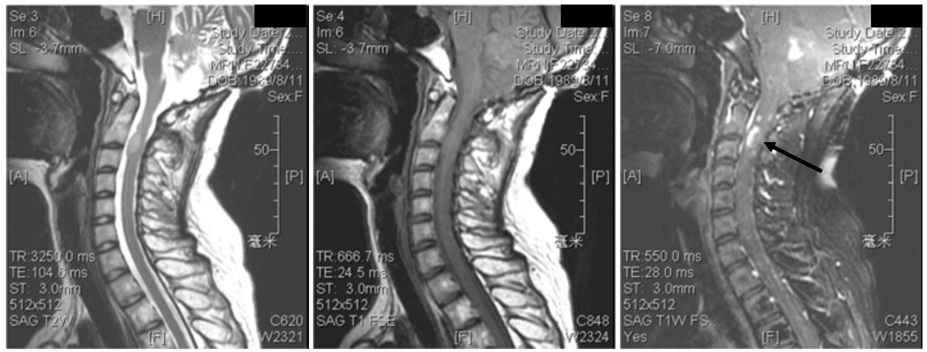
3.額外的RF加熱 (extra 90^{o} pulse)

Chemical presaturation
取信號的脈衝序列前, 以特定頻率打90^{o}RF



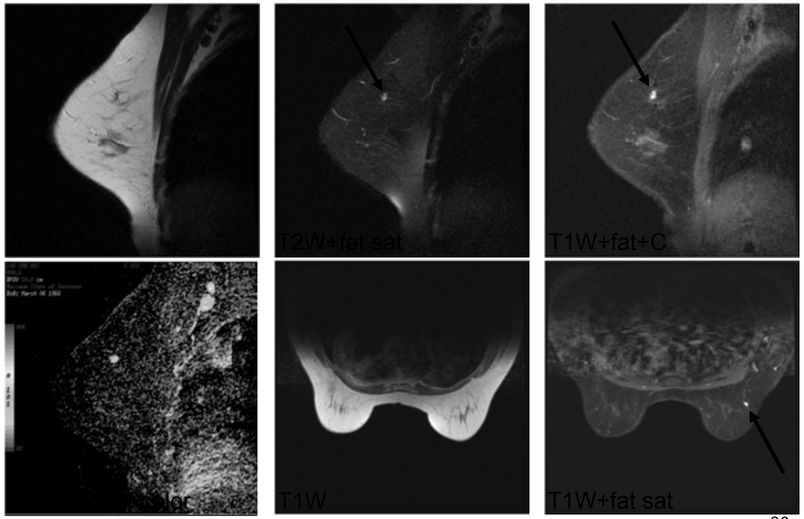
T1W　　T2W　　T1W + fat sat

Chemical pre-saturation
取信號的脈衝序列前, 以特定頻率打90^{o}RF



T2W　　T1W　　T1W + fat sat

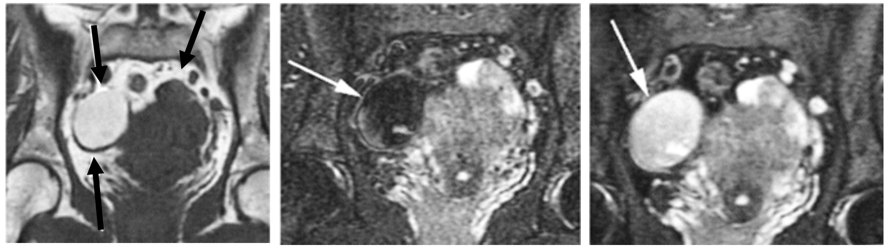
Chemical pre-saturation:
取信號的脈衝序列前, 以特定頻率打90ºRF



STIR vs. Fat Sat

- Endometrioma (aka chocolate cysts)

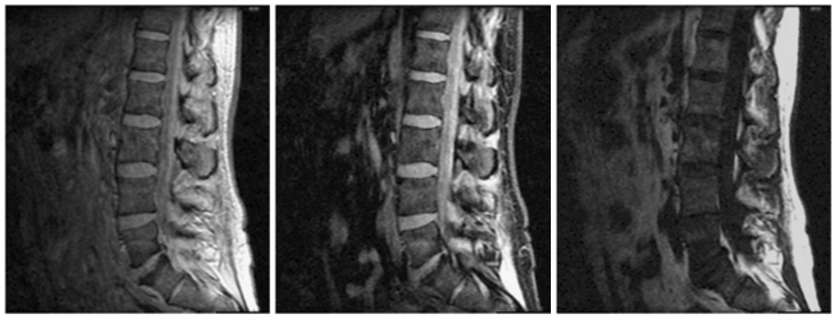
Lower SNR



Coronal T1W | Coronal STIR lesion T1 value similar fat | T2W fat sat Not include fat

MRI:The Basics,3e,p287.

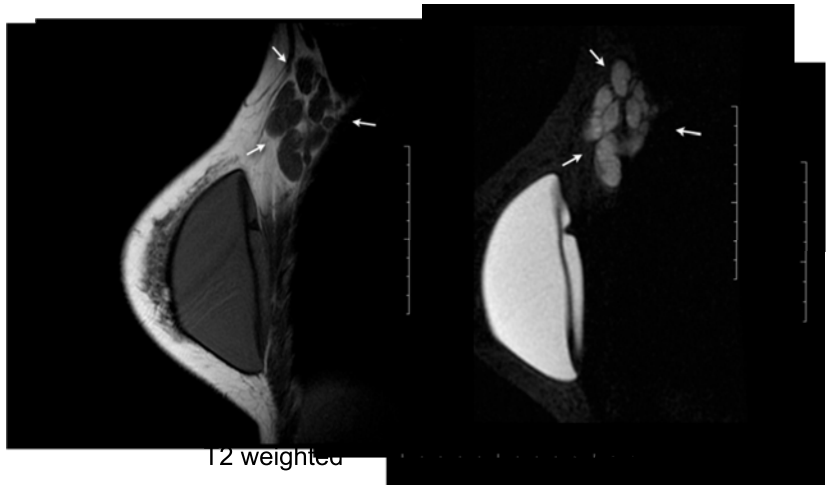
Fat Sat與Water Sat的影像比較



Raw Image | Fat Sat | Water Sat

(截錄自鍾孝文教授PPT圖檔)

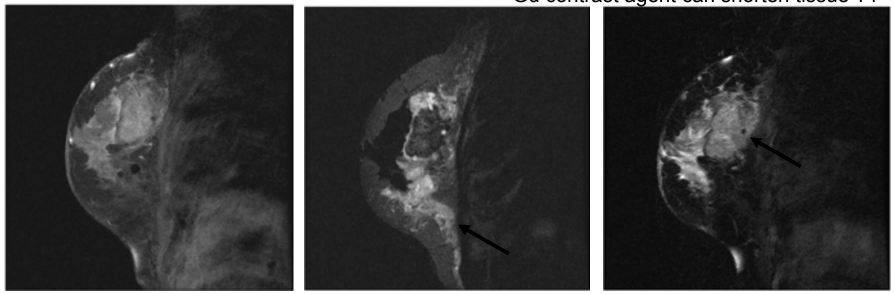
綜合saturation臨床應用
STIR、fat sat、water sat



M. H. Siqueira Mendonça, Electronic presentation online system(2013),1-60.

綜合saturation臨床應用 STIR、fat sat、water sat

T1:CSF > edema > gray > white > fat
Gd contrast agent can shorten tissue T1

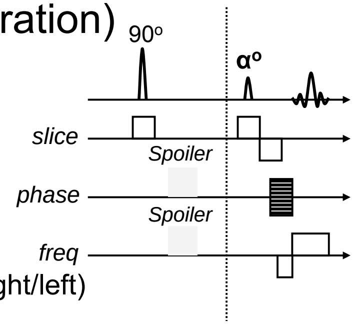


T2W + fat sat　　STIR + water sat　　T1W + fat sat + C

- **Siliconomas (小針美容)**
- **Breast tumor (r/o malignancy)**

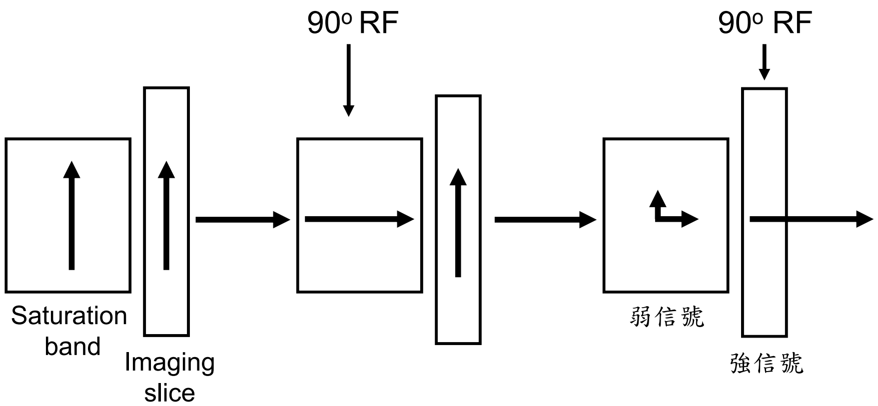
tumor血管壁比較不完整，導致比較大分子的顯影劑會滲漏到裡面去

空間預飽和(Spatial persaturation)



- 臨床上又稱為saturation band
- 90º脈衝之前多施加一個90º脈衝
- 利用空間的遮擋做為概念 (anterior/posterior, superior/inferior, right/left)
- 降低整個FOV內的運動及流動相關假影 (Motion artifacts or Flow-related artifacts)
- 針對運動和流動的組織(心臟或呼吸運動或血流等)

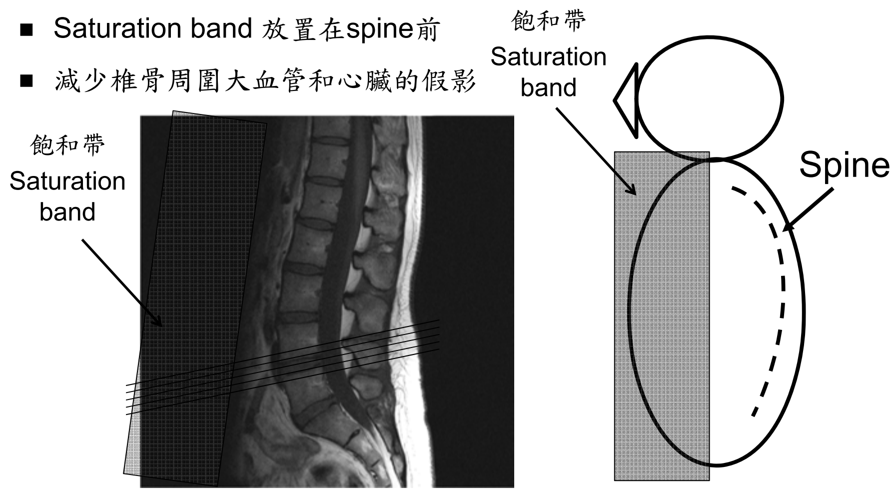
什麼是 saturation band?



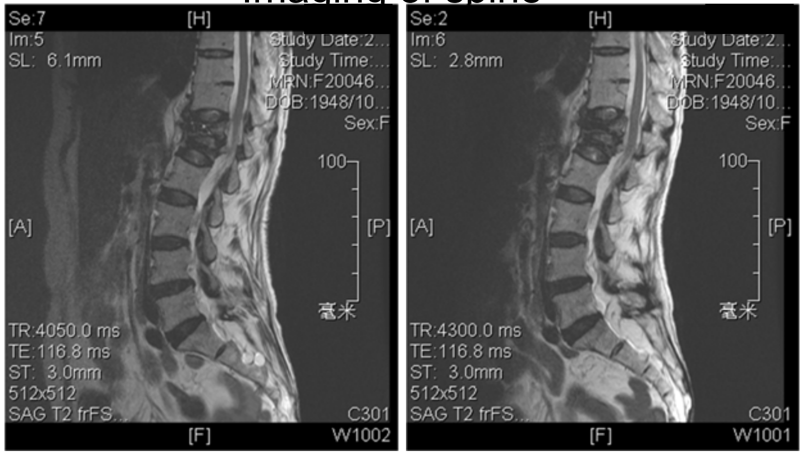
在取信號的脈衝序列之前，於特定位置打90ºRF

空間預飽和(Spatial persaturation) Imaging of spine

- Saturation band 放置在spine前
- 減少椎骨周圍大血管和心臟的假影



空間預飽和(Spatial persaturation)
Imaging of spine

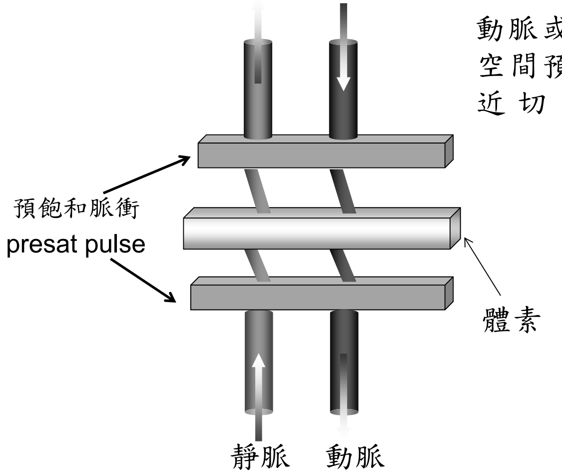


無 saturation　　有saturation

台北慈濟醫院
放射診斷科　周嘉豪

空間預飽和(Spatial persaturation)
MR angiography常用

動脈或靜脈的流動造影下，空間預飽和脈衝被放在鄰近切面的上方或下方



- MRA
- MRV

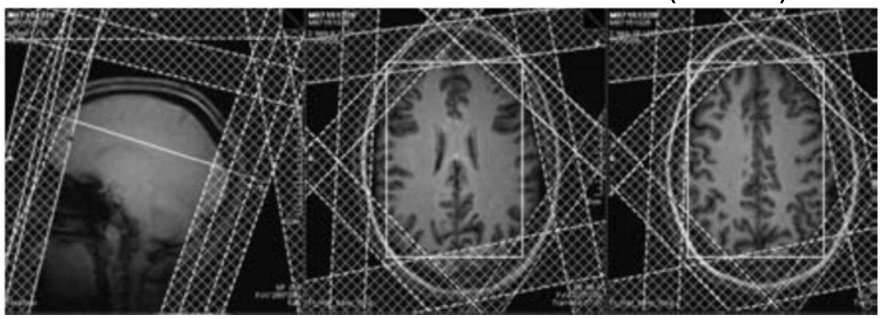
在取信號的脈衝序列之前,於特定位置打90^{o}RF

台北慈濟醫院
放射診斷科　周嘉豪

空間預飽和(Spatial persaturation)
MR spectroscopy常用

MR spectroscopy:
Saturation bands are placed on the skull regions.

No. band ↑ scan time ↑ (5-8msec)



- Fat和lactate二者太接近(1.3 ppm)
- 不適用fat sat 怕二者都被saturation

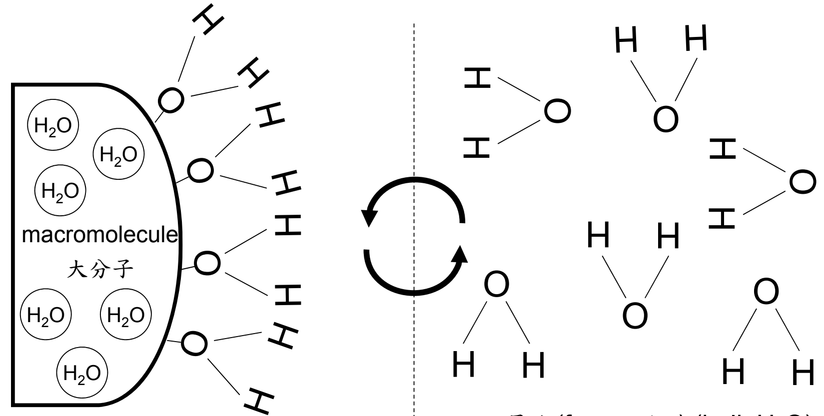
T. Li1,Proc. Intl. Soc. Mag. Reson. Med. 14 (2006)

台北慈濟醫院
放射診斷科　周嘉豪

Chemical persaturation vs. Spatial persaturation

	Chemical persaturation	Spatial persaturation
優點	1.可區別T1值相似的組織 2.未受壓抑組織訊號沒有任何影響	1.減少phase ghosts 2.減少flow artifacts
缺點	1.使用頻率選擇技術,深受磁場不均勻度的影響 2.需要額外的時間 3.造成額外的RF熱量	1.FOV內也壓抑其它部分的組織 2.延長TR，造成掃描時間增加
應用	1.適當選擇頻率可消除脂肪或水的訊號	1.Spine scan 2.MRA(MRV), TOF 3.Abdomen scan 4.Brain, MRS

磁量轉移(Magnetization Transfer , MT)
Non-fatty hydrogen nuclei



蛋白質結合水(protein-bound water)

T1: 10~100msec T2: 5~10msec

巨量水(free water) (bulk H_2O)

T1:2000~3000msec

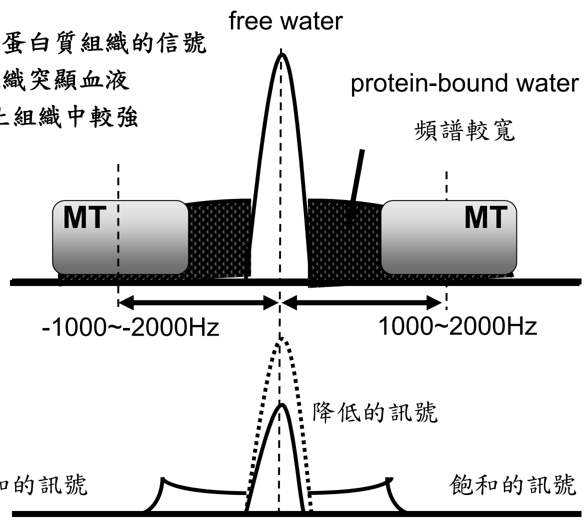
T2:1000~2000msec

磁量轉移(Magnetization Transfer , MT)

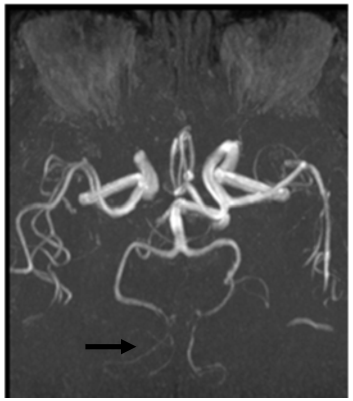
- 壓抑protein-bound water的技術
- 巨量水(bulk H_2O):氫含量高、自由運動的水分子
- 結合水(bound water) :氫含量低、蛋白質結合水分子
- protein-bound water與free water的共振頻率相差500~2500Hz
- protein-bound water的飽和傳給free water的飽和
  CSF、blood、骨髓、脂肪組織，大分子少，MT小訊號小
  腦組織、肌肉，大分子多，MT大訊號大
- 常運用在TOF-MRA抑制腦背景組織，突顯較小的腦血管
  (reduction of gray and white matter signal by 15%~40%)

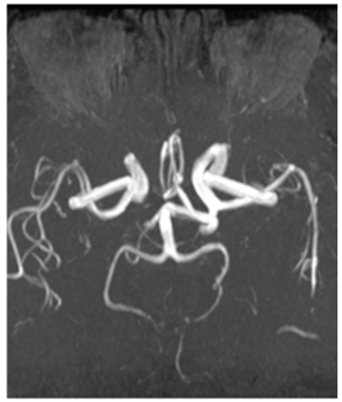
磁量轉移(Magnetization Transfer , MT)

- 抑制含有大分子蛋白質組織的信號
- 有效抑制背景組織突顯血液
- **MTS** 效應在靜止組織中較強



磁量轉移(Magnetization Transfer , MT)
TOF-MRA常用





有MT sat 無MT sat

(reduction of gray and white matter signal by 15%~40%)

Thanks for your attention!

磁振造影專業基礎課程

血流現象與磁振血管造影

Flow phenomena and MR

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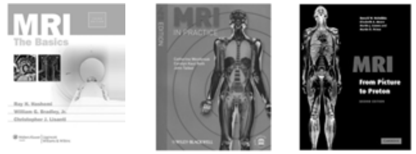
charles198217@yahoo.com.tw

本次課程內容

- 瞭解血流對 SE 和 GE 訊號的影響
- 瞭解TOF MRA, PC MRA和 CE MRA的原理
- 瞭解磁振血管造影的臨床應用與優缺點



Reference:

1.MRI The Basics (3rd)

2.MRI IN PRACTICE(4td)

3.MRI From Picture to Proton(2nd)

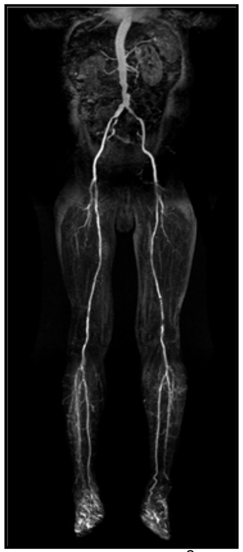
4. Laub G, Gaa J, Drobintzky M. Magnetic resonance angiography techniques. Electromedica 1998;66:68-75.

5. Graves MJ. Magnetic resonance angiography. Br. J. Radiol. 1997;70: 6-28.

有哪些血管造影 (Angiography) 的方法?

傳統血管造影

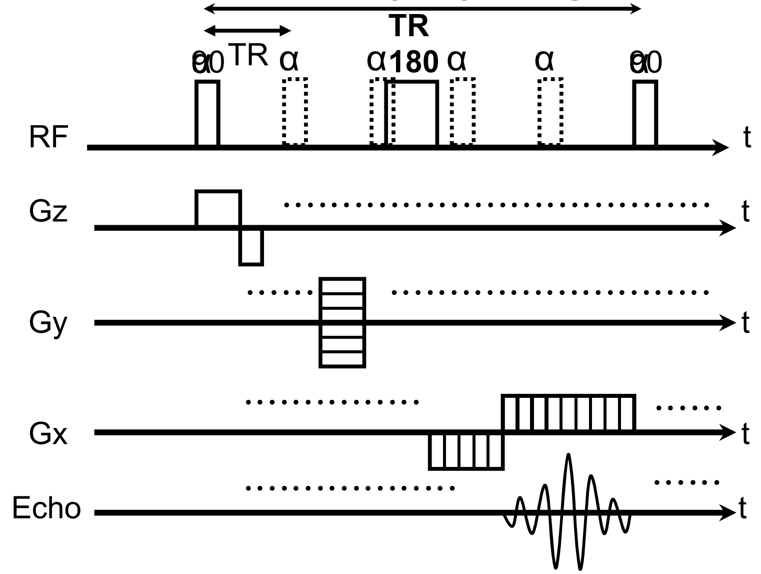
電腦斷層血管造影 **(CTA)**



磁振血管造影 **(MRA)**

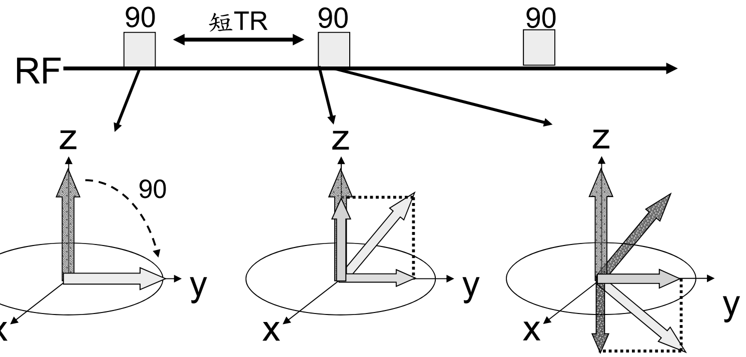
Gradient Echo (GRE) Diagram

Spin Echo (SE) Diagram

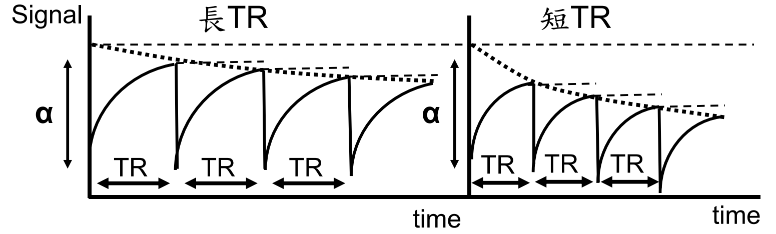


Saturation Effects(飽和效應)

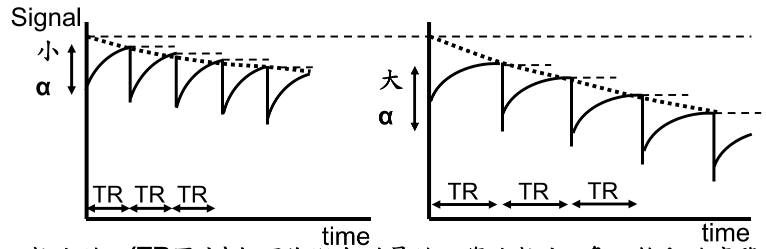
- 重複激發的RF pulse造成縱向磁量的逐漸消失
- 造成訊號的損失，也造成SNR的降低
- 短TR，大偏折角(T1W)→沒時間作T1回復→訊號低



如何減少saturation effects?



較長的**TR**會使得縱向磁量的回復比較短的**TR**好，飽和效應降低

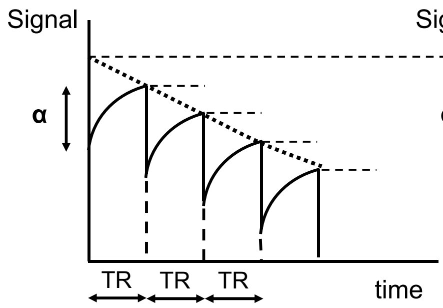


較小的**α (TR**固定**)**也可使縱向磁量的回復比較大**α**多，飽和效應降低

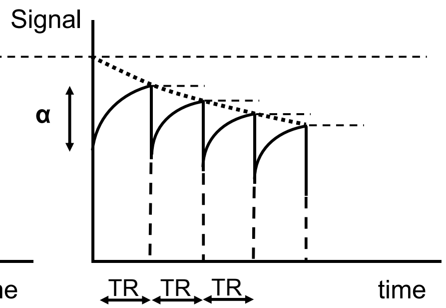
如何減少saturation effects?

注射gadolinium(使T1變短)

- 順磁性的對比劑會使T1縮短(CE-MRA)
- T1回復會比較快且飽和效應比較小

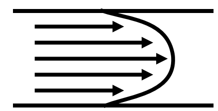
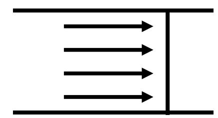
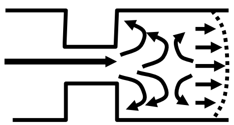
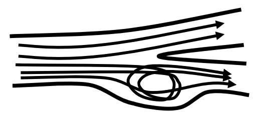


未使用**Gd**



使用**Gd**

Flow Phenomena

Laminar Flow 層流	normal vessels (大多數的血管)	parabolic profile (拋物線)	
Plug Flow 栓塞流(平流)	high velocities + large vessels (高流速的大動脈)	flat profile	
Turbulent Flow 擾流	abnormal vessels (不正常血管)	Vortex flow eddies	
Flow separation 流體分離	near the wall of vessels	streamline separated	

MRA Techniques

- **MRA Techniques** : TOF (time of flight) MRA(2D、3D) → single slab / multi slab
  PC (phase contrast) MRA(2D、3D)
  CE (contrast enhanced) MRA(3D)

Unenhanced MRA (TOF & PC)

- Amplitude effects:
  - Blood flowing into or out of a chosen slice has a different longitudinal magnetization compared to stationary spins.
  - Depend on the duration of stay (Time-Of-Flight) in the slice.
- Phase effects:
  - Blood flowing along the direction of a magnetic field gradient changes its transverse magnetization compared to stationary spins.

Time Of Flight (TOF)

- **Signal loss(訊號喪失)**
  1. high velocity(高速) **(Washout & Inflow effect有關)**
  2. turbulent flow(擾流)
  3. dephasing(失相)

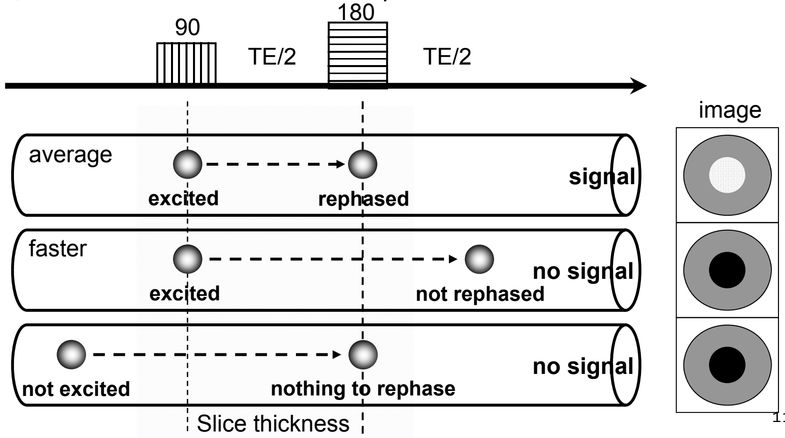
  } Flow Artifacts
- **Signal gain(訊號獲得)**
  1. flow-related enhancement (FRE)
  2. even echo rephasing(偶數回音重聚相) ← **$TE_2=2TE_1$**
  3. diastolic pseudogating(心舒期假性觸發)

  收縮期流動: 快 心舒期流動:慢 利用cardiac gating

Time-Of-Flight effect in Spin Echo

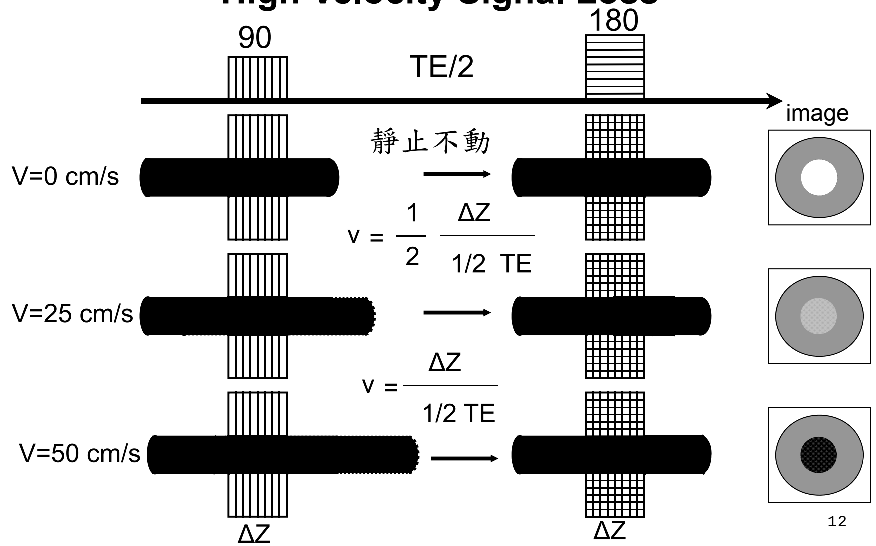
血流呈黑色或灰色(Washout effect)

- 高速的訊號損失，流動的質子在切面中停留時間不夠長
- 質子必須同時受到90º和180º RF pulse作用

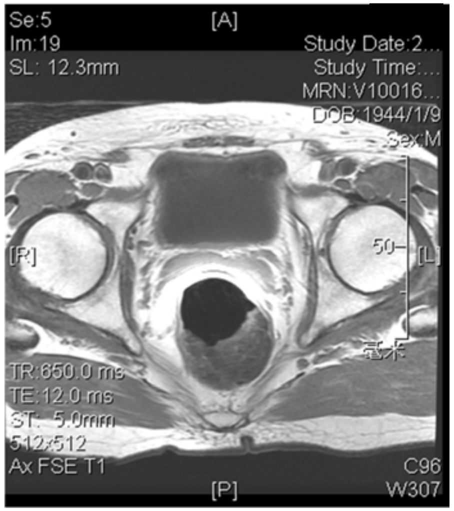


Time-Of-Flight effect in Spin Echo

High Velocity Signal Loss



Time-Of-Flight effect in Spin Echo
血流呈黑色或灰色



ex: slice thickness=1cm
TE=50msec時，
血流速度大於多少就沒有訊號?

Ans: $v = \dfrac{\Delta Z}{1/2\ TE}$

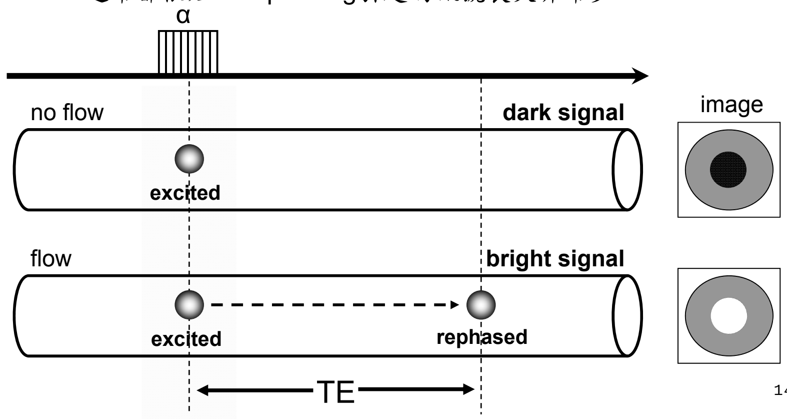
1cm/25msec = 40cm/sec

Flow blood SI↓

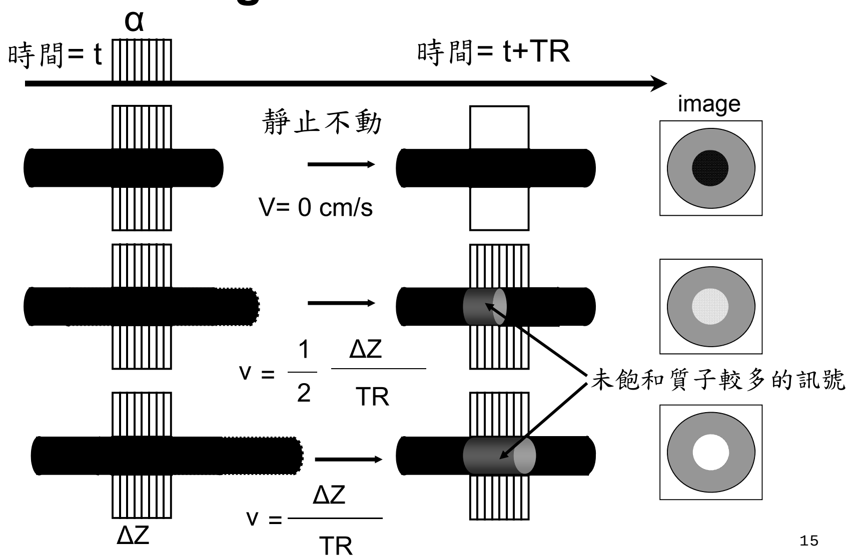
- Decreasing slice thickness
- Increasing flow velocity
- Increasing TE (T2W>T1W)
- Not occur in GRE

Time-Of-Flight effect in Gradient Echo
血流呈白色(GRE影像血流總是亮的!) (Inflow effect)

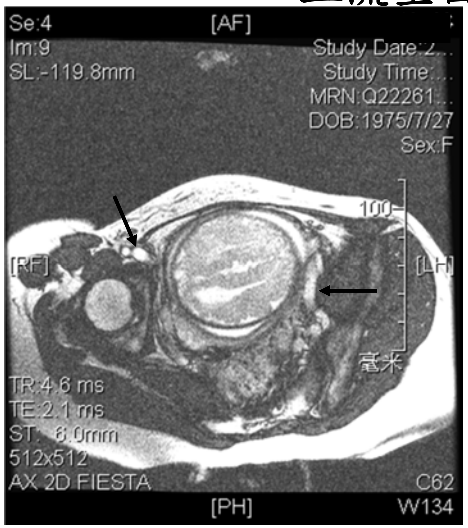
- GRE總是序列模式(一次切一張)，每一張都是血液流進來的第一張
- 沒有180ºRF，所以TOF loss影響不明顯
- TE通常都很短，dephasing引起的訊號喪失非常少



Time-Of-Flight effect in Gradient Echo



Time-Of-Flight effect in Gradient Echo
血流呈白色



ex:當slice thickness = 1 cm,
TR = 1000 msec 時,
血流速度多少就有最大訊號?

Ans: $v = \dfrac{\Delta Z}{TR}$

1 cm / 1000 msec = 1 cm/sec

Flow blood SI↑

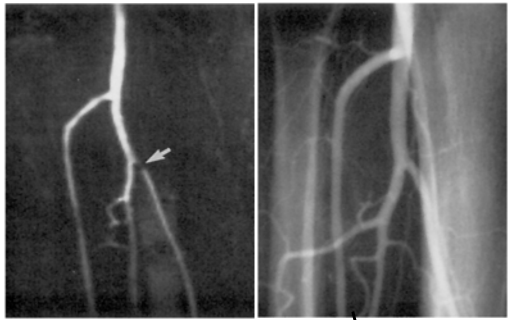
- Decreasing slice thickness
- Increasing flow velocity
- Increasing TR

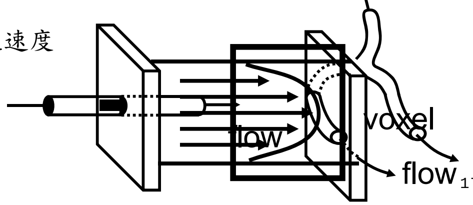
Flow Artifacts
(易造成pseudo-stenosis)

- Turbulent flow(擾流)

紊亂的流體，造成不同相位，而互相抵消而導致沒有訊號，發生在具有低或高的流速時

- Dephasing(失相)
- 體素內(intravoxel)失相:
  1.層流造成
  2.體素內存在不同血流速度
- 切面方向和血管轉彎處

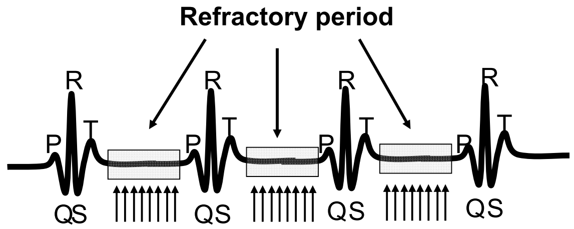




Flow 訊號獲得

Signal gain(訊號獲得)

1. flow-related enhancement (FRE) (Type of TOF)
2. diastolic pseudogating
   - 心舒期可獲得較高的血管內訊號(高速導致較多的TOF損失)
   - 使用cardiac gating來固定擷取一個時間點的心跳

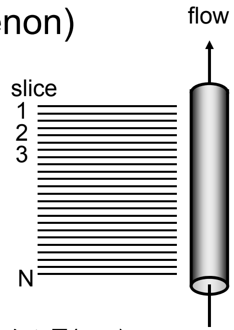


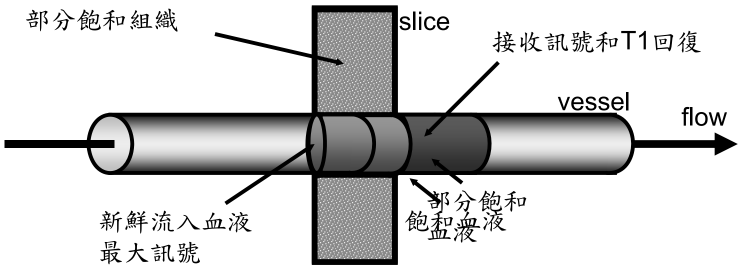
3. even echo rephasing(偶數回音重聚相)
   - SE造影有對稱回音、TE2=2TE1、偶數回音的訊號強度比奇數回音

Flow Related Enhancement (FRE)
又稱入口現象(entry phenomenon)

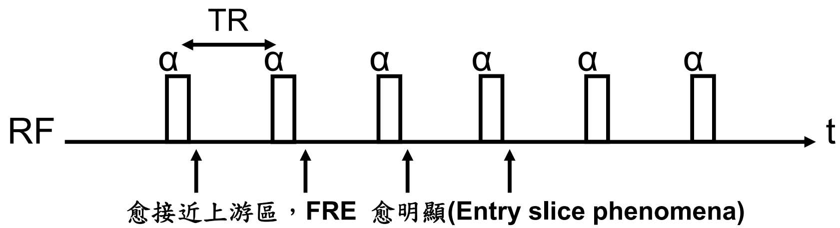
2D TOF:

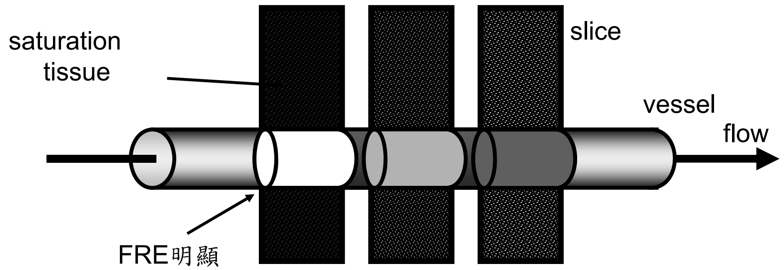
- Larger flip angle (30^o~70^o)
- Thicker slice thickness(2~3mm)(SNR↑)
- Vessels straight and perpendicular to the slices (carotid or lower extremties)
- Sensitive to slow flow





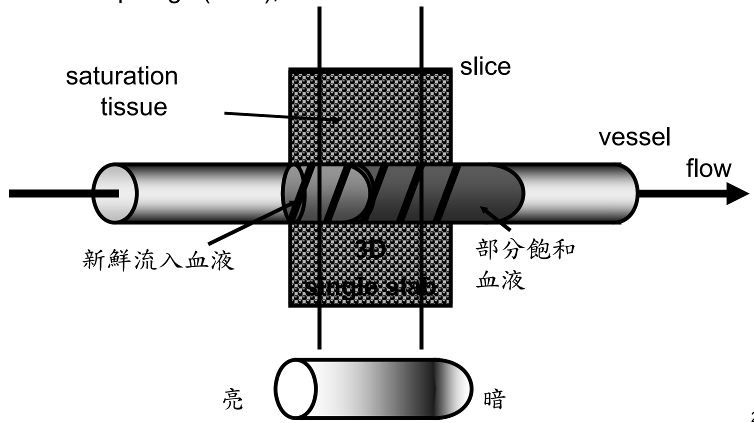
FRE & Multi-slice





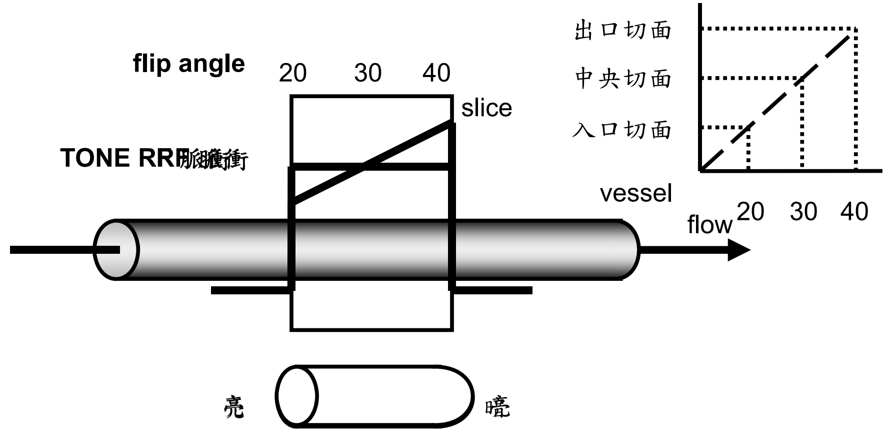
3D TOF MRA

- FRE Multi-slice Phenomena類似、3D FRE 比 2D FRE 弱
- Smaller voxels(<1mm), short TE, higher SNR
- Small flip angle(<30º), slab does not become too saturated.



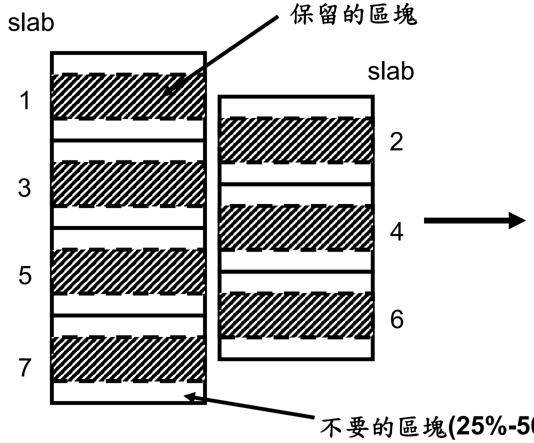
TONE

Tilted Optimized Non-saturating Excitation
傾斜最佳化的不飽和激發



MOTSA

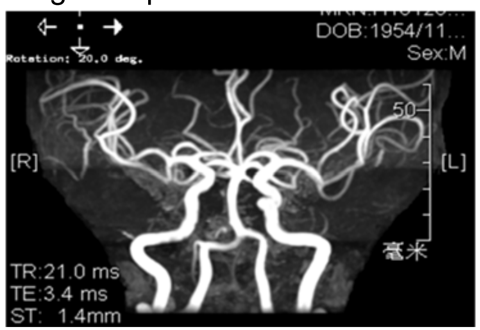
Multiple Overlapping Thin-Slab Acquisition
多重疊之薄區塊的擷取

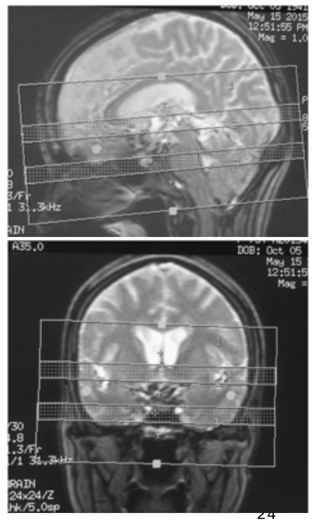


Venetian blind artifact

3D multi-slab method

- Larger vessel sections
- Reduced saturation effects (like 2D)
- MOSTA: 20%~30% overlapping
- Longer acquisition time

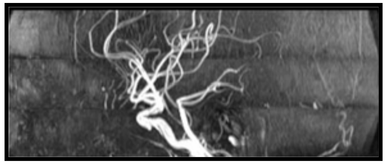




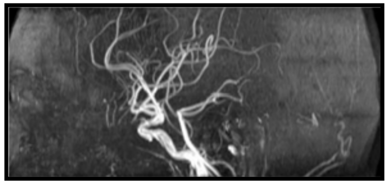
3D multi-slab method

single slab

multi-slab
non-overlapping



multi-slab
overlapping



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放射診斷科 周嘉豪

Reduce saturation effects?

- 用較小的偏折角(α) (固定TR下大α比小α有較多的Mz喪失)
- 用較長的TR (可以讓縱向磁量回復的較多)
- MOTSA (Multiple Overlapping Thin-Slab Acquisition) (較常用)
- TONE (Tilted Optimized Non-saturating Excitation)
- 注射gadolinium(使T1變短)

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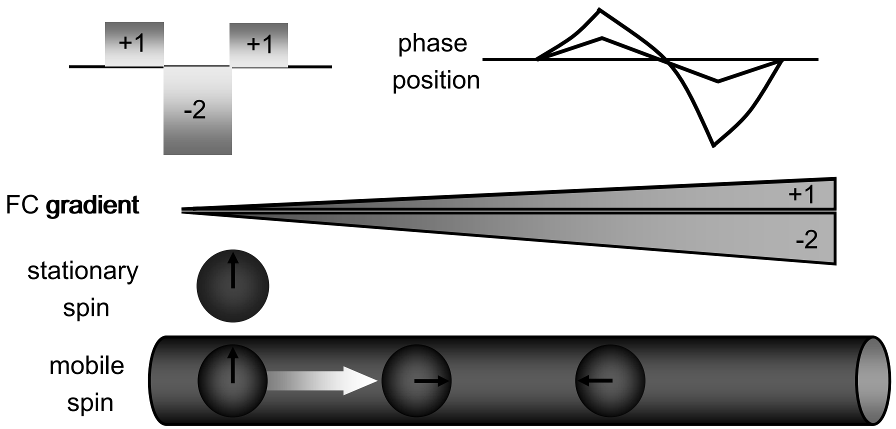
Gradient Moment Rephase (GMR)
(Flow Compensation; FC(流動補償))

- GMR:減少流體運動假影的一種方法



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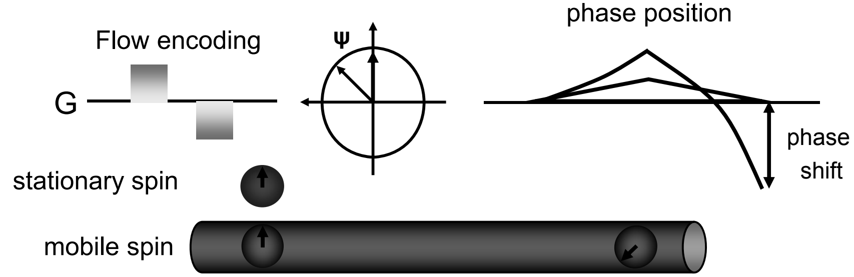
Options to improve TOF MRA

- Slices or volume perpendicular to flow direction
- 2D for slow flow , 3D for fast flow
- 3D multi-slab for larger vessel sections
- Spatial presaturation to isolate arteries and veins
- Use of minimum TE reduces signal loss due to spin dephasing
- TONE pulse or MOSTA reduces saturation effects in 3D TOF
- Magnetization transfer (MT) and fat sat improve vessel contrast
  (reduction of gray and white matter signal by 15%~40%)

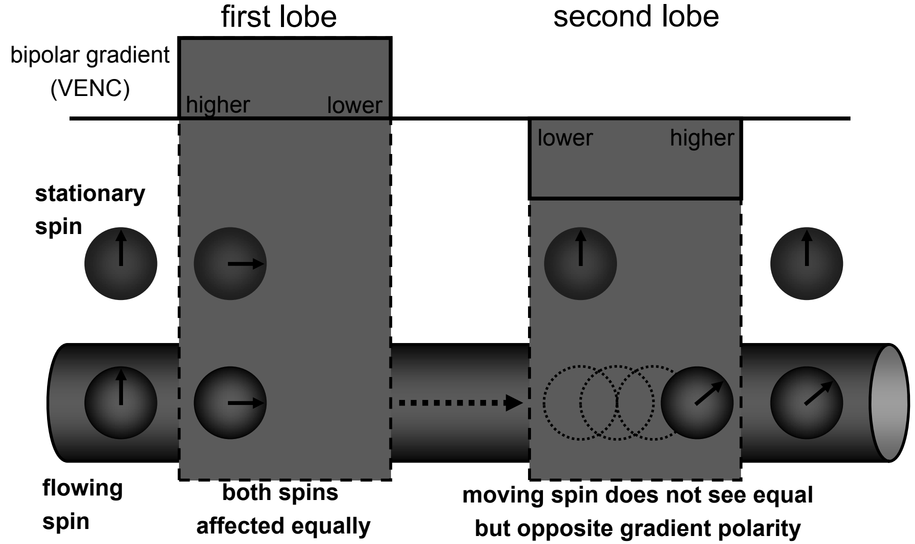
Phase Contrast MRA (PC MRA)

- Phase effects concern the transverse magnetization
- Bipolar flow-encoding gradient (strength and duration but opposite sign)
- Stationary spins = zero net phase shift
- Flowing spins = a non-zeros phase shift

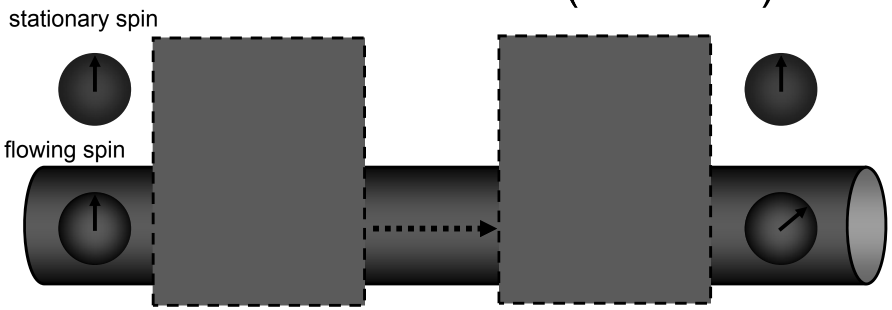


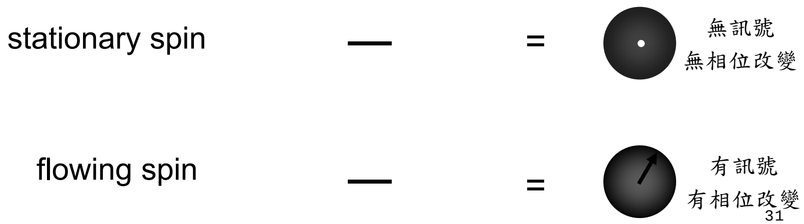
phase shift $\psi=\int \omega dt = \int(\gamma G v t)\, dt = \gamma G v \int t\, dt = 1/2\gamma G v t^2$

Phase Contrast MRA (PC MRA)



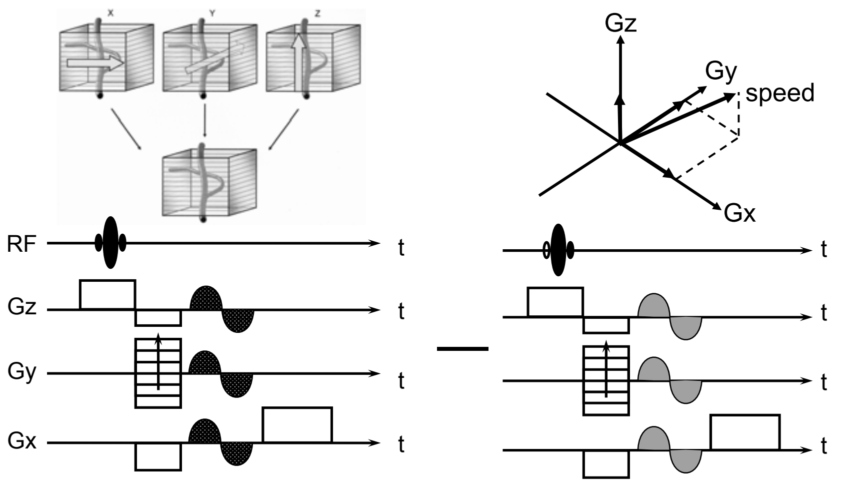
Phase Contrast MRA (PC MRA)





Magnitude & phase contrast method

- 整體流速 -- 重複三次 (Gx、Gy、Gz) + 一次參考點(flow compensation)



Flow compensation, bright blood image

Biolar gradients, dark blood image

(部分圖檔來自於鐘孝文教授之ppt檔)

What is VENC?

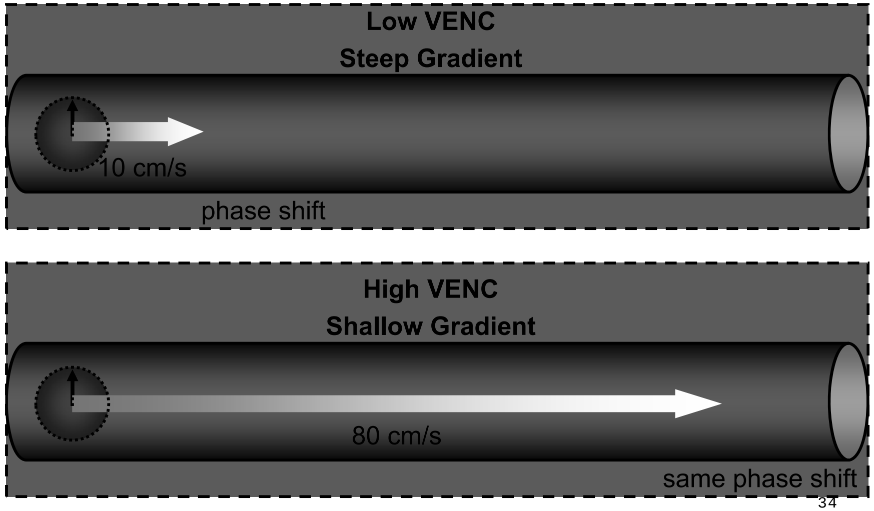
- Velocity encoding (VENC) = 速度編碼 (MR放射師可以選用)
- VENC為該梯度下所能求的最大流速
- 血流的相位改變的大小與其速度成正比
- **Aliasing velocity(假影:假的速度) = VENC – 實際速度**
- Ex:若VENC=30cm/sec，則一個具有40cm/sec之流速血管所呈現出來的流速為?

  V=30-40=-10cm/sec(朝反向的流動)

$$\psi = 1/2\gamma G v t^2$$

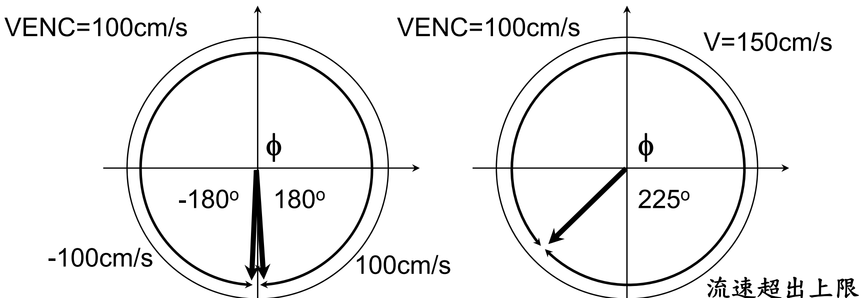
一個較小的VENC對慢速流(靜脈流)及較小的支脈較為敏感
但會對較快速的(動脈)流動造成假影

Velocity encoding (VENC)
梯度愈強(弱)，VENC愈小(大)



VENC 與 phase aliasing

- 流速與相角成正比
- 實際上VENC可分析的流速範圍
- +180⁰ ~ -180⁰ 之間，角度 360⁰ 的範圍(180⁰相角的流速)
- **Aliasing velocity= VENC – 實際速度 (V= 100 - 150 = -50cm/sec)**



180⁰ : 正向最大流速
-180⁰ : 反向最大流速

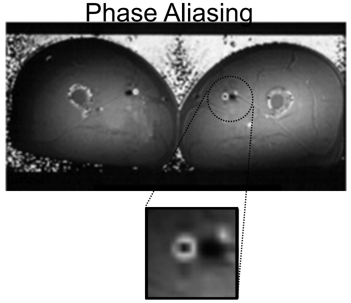
正向血流:白色
靜止組織:灰色

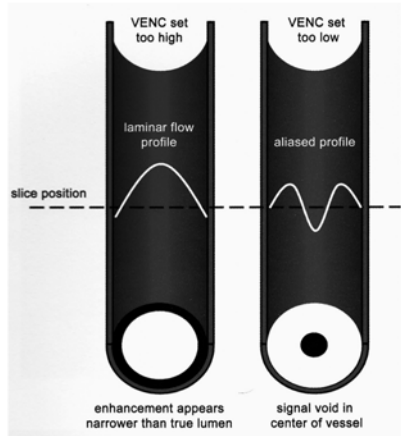
流速超出上限
正向血流:黑色

Velocity encoding (VENC)

- 調VENC = bipolar 梯度、根據流速範圍選取最適當VENC值
- VENC太高:慢血流看不清楚
- VENC太低:phase aliasing

Phase Aliasing





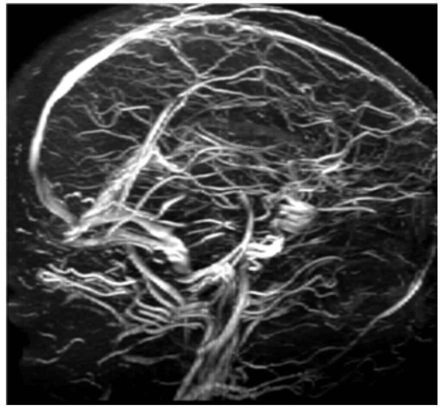
(影像來自於鐘孝文教授之ppt檔)

常見血管內的血流速度

VENC Optimization

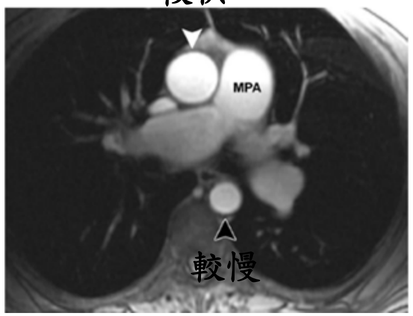
Pulmonary artery	70-130
Aorta	100-175
Carotid artery	80-120
External iliac artery	81-120
Carotid syphon	55
Common femoral artery	115
Basilar artery	40
Superficial femoral artery	90
Vertebral artery	40
Popliteal artery	70
Sagittal sinus vein	10
Peripheral veins	5-10

PC MRA: 常用於腦部靜脈造影

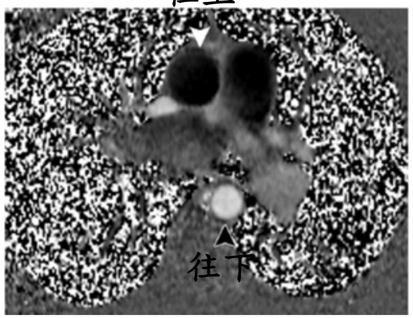


Scan time 約 6 min，時間很長

PC MRA: 定量血流速度與方向



Magnitude image
(images of blood vessels)



Phase image
(direction of flow)

(影像來自於鐘孝文教授之ppt檔)

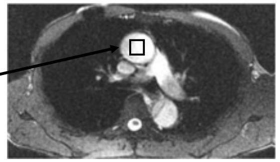
Contras-Enhanced MRA (CE MRA)

- 避免血流訊號被飽和或血流的SNR值不夠好
- 注射 Gadolinium (Gd-DTPA)使血流的T1縮短(paramagnetic)
  (0.5~4.0 ml/s、0.1~0.3 ml/kg、total 20~40 ml、GFR>30 ml/min)
- 快速靜脈注射Gd被稀釋之前,快速擷取影像
  (T1縮短最明顯時、使用GRE技術(T1W-SPGR))
- 掃描切面通常是 coronal , 而不是與血管走向垂直
  (可以在解析度增到最大的情況下，增加涵蓋範圍)
- 血流流動的失相假影像不敏感(依賴T1特性)

Contras-Enhanced MRA

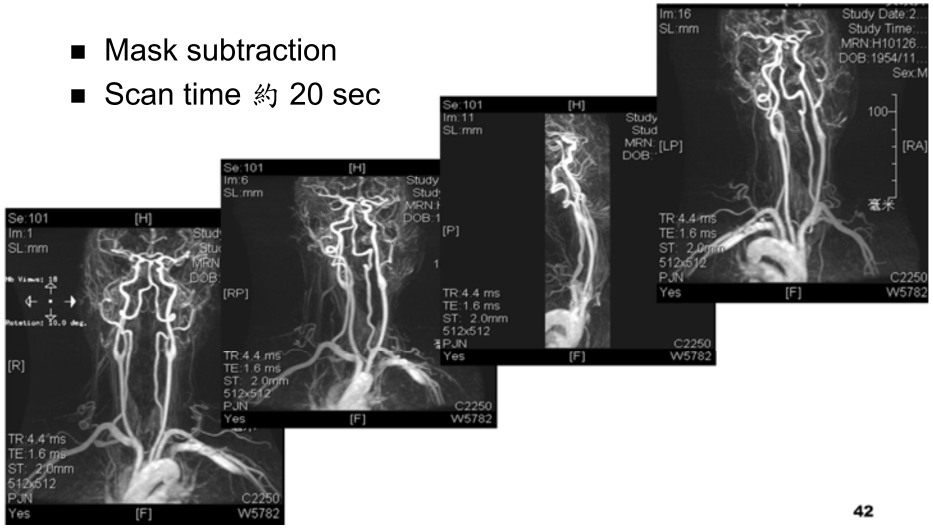
- CE-MRA: elliptical-centric 和 multiphase
- elliptical-centric:
  - 等對比劑進入感興趣的動脈後再開始擷取
- bolus對比劑的自動監測軟體
  - SmartPrep:通過將游標放置在感興趣動脈的上游來進行
  - TimeBolus:進行即時掃描以決定Gd對比劑到達感興趣動脈的時間 (觀察注射2cc的Gd對比劑後動脈到達最大亮度的時間)

2D real time
Tracker volume ROI



Contras-Enhanced MRA

- Mask subtraction
- Scan time 約 20 sec

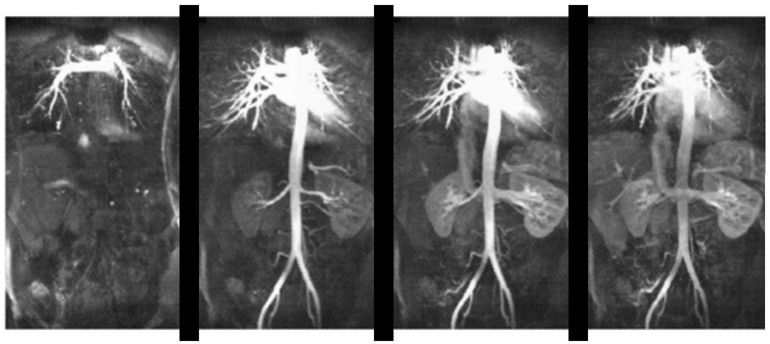


Contras-Enhanced MRA

- CE-MRA: elliptical-centric and multiphase

multiphase:

- 注入Gd之後進行多次的擷取，其中一個必定位於動脈相
- time-resolved imaging of contrast kinetics, TRICKS

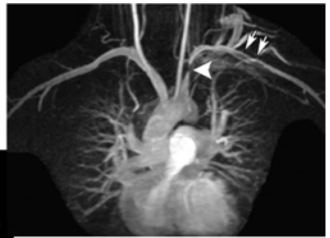


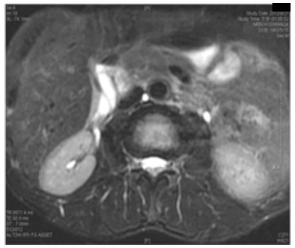
Applications areas of MRA

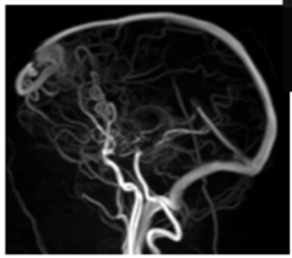
	2D-TOF	3D-TOF	2D-PC	Magnitude contrast	3D-PC	CE MRA
Intracranial Arteries		***			*	*
Intracranial Veins	***	*	*		**	*
Carotids	**	**				***
Peripheral vessels	**			*		***

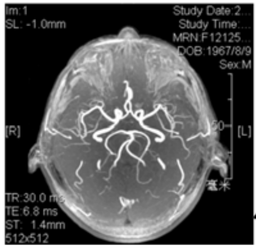
看圖說故事時間

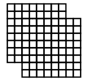
- 請各位放射師們看影像, 並想想下列二個問題:
  1. 為何血管呈黑色, 灰色 或白色?
  2. 使用何種磁振造影技術?

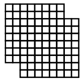












Thanks for your attention!

The Artifacts in MRI

新光吳火獅紀念醫院 放射診斷科
技術專員 李正輝

Introduction

- *All MRI images have artifacts in some degrees.*
- *Why and How ？*
- *How to remedy the artifacts encountered in MRI.*

Introduction

- *Motion artifacts*
  - *patient motion, physiological motion, flow*
- *Inhomogeneity artifacts*
  - *mental artifacts, zipper artifacts, cross talk*
- *Digital imaging artifacts*
  - *aliasing, truncation, herring-bone artifacts, halo artifacts, Gradient nonlinearities, chemical shift*

Motion Artifacts

- *Patient motion*
  - *voluntary motion, involuntary motion*
- *Physiological motion*
  - *respiration, cardiac motion, peristaltic*
- *Occurring in phase encoding direction*

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But, why are ghosts only produced in the phase-encode direction?

- *Consecutive points in the frequency-encoding direction are measured close together, typically much less than 1 ms apart.*
- *whereas consecutive phase-encoding steps are TR ms apart.*
- *Motion such as respiration and blood flow occurs slowly compared with frequency encoding but much quicker than phase encoding. So between successive phase encodings, the anatomy moves and produces a ghost signal at a different PE position.*

Motion Artifacts

- *Solution of patient motion*
  - *fixed patient, repeat scan, reduce scan time, drug-assisted*
- *Solution of physiological motion*
  - *hold on breath, respiration gating, respiratory compensation, ECG gating, fast scan technology, drug-assisted*

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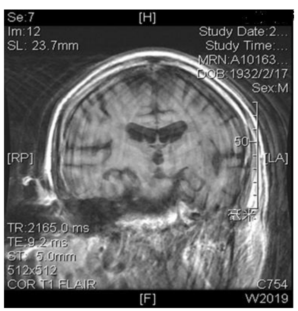
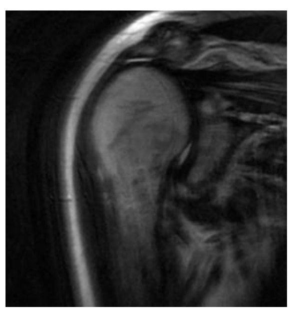
- *Patient motion*





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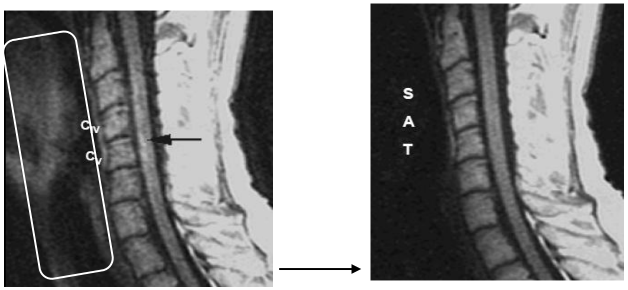
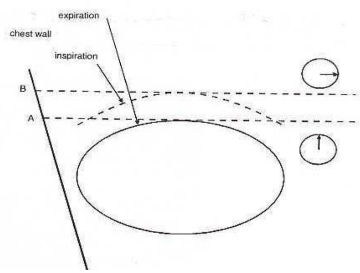
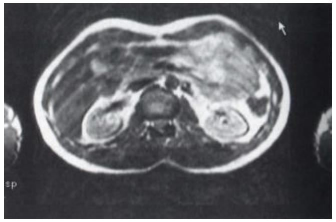
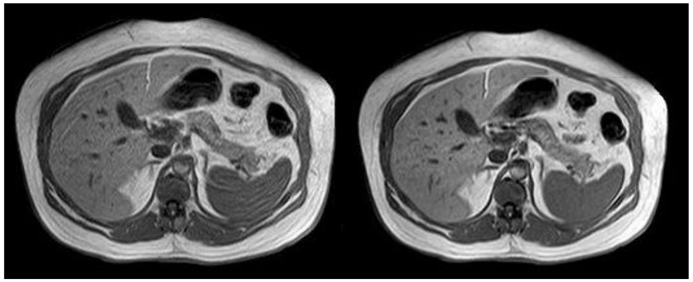


加saturation band

*Motion Artifact(**Respiratory)***

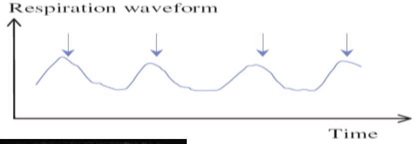


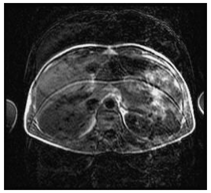


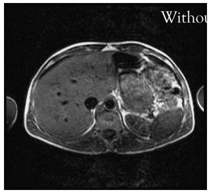


Respiration *Breath-hold*

Respiration Gating

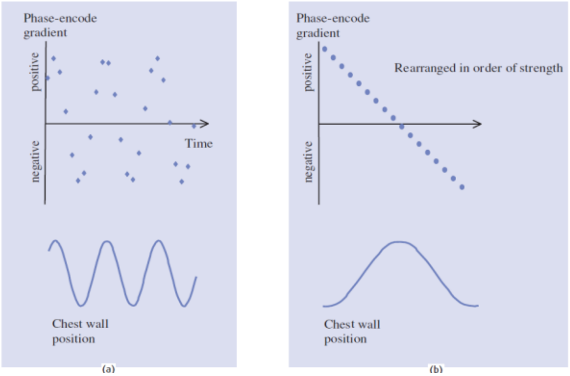




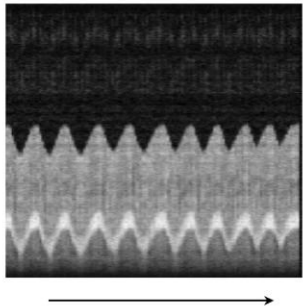


Respiratory Compensation

Respiratory-**O**rdered **P**hase **E**ncoding, ROPE



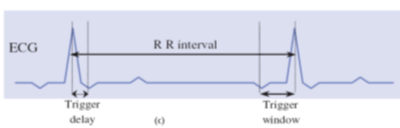
Navigator

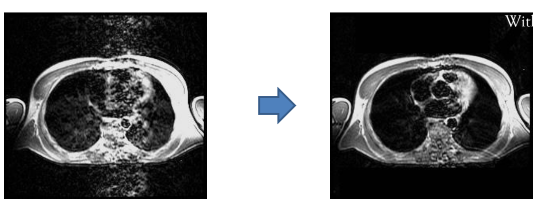


*Motion Artifact(**Cardiac pulse)***

- *Involuntary motion*
- *ECG trigger or PPU trigger are used to avoid artifacts*
- *MR compatible electrodes use carbon instead of metal to avoid causing artefact on the MR images*
- *Scan time is determined by the heart rate*
- *The TR is controlled by the R–R interval*

ECG Gating





ECG or peripheral gating?

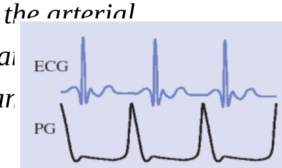
- *ECG gating is a more accurate gating method*

--- peak is usually sharp and easily recognizable

--- all the other ECG peaks can be seen too

--- good for cardiac imaging

- *Peripheral gating only detects the arterial*
- *Pulse peak is much broader thai*
- *Ease of preparing the patient ar*



*Motion Artifact(**Peristaltic motion)***

- *Causes a random continuous motion of the abdominal contents*
- *Acquiring multiple averages can reduce the ghost appearances*
- *Antiperistalsis drug such as hyoscine butylbromide (Buscopan)*
- *Ultrafast pulse sequences*

--- HASTE

*--- single-shot **F**ast**S**pin **E**cho (FSE)*

Flow Motion Artifact

- *Artifact or non artifact? There are two sides of same coin*
- *In-flow effect (flow related enhancement, FRE)*
  - *Spin echo: dark signal*
  - *Gradient echo: bright signal*
- *Velocity-induced phase effects*
  - *Resonant frequencies are changing continuously*
  - *Incorrect phase angle for their real position*
  - *Artifact on phase encoding direction*

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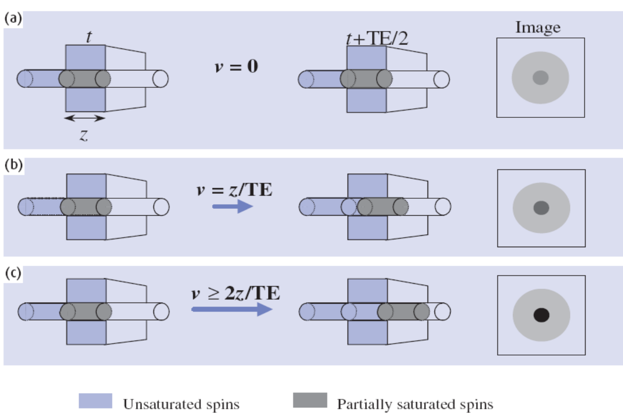
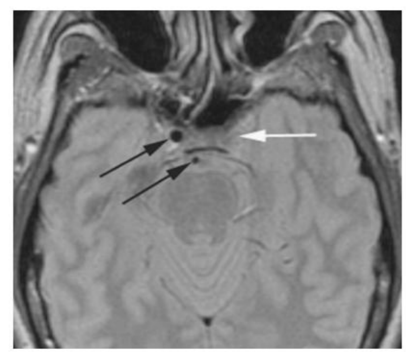
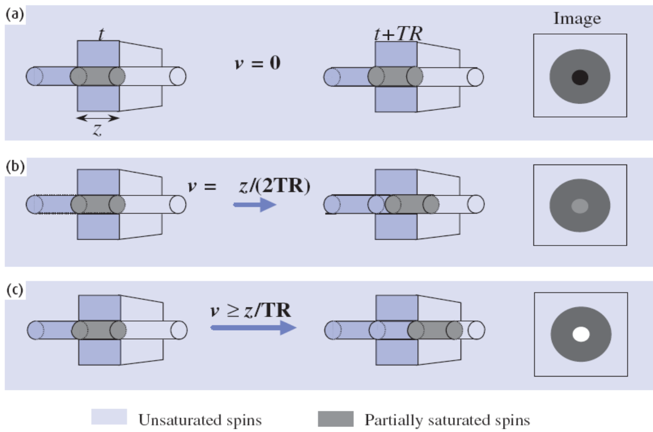
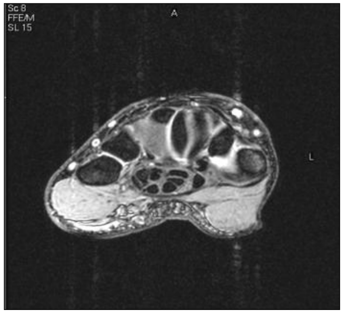


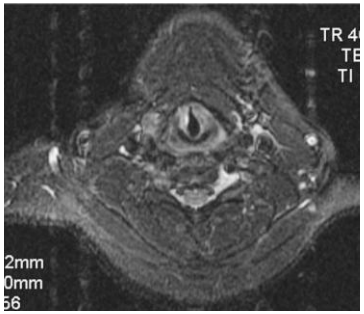
PDW axial image





Flow artifact





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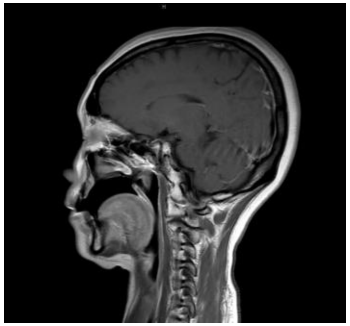
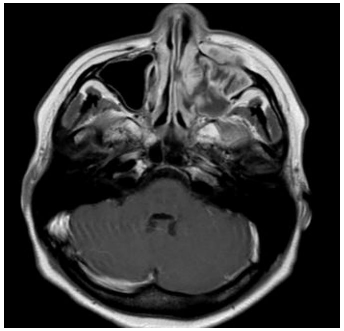
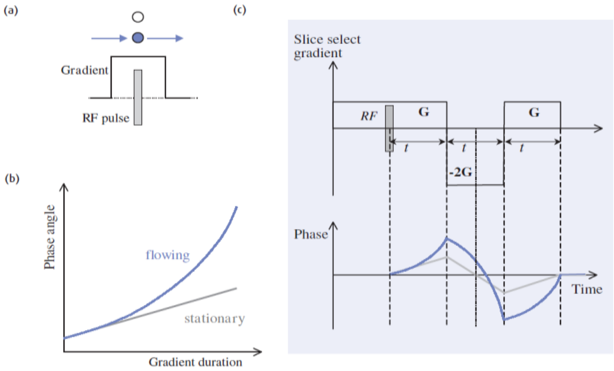


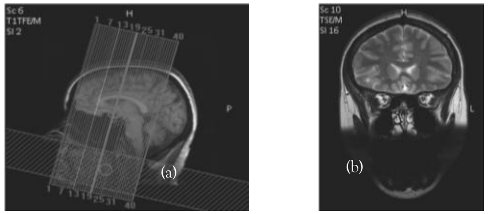


Flow compensation



Avoiding FRE Artifacts

- *Spatial saturation bands, also known as REST slabs or pre-sat bands, are simply slice selections, and can be used in many ways*



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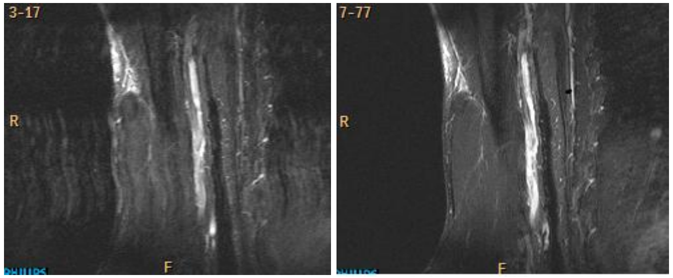


Inhomogeneity artifacts

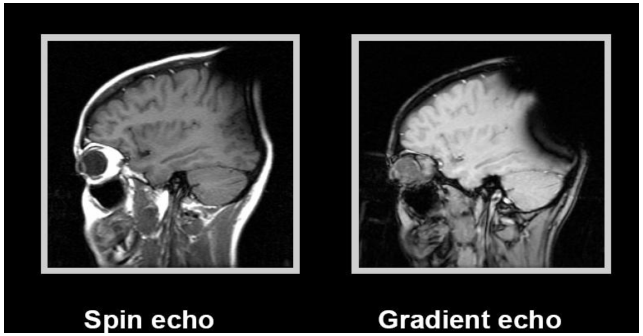
Susceptibility artifacts

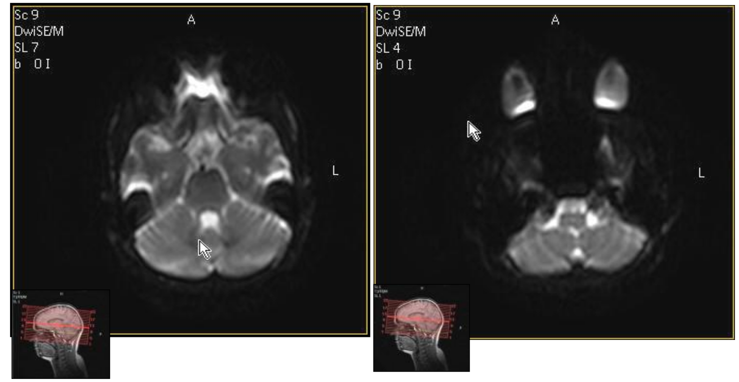
- *Susceptibility artifacts in MRI occur at interfaces of differing magnetic susceptibilities, such as at tissue-air and tissue-fat interfaces (examples include paranasal sinuses, skull base, and sella)*

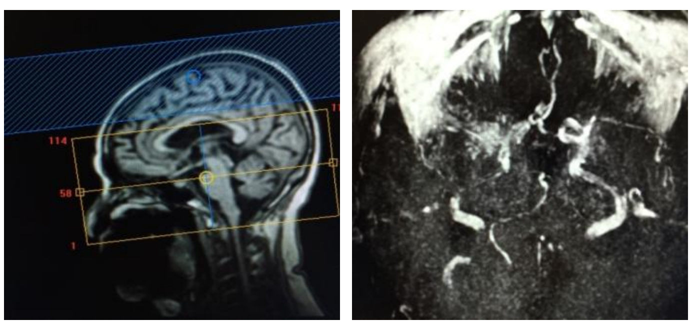
- *There are caused by inhomogeneities , susceptibility artifacts are generally worse on gradient-echo images than spin-echo images.*

Mental Artifacts

- *Metals caused homogeneity change*







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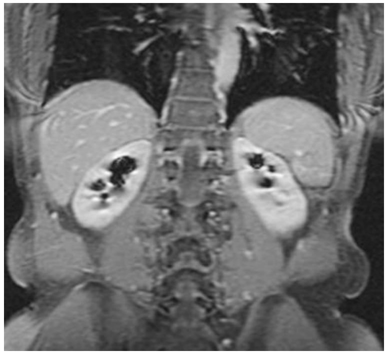


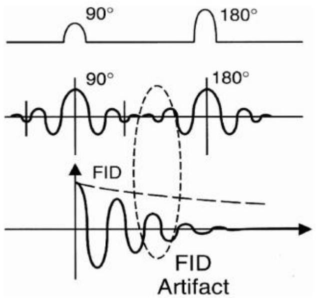
Zipper artifact (RF)

- *This artifact is one form of **central artifacts***
- *Most of zipper artifacts result from inhomogeneities of the magnetic field caused by interferences with radio frequency from various sources.*
- *Software and equipment problems can also cause zipper lines in both directions*

FID Artifacts

- *Free induction decay (FID) artifacts occur due to overlapping of side lobes of the 180^0 pulse with the FID before it has had a chance to completely decay. This overlapping causes a "zipper" artifact*
- *Along the frequency -encode direction.*

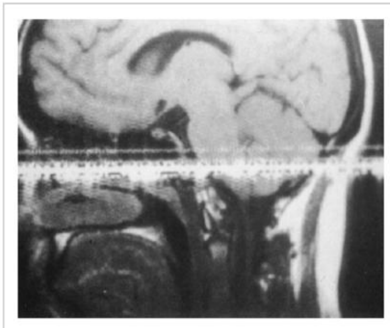
FID Artifacts



Remedy

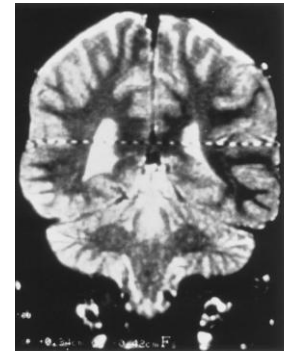
- *Increase the TE (increases the separation between the FID and the 180^0 RF pulse).*
- *Increase slice thickness. This in effect results from selecting a wide RF BW , which narrows the RF signal in the time domain, thus lowering chances for overlap.*

Zipper artifact

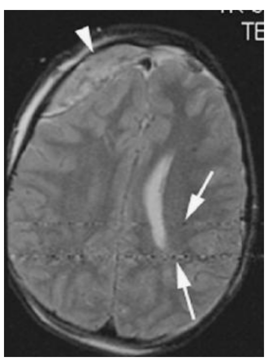


RF Feedthrough Zipper Artifact

- *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*



RF Noise

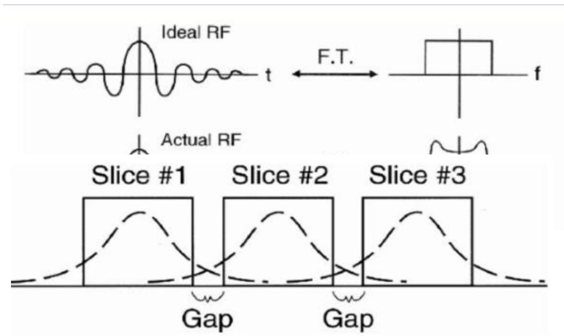


- *RF noise is caused by unwanted external RF noise (e.g., TV channel, a radio station, a flickering fluorescent light, patient electronic monitoring equipment). It is similar to RF feedthrough except that it occurs at the specific frequency (or frequencies) of the unwanted RF pulse(s) rather than at zero frequency*

Remedy for Zipper Artifact

- *Improve RF shielding.*
- *Remove monitoring devices if possible.*
- *Shut the door of the magnet room!*

Cross-talk

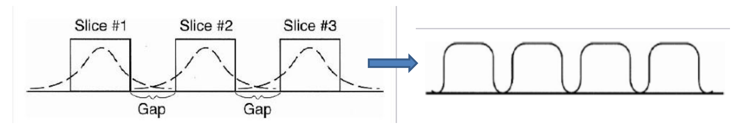


The remedy of cross talk

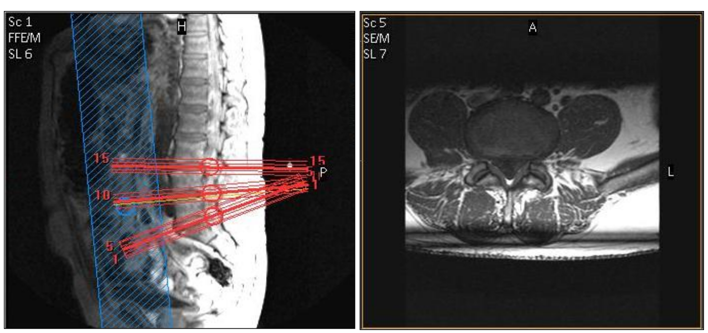
- *At least a 30% gap between the slices.*
- *Excite alternate slices (interleaved) during the acquisition.*

--- *First sequence: odd slices 1,3,5,7, …*

--- *Next sequence: even slices 2,4,6,8,*



Multi-stack artifact

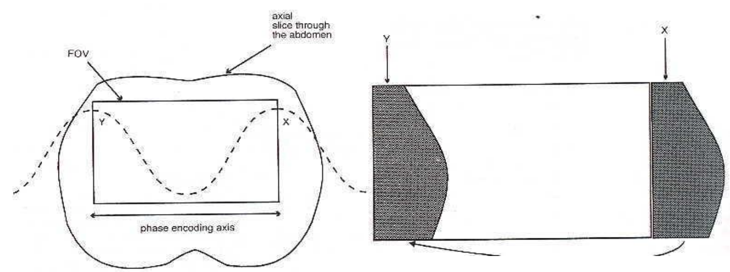


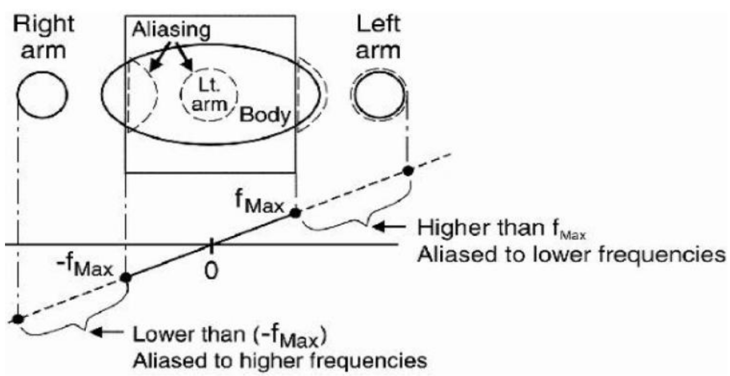
Digital imaging artifacts

Aliasing Artifacts

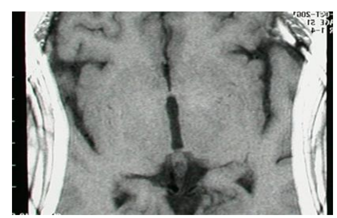
- *Aliasing artifacts, also called wrap-around artifacts*
- *Arises whenever the anatomy bigger than field of view (FOV)*

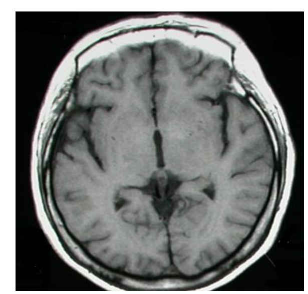
45





Aliasing Artifacts



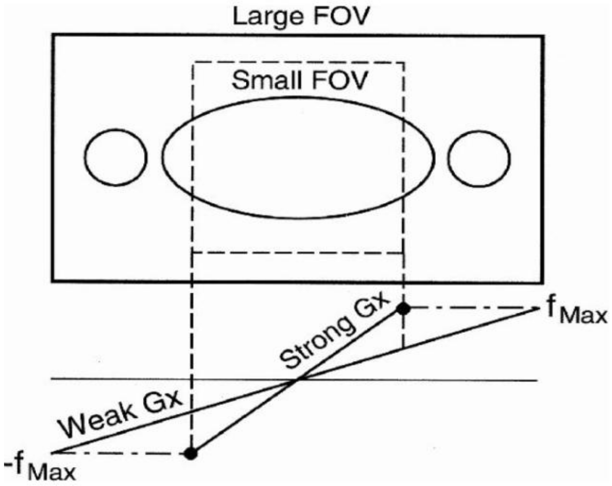


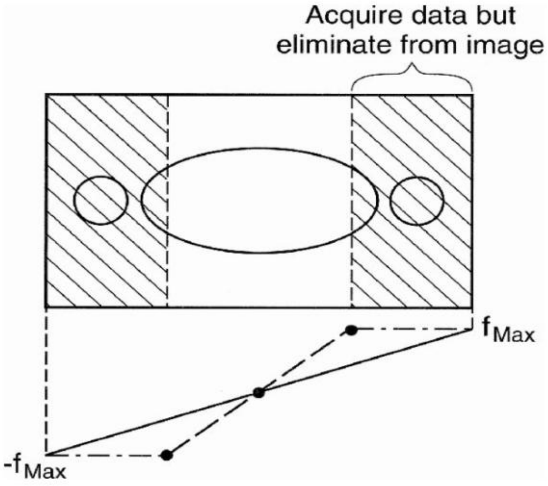
Anti-aliasing along the phase axis

- *Surface coil*
- *Increase FOV*
- *Over samples along the phase encoding axis.*

--- To increasing the number of the phase encoding,the scan time has be prolonged. So, the motion artifact may be more apparent.

- *Saturation pulse*

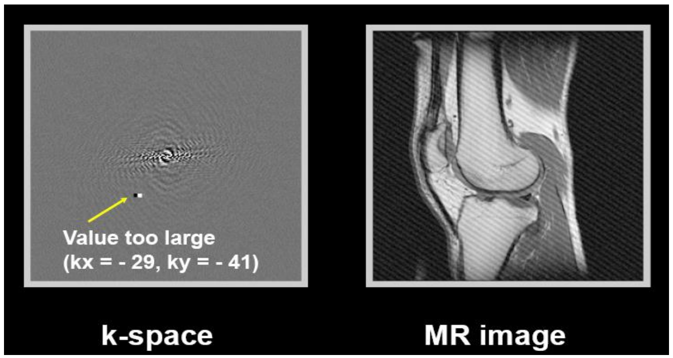


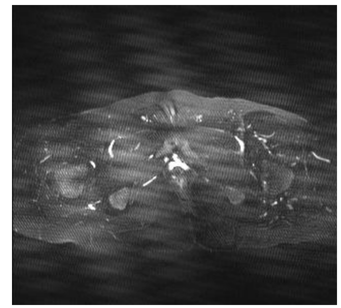


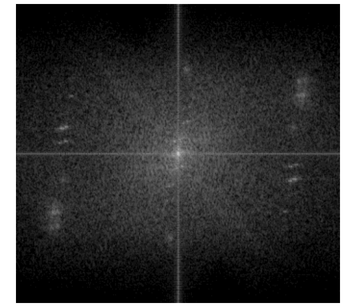
Herring-bone artifact

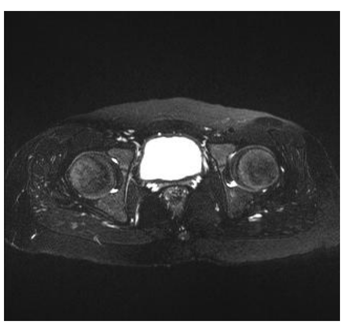
- *A regular series of high- and low-intensity stripes extending right across the image*
- *It is caused by spike noise in the raw data, whose Fourier transform is then convolved with all the image information*

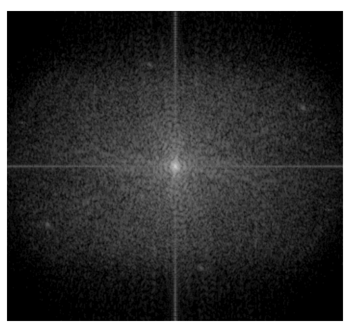
Herring-bone artifact







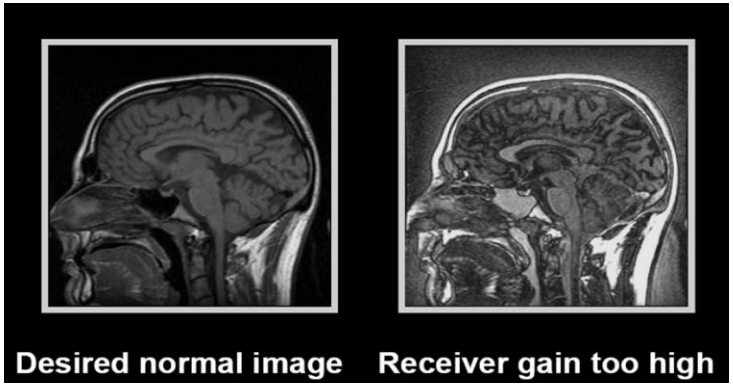




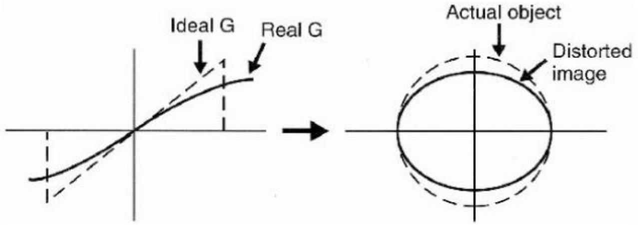
Halo artifact

- *A halo effect can be produced if the receiver gains are incorrectly set.*
- *When this happens the signal is too large for the range of the digitizer and information in the center of k-space is lost*
- *It is a rare artifact with modern automatic pre-scan systems, and is more likely to occur when receiver gains are manually set*

Halo artifact



Gradient nonlinearities

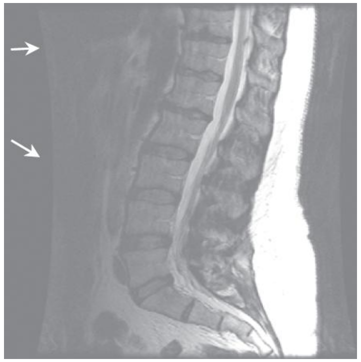


Nonlinearities in the gradient cause distortion in the image. For instance, a circle may appear elliptical.

Gradient nonlinearities

- *The effect of nonlinearities is to distort the image, tending to compress the image information at the edges of the FOV.*
- *Many systems apply a correction to the images to stretch out the pixels, and on rectangular FOV a curved edge can be seen*
- *This is quite normal and also unavoidable; if necessary the area should be re-imaged using a smaller FOV.*

Gradient nonlinearities

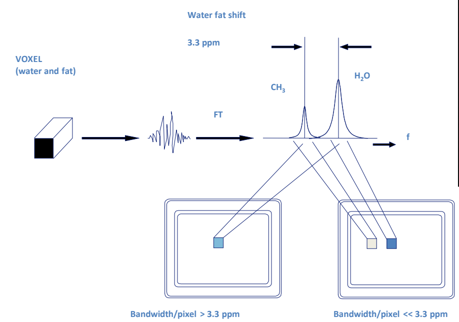


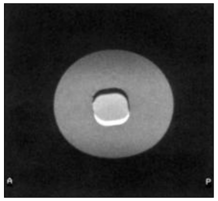
Chemical shift artifact

- *Caused by the different chemical environment of fat and water.*
- *The precessional frequency of fat＜water（depend on the main magnetic field strength）*

  ex. At 1.5T the different of precessional frequency is 220 Hz;at 1.0T is 147 Hz.But at lower field strength（o.5T or less）,it is usually insignificant.

Chemical shift





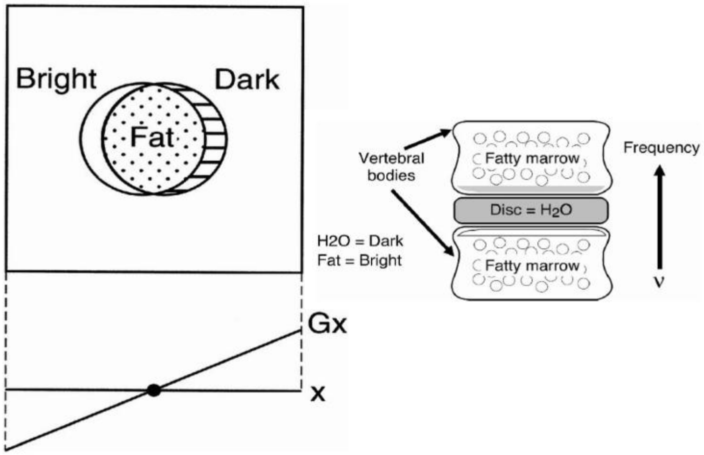
chemical shift artifact

- *For example :*

  The frequency mapped across the FOV is 32000 Hz;256 frequency samples are selected,each pixel has an individual frequency range of 125Hz（32000/256Hz）.At 1.5T,fat and water existing adjacent has a shift about 1.76 pixel which called chemical shift

chemical shift artifact

- *It depend on the size of FOV as this determines the size of each pixel.*
- *Causes a dark edge at the interface between fat and water.*
- *It occurs along the frequency encoding axis only.*



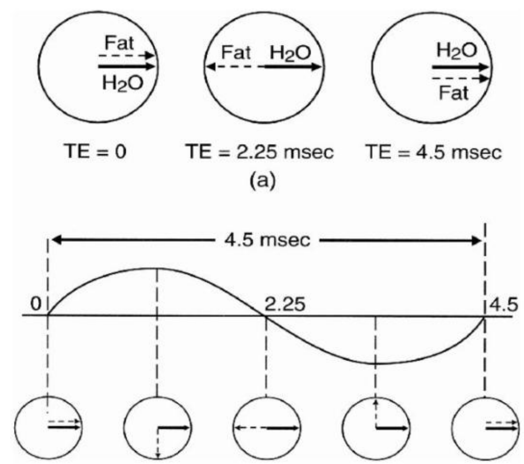
The remedy of chemical shift artifact

- *Using fat suppression.*
- *Scanning at lower filed strengths.*
- *Increase bandwidth (trade-off: lowers SNR)*
- *Switch phase and frequency directions.*
- *Use a long TE (causes more dephasing and less signal from fat).*

Chemical misregistration

Chemical Shift of the "Second Kind"

- *Also produced as a result of the precessional frequency different between fat and water.*
- *Caused because fat and water are in phase at certain times and out of phase at others.*



- *When fat & water are in phase:*
  → *signals add constructively.*
- *When fat & water are out phase:*
  → *signals cancel each other out.*

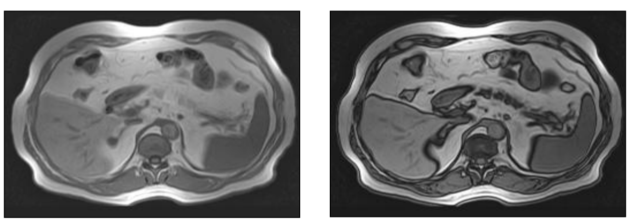
Which called ---- Chemical misregistration

- *Cause a ring of dark signal around certain organs where fat and water interfaces occur within the same voxel.*

The remedy of Chemical misregistration

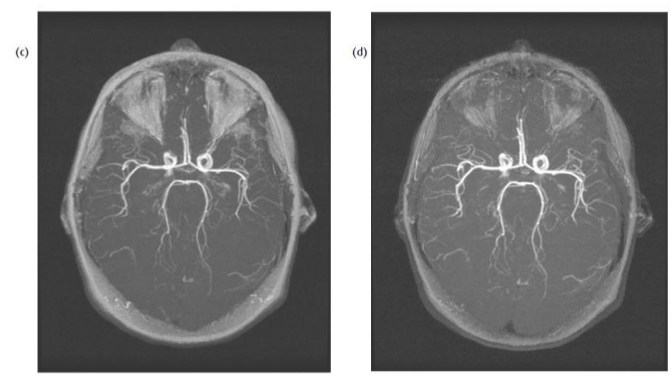
- *Use a spin echo sequence to reduce the artifact.*
- *Select a TE generates an echo when fat and water are in phase. （at 1.5T the TE is a multiple of 4.2ms）*

Chemical shift for in-out phase



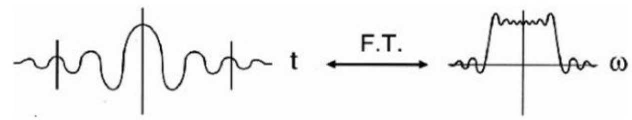
FIELD STRENGTH	In	Out	In	Out	In	Out	In
0.5 T	0	6.8 ms	13.6 ms	20.4 ms	27.2 ms	34 ms	40.8 ms
1.0 T	0	3.4 ms	6.8 ms	10.2 ms	13.6 ms	17 ms	20.4 ms
1.5 T	0	2.2 ms	4.4 ms	6.8 ms	9 ms	11.2 ms	13.4 ms
3.0 T	0	1.1 ms	2.2 ms	3.4 ms	4.5 ms	5.6 ms	6.7 ms

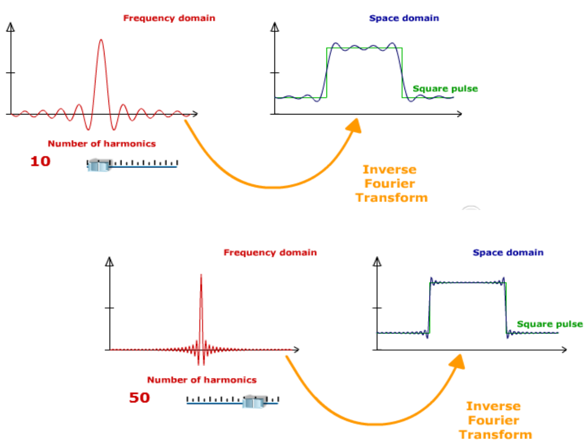
???



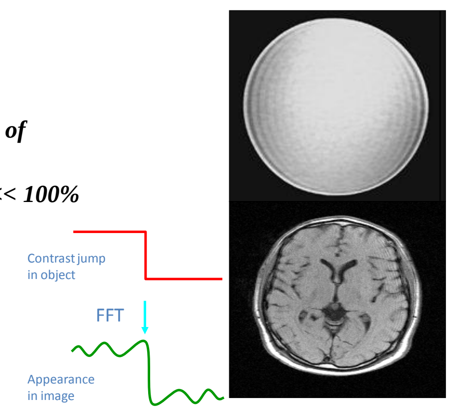
Truncation artifact

- *Caused by under sampling of data so that interfaces of high and low signal are incorrectly represented on the image.*
- *Occurs in the phase direction only.*
- *Produces a low intensity band running through a high intensity area.*





- ***Also called:***
  - ***Gibbs artefact***
  - ***Ringing***
  - ***Spectral leakage***
- ***Predominant in case of***
  - ***low matrix***
  - ***scan percentage << 100%***



The remedy of truncation artifact

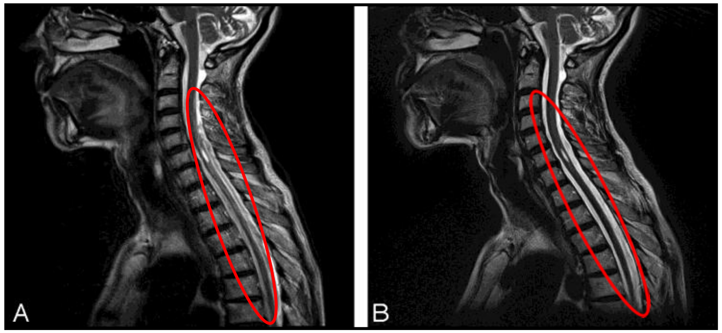
- *Increase sampling time (BW)*
- *Decrease pixel size:*

---The under sampling of data must be avoided.

→ *increase the number of phase encoding steps.*

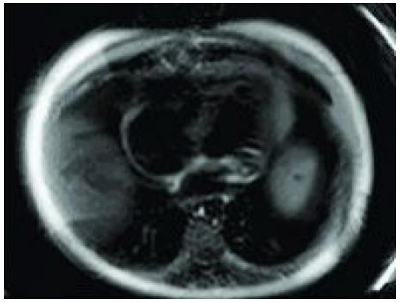
---Decreasing the FOV

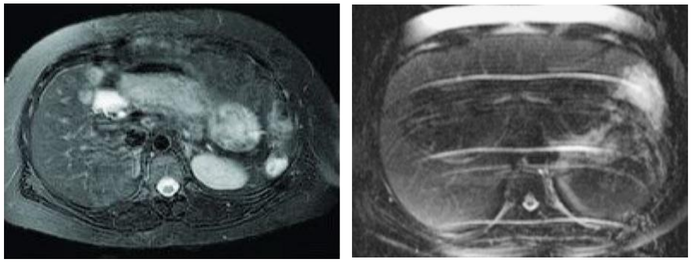
Gibbs' artifact(truncation artifacts)

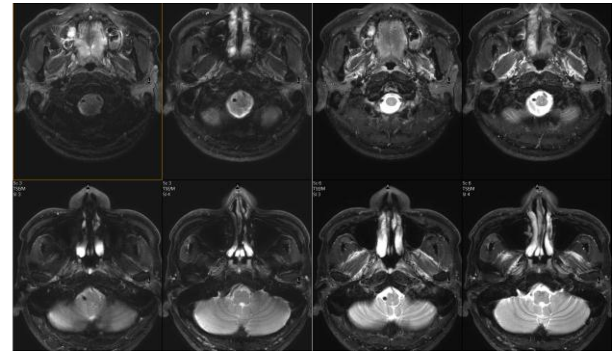


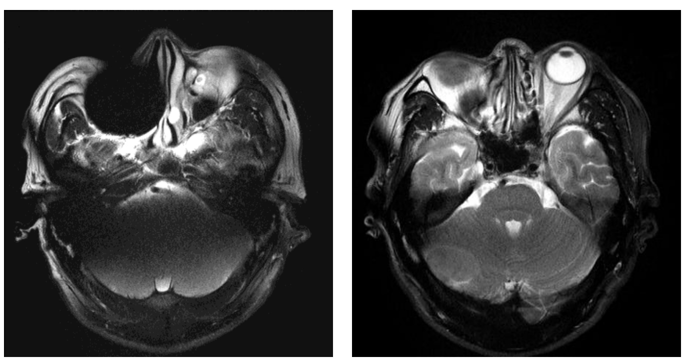
External Magnetic Field Artifacts

- *Artifacts related to B0 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 image distortion*
- *They can be reduced in SE and FSE imaging by using 180^0 refocusing pulses.*
- *They can be a source of image inhomogeneity when a fat suppression technique is used*









感謝您的聆聽，再會！！

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