磁振造影影像最佳化的設定

新光醫院放射診斷科
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討論內容

- Scan time
- Contrast
- SNR
- Resolution
- coverage

How about the scan time?

- TR
- NEX (Number of excitations)
- Number of phase encoding

Scan time=TR ×NEX ×PEs

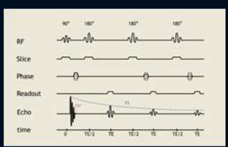
如何降低掃描時間(Scan time)

- FSE or TSE
- Gradient echo sequence
- HASTE
- EPI
- SENSE

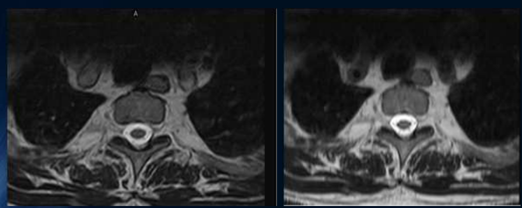
FSE or TSE (快速自旋回波序列)

- 傳統自旋回波+ 回音列車(Echo Train Length)
- 90º RF 後 實施一連串 180º RF
- 每一個180º RF 完成一條 K space 填充
- Scan time= (TR x NEX x PE) / ETL
- ETL 太長的缺點:

1. 影響T1加權的影像對比
2. 影像邊緣模糊
3. Fat訊號加強

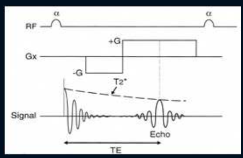


ETL 25 vs. 45



Gradient echo sequence(梯度回訊序列)

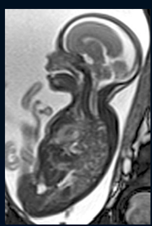
- 小於 90º 的 RF
- 沒有180º 的 RF 重聚相
- 短TR、短TE，用於快速造影
- 運用 Bi-lobed Gradient 進行重聚相
- SNR較低
- 對於磁敏性假影(susceptibility artifact) 較敏感



HASTE

- *H*alf-Fourier *A*cquisition *S*ingle-shot *T*urbo spin *E*cho imaging
- 在一次激發下取得影像(**single-shot technique**)
- ETL很長................................., 問題又出現了....................
- 訊號的特性--- 360º相位變化，正、負半波對稱
- 只取正半波，複製成負半波(一半再多一些...)
- PE下降，SNR降低
- 用於快速造影

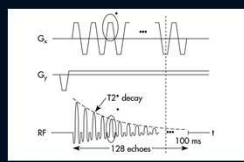
HASET影像



EPI (Echo-Planar Imaging)

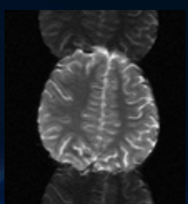
- 運用梯度磁場的快速轉換
- 在一次的激發中完成影像的擷取
- 超快速成像
- 易產生下列假影:

1. N/2 artifact
2. Susceptibility artifact
3. Chemical shift artifact

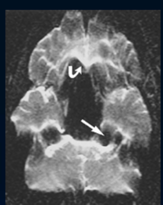


EPI假影

N/2 artifact



Susceptibility artifact

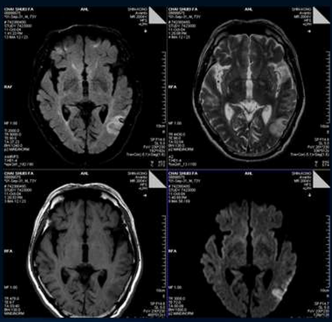


EPI 在臨床上的應用

- Diffusion imaging of the brain.
- Dynamic perfusion studies of the brain.
- Abdominal imaging.
- Cardiac imaging:

1. Coronary arteries
2. Cine cardiac imaging within a single heart beat.
3. Dynamic perfusion studies of the myocardium.

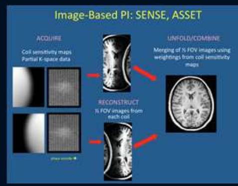
DWI成像



SENSE(平行成像技術)

簡單來說..........

- 運用多重元件(elements)組成的相位陣列線圈 (phased array coil)
- 由個別線圈的元件分別擷取部分K space的訊號，產生反摺影像
- 經由後處理運算，將反摺影像重組為正常影像



Contrast(對比)

- 定義: 不同組織或病理結構間在影像中相對亮度的差異
- 在MRI的成像中，主要因不同組織的物理特性的差異而產生，如T1及T2遲豫時間、質子密度等.......
- 公式:

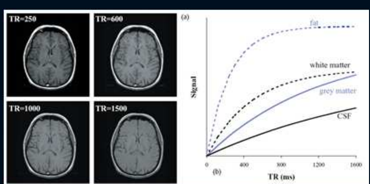
Contrast(C)= $(S_A - S_B) / (S_A + S_B)$

影響組織對比的因素

- 重複時間(TR)
- 回訊時間(TE)
- 偏轉角(FA)
- 反轉時間(TI)

重複時間(TR)與組織對比

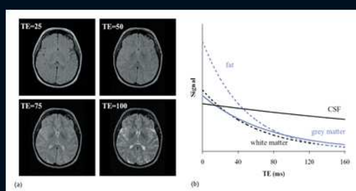
- TR的定義: 兩次 RF 之間所需要的時間
- 與掃描時間有關
- TR越長，組織間的對比越差



TE=10

回訊時間(TE)與組織對比

- TE的定義: 施加 RF後，到擷取訊號間所需要的時間
- 與掃描時間無直接的關係，和可掃描的張數有關
- TE時間越長，在長TR的狀況下，水與其他組織的對比越強



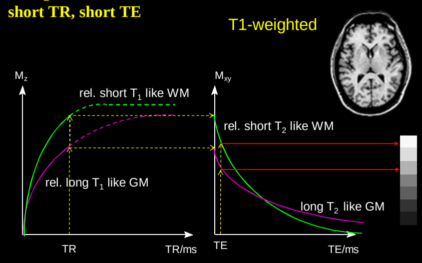
TR=1500

關於TR and TE…….

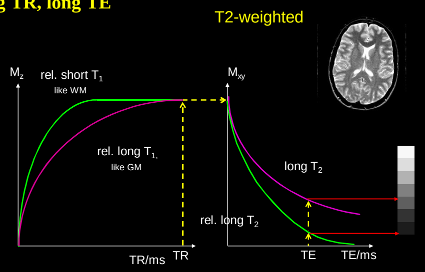
- 在訊號的擷取中，TR和TE都是重要且須同時考慮的因子
- 在波序中，TR決定T1加權的程度，而TE決定T2加權的程度
- 在不同長短TR的前提下，TE時間也需做調整以獲的最佳的影像對比
- 訊號的基本公式:

$$Signal= H(N)(1-\exp^{-TR/T1})(\exp^{-TE/T2})$$

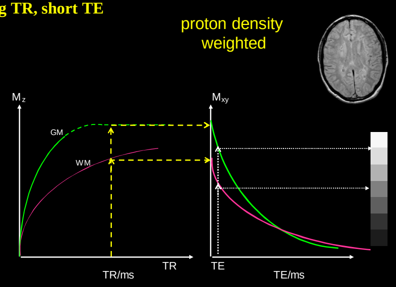
Spin Echo
short TR, short TE



Spin Echo
long TR, long TE



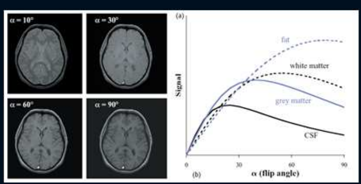
Spin Echo
rel. long TR, short TE



偏轉角(FA)與組織對比

- FA的定義: RF施加後，將M0偏轉至xy平面的角度
- 運用於梯度回訊波序(gradient echo)
- FA的角度，決定GE的影像加權:

1. 大FA → T1影像
2. 小FA → T2影像

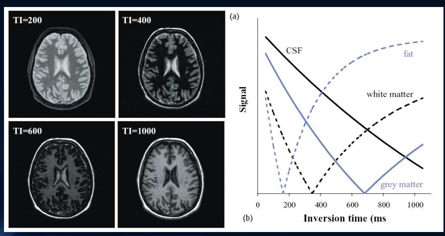


反轉時間(TI)與組織對比

- TI的定義: 施加180º RF 後，到施加90º RF 間所需要的時間
- 用於反轉回復波序(Inversion Recovery)
- 需增加TR時間
- 作用:

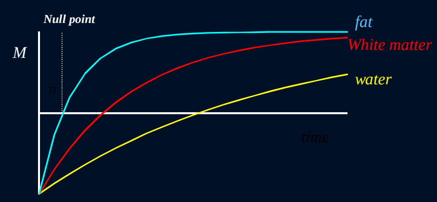
1. 壓抑特定組織訊號(null point)
2. 增強組織間的對比

TI and contrast



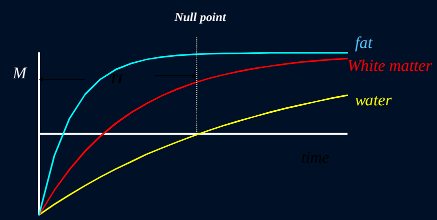
TR=4000, TE=19

STIR

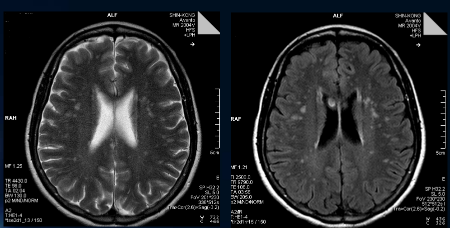


STIR image

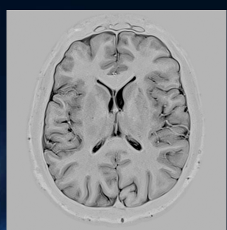
Flair

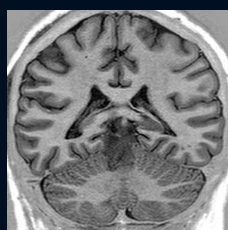


FLAIR in Brain MRI



T1 IR (TI=400ms)





Signal to Noise Ratio(訊雜比)

Signal(訊號)

- 影像中像素或體素中的相對亮度
- 與物質中的氫原子密度有關

Noise(雜訊)

- 影像中隨機出現的雜點
- 多數由病患組織中產生

影響SNR的因子

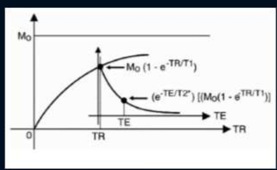
- 重複時間(TR)
- 回訊時間(TE)
- 體素大小
- 平均次數(NSA)
- 切面厚度
- 頻寬

TR, TE 與SNR

- 重複時間(TR)

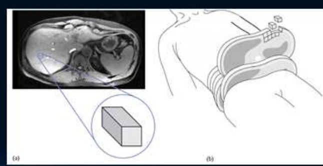
--- TR越長，SNR越高(下次可偏轉到xy平面上的越多)

- 回訊時間(TE)
- --- TE越長，SNR越低(訊號衰減越多)

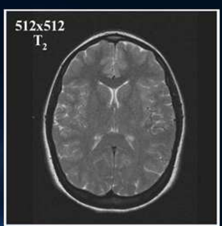


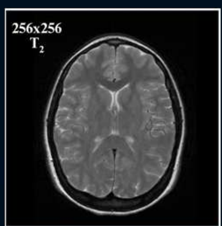
體素大小 (Voxel size)

- 大牛排理論
- 體素越大，單位體積內所含氫原子數量越多
- SNR越高



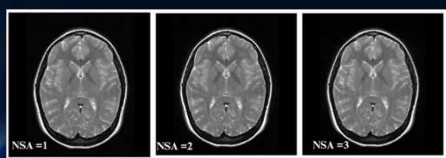
Voxel size





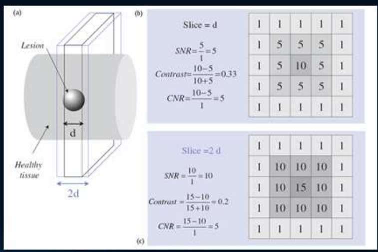
平均次數(NSA)

- 針對特定區域內組織，訊號重複擷取的次數
- 擷取的次數每增加一倍，相位編碼的次數也增加一倍
- 所需Scan time 加倍，SNR增加 $\sqrt{2}$ 倍



切面厚度

- 大牛排理論
- 切面厚度越厚，單位體積內所含氫原子數量越多
- SNR越高
- 解析度降低



頻寬(Bandwidth)

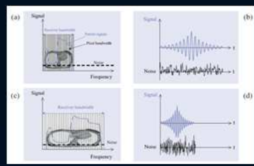
- BW 與SNR 成反比
- 當BW降低:

--- SNR 上升 (BW 降低2倍，SNR 增加 √2 倍)

--- 化學位移假影(chemical shift artifact)增加

--- TE 時間增加

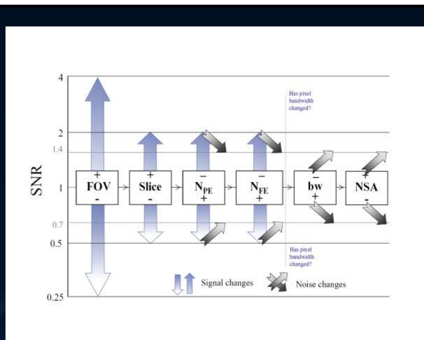
Bandwidth = 1/ΔTs =Nx /Ts



SNR???

$$SNR \propto \frac{\Delta x \cdot \Delta y \cdot \Delta z \cdot F_{sequence} \cdot \sqrt{NSA \cdot N_{PE} \cdot N_{FE}}}{\sqrt{BW}}$$

$$SNR \propto \frac{FOV_{FE} \cdot FOV_{PE} \cdot \Delta z \cdot F_{sequence} \sqrt{NSA}}{\sqrt{BW \cdot N_{FE} \cdot N_{PE}}}$$



Resolution(解析度)

- 空間解析度(Spatial resolution)

與體素大小有關

- 時間解析度(Temporal resolution)

與掃描時間有關

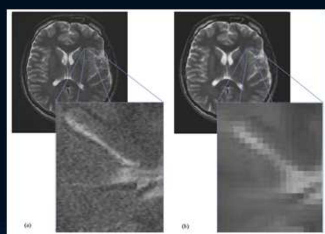
用於dynamic scan

空間解析度(Spatial resolution)

- 分辨兩點之間的最短距離
- 影響體素的大小
- 高解析度，SNR下降

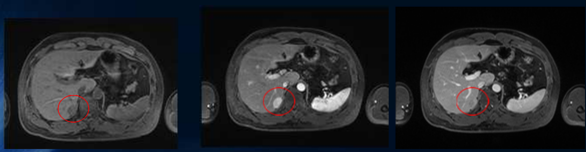
--- 增加平均次數(NSA)

--- 掃描時間增加



時間解析度(Temporal resolution)

- 運用快速掃描波序
- 針對同一掃描範圍進行多次影像擷取
- 應用在對比劑注射後的動態掃描



Coverage(涵蓋範圍)

- 定義: 在多重切面掃描波序中可以涵蓋的距離
- 影響因子:

--- Number of slices

slices = TR/(TE + Ts /2 + To)

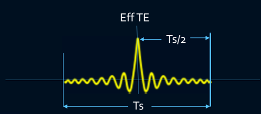
--- Slice thickness

--- Gaps

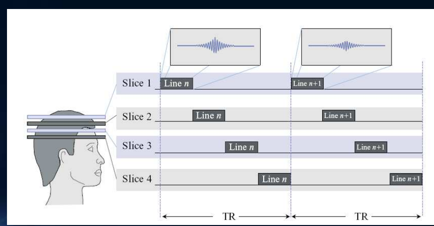
--- ETL

slices = TR/(TE + Ts /2 + To)

- TR:重複時間，TR越長，可切的切面張數增加
- TE:回訊時間， TE越長，單位TR內可切的切面張數減少
- TS:取樣時間， TE越長，單位TR內可切的切面張數減少
- TO:準備時間，用於波序的前置準備，如IR、FS 等.....

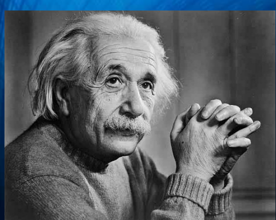


TR vs. slices



一個從未犯錯的人是因為他不曾嘗試新鮮事物

Anyone who has never made a mistake has never tried anything new.



平行成像技術-Parallel imaging

新光醫院放射診斷科
李正輝

影像醫學的巨人---MRI

速度--- MRI

- 魚與熊掌
- 速度提升--- 影像品質的犧牲
- MRI掃描速度的公式:

Scan time = TR x NEX x No. of Phase encoding

縮短檢查時間的方法

- 縮短TR

--- Gradient echo technique

- 減少掃描的次數
- 減少相位編碼的總次數

--- 影響解析度、有可能造成反摺假影

--- Haste

- 增加相位編碼的次數 / TR

--- TSE、EPI

- 平行成像技術

小學的數學題...

- 小明想要蓋一棟別墅給阿花住，一個人每天可以完成1/20的進度，請問需要幾天可以完成? 如果小華一起來幫忙，同樣每天可以完成1/20的進度，請問需要幾天可以完成?

(TSE)

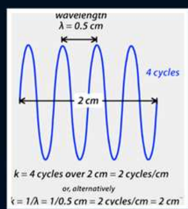
- 愚公想要挖一條隧道，如果從一側開挖 ，需要10天才能完成，於是愚公叫愚小弟從另外一側同時開挖，如果兩人的速度一樣，請問需要幾天可以完成??

(SENSE)

首先要瞭解的事--- K space

- 應用於光學、聲波、機械學，電磁學 and so on...
- 描述**波**在不同介質中的**空間頻率**
- 波長wavelength (λ): 兩個相鄰峰之間的距離
- 波數wavenumber (k): 單位距離內**波**的數目

$$k = 1/\lambda$$



空間頻率(Spatial Frequencies)

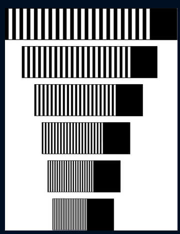
- 單位距離內的波數

--- 空間頻率高，單位距離內的波數多

--- 可分辨兩點間的距離縮短

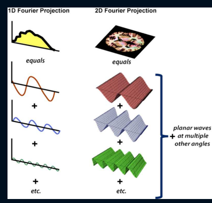
--- 高解析度(Resolution)

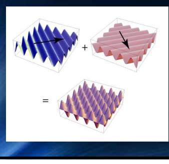
--- 以 line pairs per mm 為單位

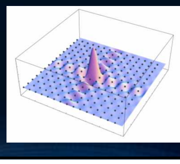


MRI vs. K space ???

- MRI的訊號--- 一系列的 sine 和 cosine波組成
- 一度空間---線，二度空間--- 平面
- 平面訊號必須由各個不同的方向進行運算





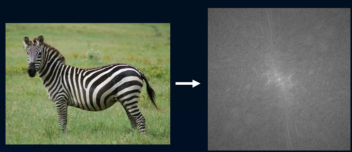


也就是說....

- 經由電腦的後處理(傅立葉轉換)，影像可以被分解成具有不同相位(phase)、頻率(frequency)、振幅(amplitude)與方向性(orientation)的訊號集合
- K space

--- 影像的空間諧波含量

--- 諧波特性的集合



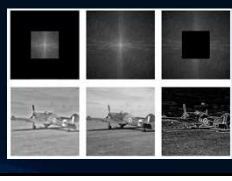
K space 中不同位置所包含的資訊??

- 中央部分--- 低頻資訊(low spatial frequency information)

--- 包含影像的對比、亮度以及外形

- 外圍部分---高頻資訊(high spatial frequency information)

---包含影像的邊界、細節以及銳利度



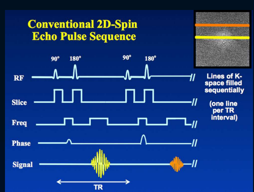
空間編碼

- 藉由不同強度與方向性的梯度磁場組合
- 利用磁度磁場產生的訊號頻率
- 將"位置"的訊號特性進行擷取
- 包含三個重要的編碼方向:

--- 切面選擇(Slice selection)

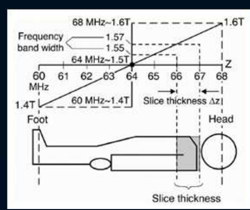
--- 相位編碼(Phase encoding)

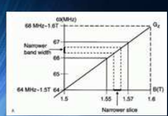
--- 頻率編碼(Frequency encoding)

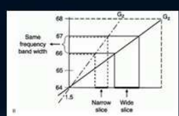


切面選擇(Slice selection)

- 決定所需擷取的切面位置與厚度
- 施加梯度磁場
- 只激發特定的頻寬內的氫原子核

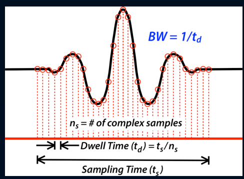






頻率編碼(Frequency encoding)

- 運用梯度磁場
- 依據所定義的取樣間隔
- 將訊號進行區隔並讀出
- 頻率編碼次數---解析度



相位編碼(Phase encoding)

- 訊號的來源
- 藉由相位的改變，將所得到的訊號，填入K space 中特定的位置
- 相位編碼次數越多，訊號越好
- 每次改變相位並填入 K space 所需的時間--- TR

掃描時間= TR x NEX x 相位編碼的次數

- 決定整體掃描時間的重要因素

常規成像技術 vs.平行成像技術

- 常規成像技術(以TSE為例，ETL=16)

掃描時間= (TR x NEX x 相位編碼的次數) / (ETL)

TR=3000ms、NEX=1、相位編碼的次數=256 、 ETL=16

Scan time= 48000ms = 48s

- 平行成像技術

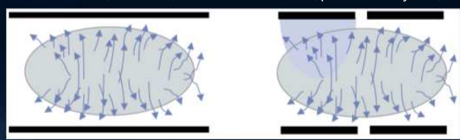
掃描時間= (TR x NEX x 相位編碼的次數) / (ETL x 加速因子)

Scan time= 48000ms/2 = 24s

- 加速因子--- 藉由減少相位編碼的次數，降低掃描時間

關於平行成像技術

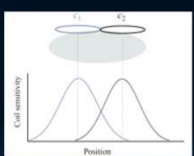
- 運用多重元件(elements)組成的相位陣列線圈(phased array coil)

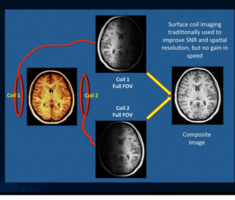


- 較大的單一元件(element)線圈相較於多重元件(elements)組成的相位陣列線圈接收較多的雜訊
- SNR 提升

關於平行成像技術

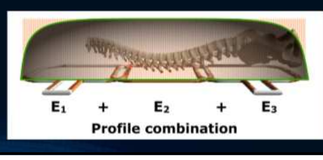
- 傳統上使用相位陣列線圈(phased array coil)中，各獨立的元件(element)接收訊號並組成最終的影像
- 每一獨立元件(element)依其位置的不同對訊號的接收效率也有所不同--- 線圈敏感度(Coil sensitivity)





關於相位陣列線圈(phased array coil)

- 為表面線圈(surface coil)的一種(具訊號接收功能，但通常不具RF發送功能)
- 由多個表面線圈組合而成
- 保留小線圈的sensitivity，但可涵蓋較大範圍的掃描空間



關於平行成像技術

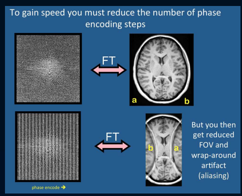
- 減少相位編碼的次數(under sampling)

---減少掃描時間

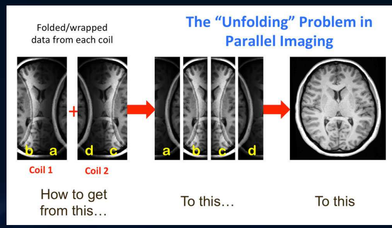
- 產生反摺假影(warp-around artifact)
- 影像重建(reconstruction)

---傅立葉轉換後(image domain)進行的重建 :SENSE, mSENSE

--- 傅立葉轉換前(frequency domain)進行的重建 :SMASH, GRAPPA)



影像的重建



傅立葉轉換後(image domain)進行的重建

- SENSE(**SENSitivity Encoding)**

---最先發展的平行成像技術

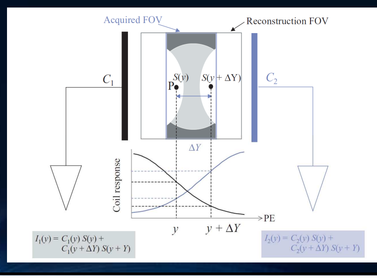
---藉由跳躍式進行相位編碼的方式以減少影像擷取的時間

---取得反摺的影像

---運用 coil sensitivity 的資訊(Reference images)將反摺影像重組

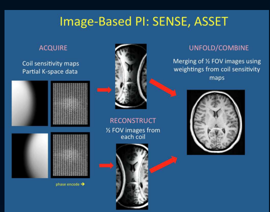
---完成無反摺假影的影像

SENSE 成像(加速因子=2)

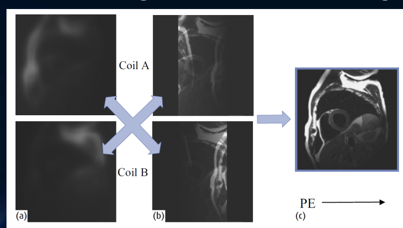


SENSE 作用的四個步驟

- 產生 coil sensitivity map
- 取得部分 K space 的 MR data
- 從個別得線圈中重建部分 K space 的 MR data
- 藉由矩陣的反推運算解反摺並組成影像

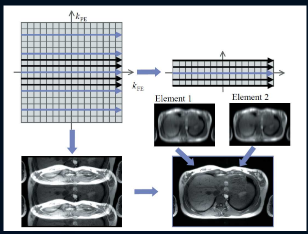


Reference images and Aliased images



mSENSE(Modified SENSE)

- 不另外產生Reference images
- 在取正常診斷用影像時，在K空間的中央部分額外取數條訊號
- 此額外取得之訊號由各獨立之線圈(or element)分別取得
- 產生低解析度、無反摺之影像作為Reference images

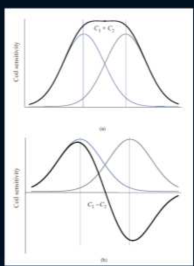


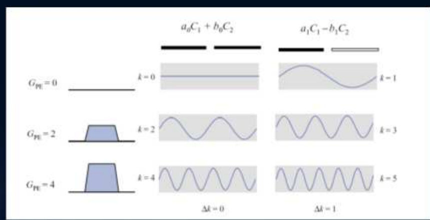
傅立葉轉換前(frequency domain)進行的重建

- SMASH (**Si**multaneous **A**cquisition of **S**patial **H**armonics)
- **Auto-calibrating SMASH**
- Variable density Auto-SMASH (VD-Auto-SMASH)
- GRAPPA (**Gene**Ralized **A**utocalibrating **P**artially **P**arallel **A**cquisitions)

SMASH (SiMultaneous Acquisition of Spatial Harmonics)

- 第一代的平行成像技術
- 作用在 K space 的層面
- 藉由發送RF並接收相關的空間資訊，產生一組與真實訊號相似的假性k-space 相位編碼訊號
- 也就是說，在適當的運用來自不同線圈元件的信號的組合，以產生均勻(相加)或不均勻(相減)的空間頻率
- 同時使用coil sensitivities 來產生不同權重的組合





結合梯度改變與線圈相位編碼排列在3次訊號擷取中產生6條Kspace lines

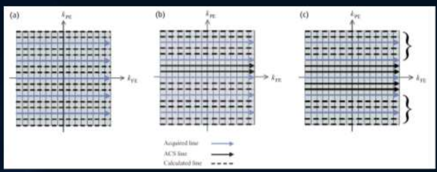
Auto-calibrating SMASH

- 取得部分 K space 的 MR data
- 在接近中心部分取得 auto-calibrating signal' (ACS) lines
- 計算最接近 ACS 的K space 訊號並與ACS 訊號進行對比
- 將所得到的有效參數結果套用到其他取得的部分 K space MR data中

Variable density Auto-SMASH (VD-Auto-SMASH)

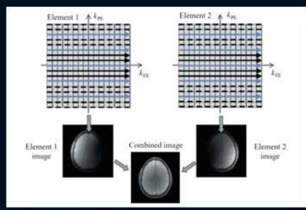
- 在中心部分取得更多的 auto-calibrating signal' (ACS) lines
- 減少假影(artifacts)及重組時的錯誤
- 相較於**Auto-calibrating SMASH**，**scan time 較長**

SMASH vs. Auto- SMASH vs. VD-auto SMASH



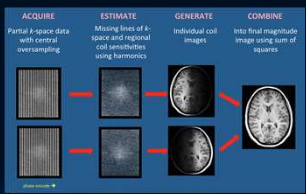
GRAPPA (GeneRalized Autocalibrating Partially Parallel Acquisitions)

- Auto-SMASH 進階版
- 由個別線圈各自產生獨立的 ACS line
- 各自對個別線圈所產生的訊號進行校正並進行重組



GRAPPA 作用的四個步驟

- 部分 K space MR data 的取得
- 建立遺失的 K space MR data
- 產生由個別線圈所組成的影像
- 組成完整的影像



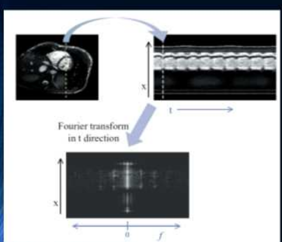
k-t BLAST (Broad-use Linear Acquisition Speed-up Technique)

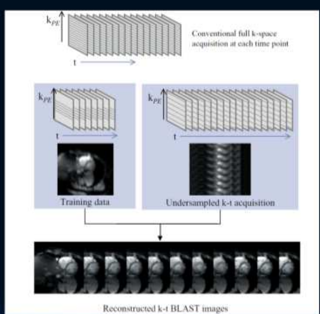
- 運用於單一切面中的動態造影(dynamic techniques)
- 原理: (以心臟動態造影為例)

--- 胸壁、肺臟、脊椎、肌肉等組織在造影過程中並無太大變化

--- 偷懶一下.... 周圍組織只做一次，只針對移動的心臟進行完整掃描

- 快速、高時間及空間解析度
- 可與SENSE技術結合，實施快速、多切面、高時間及空間解析度造影
- Training data: 一組低解析度的連續影像(只要幾張就夠了....)

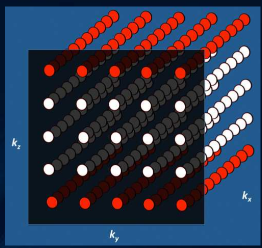




CAIPIRINHA

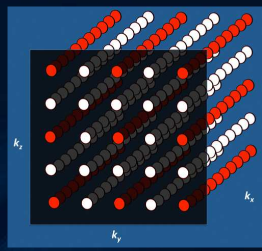
- Controlled Aliasing in Parallel Imaging Results in Higher Acceleration
- 用於3D 閉氣快速造影
- 運用特殊的K space 填充方式來增加掃描速度

- 在單一方向進行4倍加速的影像擷取:
  快速但雜訊較高，同時仍有殘存的反摺假影



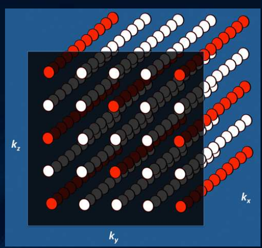
Acceleration in a single direction (R=4)

在兩個方向進行4倍加速的影像擷取:
具有較佳的相位排列方式，仍維持4倍的加速因子



Acceleration in two directions (R = 2 x 2 = 4)

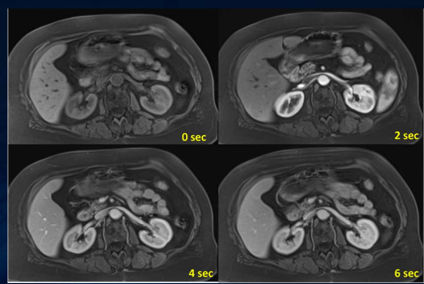
- CAIPIRINHA
  在兩個方向進行2倍加速的影像擷取
  藉由相位偏移維持4倍加速效果
  降低雜訊與反摺假影
  常用於3D 肝臟造影、DWI、MRA



CAIPIRINHA pattern with acceleration in two directions plus phase shift of alternate rows

3D 肝臟造影

3D T1-VIBE using CAIPIRINHA with acceleration factor (R) = 4
A set of 72 slices (3-mm-thick) through the entire liver are repeatedly acquired every 2 sec in a single breath hold during dynamic passage of contrast.



平行造影技術中的雜訊

- 減少相位編碼的次數 ⟶ SNR降低
- 加速因子(R)增加，SNR下降

例子: R=2 ，掃描時間降為1/2

，SNR 降為 1/√2 ≈ 0.71

- 幾何因子(g): 與反摺點的數量與線圈敏感度的差異有關

--- 表面線圈的數量與位置

--- 切面選擇與phase encoding 的方向

$$SNR_{parallel} = \frac{SNR}{g\sqrt{R}}$$

R*: *reduction* or *acceleration factor
g* : *geometry factor

加速因子(R)與雜訊

平行造影技術中的假影

- 傳統非平行造影技術中的假影

--- noise, motion, aliasing, chemical shift, Gibbs, susceptibility, RF-interference, etc.

- 不同於傳統非平行造影技術中的假影

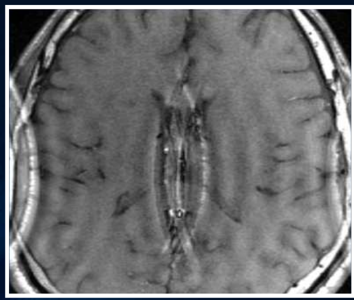
--- **SENSE ghost:** 取樣不足產生的反摺假影，發生於影像中央

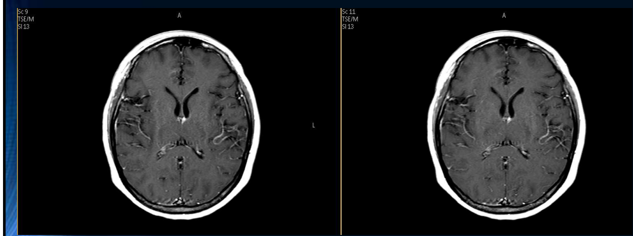
--- **Motion artifacts:** 發生於calibration scan 與 image acquisition間

--- **Noise:** 因幾何因子(g)產生的假影

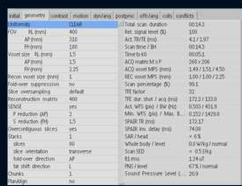
SENSE ghost

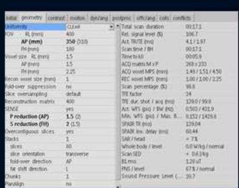
- 加大FOV
- Over Sampling
- 降低加速因子(R)





PE=200, SENSE=2　　PE=200, SENSE=1.25





Motion artifacts

- 常見於高訊號組織(FAT)
- 重新固定病患後再次執行掃描
- 使用自動校準平行加速技術(mSENSE)

Noise in SENSE

- 重新進行校正(calibration)

---重覆coil sensitivity mapping

--- Acquiring more ACS lines

- 改變相位編碼方向
- 降低加速因子(R)

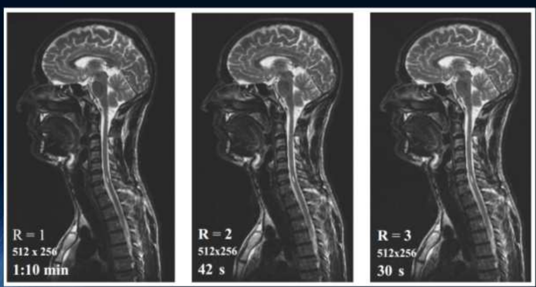
平行造影技術在臨床使用的優勢

- 減低掃描時間
- 閉氣時間減少，降低移動假影
- 合理時間內取得高解析度影像
- EPI成像的應用

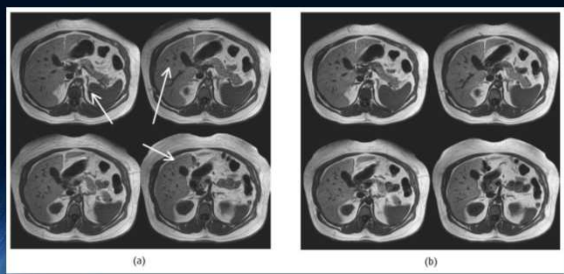
--- 縮短echo train length (increasing the PE bandwidth)

--- 減少susceptibility distortions

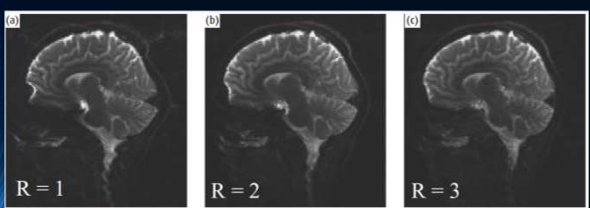
- 降低HASTE影像中的邊緣霧化(減少echo spacing)



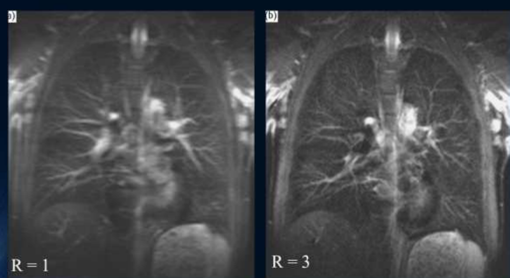
不同加速因子對掃描時間的影響



SENSE 在降低呼吸假影的應用(25s vs. 13s)

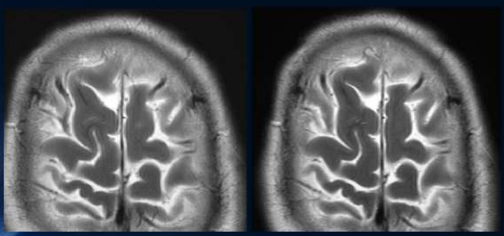


EPI成像的應用---減少susceptibility distortions



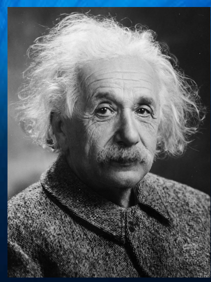
降低HASTE影像中的邊緣霧化

SENSE在流動假影抑制的應用

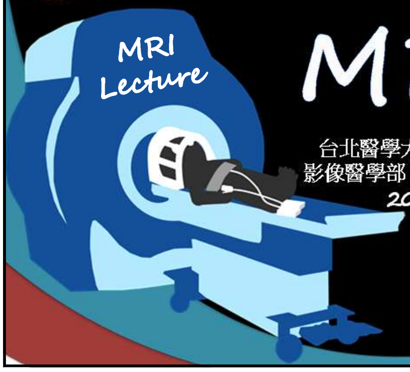


科學的全部不過就是日常思考的提煉

The whole of science is nothing more than a refinement of everyday thinking.



中華民國醫事放射學會
Taiwan Society of Radiological Technologists (TWSRT)

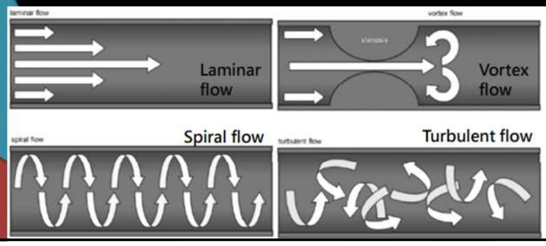


MRA

台北醫學大學—部立雙和醫院
影像醫學部 醫學物理師 李宜恬
2018.03.24

Flow phenomena

- Sources: Flowing blood or cerebrospinal fluid (CSF)
- Phenomena: become dark or bright in MRI



Characteristic Factors

- Velocity
- Pulse sequence (SE vs. GRE)
- Slice Position (containing the vessel relative to the rest of the slices)
- Contrast (TR and TE)
- Echo number (even or odd)
- Slice thickness
- Flip angle
- Gradient strength and rise time
- Use of gradient moment nulling (flow compensation) techniques
- Use of spatial saturation
- Use of cardiac gating
- Chance of cardiac gating (pseudogating)

Velocity vs. Flow

- Velocity (cm/sec), v
- Volumetric flow (cm^3/sec), Q
- Cross-section area of the vessel (cm^2), A
- **V=Q/A**

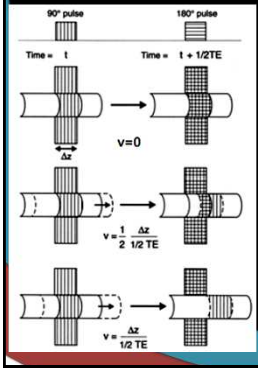
Appearance of flowing blood

- **Flow effects** can be attributed to
  - Time of flight (TOF) effects, Mz
  - Motion-induced phase changes (phase contrast), Mxy
- **TOF effects**
  - Signal loss (high-velocity signal loss or TOF loss)
  - Signal gain (flow-related enhancement, FRE)

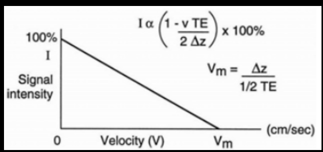
TOF effects

- Decrease signal intensity (dark)
  - High velocity
  - Turbulent flow
  - Dephasing (odd-echo dephasing and intravoxel dephasing)
- Increased signal intensity (bright)
  - Even-echo rephasing
  - Diastolic pseudogating
  - Flow-related enhancement (FRE)

High velocity signal loss



- **Spin echo** imaging
- Slice-selective refocusing RF

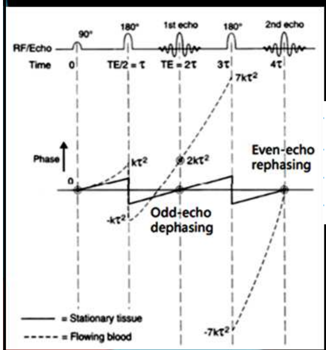


Velocity (v) and phase (ϕ)

$$\varphi = \int \omega dt = \int (\gamma G v t) dt = \gamma G v \int t dt = \gamma G v (t^2/2)$$

- Phase and velocity are proportional
- A quadratic relationship exists between phase and time $\varphi = kt^2$

Echo number effects



$t = \tau = \frac{1}{2}TE$ → assume $\varphi = k\tau^2$

$t = [\tau, 2\tau]$ → $\varphi = [k(2\tau)^2 - k\tau^2] = 3k\tau^2$

$t = [2\tau, 3\tau]$ → $\varphi = [k(3\tau)^2 - k(2\tau)^2] = 5k\tau^2$

$t = [3\tau, 4\tau]$ → $\varphi = [k(4\tau)^2 - k(3\tau)^2] = 7k\tau^2$

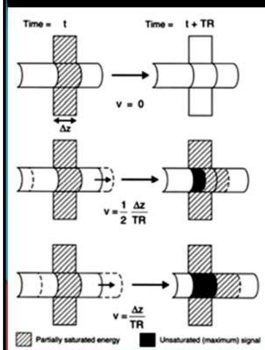
Echo number effects

- Assume velocity of blood flow is constant
- Odd echo → dephasing → signal decrease
- Even echo → rephasing → signal gain

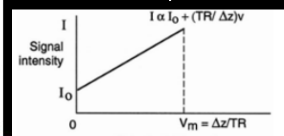
Diastolic pseudogating

- Systole (rapid) vs. diastole (slower)
- In diastolic, the TOF effects result in higher intravascular signal.
- Use cardiac gating to acquire slice at a fixed point in the cardiac cycle.
  - **TR = 1/(heart rate)**

Flow-related enhancement (FRE)

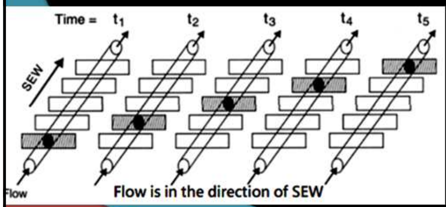


- **Gradient echo**
- Entry phenomenon
- The fresh inflowing blood that enters the first slice is totally un saturated (by last RF excitation)



Cocurrent Flow

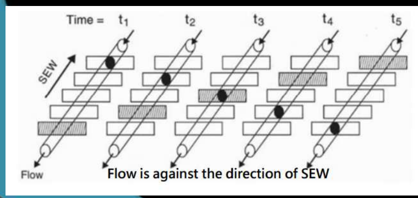
- The slice excitation wave (SEW) is the the direction of successive 90 deg. Excitation pulses.
- If flow is perpendicular to the slice…..



Increase the possibility of saturation →Weak FRE

Countercurrent Flow

- If flow is perpendicular to the slice…..



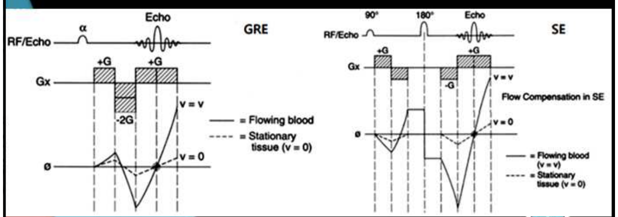
Reduce the possibility of saturation →Strong FRE

Flow compensation (FC)

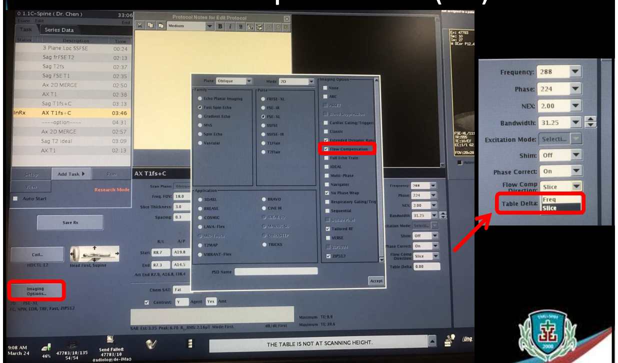
- Gradient moment nulling (GMN)：can minimizing flow artifacts
- Add extra gradient pulses to produce the even-echo rephasing effect on the first echo
  - 1st-order FC：Constant velocity
  - 2nd-order FC：Constant acceleration
  - 3rd-order FC：Turbulent (jerk) flow
- Can be applied to all the three coordinates
- FC lobes lengthen the TR/minimum TE

Flow compensation (FC)

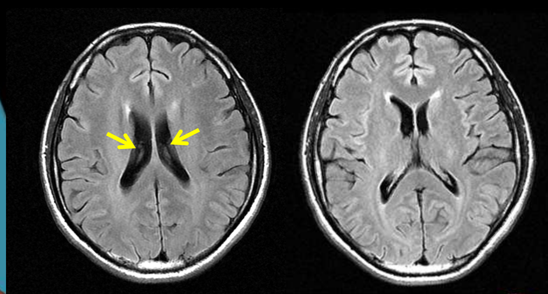
- 1 2 1 gradient lobes (constant velocity)



Flow compensation (FC)



Flow compensation (FC)



Unenhanced MRA

- Rely solely on flow effects (the movement of blood)
- Amplitude effects (through-plane flow)
  - Blood flowing into or out of a chosen slice has a different **longitudinal magnetization (Mz)** compared to stationary spins.
  - Depend on the duration of stay (time-of-flight; TOF) in the slice
- Phase effects (in-plane flow)
  - Blood flowing along the direction of a magnetic field gradient changes its **transverse magnetization (Mxy)** compared to stationary spins.

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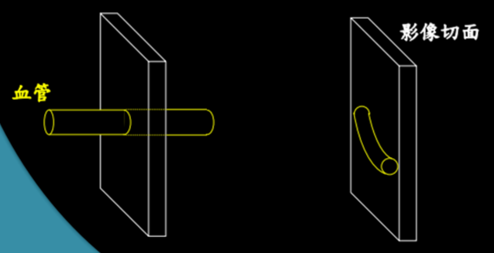
Through-plane vs. in-plane vessel



Through-plane (amplitude effect)

In-plane (phase effect)

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Flow-related signal enhancement

- The FRE occurs **both with SE and GRE** sequences.
- The competing **TOF loss in SE** tends to overbalance the FRE at higher flow velocities, leading to decreased flow signal.
- **TOF angiography**
  - GRE sequences
  - Bright-blood images

Factors affect FRE

- Through-plane or in-plane flow (slice direction)
- Ratio between flow rate and slice thickness
- TR
- Flip angle

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Slice direction

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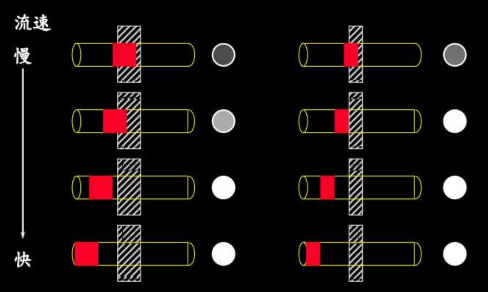
Flow rate and slice thickness



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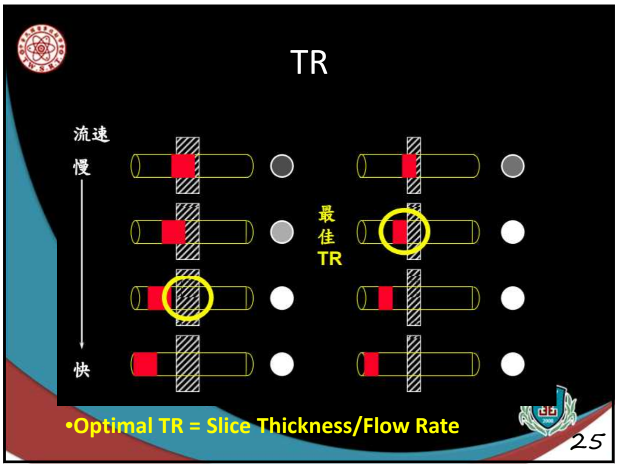
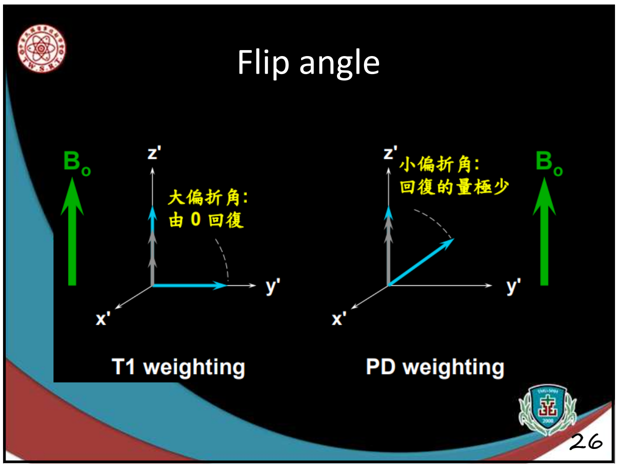
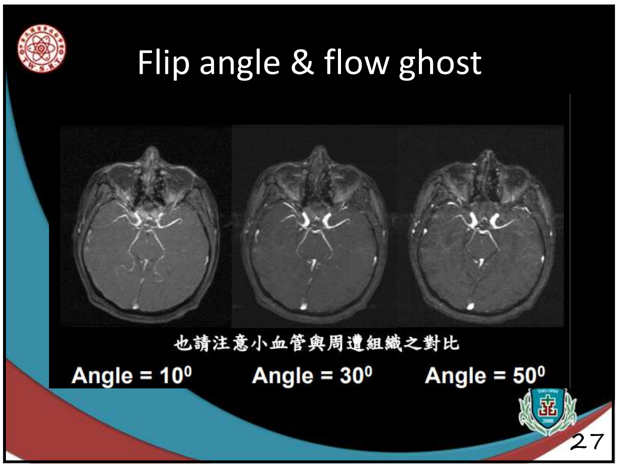
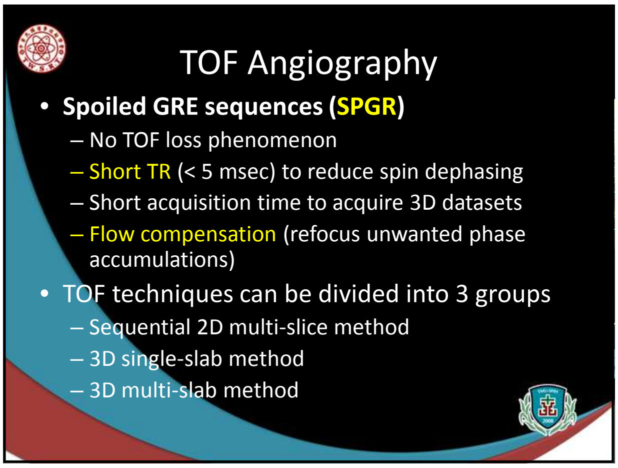
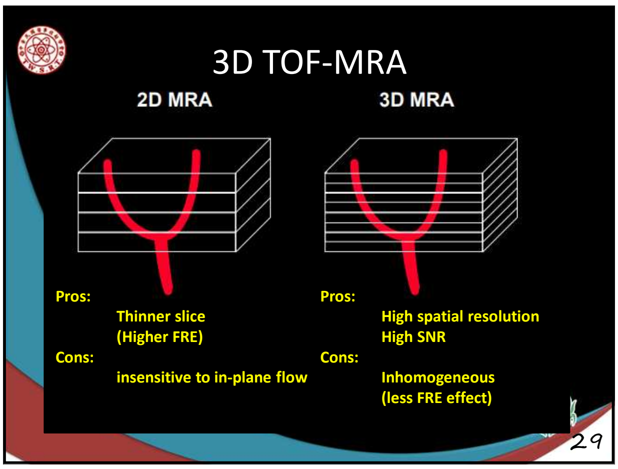
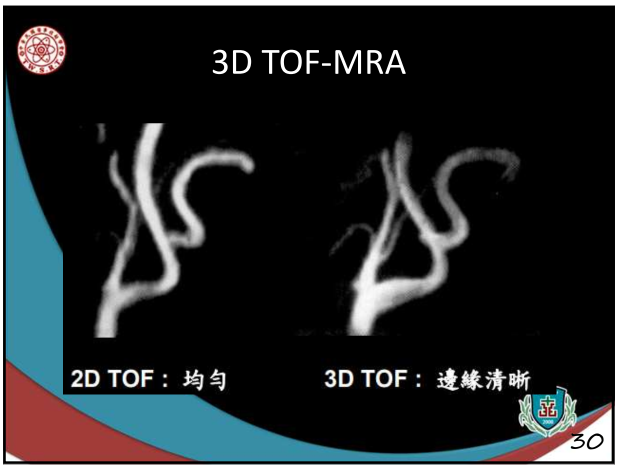

TR
流速
慢
快
最
佳
TR
•Optimal TR = Slice Thickness/Flow Rate
25


Flip angle
B_o
z'
y'
x'
大偏折角:
由 0 回復
小偏折角:
回復的量極少
T1 weighting
PD weighting
26

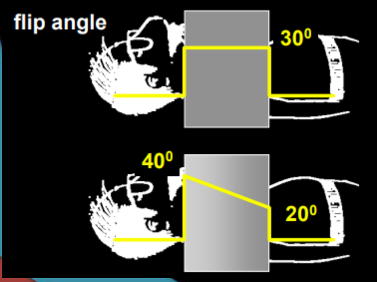

Flip angle & flow ghost
也請注意小血管與周遭組織之對比
Angle = 10^0
Angle = 30^0
Angle = 50^0
27


TOF Angiography
• Spoiled GRE sequences (SPGR)
– No TOF loss phenomenon
– Short TR (< 5 msec) to reduce spin dephasing
– Short acquisition time to acquire 3D datasets
– Flow compensation (refocus unwanted phase accumulations)
• TOF techniques can be divided into 3 groups
– Sequential 2D multi-slice method
– 3D single-slab method
– 3D multi-slab method


3D TOF-MRA
2D MRA
3D MRA
Pros:
Thinner slice
(Higher FRE)
Cons:
insensitive to in-plane flow
Pros:
High spatial resolution
High SNR
Cons:
Inhomogeneous
(less FRE effect)
29


3D TOF-MRA
2D TOF： 均匀
3D TOF： 邊緣清晰
30

TONE (Ramp Pulse) Technique



傳統 sinc RF 脈衝

TONE RF 脈衝

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TONE (Ramp Pulse) Technique

MRA: I -> S
MRV: A -> P

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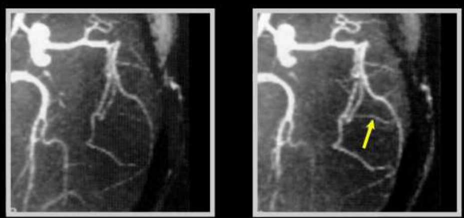
TONE (Ramp Pulse) Technique



傳統 RF pulse | TONE pulse

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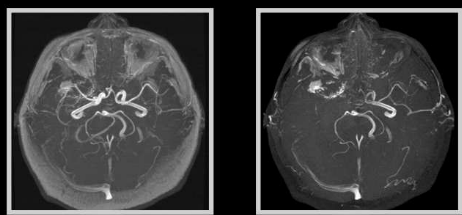
Background-blood contrast

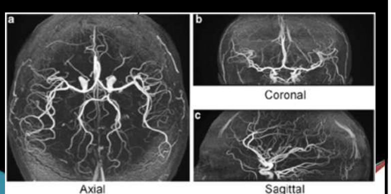
- Fat suppression



No Fat SAT | Fat SAT

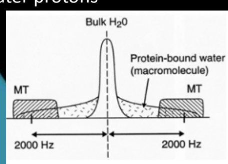
Background-blood contrast

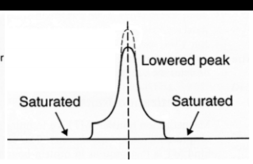
- Magnetization transfer contrast (MTC)
- MTC can further suppress background signal.
  - Reduction of gray and white matter signal by 15-40%
  - But not in blood
- Fat suppression



Magnetization Transfer (MT)

- To suppress protein-bound water
- Protons in protein-bound water exhibit a resonant frequency that is approximately 500 to 2500 Hz away from that of bulk water protons





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Magnetization Transfer (MT)

- MT is similar to spectral fat suppression techniques except that here, the off-resonant frequency is up to 2000 Hz as opposed to 220 Hz in the case of fat suppression.
- Used in time of flight (TOF) MRA to suppress the background brain tissue and enhance visualization of smaller vessels

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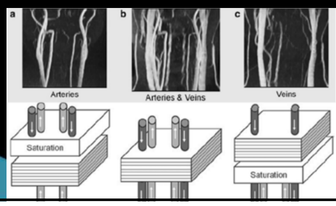
Spatial saturation pulse

- Superior saturation pulses are used suppress the signal from veins above the heart, and arteries below the heart
- Inferior saturation pulses are used to suppress the signal from arteries above the heart and veins below the heart

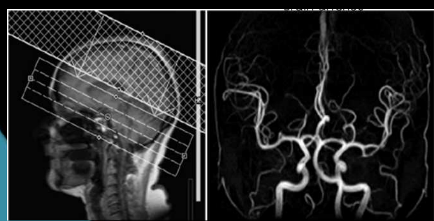


Sequential 2D technique

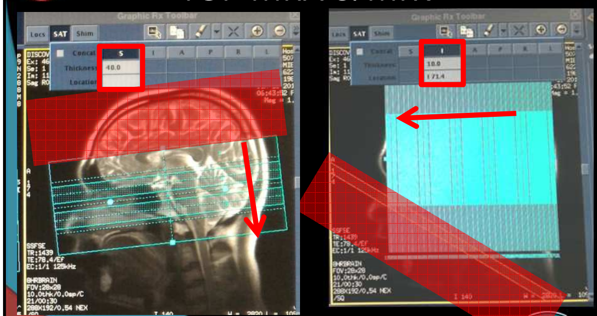
- Larger flip angle (30 deg.~70 deg.)
- Thicker slice thickness (2~3 mm) to achieve better SNR
- Best suited for imaging vessels that are straight and perpendicular to the slices.
  - Carotid arteries or vessels in the lower extremities.
- It is necessary to synchronize the acquisition of data to the cardiac cycle (**ECG gating**).

3D multi-slab method

- Presaturation slab above the imaging volume suppresses the signal of venous flow.

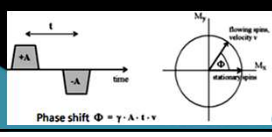


TOF-MRA & MRV



Phase effects

- Phase effects concern the transverse magnetization.
- Apply a pair of gradients with identical strength and duration but opposite **sign (bipolar flow-encoding gradient).**
- Stationary spins → zero net phase shift
- Flowing spins → a non-zeros phase shift

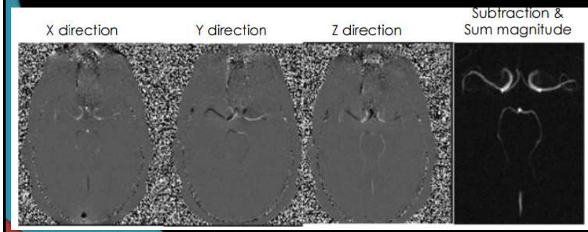


Phase contrast method

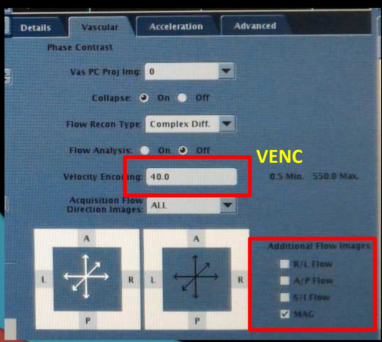
- A direct quantitative measure of the velocity of the flowing blood
- No restriction on image orientation (not dependent on inflow effects)
- **Velocity encoding (VENC)**
  - The velocities between –VENC and +VENC are encoded by the phase shifts between -180° and +180°.
  - The flow velocity exceeded the VENC value → aliasing
- **General velocity**
  - Arterial flow 40~60 cm/s
  - Venous flow 20~30 cm/s

Phase contrast method

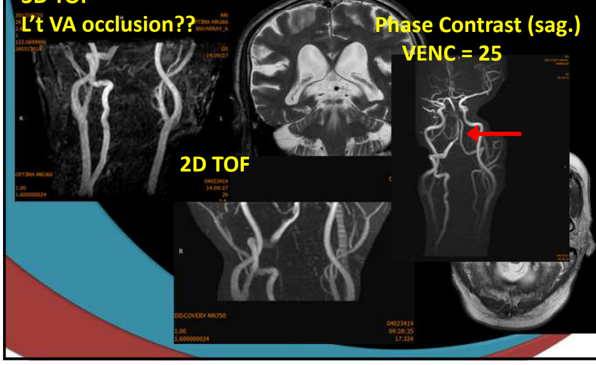
- Phase-encoded images



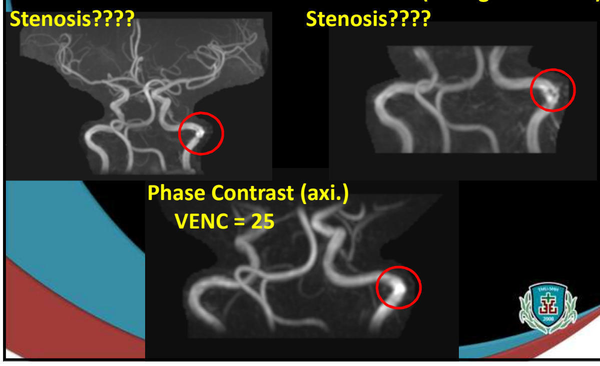
Phase contrast = Inhance 3D Velocity



Phase contrast (Example. 1)



Phase contrast (Example. 2)



TOF vs. phase contrast MRA

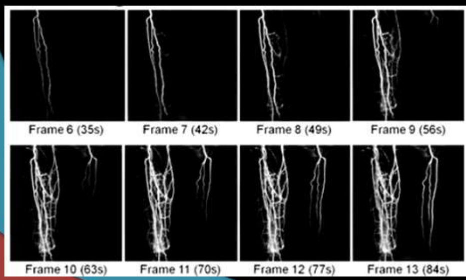
	TOF-MRA	Phase contrast MRA
Advantages	Simple to implement, robust	No saturation effects
	High spatial resolution	Excellent background suppression
	Shorter acquisition time (in 3D)	Enables quantitative flow measurement
Disadvantages	Reduced sensitivity to slow flow	Prior knowledge about flow rates
	Restrictions to size and orientation of the imaging volume	Very long acquisition times for 3D techniques
	Short T1 tissue may be mistaken for flowing blood	Susceptible to phase errors

Contrast-enhanced MRA

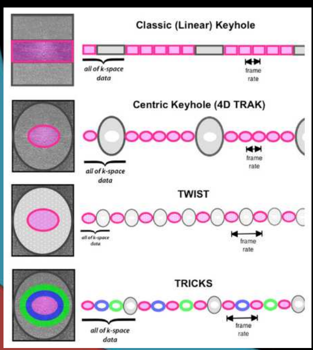
- Avoidance of blood signal saturation
- Better turbulent flow imaging
- Injection a contrast material intravenously (IV) to selectively shorten the T1 of the blood → brighter signal in T1W images.
- Gadolinium-chelate (Gd) contrast agents
  - Seven unpaired electrons → paramagnetic, shorten T1 and T2
  - Injection rate: 0.5~4.0 ml/s
  - Injection volume: 0.1~0.3 mmol/kg body weight, typically 20~40 ml
  - Computer-controlled power injector
  - Examine the patient's renal function before scanning!

Contrast-enhanced MRA

- 3D, RF-spoiled, fast gradient-echo imaging sequences → T1W images (FSPGR, FLASH)

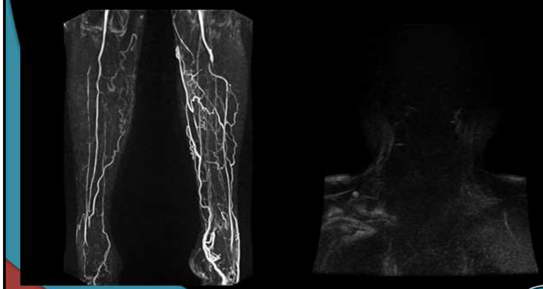


Time-Resolved MRA



- Keyhole imaging.
- Siemens: TWIST (Time-resolved angiography With Stochastic Trajectories)
- GE: TRICKS (Time-Resolved Imaging of Contrast KineticS)

Time-Resolved MRA



Susceptibility Weighted Imaging (SWI)

- SWI is an MR technique that utilizes the magnetic susceptibility differences
  - Visualize small veins in the brain
  - Microbleed
  - Sensitive to iron & calcification

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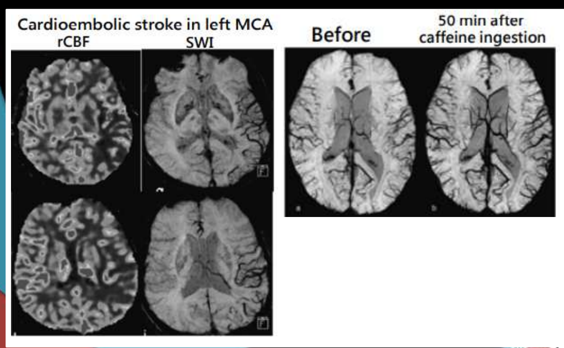
Susceptibility Weighted Imaging (SWI)



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History of SWI

- Originally proposed by Reichenbach et al. as "MRV" or "BOLD venographic imaging"
  – Small vessels in the human brain: MR venography with deoxyhemoglobin as an intrinsic contrast agent. Radiology, 1997.
- Haacke et al. 2004
  – Susceptibility weighted imaging (SWI)

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Magnetic Susceptibility

- When an object is placed in an external magnetic field H, magnetization is induced in the object.
- Magnetic susceptibility is the magnetic response of a material when it is placed in a magnetic field.
  – $M = \chi H$
  – χ = susceptibility (ppm)
  – M = induced magnetization
  – H = applied field
- If diamagnetic, $\chi < 0$
- If paramagnetic, like deoxygenated blood, $\chi > 0$

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Image Acquisition

- High-resolution 3D gradient echo imaging with 3-direction flow compensation
  – Long TR
  – Long TE (~40 ms at 1.5T, ~25 ms at 3.0T) to get T2* weighting
- Utilize both magnitude and phase images
- GE: SWAN, Siemens: SWI

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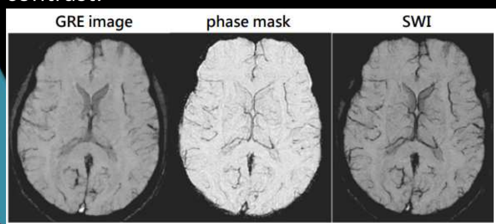
SWI vs. conventional GRE

- The use of the filtered phase to enhance contrast.



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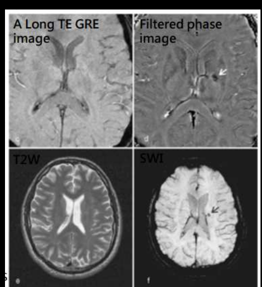
SWI vs. conventional GRE

- Tissues that have very low and uniform iron distribution will show a phase effect, but not a T2* effect.
  – Without phase dispersion -> no T2* effect.



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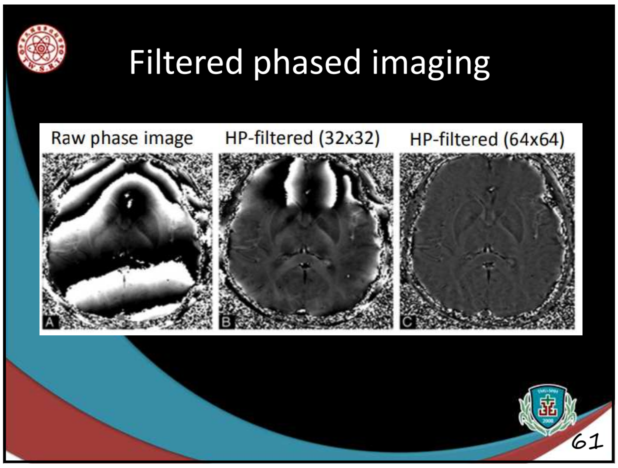
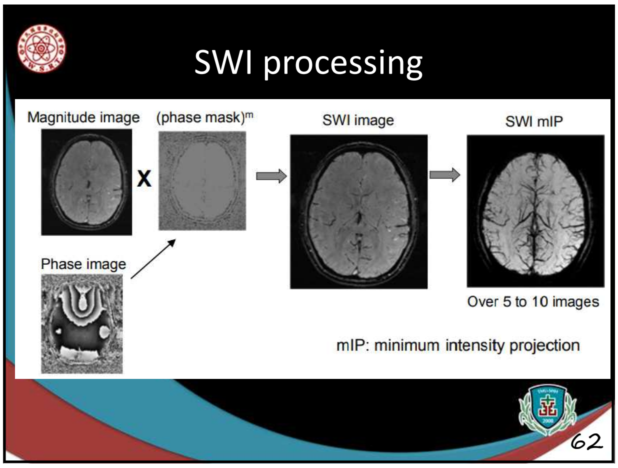
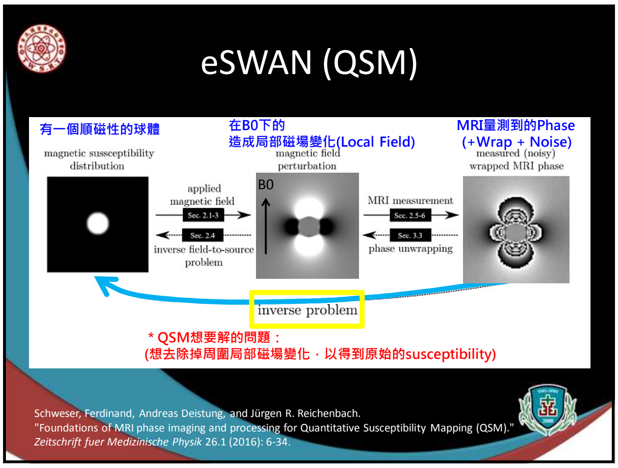
SWI processing

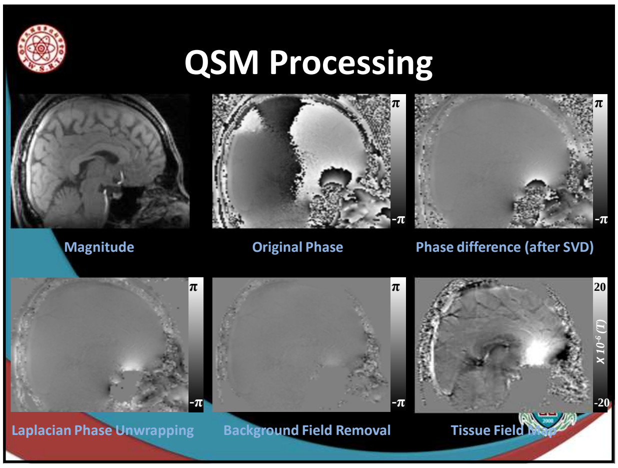
- Acquire high-resolution 3D GRE with flow compensation.
- Apply HPF to phase image to obtain SWI filtered phase data.
- Create phase mask depending on sign.
- Multiply phase mask by original magnitude image to obtain "merged SWI magnitude data."
- Perform a minimum intensity projection (mIP) over neighboring slices

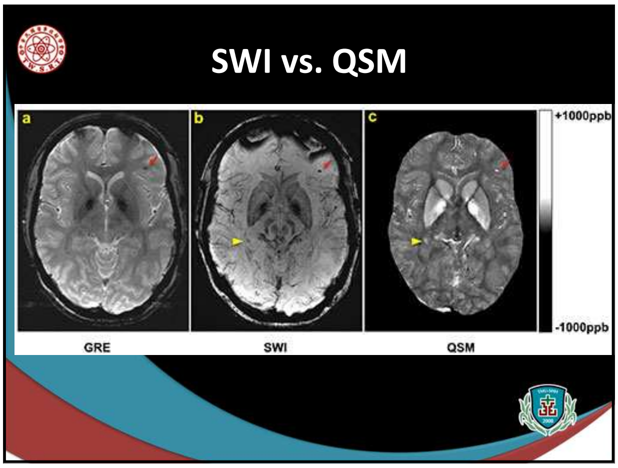
60


Filtered phased imaging
Raw phase image
HP-filtered (32x32)
HP-filtered (64x64)
A
B
C
61


SWI processing
Magnitude image
(phase mask)^m
X
SWI image
SWI mIP
Phase image
Over 5 to 10 images
mIP: minimum intensity projection
62


eSWAN (QSM)
有一個順磁性的球體
magnetic susceptibility distribution
在B0下的
造成局部磁場變化(Local Field)
magnetic field perturbation
MRI量測到的Phase
(+Wrap + Noise)
measured (noisy) wrapped MRI phase
applied magnetic field
Sec. 2.1-3
Sec. 2.4
inverse field-to-source problem
B0
MRI measurement
Sec. 2.5-6
Sec. 3.3
phase unwrapping
inverse problem
* QSM想要解的問題：
(想去除掉周圍局部磁場變化，以得到原始的susceptibility)
Schweser, Ferdinand, Andreas Deistung, and Jürgen R. Reichenbach.
"Foundations of MRI phase imaging and processing for Quantitative Susceptibility Mapping (QSM)."
Zeitschrift fuer Medizinische Physik 26.1 (2016): 6-34.


QSM Processing
Magnitude
Original Phase
Phase difference (after SVD)
Laplacian Phase Unwrapping
Background Field Removal
Tissue Field


SWI vs. QSM
a
b
c
+1000ppb
-1000ppb
GRE
SWI
QSM


Thank you...