Non-Cartesian MRI

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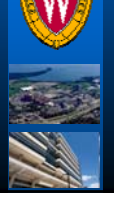
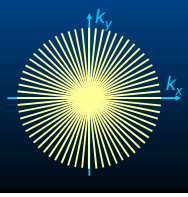
University of Wisconsin - Madison





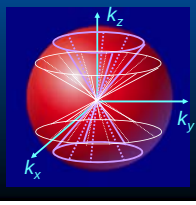
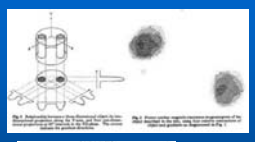
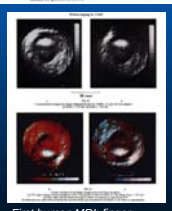


Image Encoding – The early days



First 'image': Lauterbur, 16 Mar 1973, Nature
2 water-filled capillary tubes

First human body MRI Chest: Damadian et al. 1977



First human MRI: finger: Mansfield et al, 1976

First human body MRI with line-scan: Mansfield et al, 1978

Spinwarp Imaging

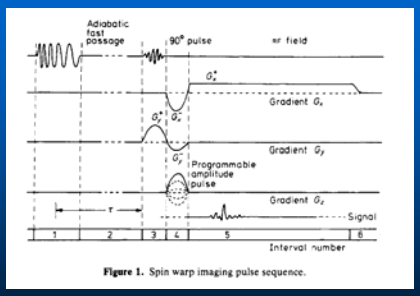
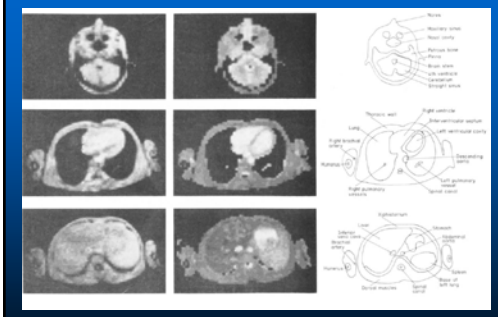


Figure 1. Spin warp imaging pulse sequence.

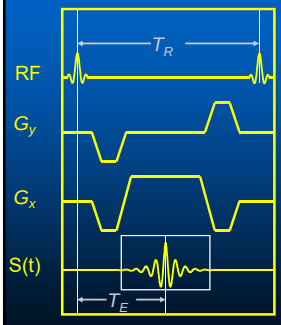
W. A. Edelstein et al.
Spin warp NMR imaging and applications to human whole-body imaging
Physics in Medicine and Biology 25: 751-6, 1980

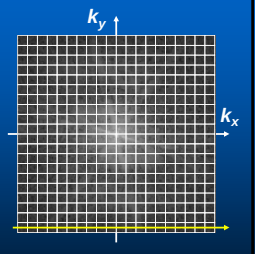
Spinwarp Imaging



W. A. Edelstein et al.,1980
0.04 T; Image matrix: 64x64
Scan time: 128 s

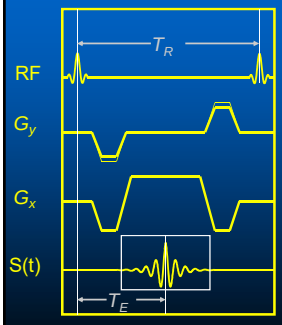
Data Acquisition – k-space

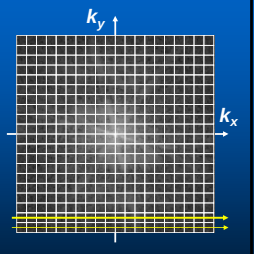




2D Gradient Echo

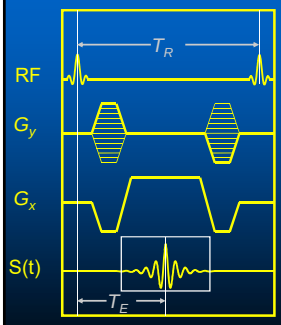
Data Acquisition – k-space

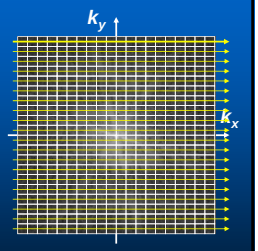




2D Gradient Echo

Data Acquisition – k-space

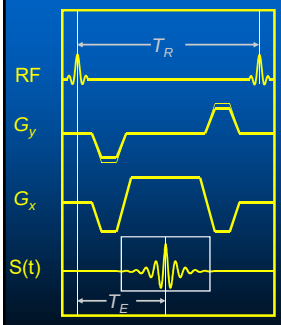


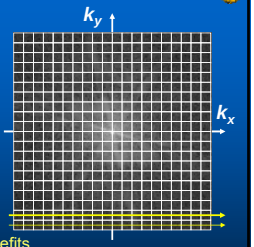


2D Gradient Echo

Data Acquisition – k-space

2D Gradient Echo

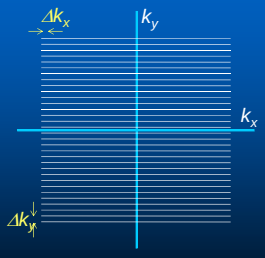




Benefits
- Robust to timing errors
  - Linear phase variation only
- Rapid reconstruction (Fast FT)
- Sampling theory is straightforward
  - equidistant sampling
- Parallel Imaging – well suited
- Imaging volume can be tailored

Limitations of 2D Spinwarp

FT (Spin-warp)

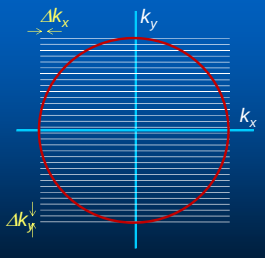


- One line of k-space per excitation
  - Many lines needed for high res image
- Limited Scan efficiency
  - Time data are acquired vs. TR

Nyquist: $N_y = N_{readout}$

Limitations of 2D Spinwarp

FT (Spin-warp)

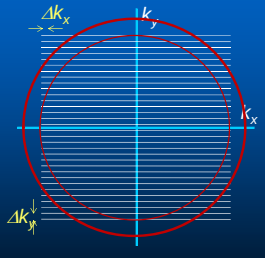


- One line of k-space per excitation
  - Many lines needed for high res image
- Limited Scan efficiency
  - Time data are acquired vs. TR
- K-space corners are sampled
  - 20% scan time penalty for limited value

Nyquist: $N_y = N_{readout}$

Limitations of 2D Spinwarp

FT (Spin-warp)



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  - Many lines needed for high res image
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- K-space corners are sampled
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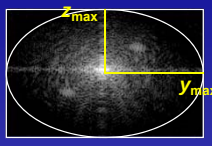
Nyquist: $N_y = N_{readout}$

K-space corners

- Goal: reduce scan time



$A_{rectangle} = 2y_{max} \times 2z_{max} = 4y_{max}\, z_{max}$

$A_{ellipse} = \pi\, y_{max}\, z_{max}$

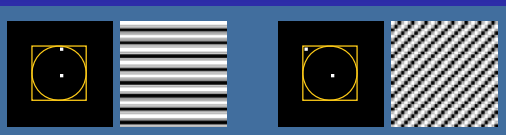
reduction: $\pi/4$ = 78.5 %

- Sampling of corners
  - Higher spatial frequencies in the diagonales
  - Zero-filling becomes beneficial
  - Watch for Fermi filter (switched off)

M. Bernstein, MRM (2001)

K-space corners

Without Zero-filling (16x16)



With zero-filling (32x32)

JOURNAL OF MAGNETIC RESONANCE IMAGING 14:270–280 (2001)

Original Research

Effect of Windowing and Zero-Filled Reconstruction of MRI Data on Spatial Resolution and Acquisition Strategy

Matt A. Bernstein, PhD,* Sean B. Fain, PhD, and Stephen J. Riederer, PhD

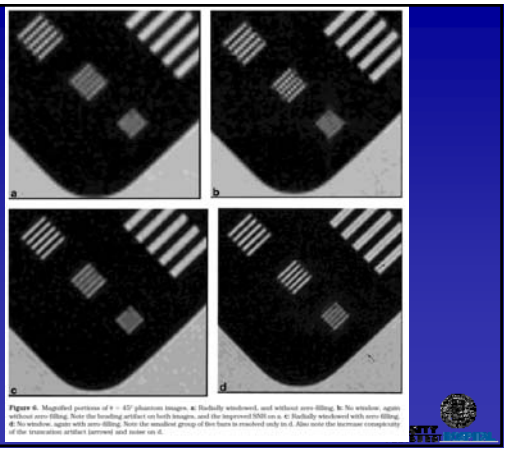
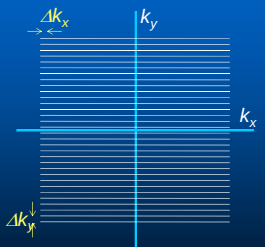


Figure 6. Magnified portions of $\pi/4$ phantom images. a: Radially windowed, and without zero-filling. b: No window, again without zero-filling. Note the bending artifact on both images, and the improved SNR on a. c: Radially windowed with zero-filling. d: No window, again with zero-filling. Note the smallest group of five bars is resolved only in d. Also note the increase conspicuity of the truncation artifact (arrows) and noise on d.

Limitations of 2D Spinwarp

FT (Spin-warp)



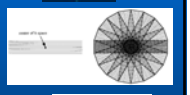
Nyquist: $N_y = N_{readout}$

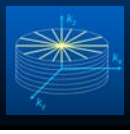
- One line of k-space per excitation
  - Many lines needed for high res image
- Limited Scan efficiency
  - Time data are acquired vs. TR
- K-space corners are sampled
  - 20% scan time penalty for limited value
- Poorly suited for undersampling
  - Foldover artefacts
  - Hard to ‚randomize' pattern (CS)
  - ....

Alternative Trajectories

Alternative Trajectories

- Radial Sampling
  - *Every k-space line traverses through center of k-sapce*
- Propeller
  - *Thin k-space slabs rotating around origin*
- Spiral imaging
  - *Long readouts in spiral trajectory*
- More excotic trajectories
  - *Cones, Rosettes, Shells, Hybrids, ...*



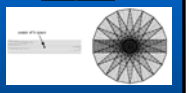


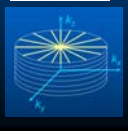
Design Goals

- Reduced total scan times / larger coverage
  - *Improved efficiency (higher readout/TR ratio)*
  - *Undersampling with tolerable artefacts*
- Short TE (ZTE, UTE)
  - *Reduced flow artefacts (small 1st moment around TE)*
  - *Reduced intravoxel dephasing and displacement*
- Dealing with Motion
  - *Reduced motion artfefacts, Self gating, Double Gating*
- ‚Random' sampling
  - *Incoherent sampling artfeatcs for compressed sensing*

Alternative Trajectories

Radial Sampling
Every k-space line traverses through center of k-sapce
Propeller
Thin k-space slabs rotating around origin
Spiral imaging
Long readouts in spiral trajectory
More excotic trajectories
Cones, Rosettes, Shells, Hybrids, ...



Design Goals

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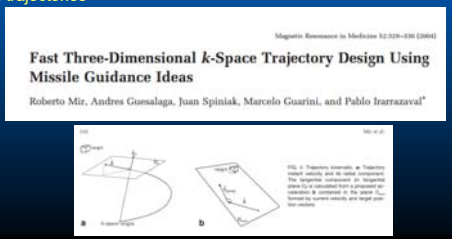
Incoherent sampling artfeatcs for compressed sensing

Alternative Trajectories

Radial Sampling
Every k-space line traverses through center of k-sapce
Propeller
Thin k-space slabs rotating around origin
Spiral imaging
Long readouts in spiral trajectory
More excotic trajectories


Magnetic Resonance in Medicine 52:329–336 (2004)

Fast Three-Dimensional *k*-Space Trajectory Design Using Missile Guidance Ideas

Roberto Mir, Andres Guesalaga, Juan Spiniak, Marcelo Guarini, and Pablo Irarrazaval*

Alternative Trajectories

Radial Sampling
Every k-space line traverses through center of k-sapce
Propeller
Thin k-space slabs rotating around origin
Spiral imaging
Long readouts in spiral trajectory
More excotic trajectories

Design goals

Design Limitations

Alternative Trajectories

Radial Sampling
Every k-space line traverses through center of k-sapce
Propeller
Thin k-space slabs rotating around origin
Spiral imaging
Long readouts in spiral trajectory
More excotic trajectories

Design goals
Faster scan times
Improved efficiency
Small Flow artefacts (small 1st moment around TE)
Short TE
Design Limitations
Gradient strength
Gradient slew rate
Artefacts from ‚jumps' in k-space

Spiral Imaging

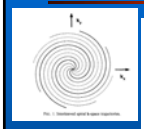
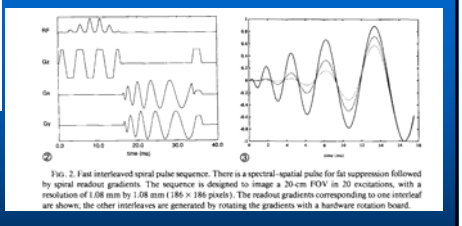
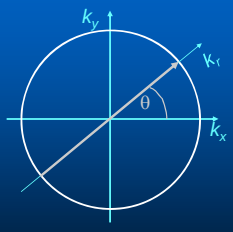



FIG. 2. Fast interleaved spiral pulse sequence. There is a spectral–spatial pulse for fat suppression followed by spiral readout gradients. The sequence is designed to image a 20-cm FOV in 20 excitations, with a resolution of 1.08 mm by 1.08 mm (186 × 186 pixels). The readout gradients corresponding to one interleaf are shown; the other interleaves are generated by rotating the gradients with a hardware rotation board.

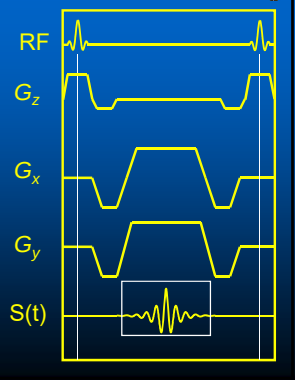
Prior GE System MR 4, 1.5T: *whole body (48 cm)* *zoom (35 cm)*
max gradient strength = *23 mT/m* *40 mT/m*
max gradient slew rate = *77 mT/m/s* *150 mT/m/s*

C. Meyer et al., MRM 1992

2D Radial Trajectory



$P(k_r,\theta)$



Radon Transform

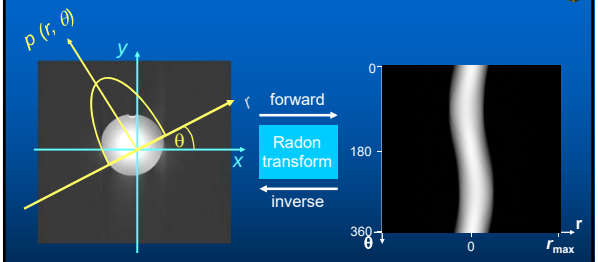


Image space

f(x,y)

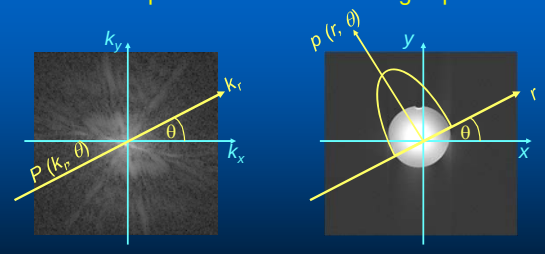
Sinogram

p (r, θ)

Projection Slice Theorem

Fourier space

Image space

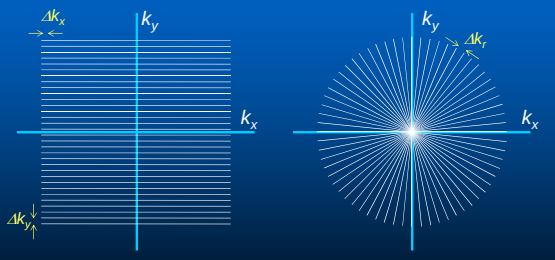


$p\,(r,\theta)$: projection = 1D $FT^{-1}\{P\,(k_r,\theta)\}$

2D Projection Reconstruction

FT (Spin-warp)

2D PR



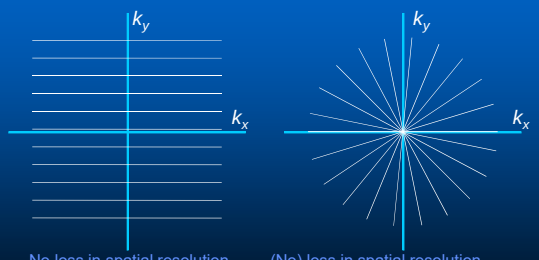
Nyquist: $N_y = N_{readout}$

Nyquist: $N_p = \pi/2\ N_{readout}$

Undersampling

FT (Spin-warp)

2D PR

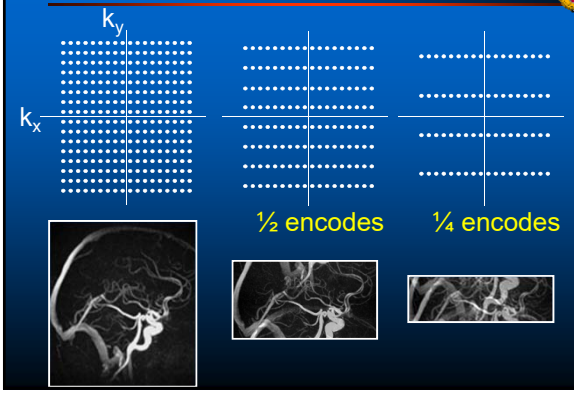


No loss in spatial resolution
Foldover Artifacts
Decrease in SNR

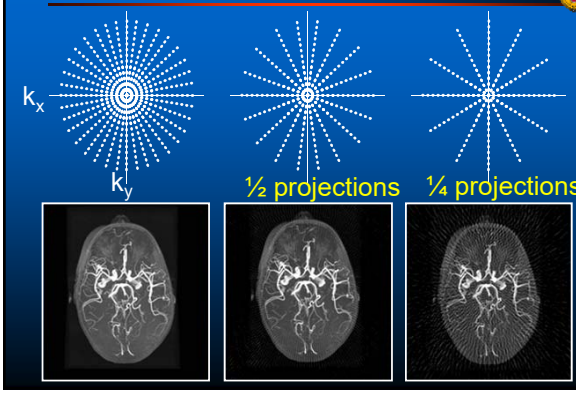
(No) loss in spatial resolution
Undersampling Artifacts
Decrease in SNR

D. Peters et al., ISMRM 1998

Undersampling (Spinwarp)



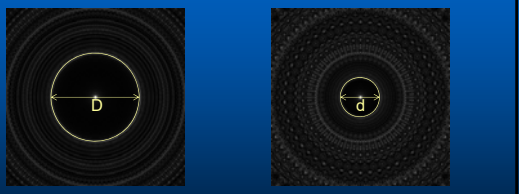
Radial Sampling Density: $1/k_r$



rFOV concepts: PSF

Fully sampled
$N_p = \pi/2 \times N_r$

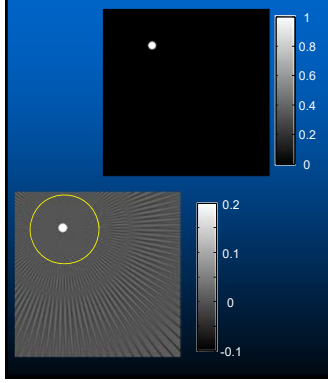
Undersampled
$N_p = \pi/2 \times N_r / 2.5$



$$\frac{D}{d} = \frac{2 \times N_p}{\pi \times N_r}$$

K. Scheffler and J. Hennig, MRM, 1998

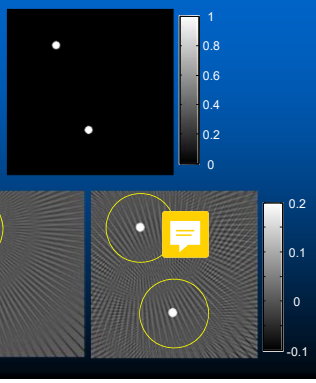
Streak Artifacts



Original:
256 x 256 matrix

PR:
$N_{readout}$ = 256
N_p = 64
radial us = 6.3
cartesian us = 4

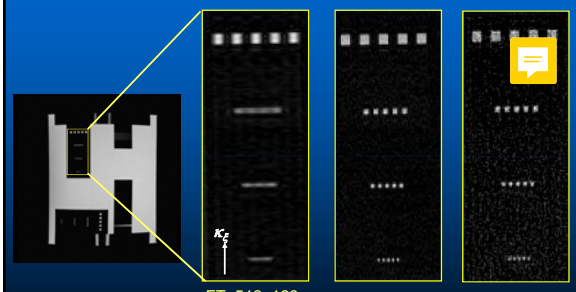
Streak Artifacts



Original:
256 x 256 matrix

PR:
$N_{readout}$ = 256
N_p = 64
radial us = 6.3
cartesian us = 4

Resolution in FT und PR



FT: 512×128

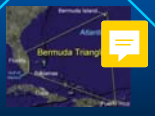
FT: 512x512

PR: N_p=128
$N_{readout}$ = 512

Courtesy of Dana Peters, Harvard

MRI Tradeoffs

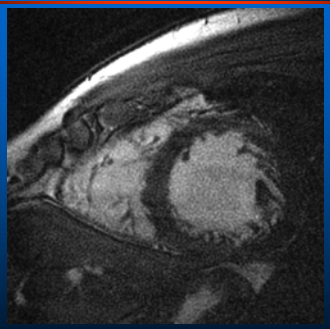
Spatial resolution ↑
Imaging Time ↑
SNR ↓



Imaging Time ↓
Spatial Resolution ↓
SNR ↓

SNR ↑
Spatial Resolution ↓
Imaging Time ↓

True FISP Cine



Nr = 512
temp resolution = 40 ms
scan time = 23 s

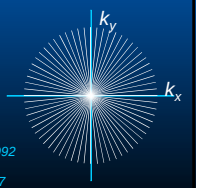
Courtesy of Dana Peters, Harvard

Radial Acquistions

Rapid imaging with undersampling
- Interventional MR — K. Scheffler, MRM 1998; S. Weiss, MRI 1999
- Contrast-enhanced MRA — D. Peters, MRM 2000; A. Larson, MRM 2001
- bSSFP cardiac cine — T. Schaeffter, MRM 2001; Shankaranarayanan, Radiology 2001

Robust to motion artifacts
- Motion is 'averaged' — G. Glover, MRM 1992
- No ghosting artifacts
- Navigatorlike information — N. Gai, Med Phys 1996

UTE (Ultrashort TE imaging)
- Minimal TE times for short T2 — G.Glover, JMRI 1992
- CSI imaging — J. Du, MRM 2007



Non-Cartesian – 3D Trajectories

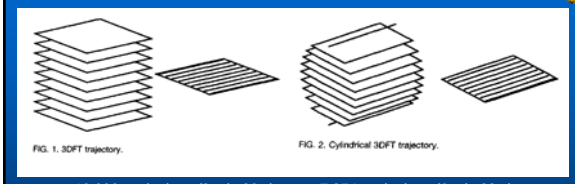


FIG. 1. 3DFT trajectory.

FIG. 2. Cylindrical 3DFT trajectory.

10,000 excitations (8 min 20 s)
Prep time = 1.43 ms
TE = 2.3 ms
Significant first moment at TE
Flow Comp would be best

7,854 excitations (6 min 33 s)
Prep time = 1.43 ms
TE = 2.3 ms
Significant first moment at TE
Flow Comp would be best

100x100x100 pixels
TR = 50 ms
Readout time = 14.3 ms

From P. Irarrazabal and D. Nishimura, MRM 1995

Non-Cartesian – 3D Trajectories

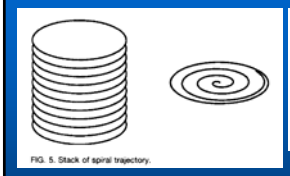


FIG. 5. Stack of spiral trajectory.

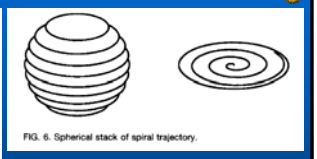


FIG. 6. Spherical stack of spiral trajectory.

900 excitations (45 s)
Prep time = 1.43 ms
TE = 0 ms
Low first moment at TE
but slice encoding

628 excitations (31 s)
Prep time = 1.43 ms
TE = 0 ms
Low first moment at TE
but slice encoding

From P. Irarrazabal and D. Nishimura, MRM 1995

Non-Cartesian – 3D Trajectories

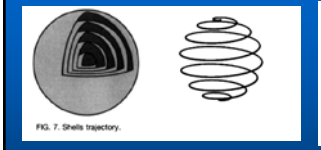


FIG. 7. Shells trajectory.

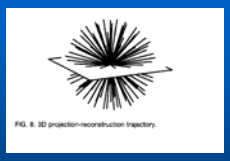


FIG. 8. 3D projection-reconstruction trajectory.

629 excitations (31 s)
Prep time = 1.43 ms
TE = 0 ms
Low first moment at TE
but 'slice' encoding (pole)

31,416 excitations (26 min 10 s)
Prep time = 0 ms
TE = 0 ms
zero first moment at TE
excellent flow properties

From P. Irarrazabal and D. Nishimura, MRM 1995

Non-Cartesian – 3D Trajectories

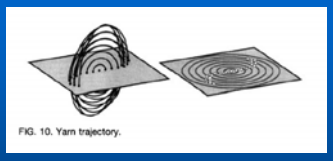


FIG. 10. Yarn trajectory.

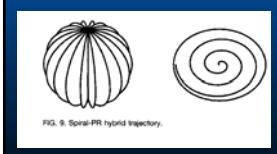


FIG. 9. Spiral-PR hybrid trajectory.

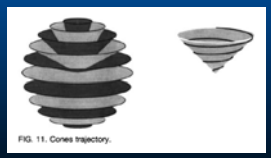
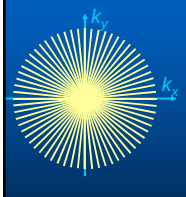


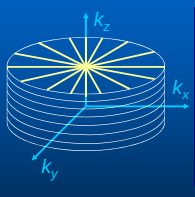
FIG. 11. Cones trajectory.

From P. Irarrazabal and D. Nishimura, MRM 1995

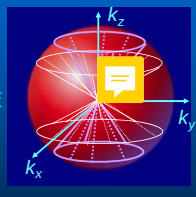
Radial Sampling



2D



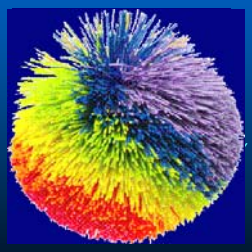
3D hybrid – Stack of stars



3D radial - VIPR

VIPR

Vastly Undersampled Isotropic PRojection Imaging



3D Radial acquisition

Nyquist: $N_p = \pi/2\ N^2_{readout}$

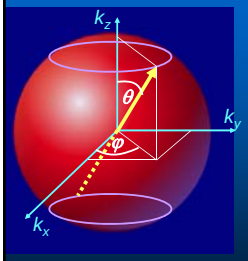
Large FOV

Isotropic spatial resolution

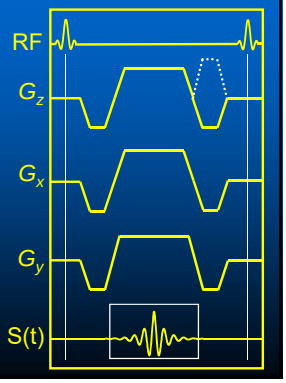
Undersampling

A Barger et. al, MRM 2002

VIPR

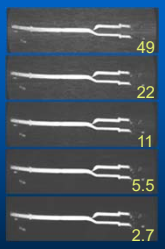


Vastly undersampled Isotropic Projection Reconstruction



A Barger et. al, MRM 2002

Relative VIPR Noise vs Speedup



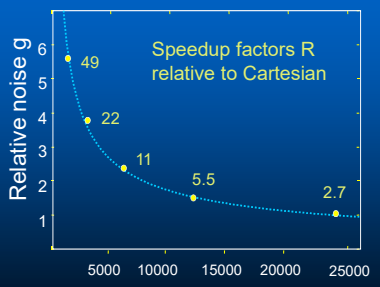


Image volume:
256 x 256 x 256

Courtesy of A.V. Barger

PC VIPR – Cranial

Normal Volunteer

PC VIPR Parameters

- Dual Echo
- FOV: 20 x 20 x 20 cm
- Res: 0.6 x 0.6 x 0.6 mm
- 9000 Projections (36x)
- TR=15.9
- Bandwidth = 31.25
- VENC = 50 cm/s
- 5:07 min Scan Time

Same Cartesian PC

- 48+ min Exam (Partial)

Same TOF

- 24+ min Exam (Partial)

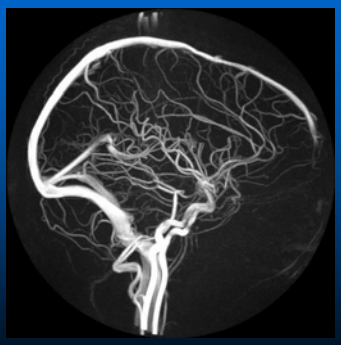


Image Reconstruction

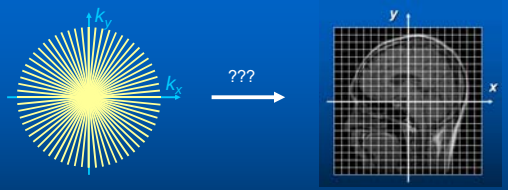


Image Reconstruction

Projection Reconstruction

- Matrix inversion
- Iterative Reconstruction
- Filtered back projection (Radon)
  - (magnitude or complex)

Gridding (Regridding)

- Weight data according to sampling density
- grid polar (spherical) samples onto Cartesian grid
- 2D or 3D inverse Fourier Transform
- Divide by FT of the interpolation kernel to remove image apodization from convolution

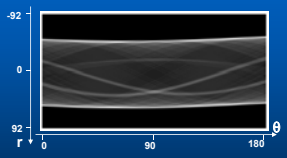
- Non-uniform FFT (NUFFT)

Backprojection

Image

f(x,y)

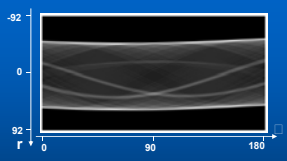
Sinogram

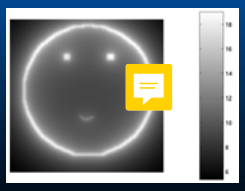


p (r, θ)

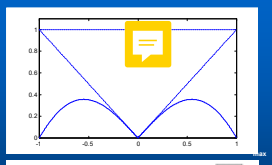
Backprojection

Image (x,y)

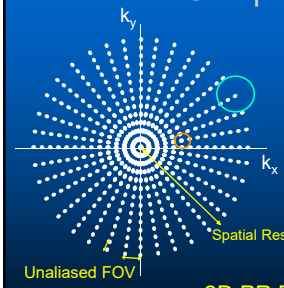


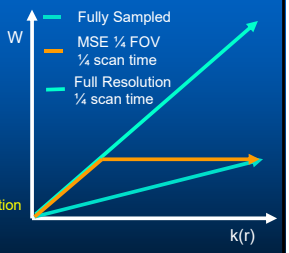


Filtered Backprojection



Determining the Density Compensation

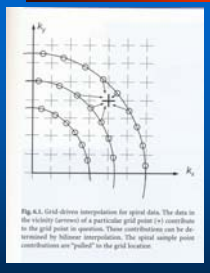




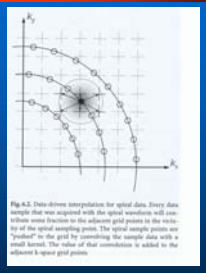
2D PR Example

J. G. Pipe, MRM, 43:867-875, 200

Gridding

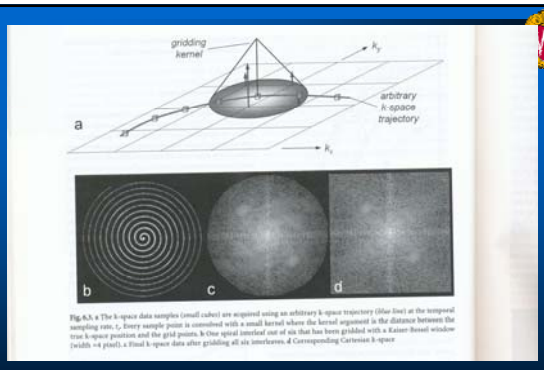


Grid-driven interpolation



Data-driven interpolation

From R. Bammer et al. 'Parallel-Imaging Reconstruction of Arbitrary k-Space Sampling Data' in S. Schoenberg, O. Dietrich, and M. Reiser – 'Parallel Imaging in Clinical MR Applications

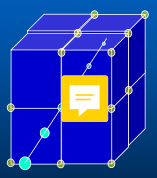


Example: Spiral (6 interleaves)

From R. Bammer et al. 'Parallel-Imaging Reconstruction of Arbitrary k-Space Sampling Data' in S. Schoenberg, O. Dietrich, and M. Reiser – 'Parallel Imaging in Clinical MR Applications

Gridding

1. Weight data according to sampling density
   As simple as ρ for 2D or ρ^2 for 3D
2. Grid onto Cartesian Coordinates
   Kaiser-Bessel or 2D/3D separable triangular convolution kernel
3. 2D or 3D inverse Fourier Transform
4. Divide by FT of the interpolation kernel to remove image apodization from convolution



-> much faster than Radon in 3D

J. Jackson et al., IEEE TMI 1991
M. Lauzon and B. Rutt, MRM 1998
J. Pipe, MRM 2000

Ultra-Short TE (UTE) Imaging

Tissues with short T2

T2 of protons in water
Usually clinical range

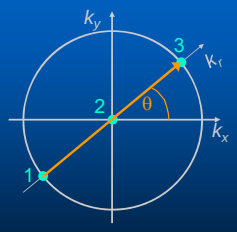
T2 of protons in proteins and in water tightly bound to protein
10 μs range

Standard MR sequences
Detectable T2s: 10 ms, modern systems: 2 ms

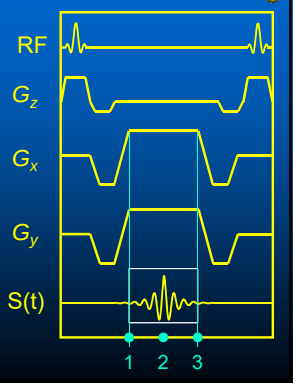
Table 2
Tissues With a Majority of Short T_2 Components

Meninges (dura)	Falx	Tentorium
Membranes	Capsules	Bands
Retinaculi	Septae	Fascae
Sheaths	Nails	Hair
Aponeuroses	Tendons	Ligaments
Menisci	Labrii	Periosteum
Bone	Dentine	Enamel

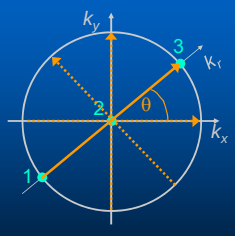
2D Radial Trajectory



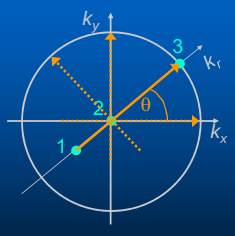
$P(k_r,\theta)$



2D Radial Trajectory

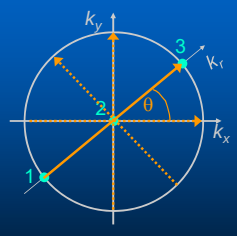


Full echo radial

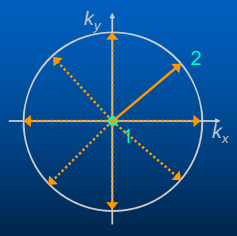


Fractional echo radial

2D Radial Trajectory



Full echo radial



UTE radial

2D UTE Sequence

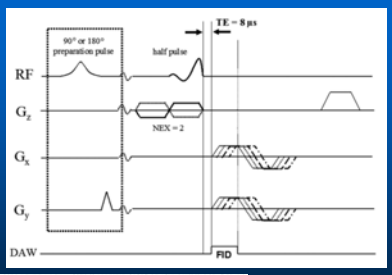


FIG. 1. UTESI sequence employs half-pulse excitation with radial ramp sampling at variable TEs, with a minimal achievable TE of 8 μs. The data acquisition window (DAW) shifts with the read pulses. This sequence is preceded by either a long duration 90° preparation pulse for saturation or 180° preparation pulse for inversion of the long T_2* magnetization.

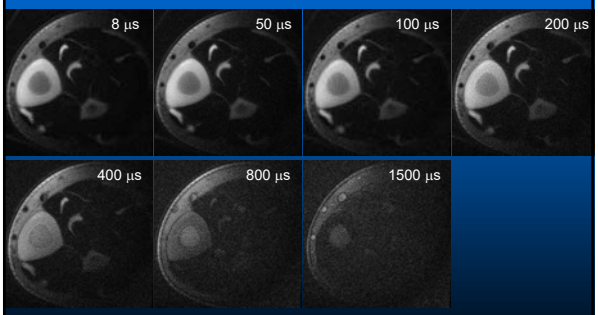
Jiang Du et al. MRM, 2007

Lewa

TE = 8 μs

* Phantom graciously provided by Peter Nishimura, Age 6

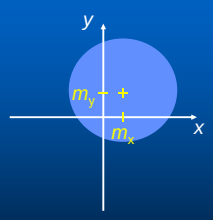
2D UTE Imaging – Half Pulse

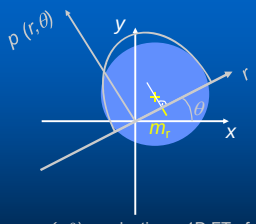


Courtesy of Jiang Du, UCSD

Motion Correction

Center of Mass (COM)





$p(r,\theta)$: projection = 1D FT of echo

m_x and m_y : first moments

$m_r(\theta) = \Sigma\, p(r,\theta)\, r$

$m_r(\theta) = m_x \cos\theta + m_y \sin\theta$

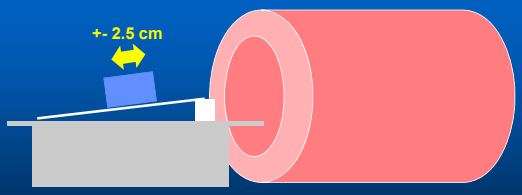
G. Glover, MRM 1995

Correction for Translation

1. Define a reference COM
   Origin of coordinate system, average position, or subset of reference projections
2. Estimate translational motion for each projection $m_{motion} = m_{r,p} - m_{r,m}$
3. Invert the effects of motion in *k*-space
   Fourier Shift-Theorem:
   $S'(k_r, \boldsymbol{\theta}, \boldsymbol{\varphi}) = S(k_r, \boldsymbol{\theta}, \boldsymbol{\varphi}) \exp(-i2\pi k_r\, m_{motion})$

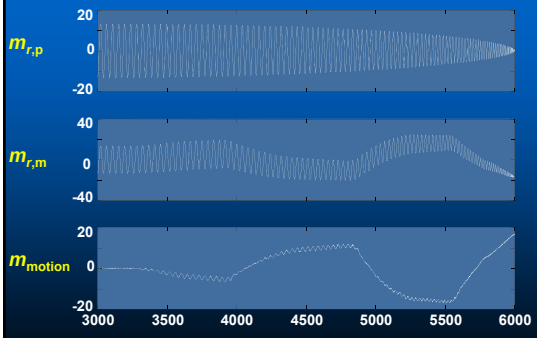
O. Wieben, ISMRM 2001

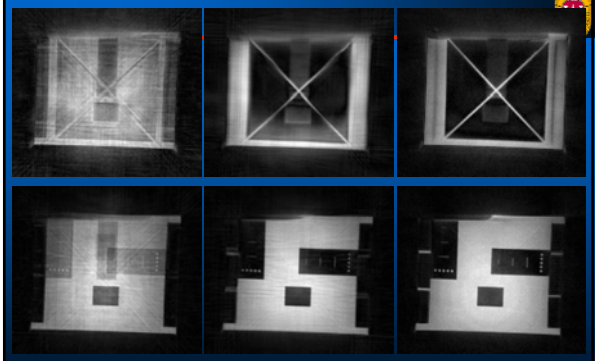
Phantom Experiment



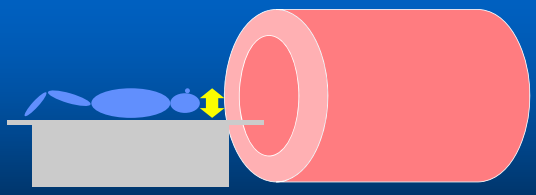
Amplitude: +-2.5 cm
Frequency: mimick breathing
Orientation: mainly S/I (partially A/P)
Manual control

COM – Interleave 2

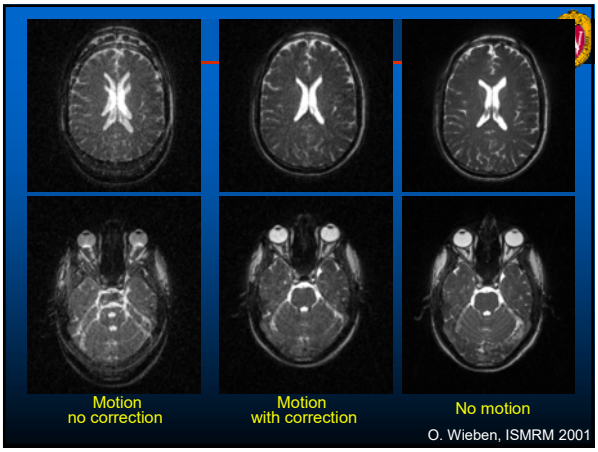




In Vivo



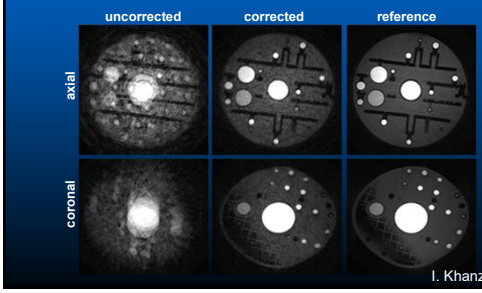
Amplitude: 1 cm
Frequency: once only
Orientation: A/P (in-plane)



Motion Correction

Phantom study

- Continuous translational motion applied to spherical phantom on table rocker of a clinical 3T scanner (GE Healthcare, Waukesha, WI).
- Sinusoidal motion, with an amplitude of ±20 mm and a maximum velocity of 15 mm/s during VIPR SPGR acquisition.



Trajectory Errors

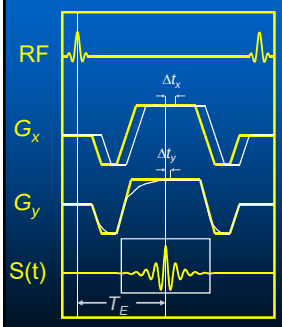
The desired and the actual trajectory differ

- Gradient anisotropy
  - different system delays for x, y, and z gradients
  - -> echo shifts: first order trajectory errors

B. Aldefeld, MRM 1998
S. Reeder, MRM 1999

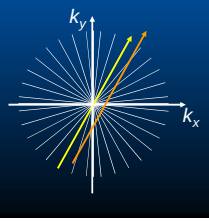
- Eddy currents
  - Short TRs and steep gradients
    - -> pre-emphasis corrections are for longer time constants
    - -> first and higher order trajectory errors

Data Acquisition - 2D GRE



Off-center echoes

- Leads to phase twist in image space
- FT imaging: magnitude is not affected
- Problems in oblique EPI and non-cartesian sampling



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Solutions

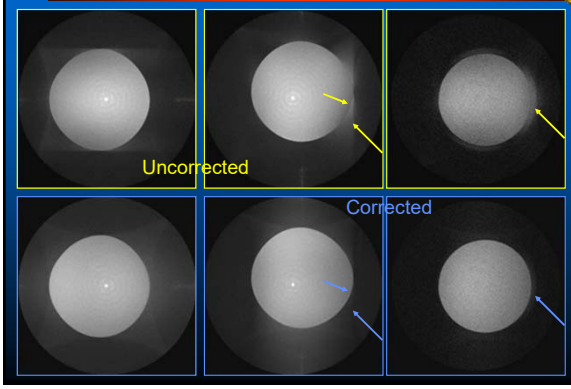
- Correct first order trajectory errors
  - Measure and analyze antiparallel echoes
  - Correct
    - During gridding
    - During acquisition: modify gradient area

D. Peters, ISMRM 2003
O. Wieben, ISMRM 2003

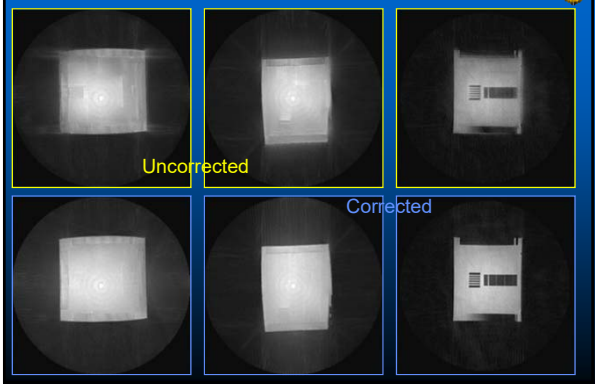
- Measure actual trajectory

Duyn, JMR 1998
B. Dale, ISMRM 2003

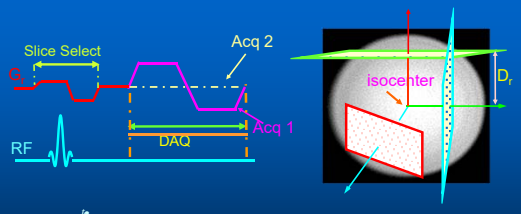
Results: Ball Phantom



Results: High Resolution Phantom



K-space Trajectory Measurement

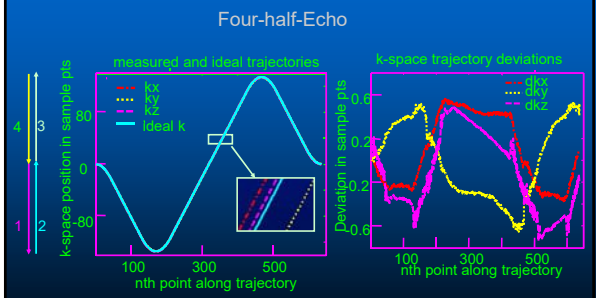


$$\phi_r(t) = \int_0^t \gamma G_r(t) D_r dt = D_r k_r(t) \qquad K_r = (\phi_{r1}(t) - \phi_{r2})/D_r$$

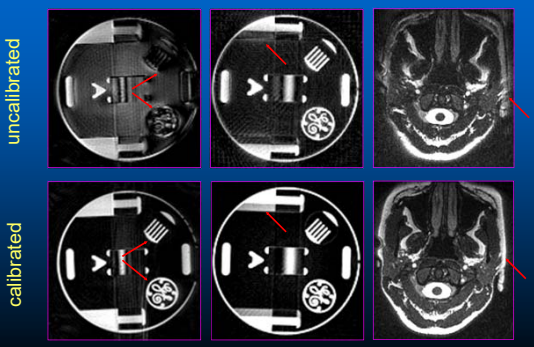
- Slice thickness: 1mm~1.5 mm
- Off-center distance (D_r): 3 cm
- Can be performed during the imaging scan

Duyn, et al. JMR 132 150, 1998 *B. Dale, 10th ISMRM 2003, 2334*

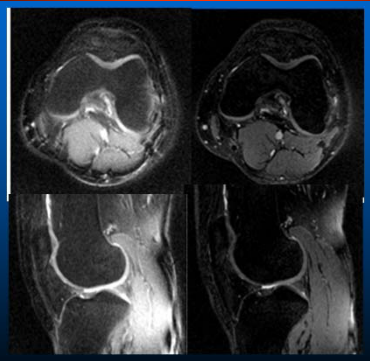
Measured K-space Trajectories



Gridding based on actual k-space location



Gridding – Knee scan



Uncalibrated Calibrated

Non-Cartesian PC MRI

Benefits
- + Accelerated Imaging
- + High spatial resolution (isotropic)
- + Favorable Motion artefacts
- + UTE Flow Imaging
- + (Large Coverage)

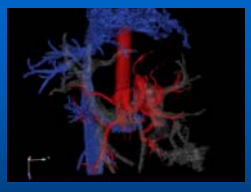
Non-standard Acquisition
Reconstruction is more complex
- Gridding
- -> Longer recon times
- Parallel Imaging Recon is challenging

Sensitivity to Artefacts
- Trajectory errors (calibration required)
- Off-resonance

Artefacts
- Streaks/Blurring unfamiliar to clinicians
- Displacement in readout (changing dir)

Summary

Cartesian
- Robust image quality
- Proven
- Fast reconstruction
  - works well with parallel imaging
- Tailor Image Volume



Non-Cartesian
- Can overcome some of current limitations
  - Scan time / spatial resolution / coverage
  - Short TE
  - ‚Random Sampling'
- Acquisition more complex
  - calibrations
- Recon more complex
  - Feasible (see Propeller)