MR Spectroscopic Imaging

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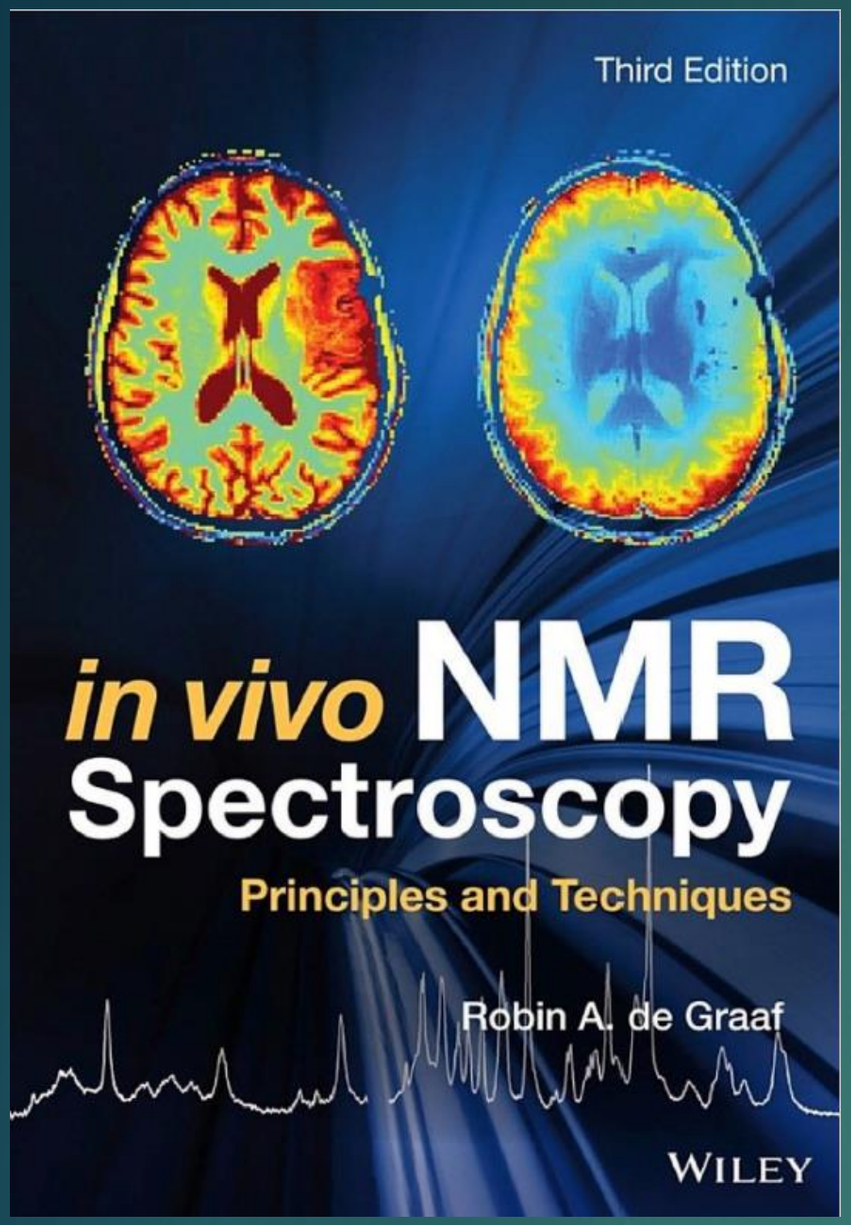
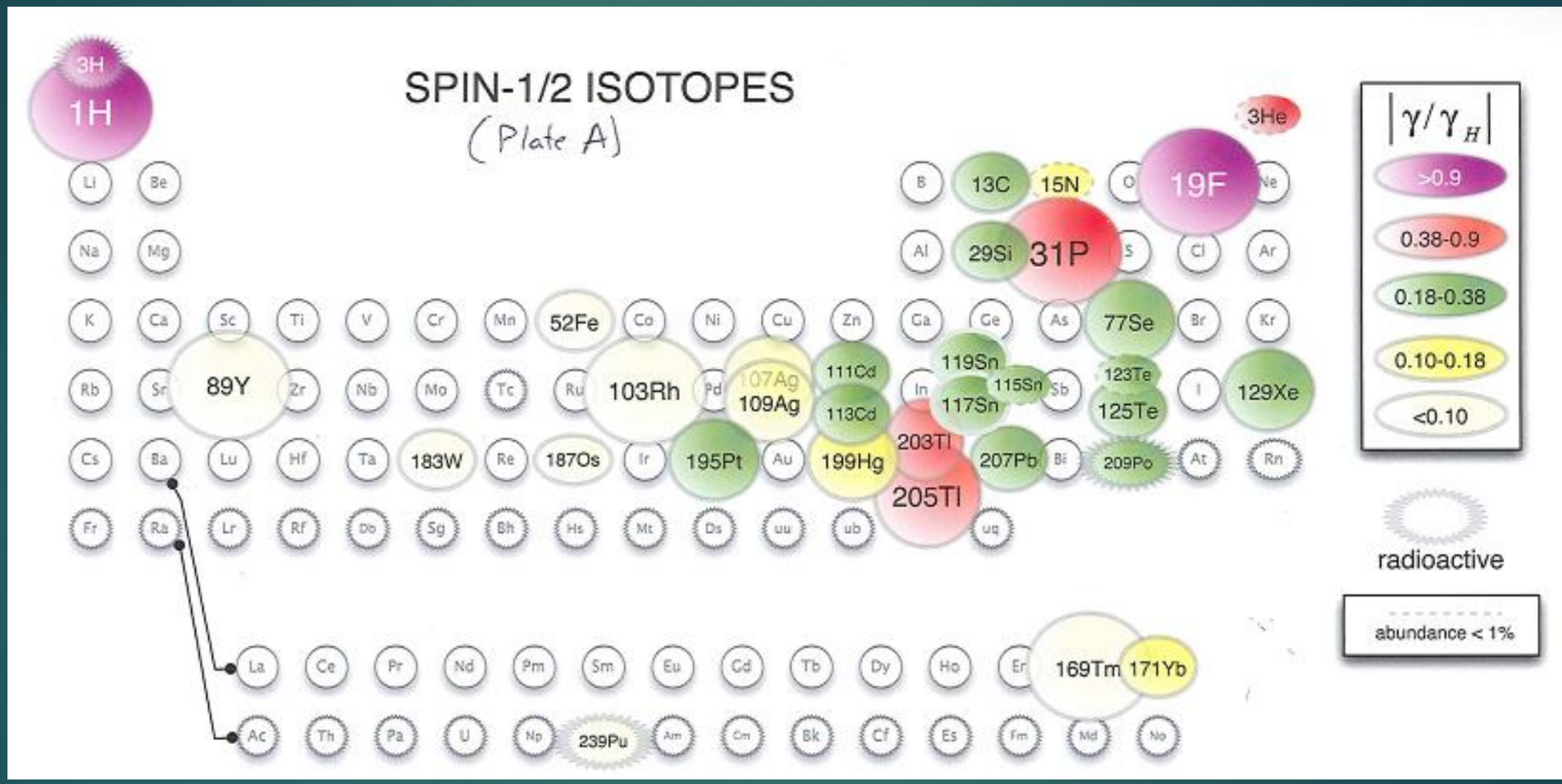


Contents

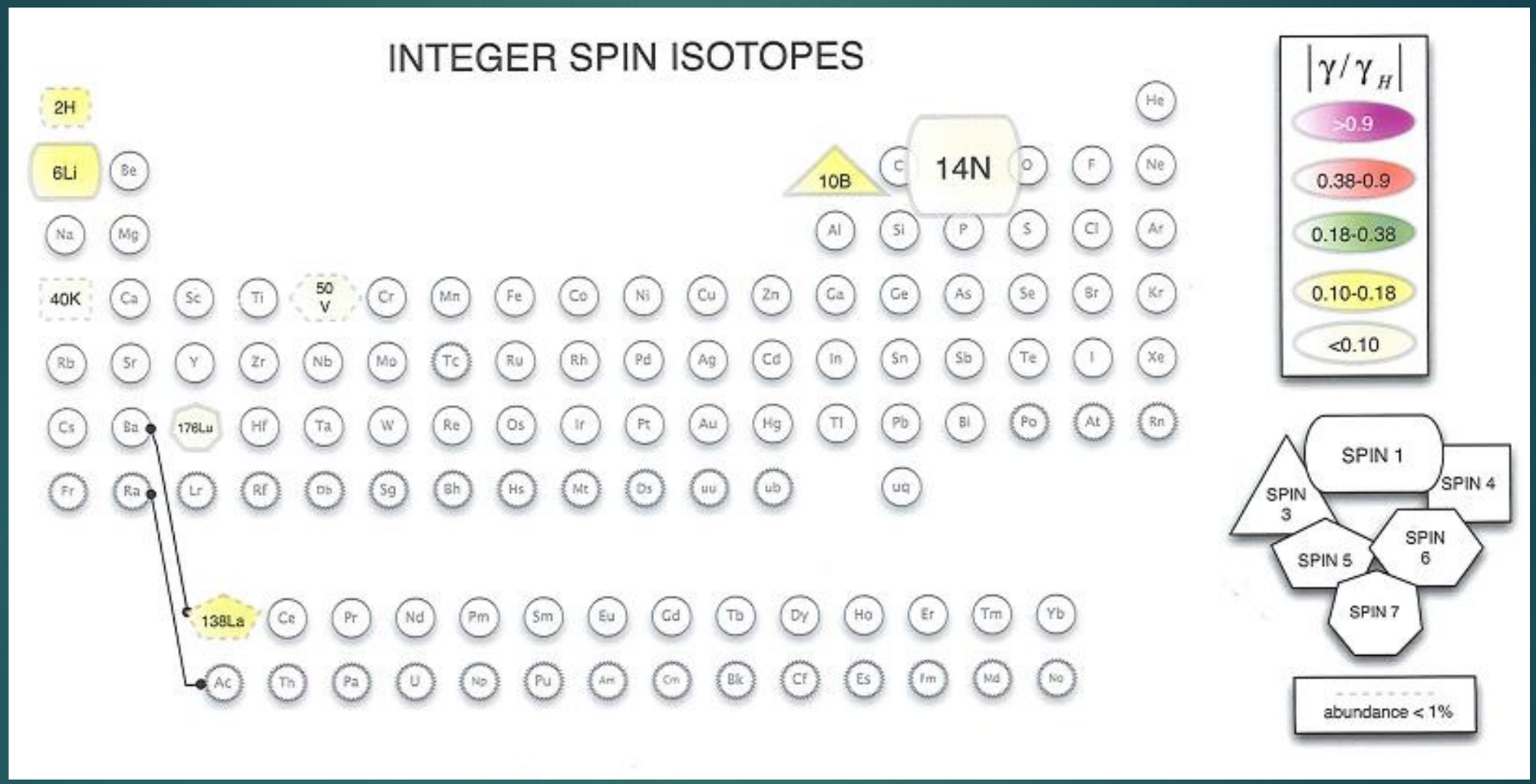
Important Nuclei for Biomedical MR

Nucleus	Spin	γ, MHz/T	Natural Abundance	Relative Sensitivity
^{1}H	1/2	42.576	99.985	100
^{2}H	1	6.536	0.015	0.96
^{3}He	1/2	32.433	.00013	44
^{13}C	1/2	10.705	1.108	1.6
^{17}O	3/2	5.772	0.037	2.9
^{19}F	1/2	40.055	100	83.4
^{23}Na	3/2	11.262	100	9.3
^{31}P	1/2	17.236	100	6.6
^{39}K	3/2	1.987	93.08	.05

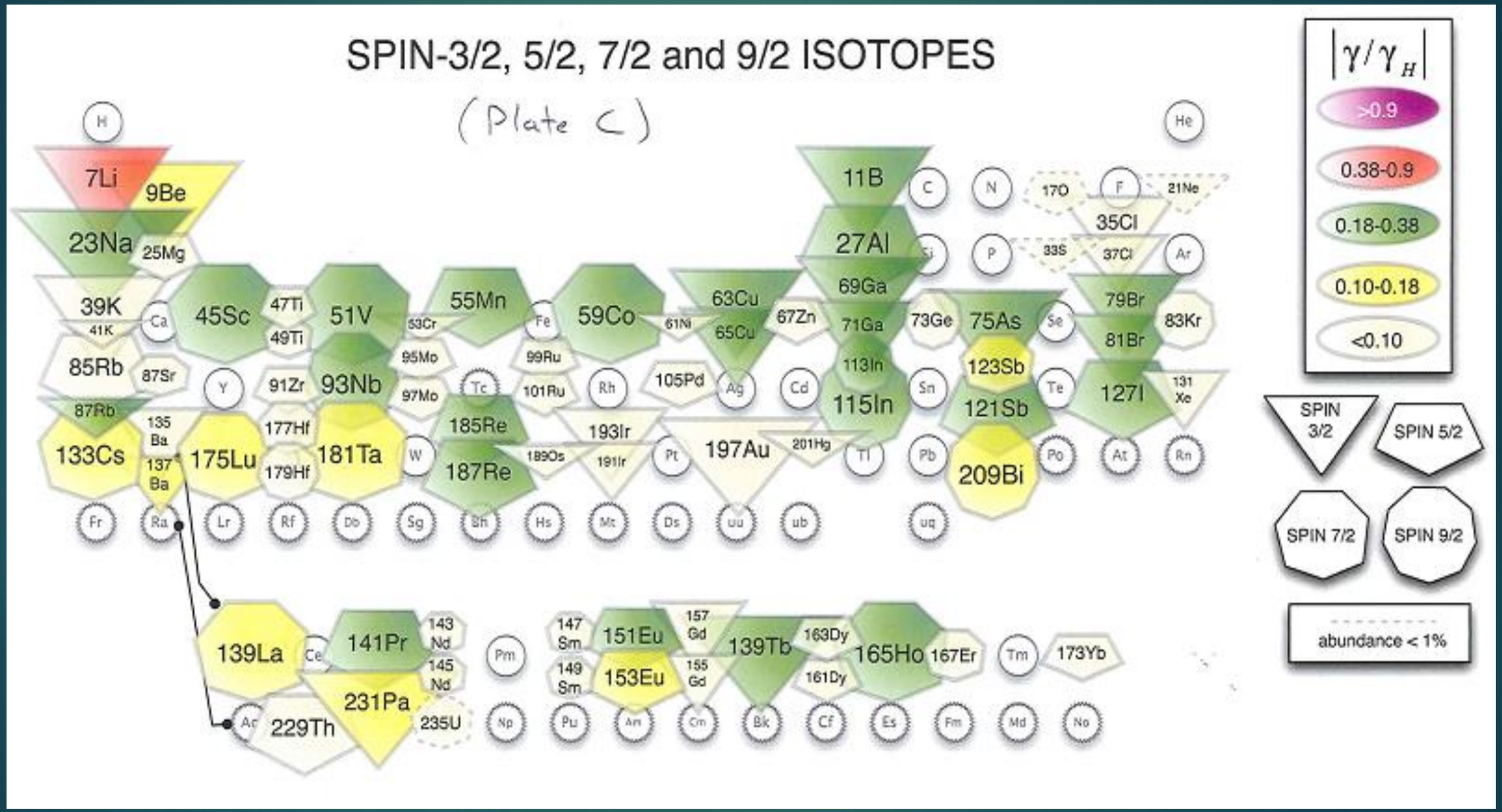
Spin Basics


SPIN-1/2 ISOTOPES
(Plate A)
3H
1H
3He
19F
13C
15N
29Si
31P
77Se
119Sn
117Sn
115Sn
123Te
125Te
129Xe
111Cd
113Cd
107Ag
109Ag
103Rh
52Fe
89Y
183W
187Os
195Pt
199Hg
203Tl
205Tl
207Pb
209Po
169Tm
171Yb
239Pu
|γ/γH|
>0.9
0.38-0.9
0.18-0.38
0.10-0.18
<0.10
radioactive
abundance < 1%

Spin Basics


INTEGER SPIN ISOTOPES
2H
6Li
10B
14N
40K
50 V
176Lu
138La
$|\gamma/\gamma_H|$
>0.9
0.38-0.9
0.18-0.38
0.10-0.18
<0.10
SPIN 1
SPIN 3
SPIN 4
SPIN 5
SPIN 6
SPIN 7
abundance < 1%

Spin Basics


SPIN-3/2, 5/2, 7/2 and 9/2 ISOTOPES
(Plate C)
$|\gamma/\gamma_H|$
>0.9
0.38-0.9
0.18-0.38
0.10-0.18
<0.10
SPIN 3/2
SPIN 5/2
SPIN 7/2
SPIN 9/2
abundance < 1%
H
He
7Li
9Be
11B
C
N
17O
F
21Ne
23Na
25Mg
27Al
35Cl
Si
P
33S
37Cl
Ar
39K
41K
Ca
45Sc
47Ti
49Ti
51V
53Cr
55Mn
Fe
59Co
61Ni
63Cu
65Cu
67Zn
69Ga
71Ga
73Ge
75As
Se
79Br
81Br
83Kr
85Rb
87Sr
Y
91Zr
93Nb
95Mo
97Mo
Tc
99Ru
101Ru
Rh
105Pd
Ag
Cd
113In
115In
Sn
123Sb
121Sb
Te
127I
131Xe
87Rb
133Cs
135Ba
137Ba
175Lu
177Hf
179Hf
181Ta
W
185Re
187Re
189Os
193Ir
191Ir
Pt
197Au
201Hg
Tl
Pb
209Bi
Po
At
Rn
Fr
Ra
Lr
Rf
Db
Sg
Bh
Hs
Mt
Ds
uu
ub
uq
139La
Ce
141Pr
143Nd
145Nd
Pm
147Sm
149Sm
151Eu
153Eu
157Gd
155Gd
139Tb
163Dy
161Dy
165Ho
167Er
Tm
173Yb
Ac
229Th
231Pa
235U
Np
Pu
Am
Cm
Bk
Cf
Es
Fm
Md
No

MR Spectroscopic Imaging

- MRI- Basics and k-Space Encoding
- Single Voxel Spectroscopy
- Multi-voxel Spectroscopy/Spectroscopic Imaging
- *Acceleration Techniques*: Phase-encoding, parallel Imaging, Echo-planar Imaging, Concentric Rings, Radial Imaging and more
- Multi-dimensional MR Spectroscopic Imaging (2D spectral+3D spatial)
- Conclusions

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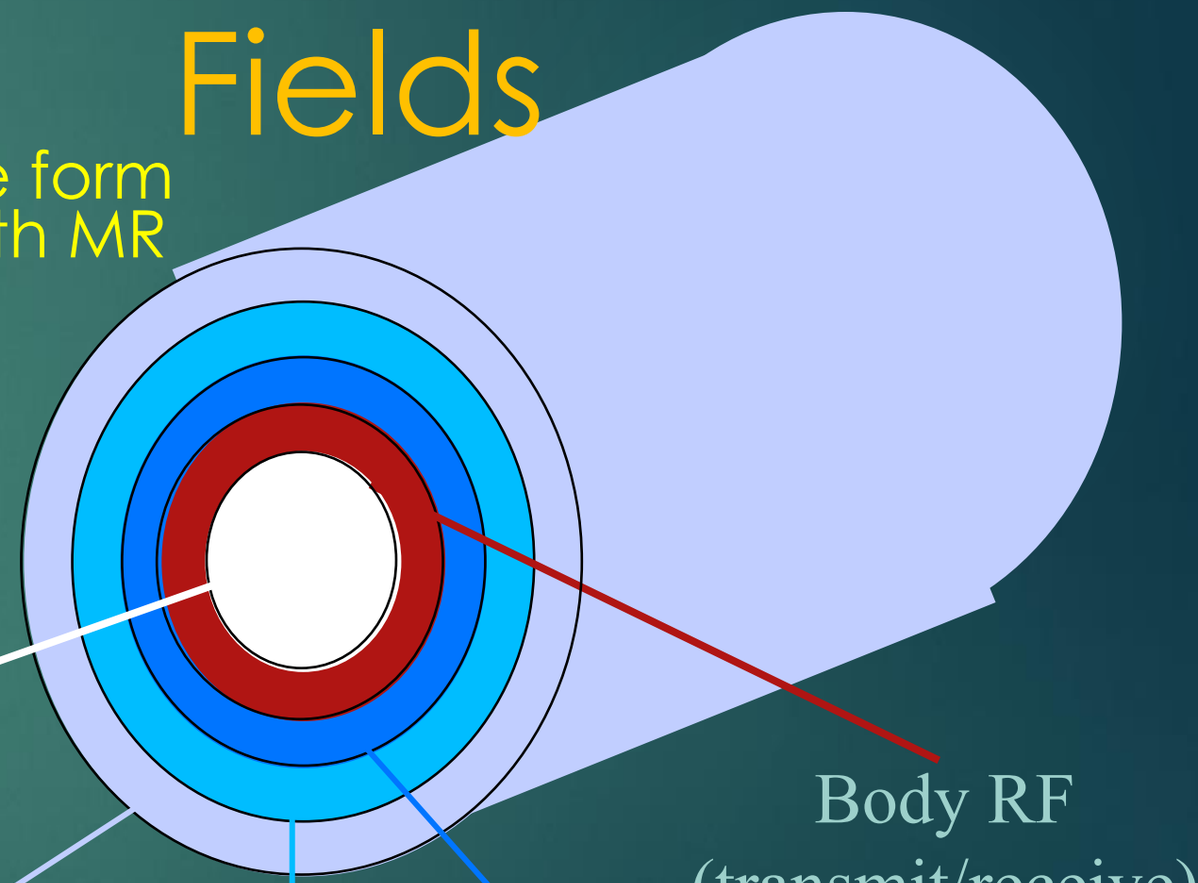
MRI Uses Three Magnetic Fields

Static High Field (B_0)
Creates or polarizes signal
1000 Gauss to 110,000 Gauss
(Earth's field is 0.5 G)

Gradient Fields
1-4 G/cm
Used to image: determine spatial position of MR signal

Radiofrequency Field (B_1)
Excites or perturbs signal into a measurable form
On the order of O.1 G but in resonance with MR signal
RF coils also measure MR signal
Excited or perturbed signal returns to equilibrium
Important contrast mechanism



Lauterbur 1973
KWE 1975

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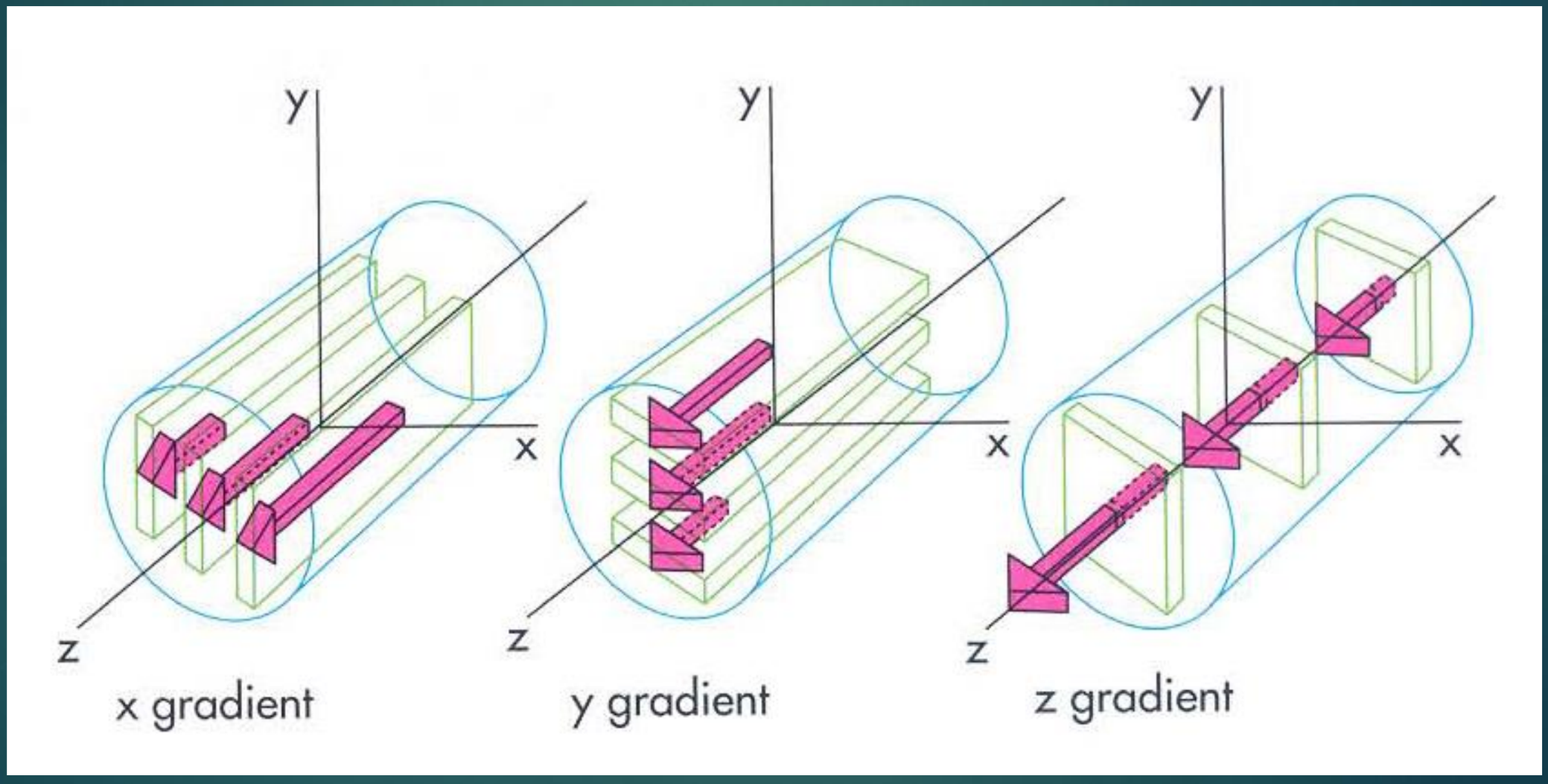
UCLA

Effect of pulsed field gradients (X, Y, Z)- Spatial Localization

Every imaging system will have three gradient coils that can modify the static field strength (Bo) in X, Y,Z directions. Thus you have the control over changing the Larmor frequencies of nuclear spins in X,Y,Z directions



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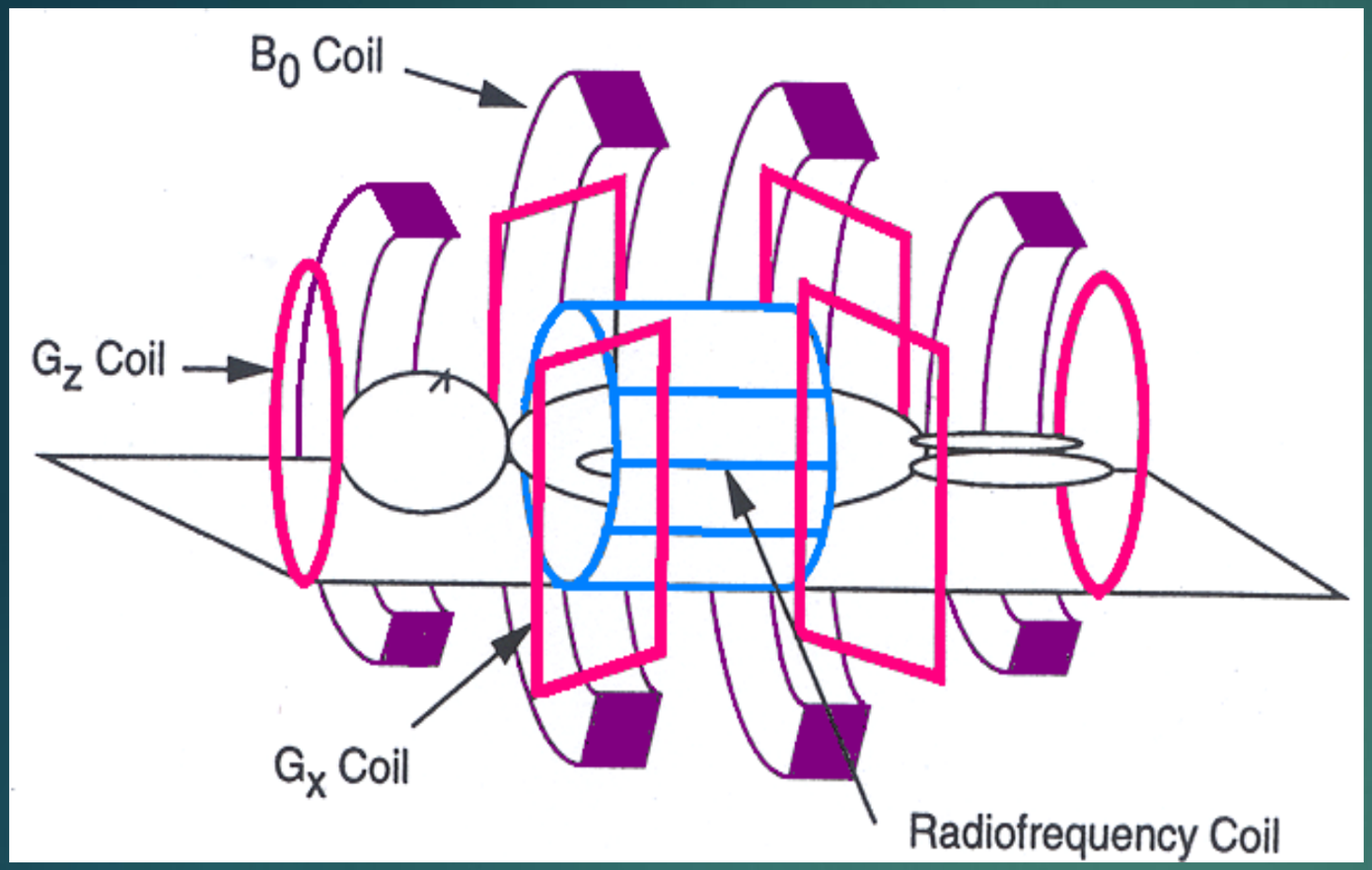
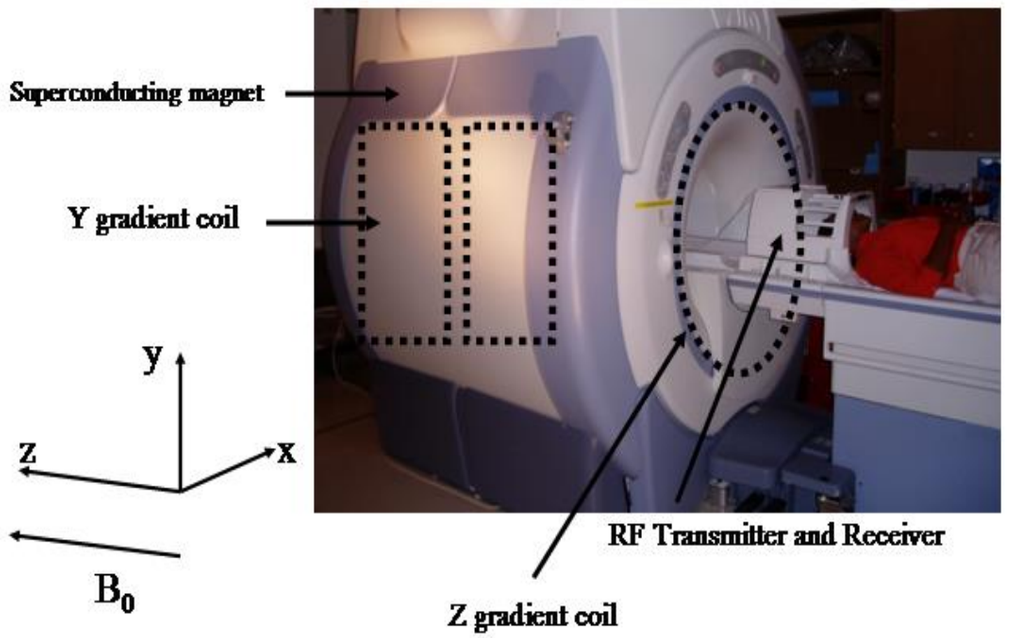
Gradient Coils



Nishimura, MRI Principles



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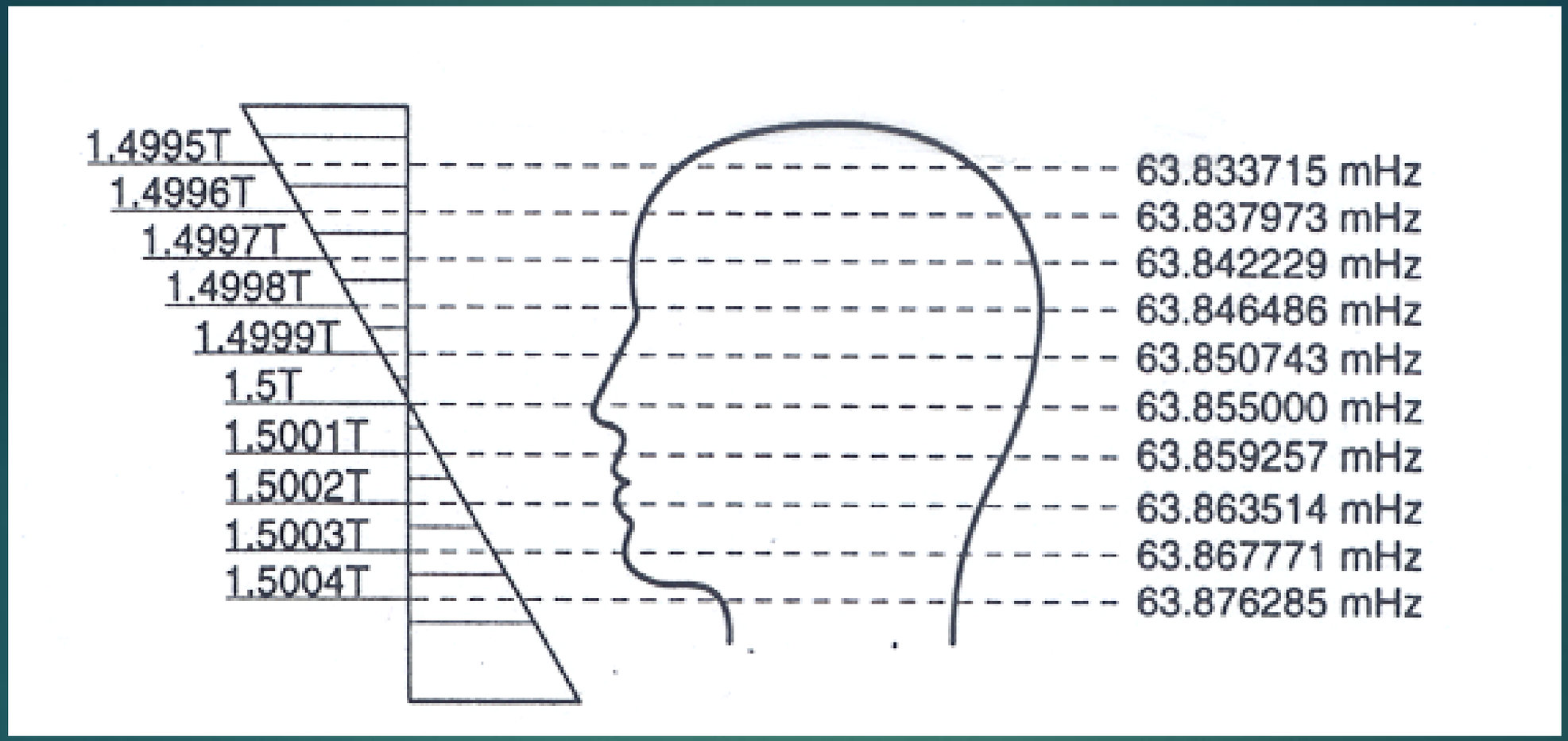
Spatial Encoding/Slice Selection



- The effects of the main magnetic field and the applied slice gradient. In this example, the local magnetic field changes in one-Gauss increments accompanied by a change in the precessional frequency from chin to the top of the head.

k-Space Acquisition

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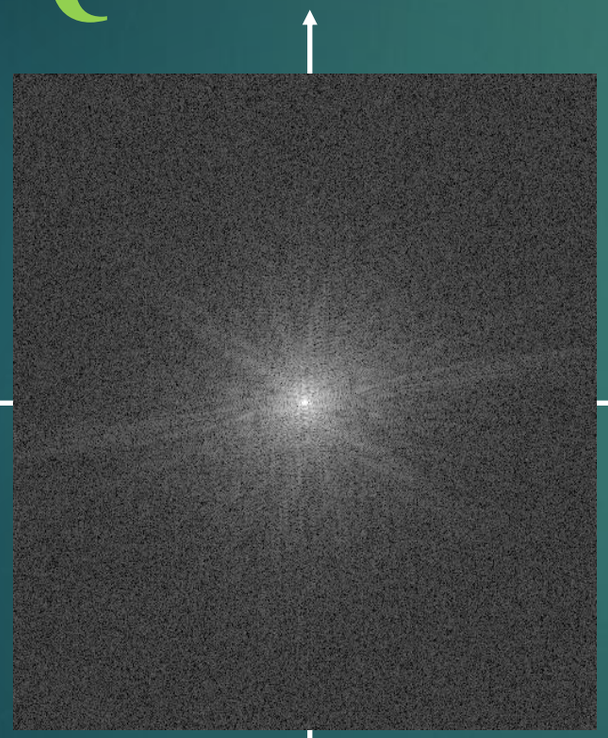
Phase Encode

Sampled Signal

DAQ

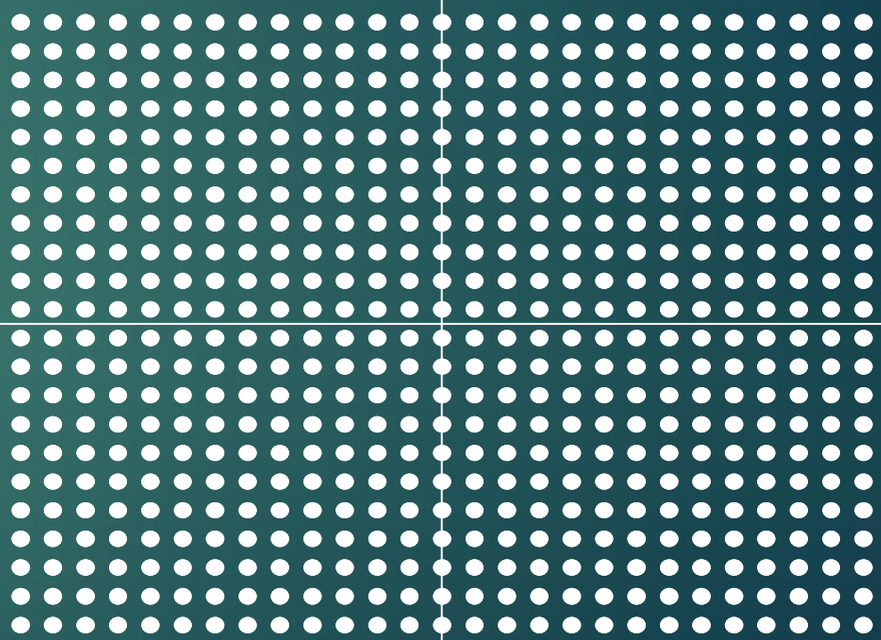


k_y

k_x

Phase Encoding

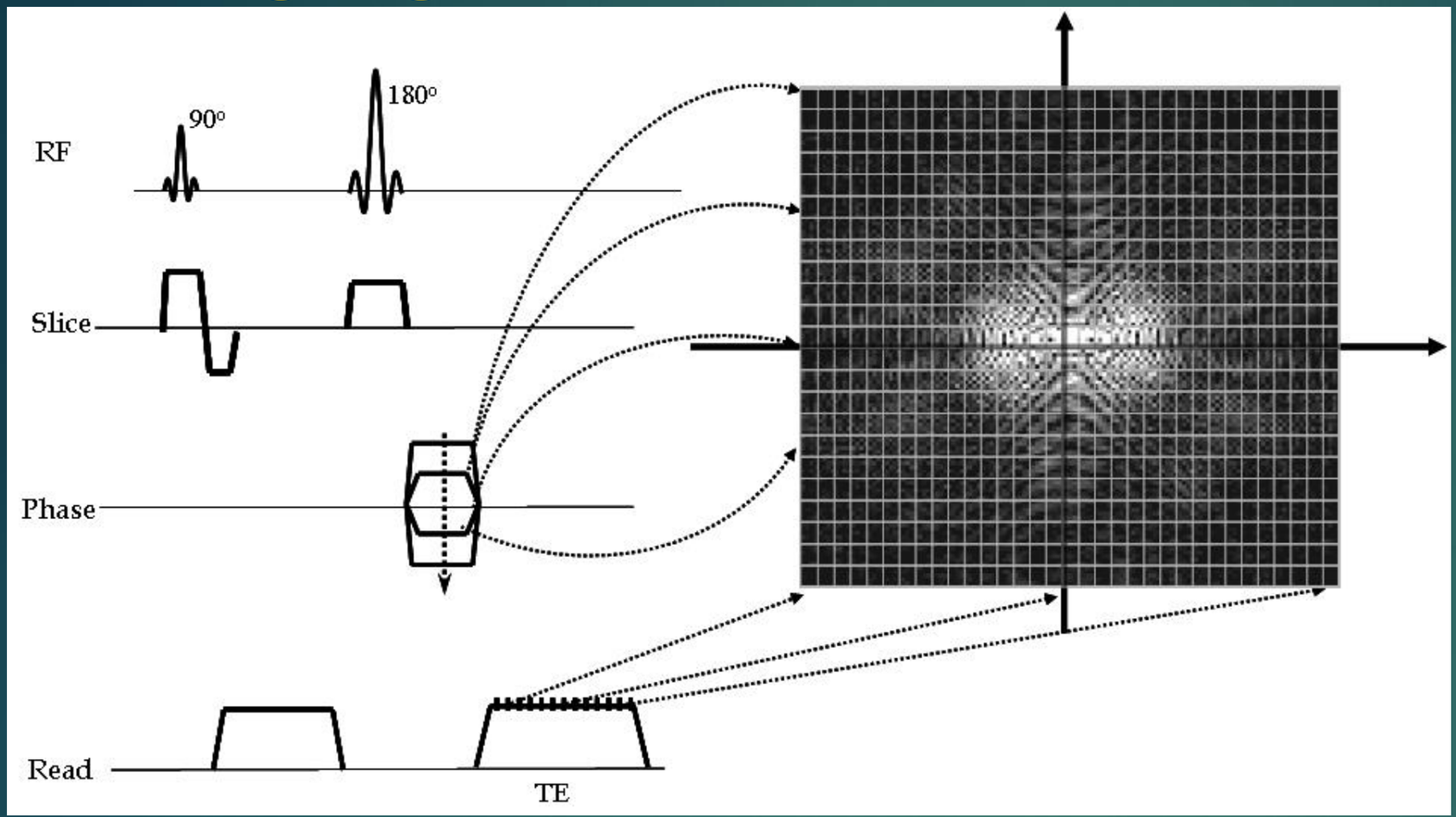
One line of k-space acquired per TR



Frequency Encoding

Kumar Welti Ernst JMR 18;69-83 1975; Edelstein et al. Spin Warp Imaging. PBM 1980

Fourier Zeugmatography/Spin-Warp Imaging



Gradient applied along the y-axis will cause the spins to precess at a frequency determined by their y position, and is called phase encoding. Next a gradient is applied along the x-axis and the spin-echo is collected. The frequency components of the echo gives information of the x-position and the phase values give information of the y-position.

$$S(t_x, t_y) = \int\int \mathcal{A}(x,y) \exp\left[i\,\gamma(G_x x t_x + G_y y t_y)\right] dx\, dy$$

$k_x = \gamma G_x t_x$ and $k_y = \gamma G_y t_y$

$$S(k_x, k_y) = \int\int \mathcal{A}(x,y) \exp\left[i(k_x x + k_y y)\right] dx\, dy$$

$$\mathcal{A}(x,y) = \int\int S(k_x, k_y) \exp\left[-i(k_x x + k_y y)\right] dx\, dy$$

Kumar Welti Ernst JMR 18;69-83 1975; Edelstein et al. Spin Warp Imaging. PBM 1980

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K-Space

For a given data point in k-space, say (kx, ky), its signal S(kx, ky) is the sum of all the little signal from each voxel I(x,y) in the physical space, under the gradient field at that particular moment

$$S(k_x, k_y) = \int \int I(x,y) e^{-i2\pi(k_x x + k_y y)} dx dy$$

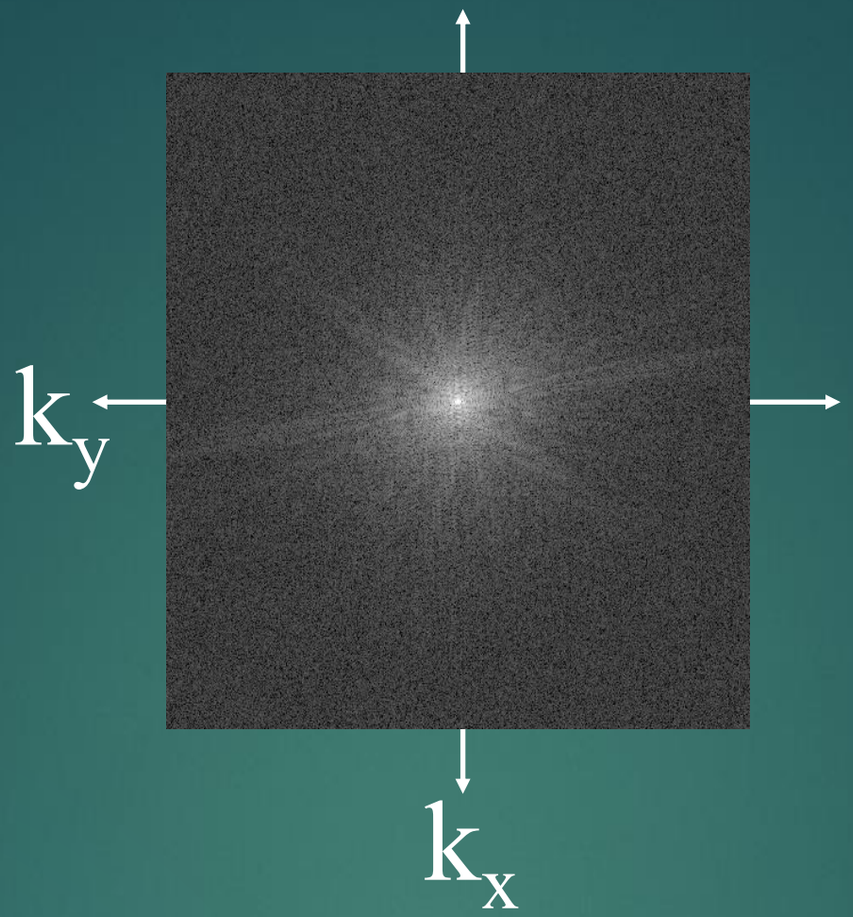
From this equation, it can be seen that the acquired MR signal, which is also in a 2-D space (with kx, ky coordinates), is the Fourier Transform of the imaged object.

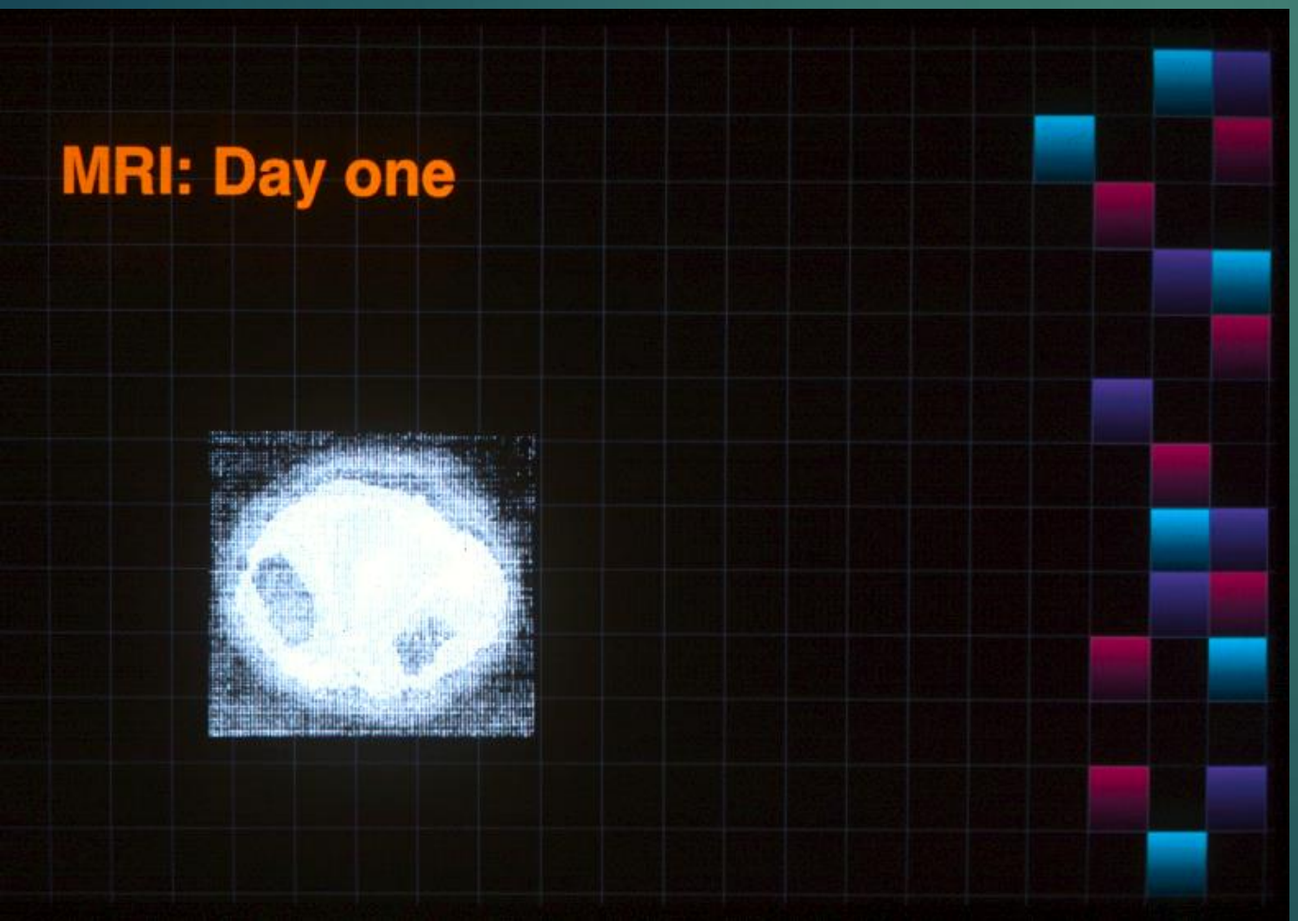
$$Kx = \gamma/2\pi \int_0^t Gx(t)\, dt$$

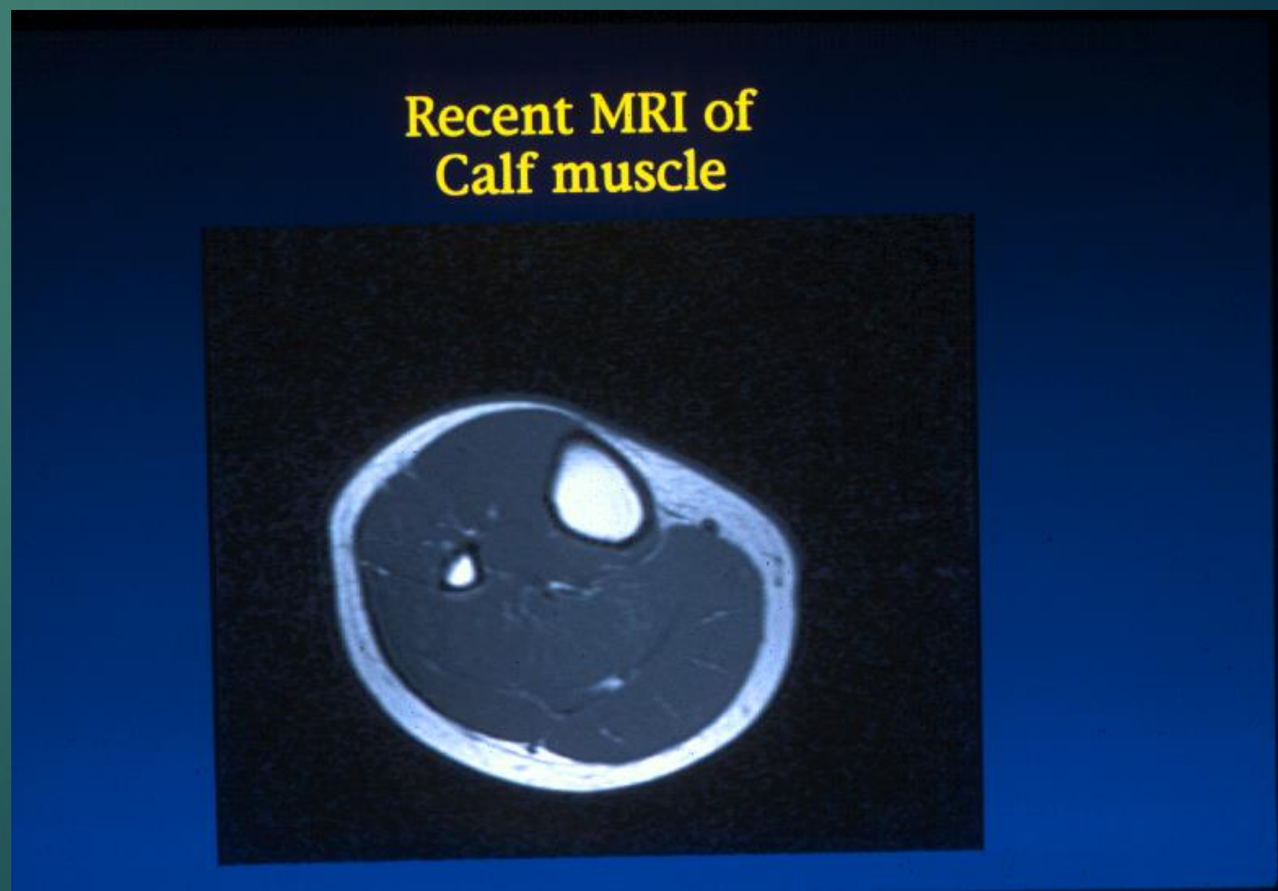
$$Ky = \gamma/2\pi \int_0^t Gy(t)\, dt$$

The frequency and phase encoding gradients control the imaging trajectory in k-space

MRI, Brown et al. 1999; Principles of MRI Liang and Lauterbur 2000; MRI Pulse Sequences, Zhoe et al. 2004; EPI Theory, Technique and Application, Stehling et al. 1998

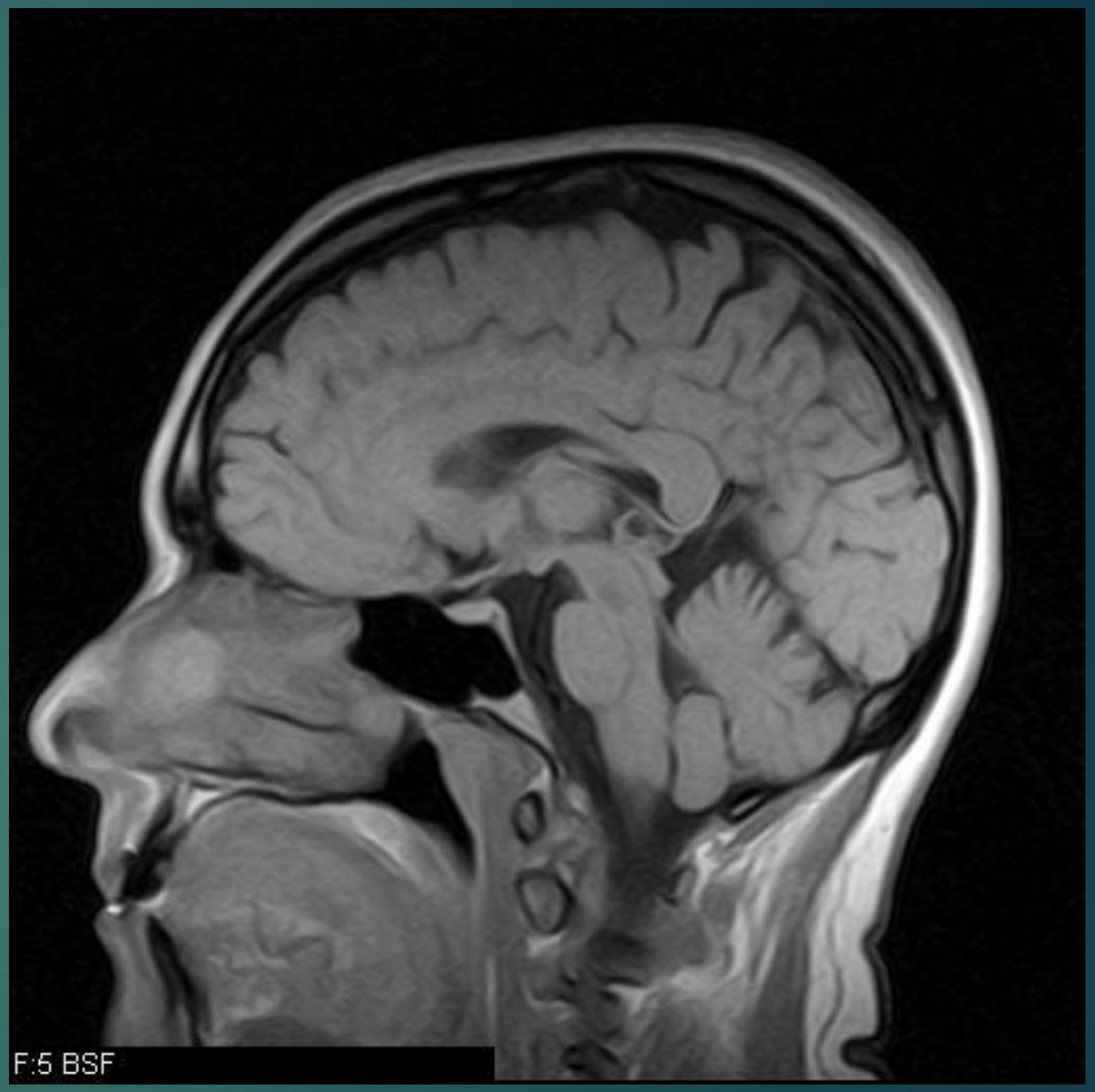

k_y
k_x


MRI: Day one


Recent MRI of
Calf muscle

Magnetic Resonance Imaging (MRI)

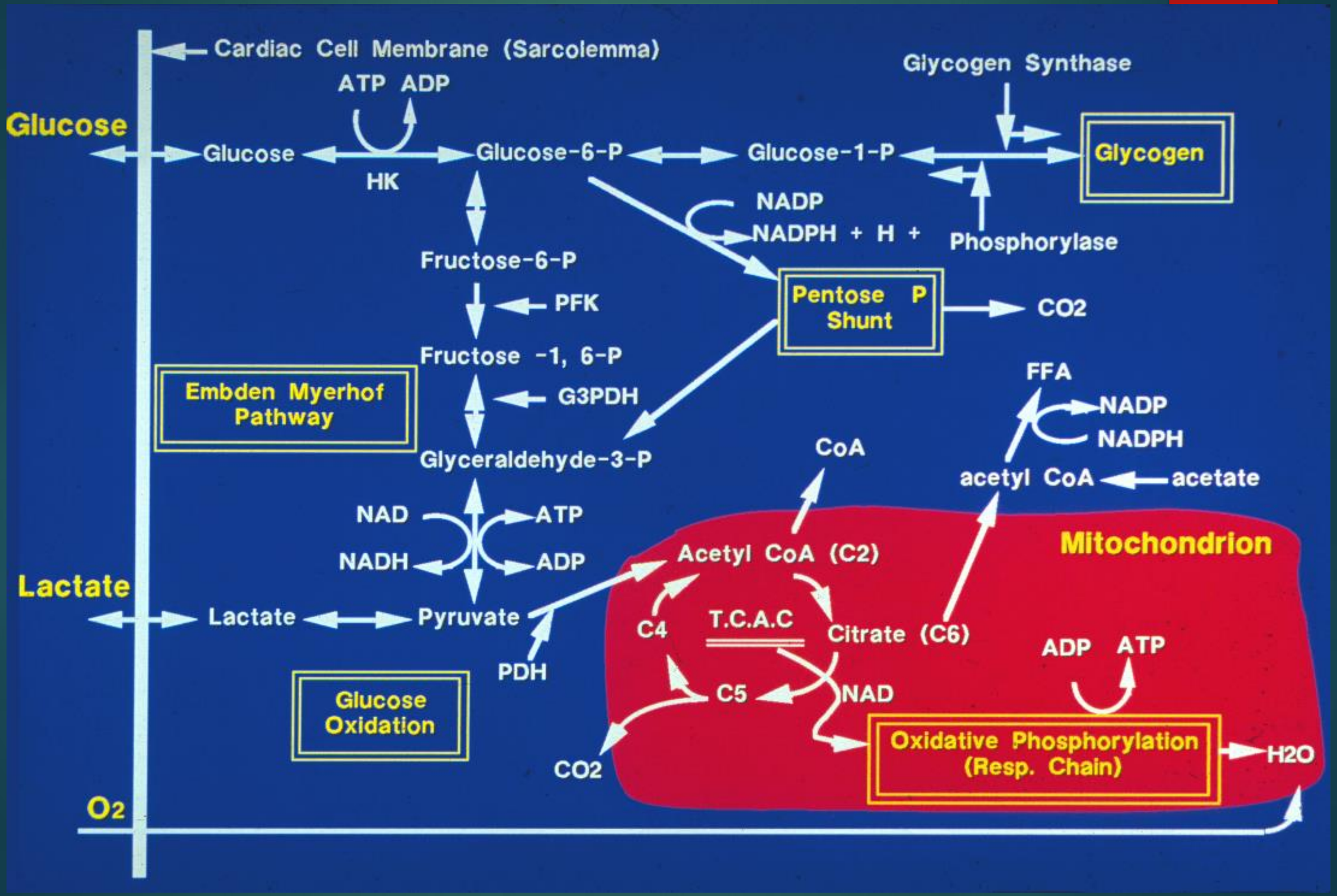
- MRI exploits Nuclear Magnetic Resonance (NMR) to produce water-based images
  - Signal from ^{1}H in water
  - Gray scale caused by T1/T2 relaxation and ^{1}H density within a voxel
- MRI resolution
  - 512x512 voxels in a slice
  - Sub-millimeter voxel volume
- Structural differences cause T1/T2 relaxation variation among voxels



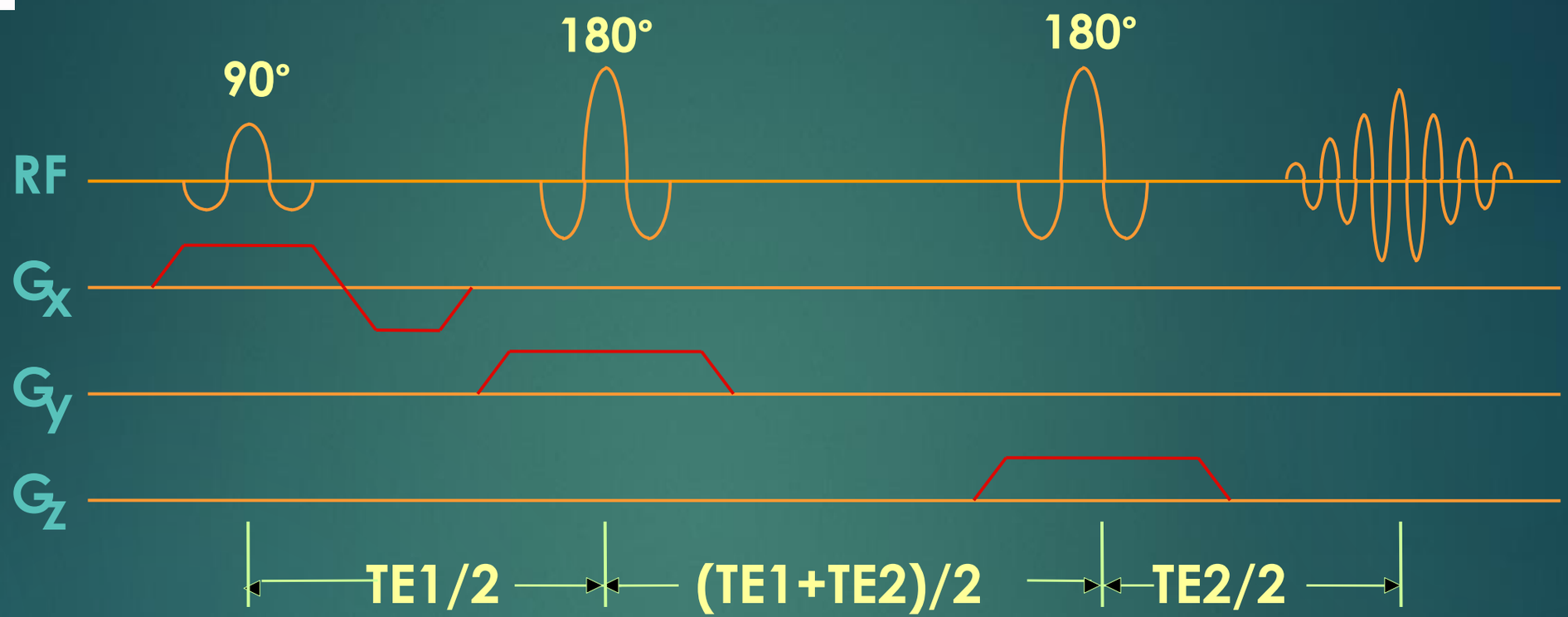
Problems with Anatomical Imaging

- Despite its superb soft tissue contrast and multiplanar capability, anatomical MRI is largely limited to depicting morphological abnormality.
- Anatomical MRI suffers from nonspecificity. Different disease processes can appear similar upon anatomic imaging, and in turn a single disease entity may have varied imaging findings.
- The underlying metabolic or functional integrity of brain cannot be adequately evaluated based on anatomical MRI alone.
- To that end, several physiology-based MRI methods have been developed to improve tumor characterization.

- Diffusion Weighted (DW) MRI/Diffusion Tensor Imaging (DTI): In addition to early diagnosis of cerebral ischemia, DW MRI is extremely sensitive in detecting other intracranial disease processes, including cerebral abscess, traumatic shearing injury, etc.
- Perfusion Imaging: Dynamic susceptibility-weighted contrast-enhanced (DSC) perfusion MRI of the brain provides hemodynamic information.
- MR Spectroscopy for biochemical characterization, Improving Specificity of cancer and more


Cardiac Cell Membrane (Sarcolemma)
ATP ADP
Glucose
Glucose
HK
Glucose-6-P
Glucose-1-P
Glycogen Synthase
Glycogen
NADP
NADPH + H +
Phosphorylase
Fructose-6-P
PFK
Pentose P Shunt
CO2
Fructose -1, 6-P
Embden Myerhof Pathway
G3PDH
FFA
NADP
NADPH
CoA
Glyceraldehyde-3-P
acetyl CoA
acetate
NAD
ATP
NADH
ADP
Mitochondrion
Acetyl CoA (C2)
Lactate
Lactate
Pyruvate
C4
T.C.A.C
Citrate (C6)
ADP ATP
PDH
C5
NAD
Glucose Oxidation
Oxidative Phosphorylation (Resp. Chain)
H2O
CO2
O2

Point Resolved Spectroscopy, PRESS

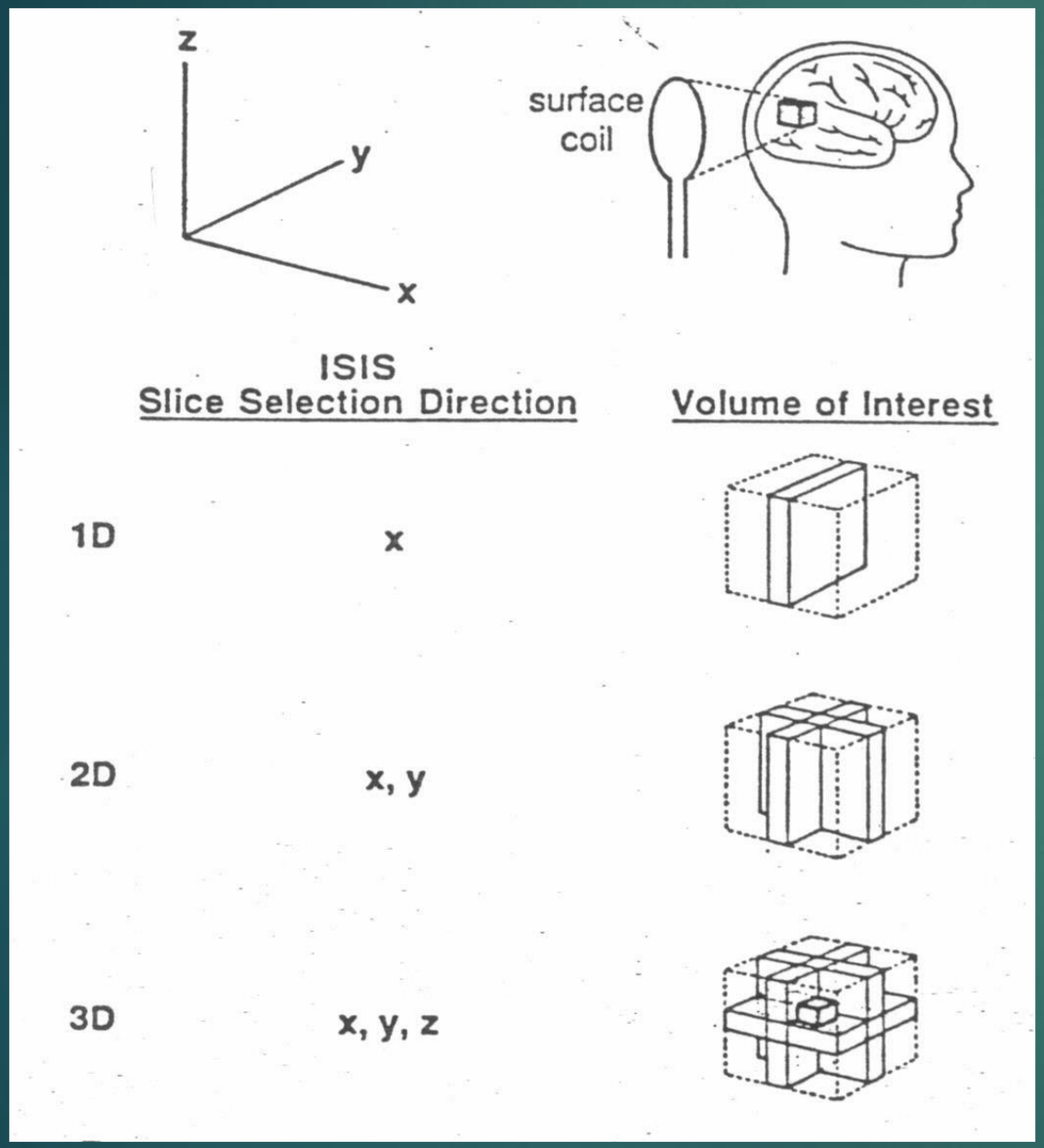


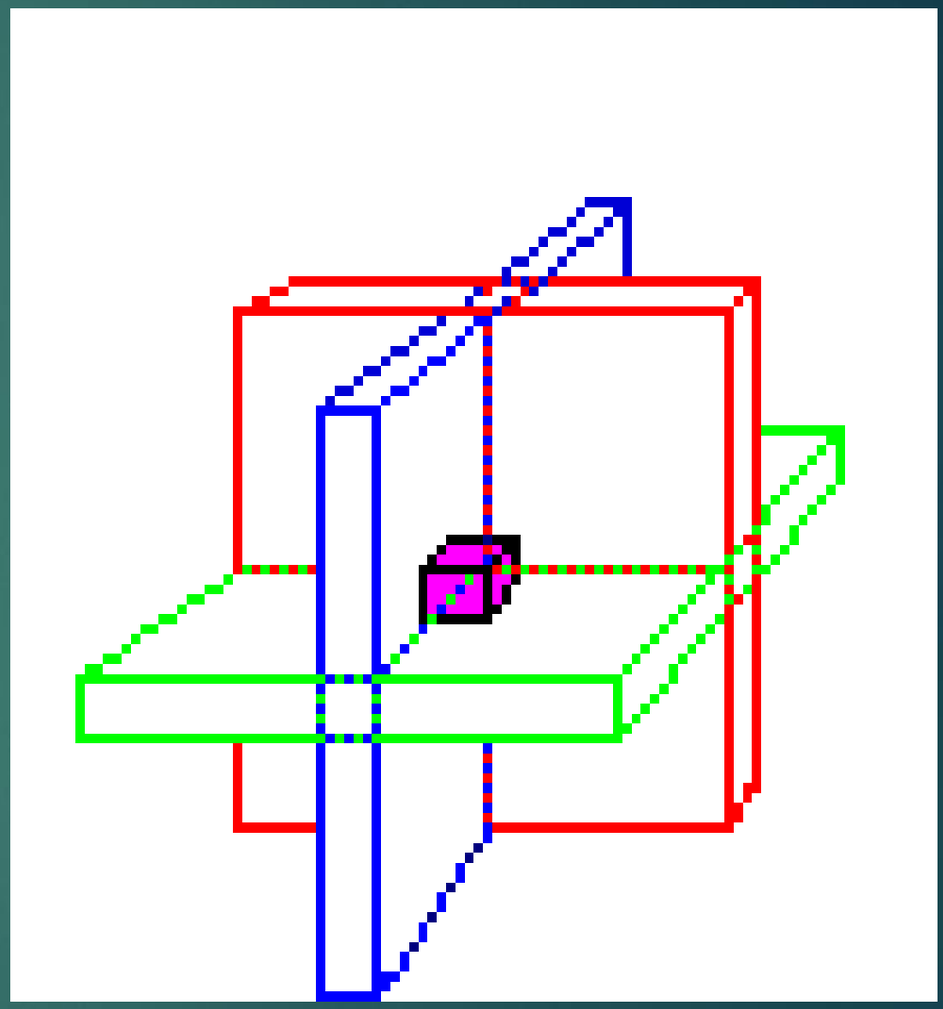
- A slice-selective 90° pulse is followed by two slice-selective 180° refocusing pulses
- Achieves localization within a single acquisition
- Suitable for signals with long T_2 – ^1H MRS

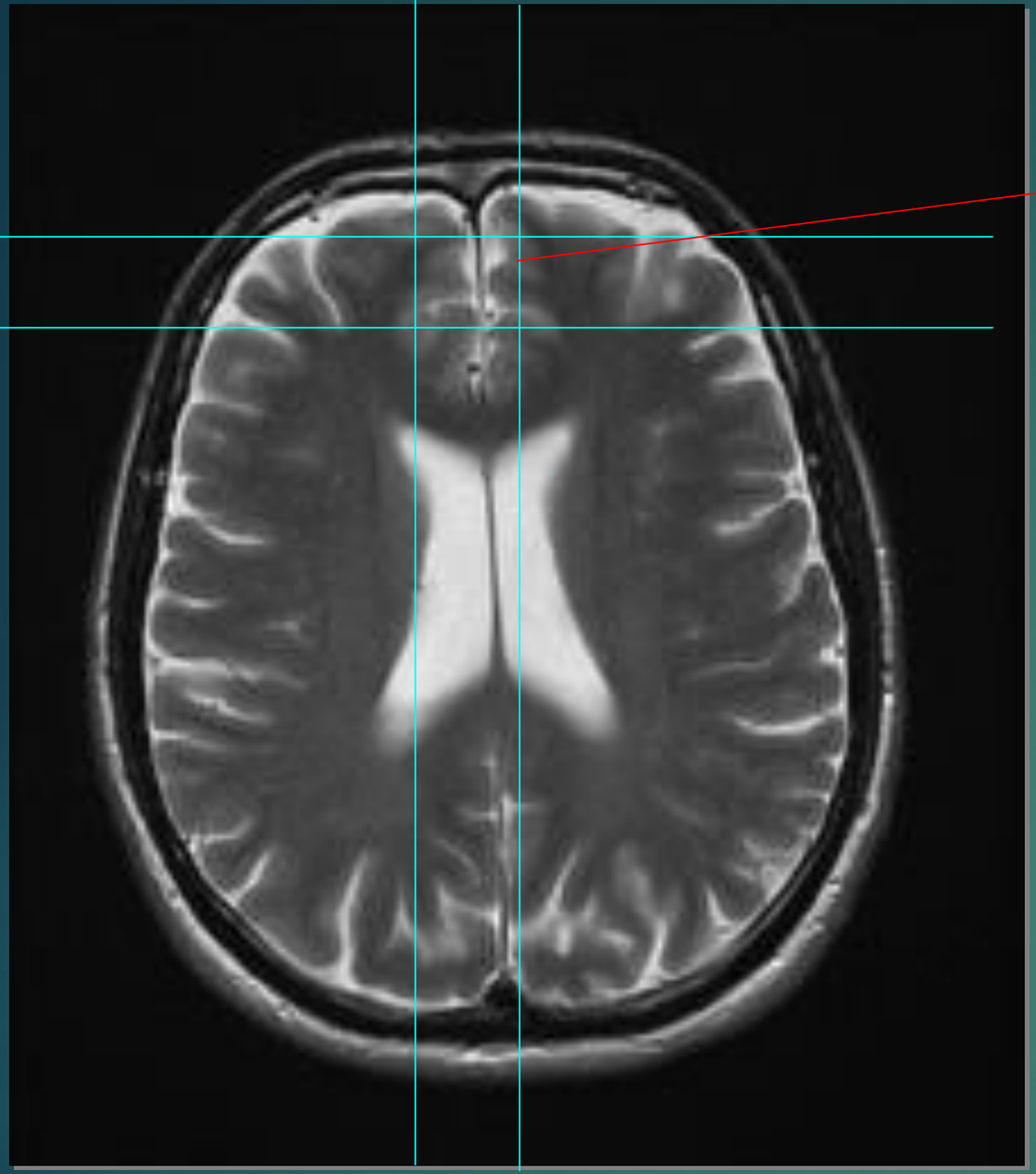
Bottomley PA. Annal NY Acad Sci 1987; 508: 333-348.

STEAM: Frahm MRM 1989

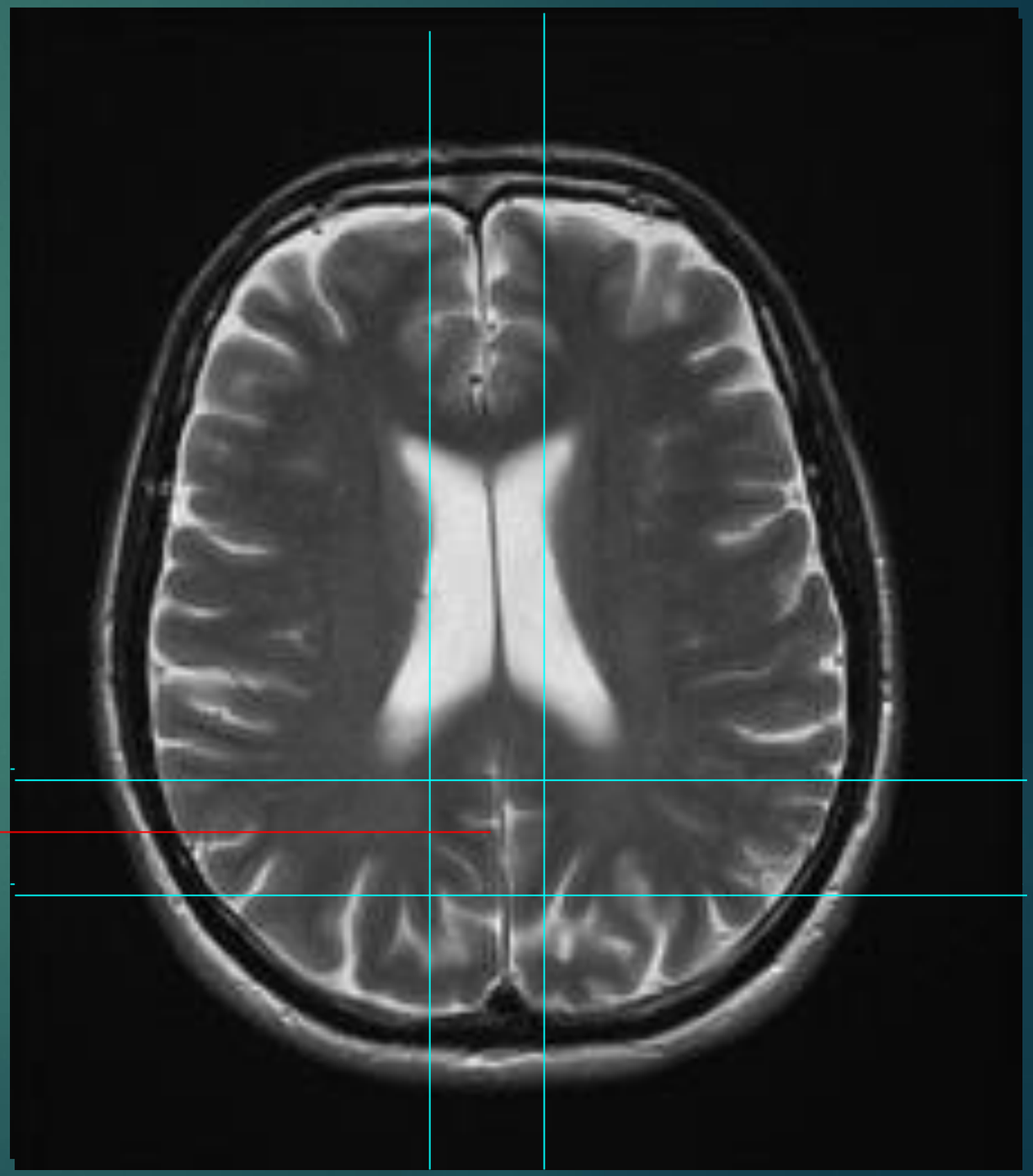
Localization







Frontal Gray



Occipital Gray

5-10 minutes for each voxel MRS
Total duration = Nx10 minutes for N voxels???

Single Voxel Spectroscopy

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disadvantages

- requires large sample volume (2x2x2 cm^3 or more)
- requires many averages for adequate SNR
- limited coverage
- can only cover a small region in one experiment

Multi-Voxel Spectroscopy

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the problem of limited coverage can be fixed by taking conducting multiple experiments from different locations

this is problematic as the total experimental time will scale with the number of different voxels you wish to measure

Spectral Characteristics from different zones in Healthy (<40 years) Volunteer

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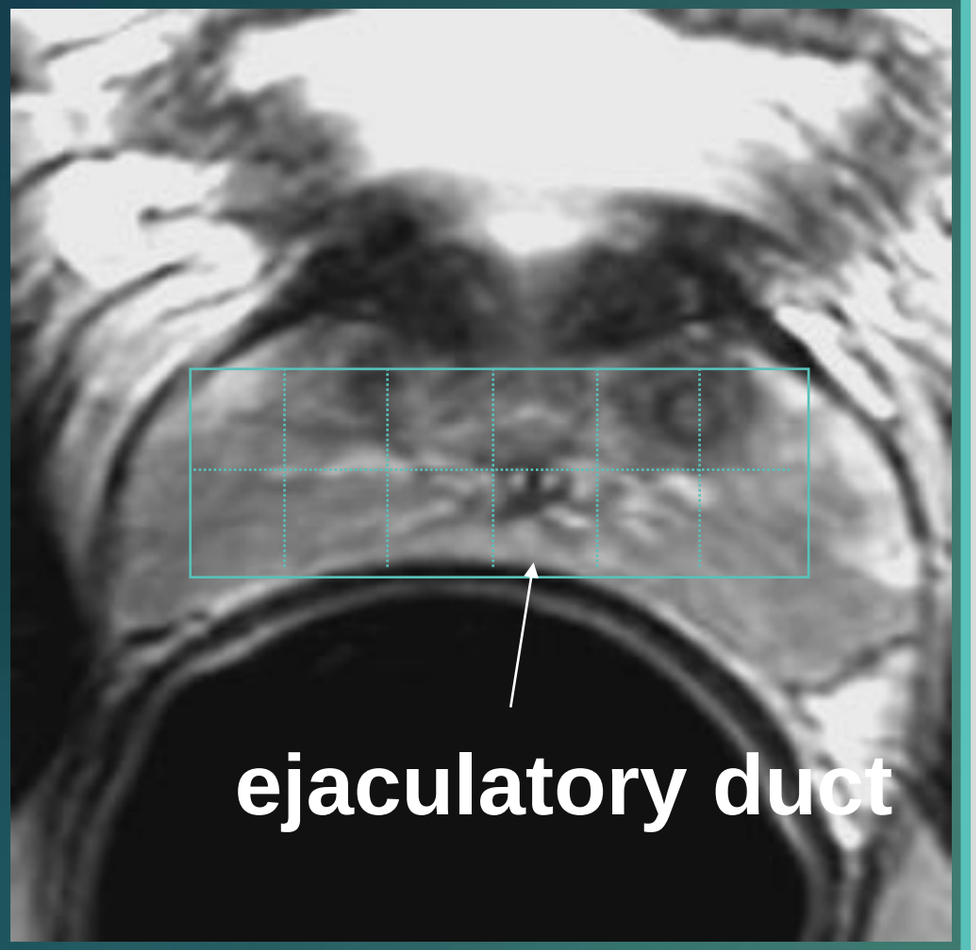
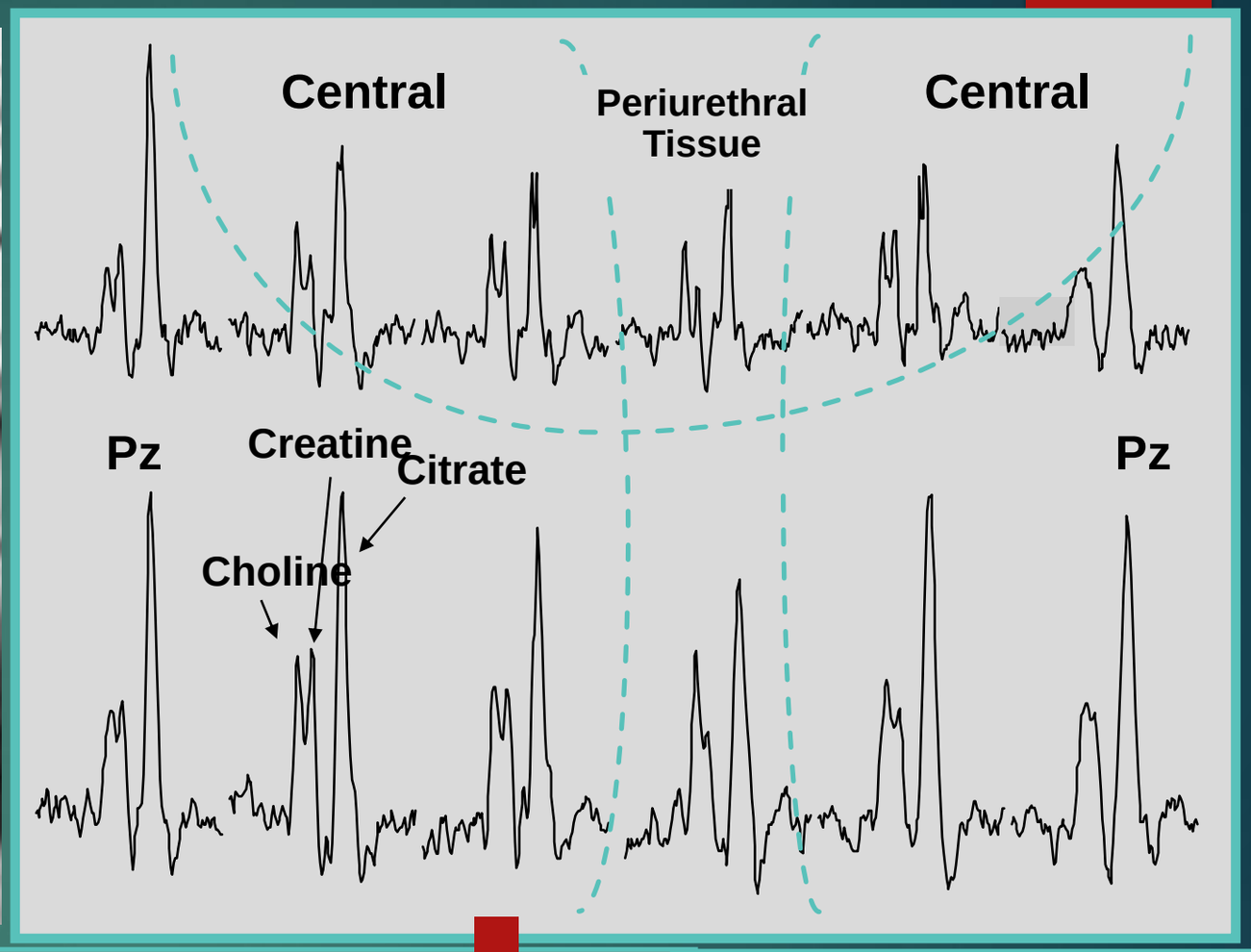
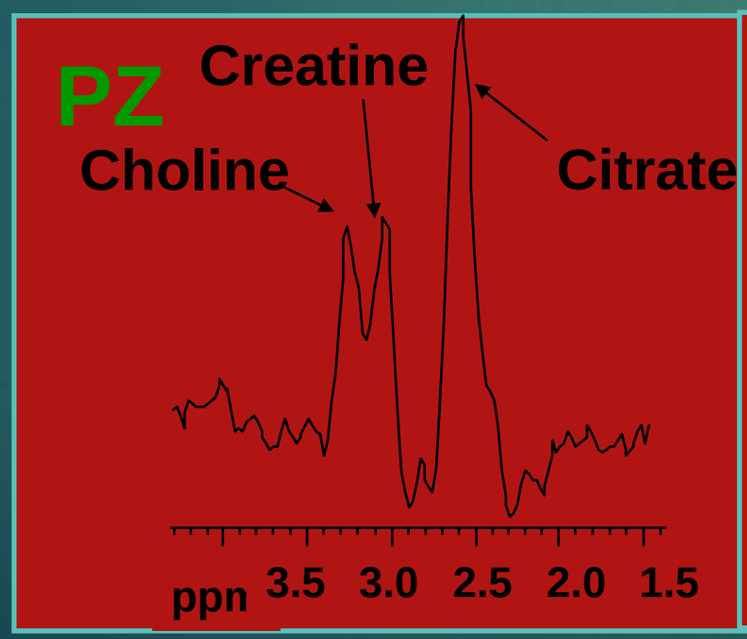
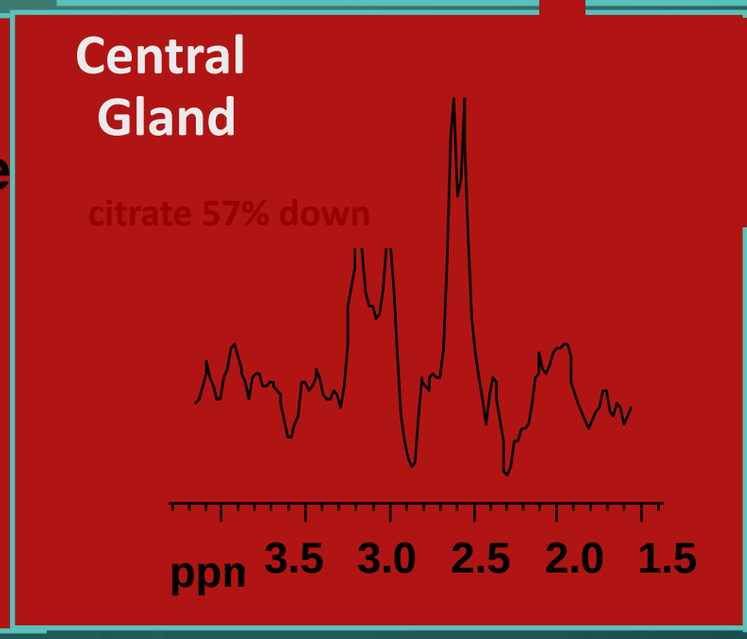
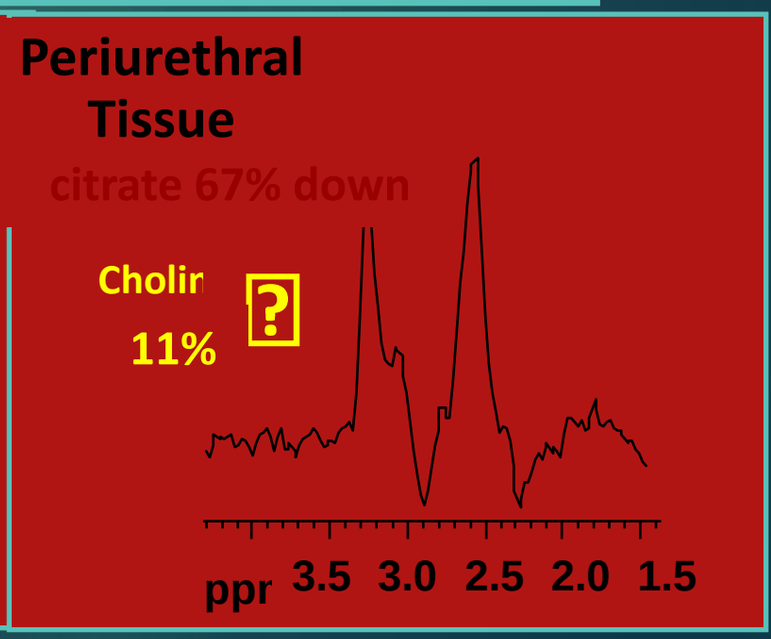




Three Distinctive Metabolic Patterns







Courtesy: Prof. John Kurhanewicz

Proc. Natl. Acad. Sci. USA
Vol. 79, pp. 3523–3526, June 1982
Biophysics

NMR chemical shift imaging in three dimensions

(*in vivo* biochemistry/^{31}P imaging/metabolite mapping)

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Communicated by John J. Hopfield, March 10, 1982

ABSTRACT **A method for obtaining the three-dimensional distribution of chemical shifts in a spatially inhomogeneous sample using Fourier transform NMR is presented. The method uses a sequence of pulsed field gradients to measure the Fourier transform of the desired distribution on a rectangular grid in (k,*t*) space. Simple Fourier inversion then recovers the original distribution. An estimated signal/noise ratio of 20 in 10 min is obtained for an "image" of the distribution of a 10 mM phosphorylated metabolite in the human head at a field of 20 kG with 2-cm resolution.**

the resonant frequency of the spins) varying linear gradient, $[\mathbf{G}(t)\cdot\mathbf{x}]\hat{z}$, as shown in Fig. 1, how will this affect the FID? Under these conditions, the phase of each spin at time t after a rf pulse will depend on both $\mathbf{x}$ and δ as its instantaneous frequency is now given by $\frac{d\phi}{dt} = \gamma H_T(t)$, where γ is the gyromagnetic ratio for the species under observation and $H_T(t) = [H_o + \mathbf{G}(t)\cdot\mathbf{x}](1 + \varepsilon)$. Here we have just augmented the externally applied field, $H_o + \mathbf{G}(t)\cdot\mathbf{x}$, by $(1 + \varepsilon)$ to take into account the electronic shield-

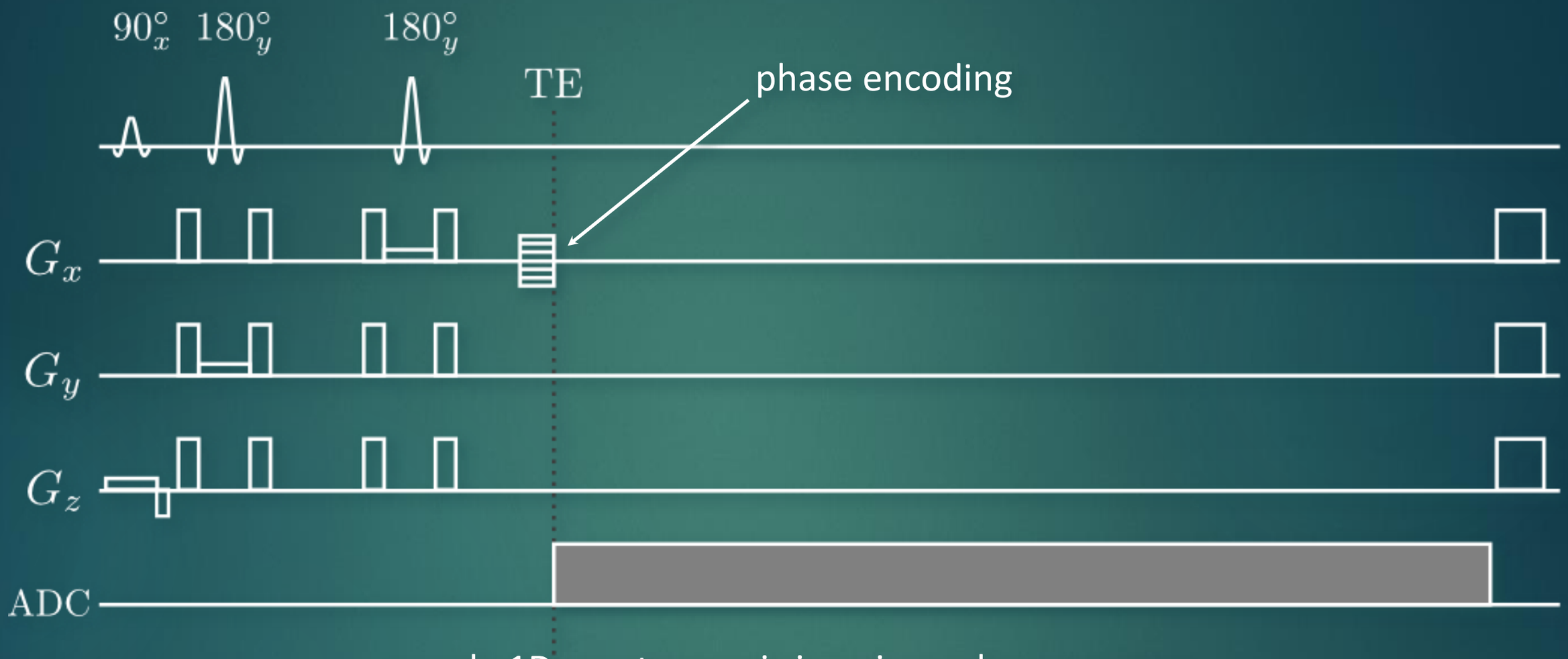
MRSI/CSI

greater coverage can be obtained by spatially encoding time signals with phase-encoding gradients

phase-encoded data is encoded in k-space

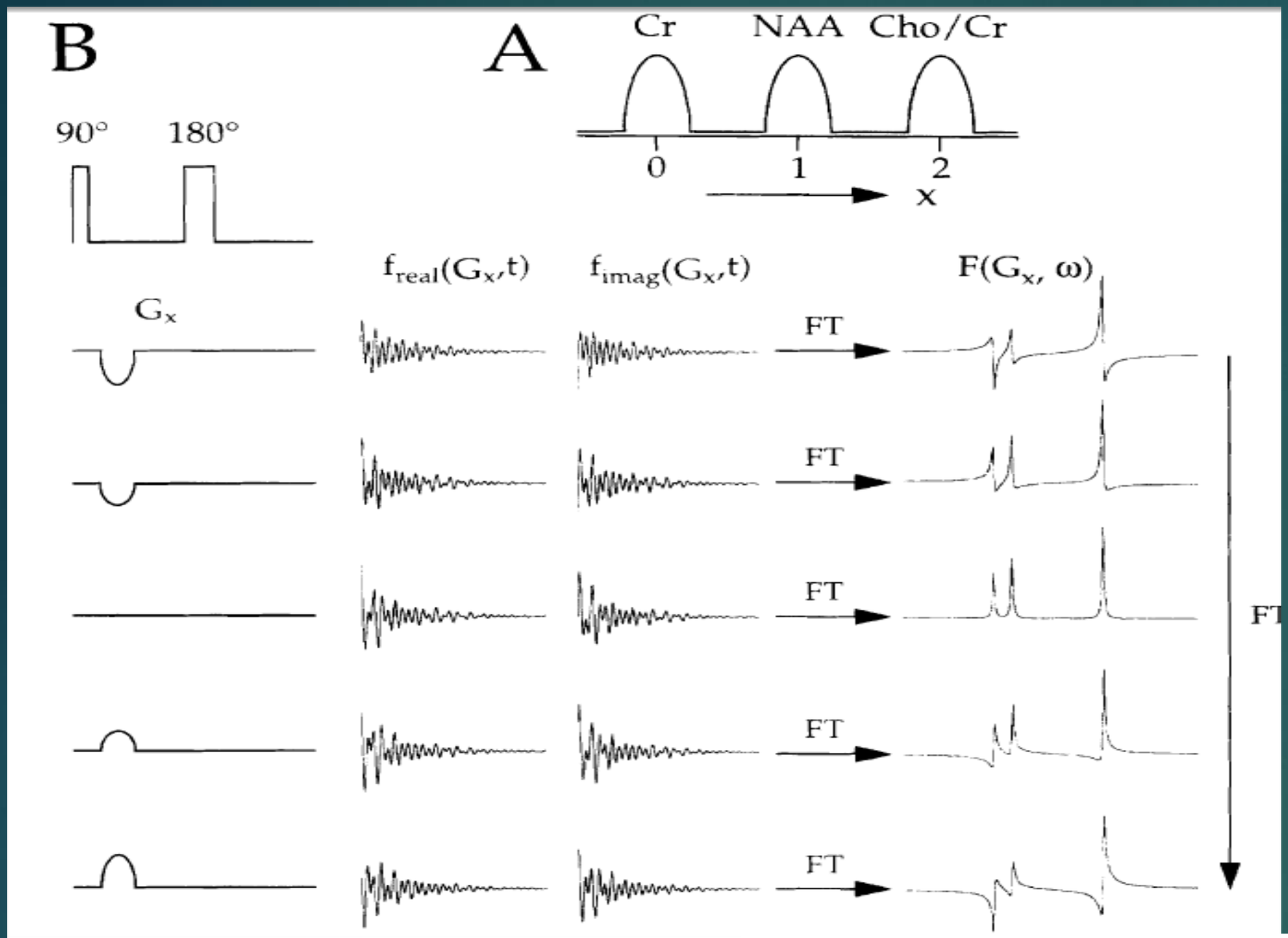
this allows for more voxels to be collected in a single experiment as well as smaller voxels

1D Spatial Encoding

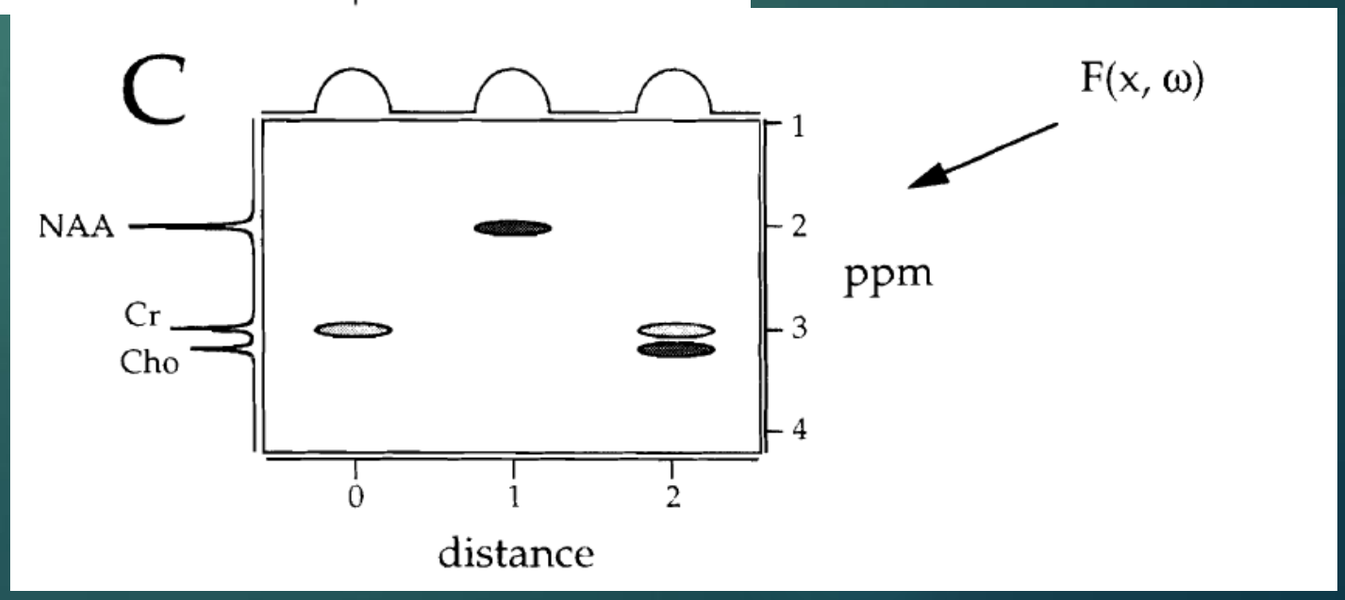


sample 1D spectroscopic imaging pulse sequence

each FID is phase encoded along one dimension



Resulting SI data set
After FFT along x and t we get



2D Spectroscopic Imaging

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how do we fill out 2D k-t-space?

same as before except phase encoding happens in two different dimensions now

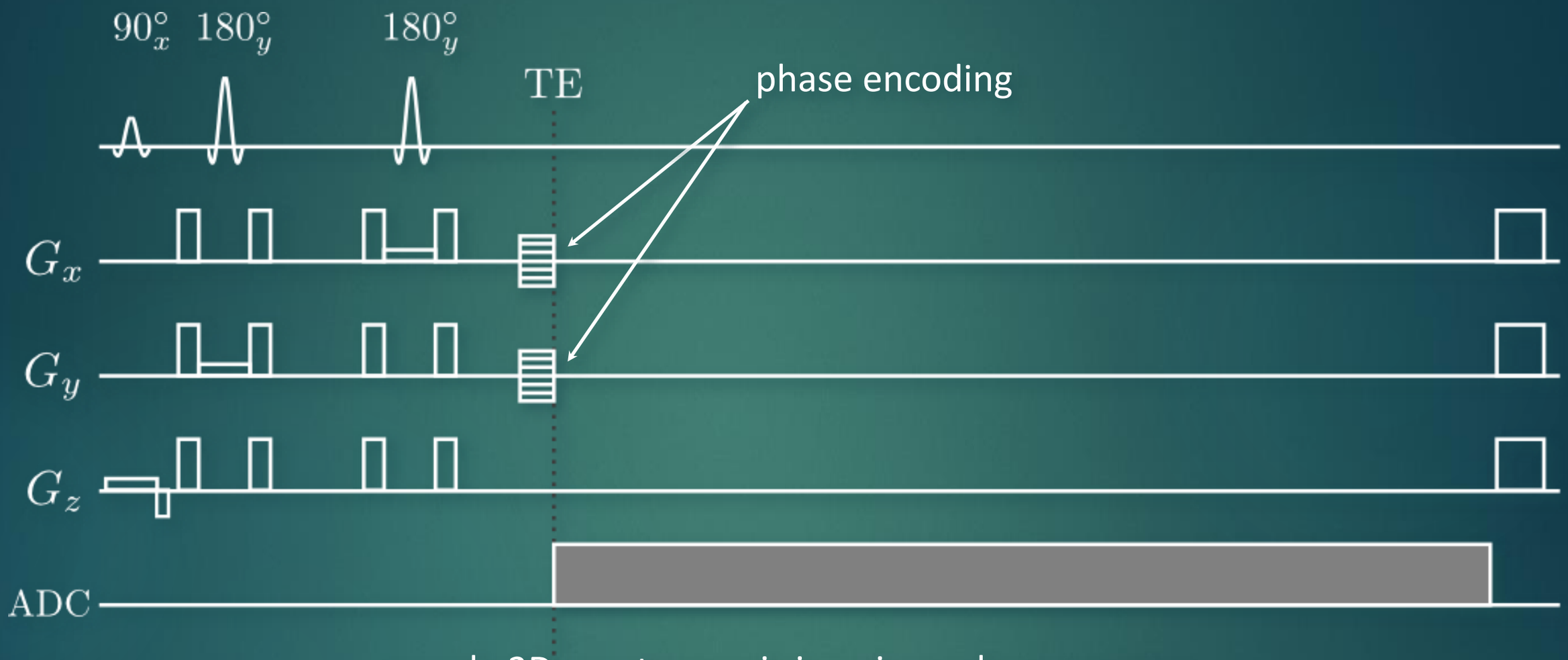
Spatial Encoding

how do we use gradients to move around k-space?

$$\vec{k}(t) = \frac{\gamma}{2\pi} \int_0^t \vec{\mathrm{G}}(\tau) \mathrm{d}\tau$$

we can move anywhere in k-space so long as we program our gradients correctly

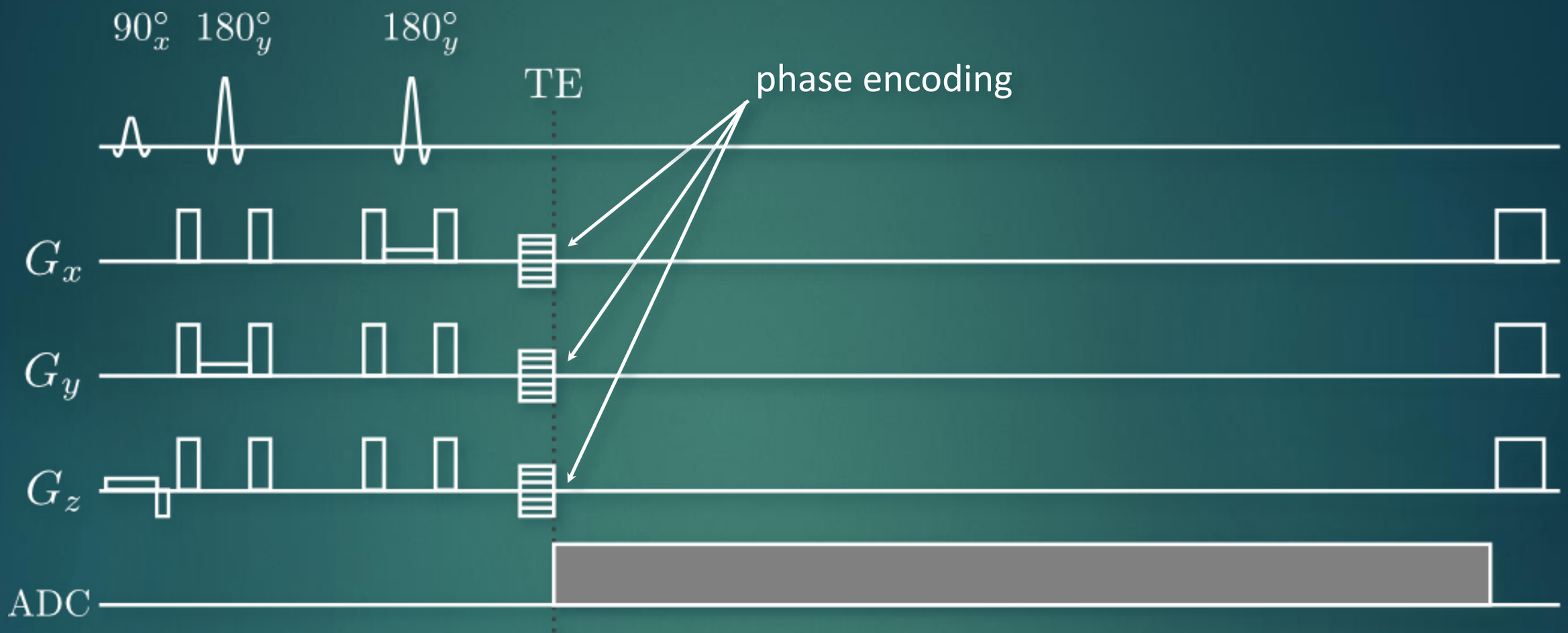
2D Spatial Encoding



sample 2D spectroscopic imaging pulse sequence

each FID is phase encoded along one dimension

3D Spatial Encoding



sample 3D spectroscopic imaging pulse sequence

each FID is phase encoded along one dimension

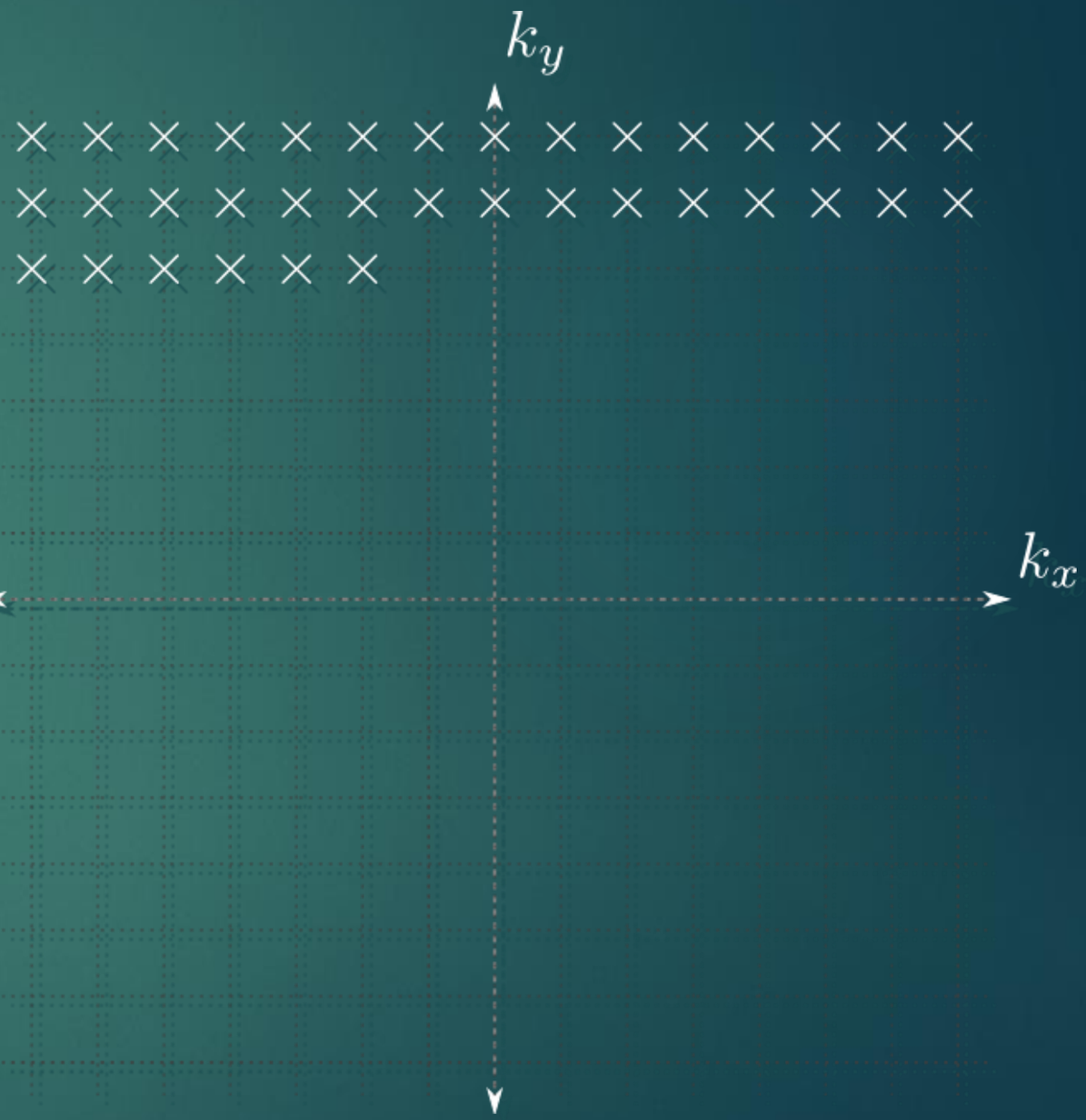
Brown 1982

Maudsley 1984

Sampling Considerations

constant time, vary amplitude

$$\Delta k_x = \frac{1}{\mathrm{FOV}_x} \quad \Delta k_y = \frac{1}{\mathrm{FOV}_y}$$



$$k_x = n\Delta k_x = \frac{n_x}{\mathrm{FOV}_x} = \gamma \mathrm{G}_x t$$

$$k_y = n\Delta k_y = \frac{n_x}{\mathrm{FOV}_y} = \gamma \mathrm{G}_y t$$

$$\mathrm{G}_x = \frac{n_x}{\gamma \mathrm{FOV}_x t} \quad \mathrm{G}_t = \frac{n_t}{\gamma \mathrm{FOV}_y t}$$

MRSI

Each k-space point is individually collected on a cartesian grid

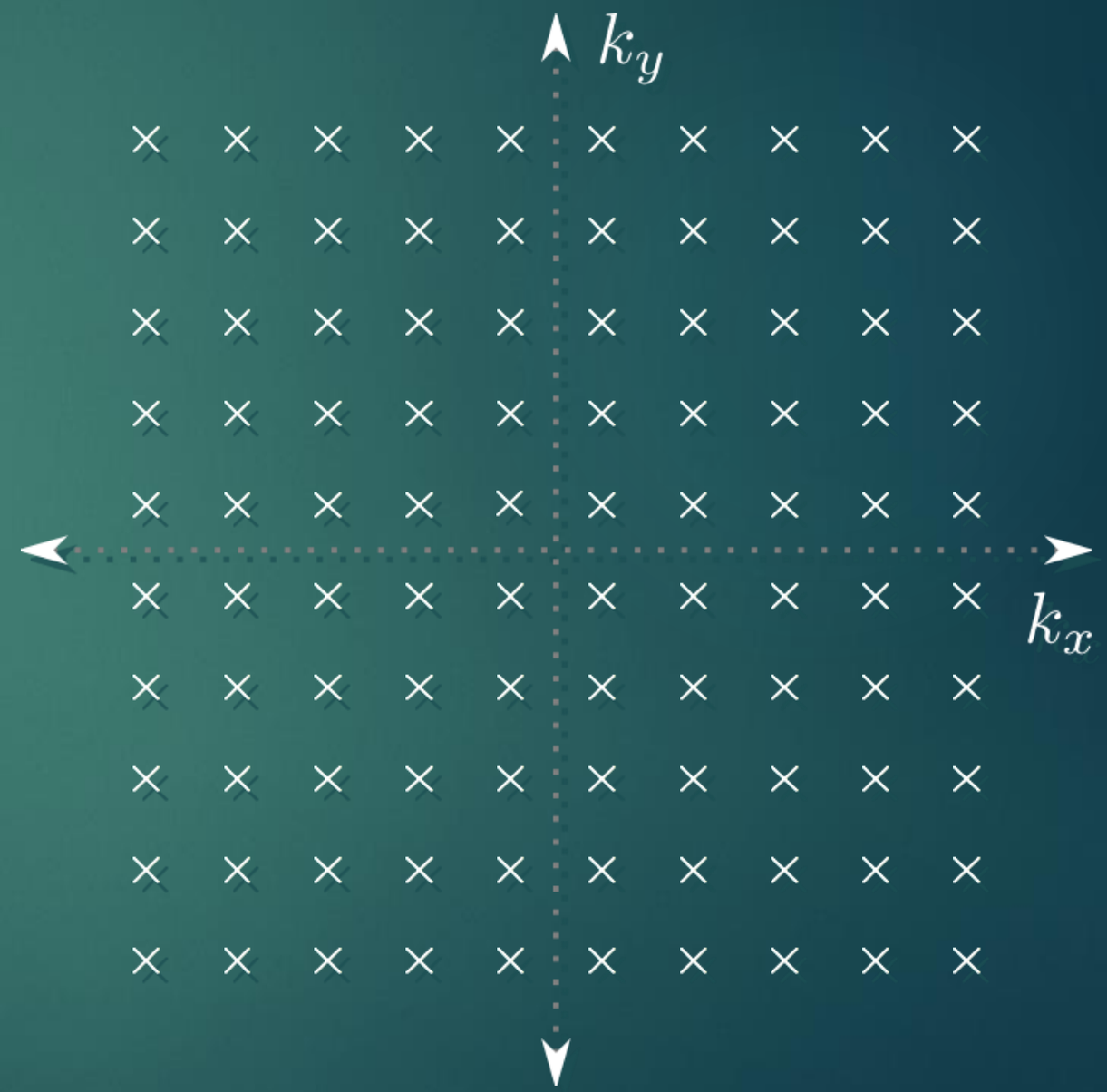


Image data is obtained by applying 2D FFT along spatial dimensions

Total acquisition time is thus

$$N_x \times N_y \times \mathrm{TR} \times \mathrm{NEX}$$

3D Spatial Encoding

the amount of time for a CSI scan is thus

$$N_x \times N_y \times N_z \times \mathrm{TR} \times \mathrm{NEX}$$

for a 32x32x1 scan (2D) with a TR = 1s and 1 average, the scan time is 17 minutes

for a 32x32x16x1 scan (3D) with a TR = 1s and 1 average, the scan time is 4.5 hours

without acceleration techniques, CSI is very slow and inefficient (and low res)

CHESS (global)

OVS (slice localized)

3D STEAMCSI PRESSCSI

Data Acquisition (N*Δt)

Recovery Time (TR)

MR Spectroscopic Imaging (2-3 Phase Encoded)

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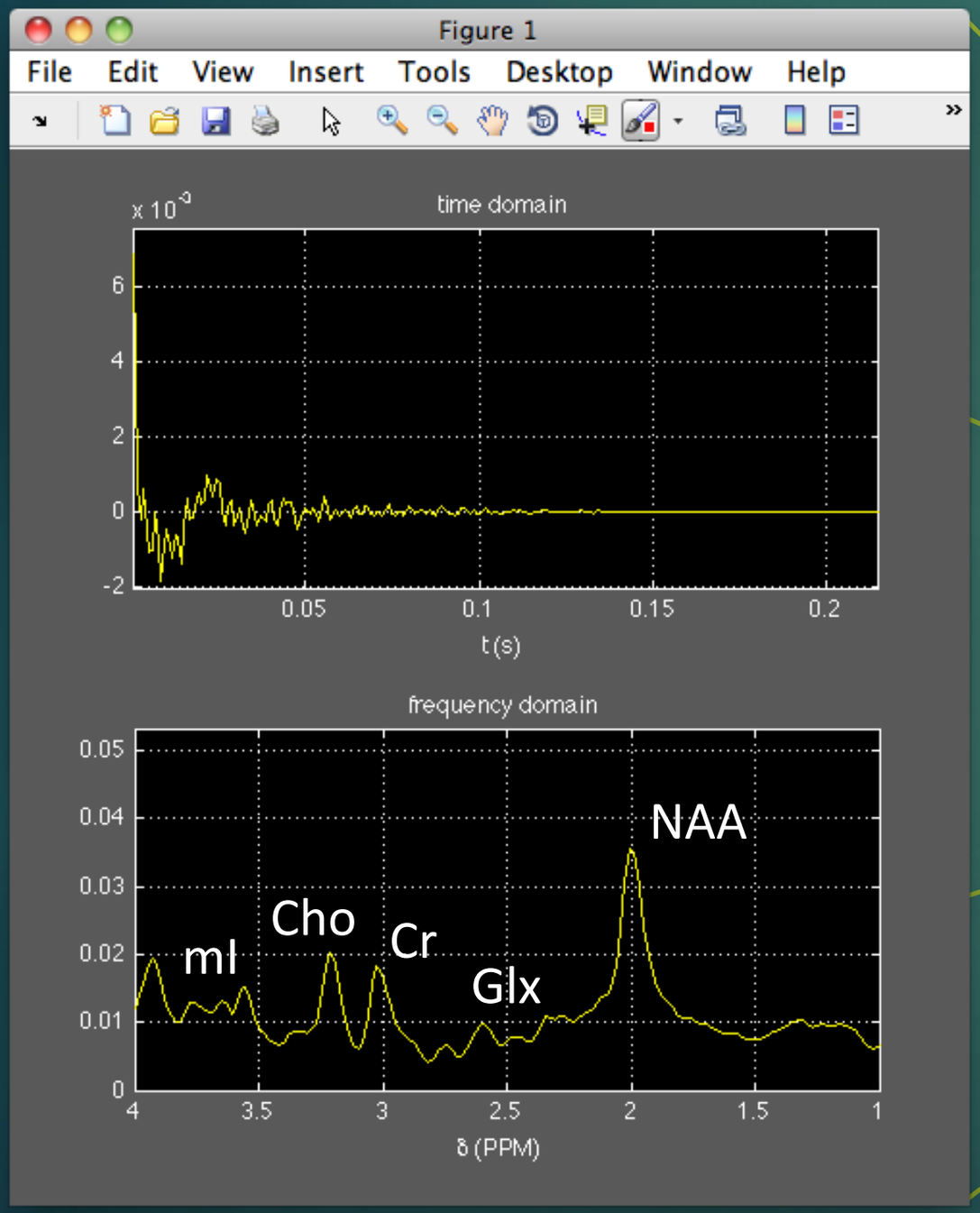
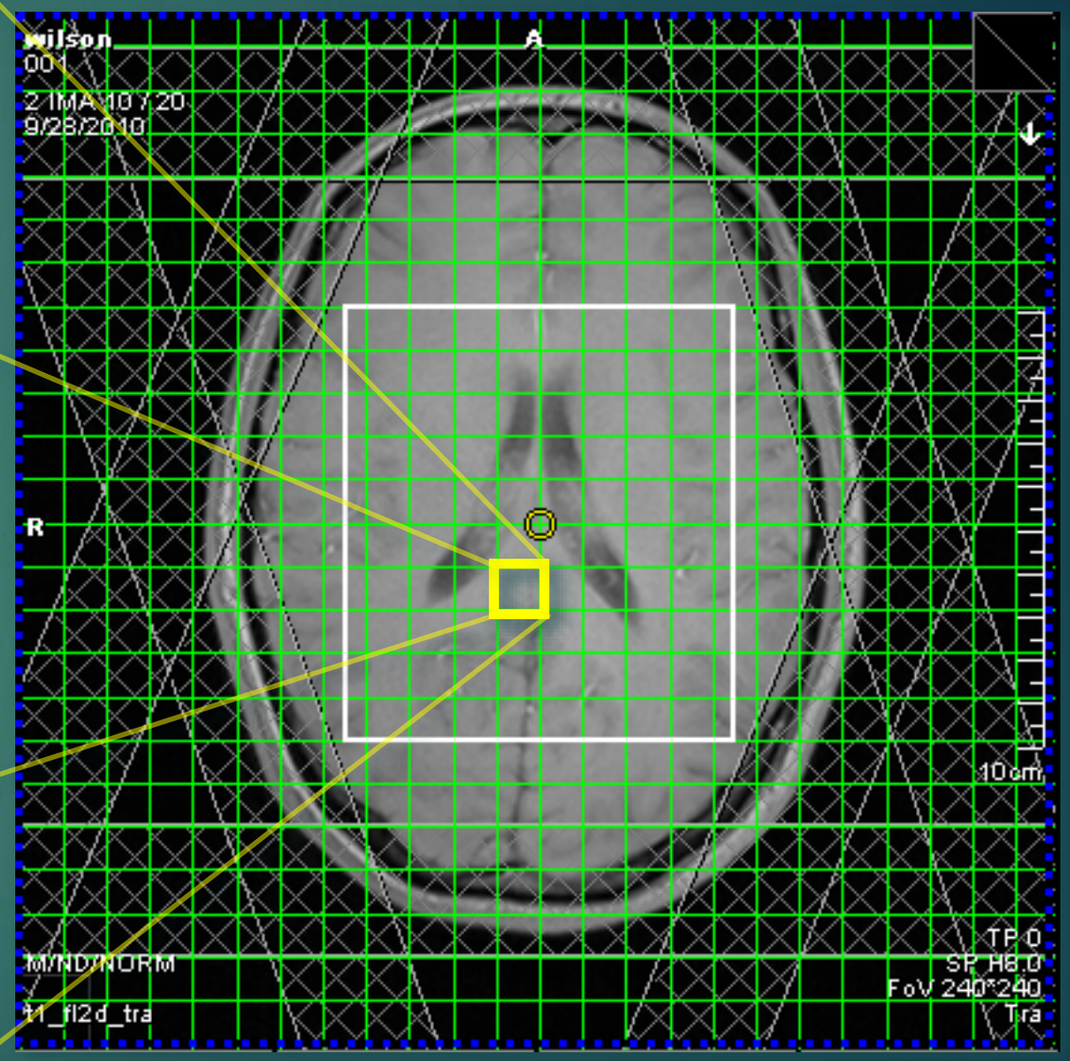


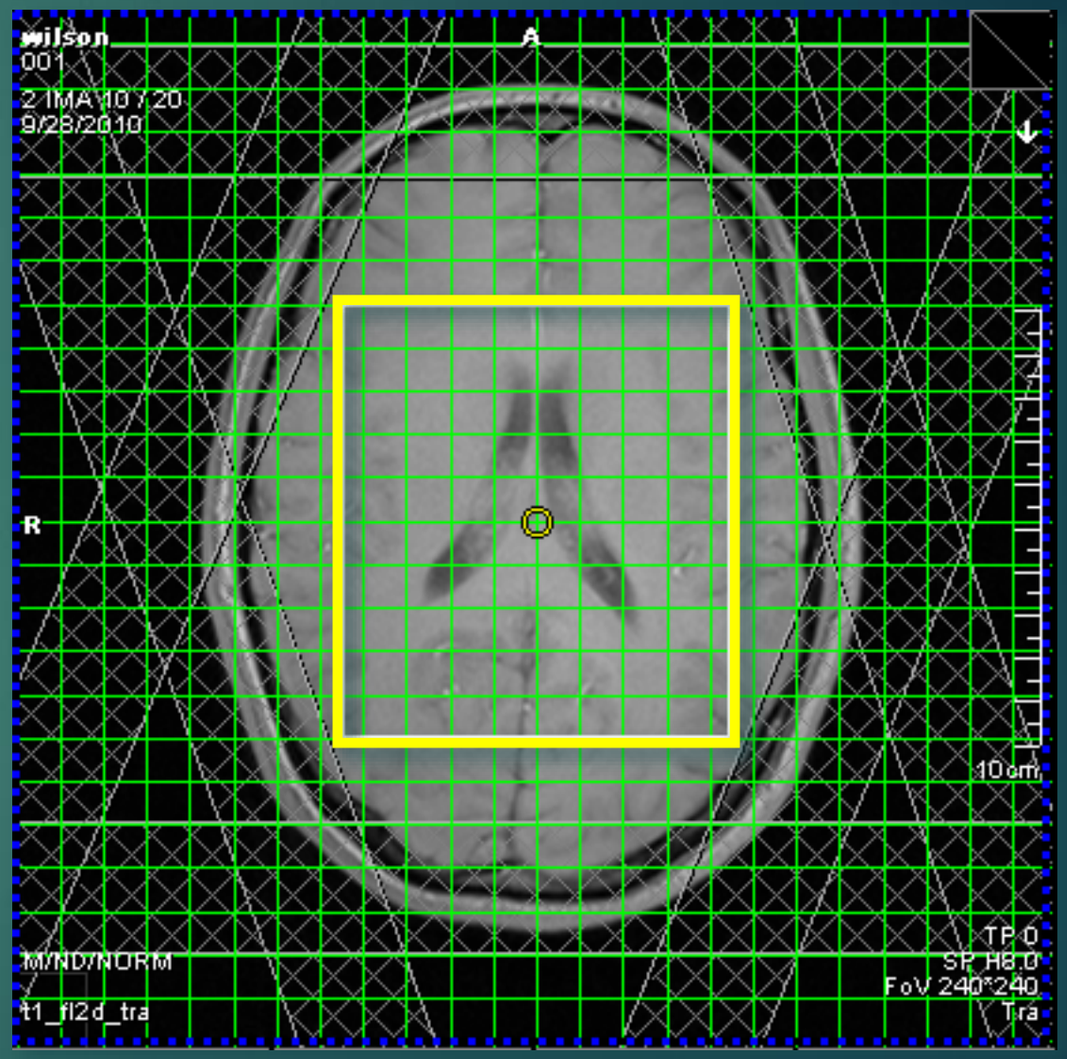
localized spectra



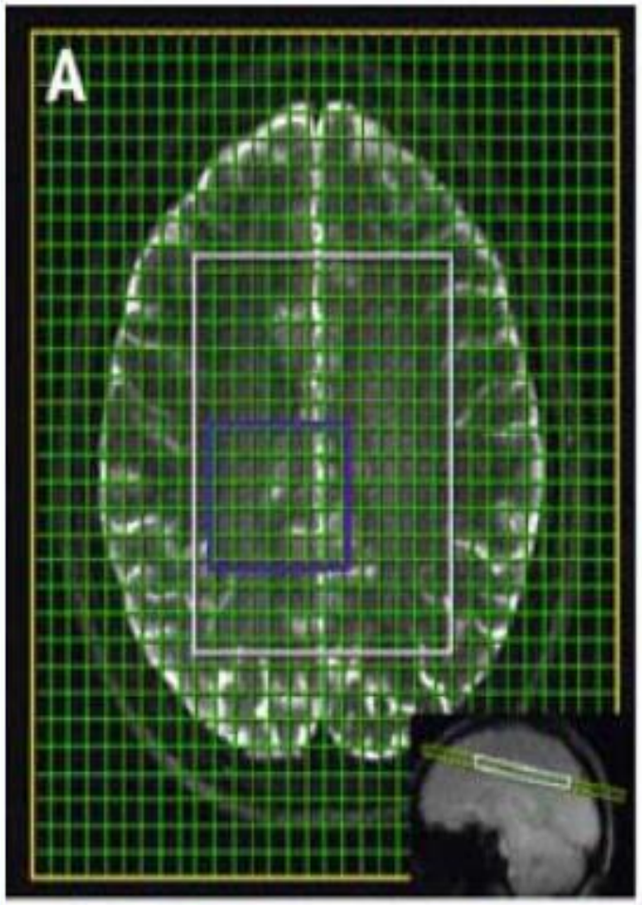
MR Spectroscopic Imaging (Cont'd.)

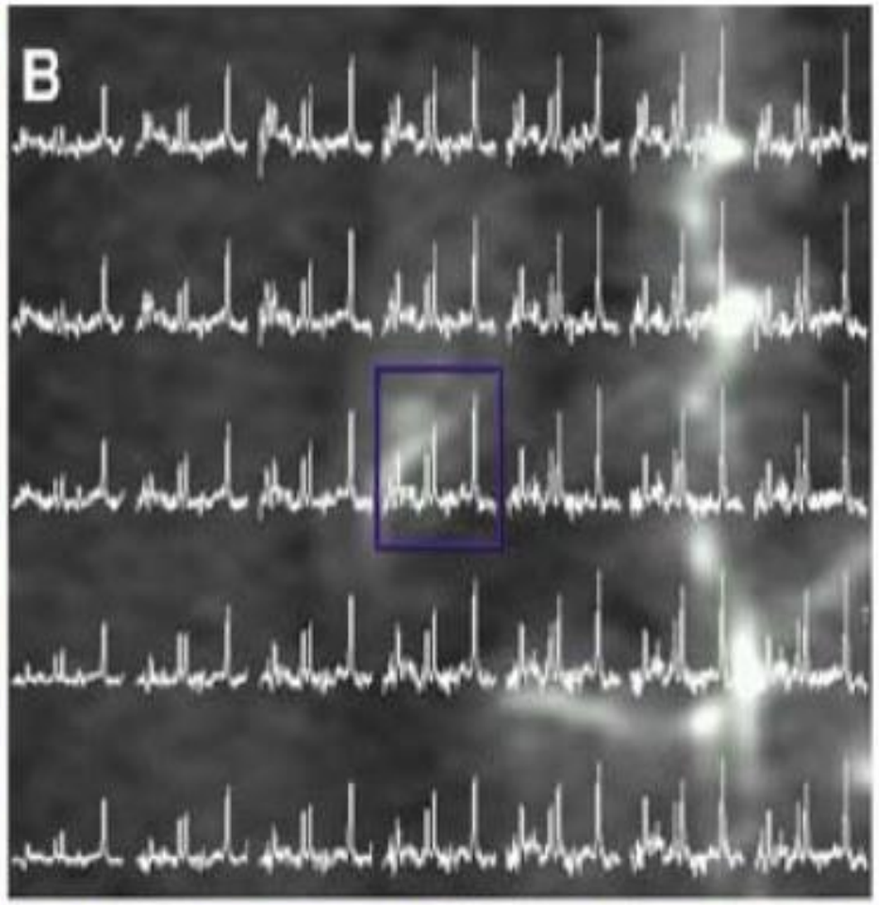
David Geffen School of Medicine

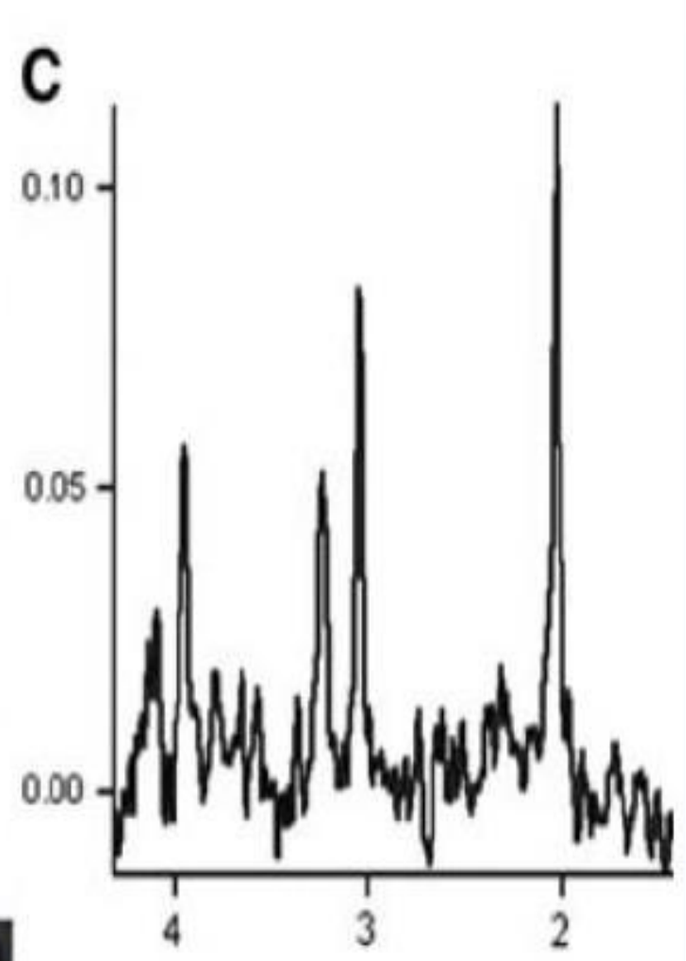
metabolite maps

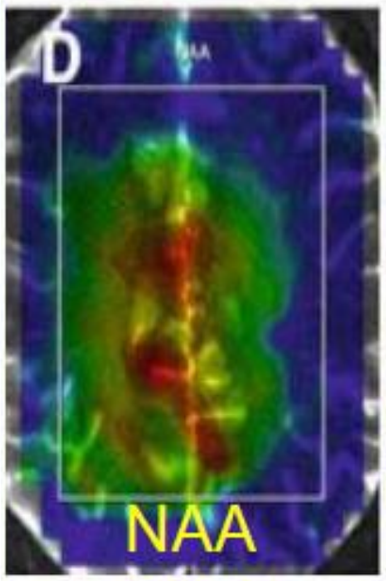


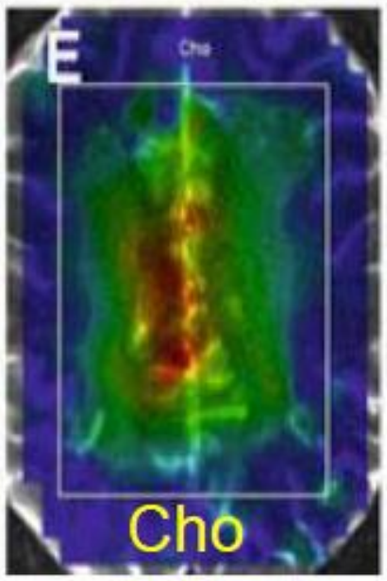
High resolution metabolite maps

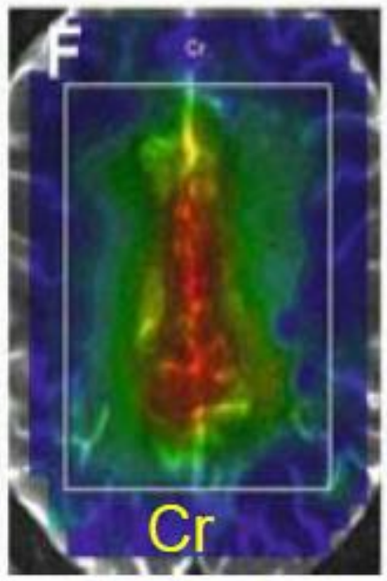


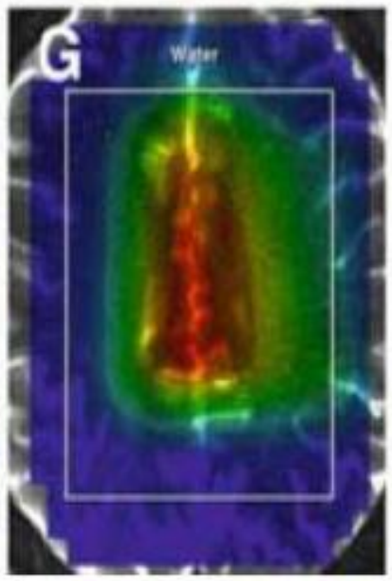












3D MRSI, TE 50,
0.14 cc voxel,
10mins
LASER seq, CP
head coil

Scheenen TWJ et al. Magn Reson Mat Phy Biol Med 21:95-101, 2008

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Metabolic mapping quality – 3T vs 7T

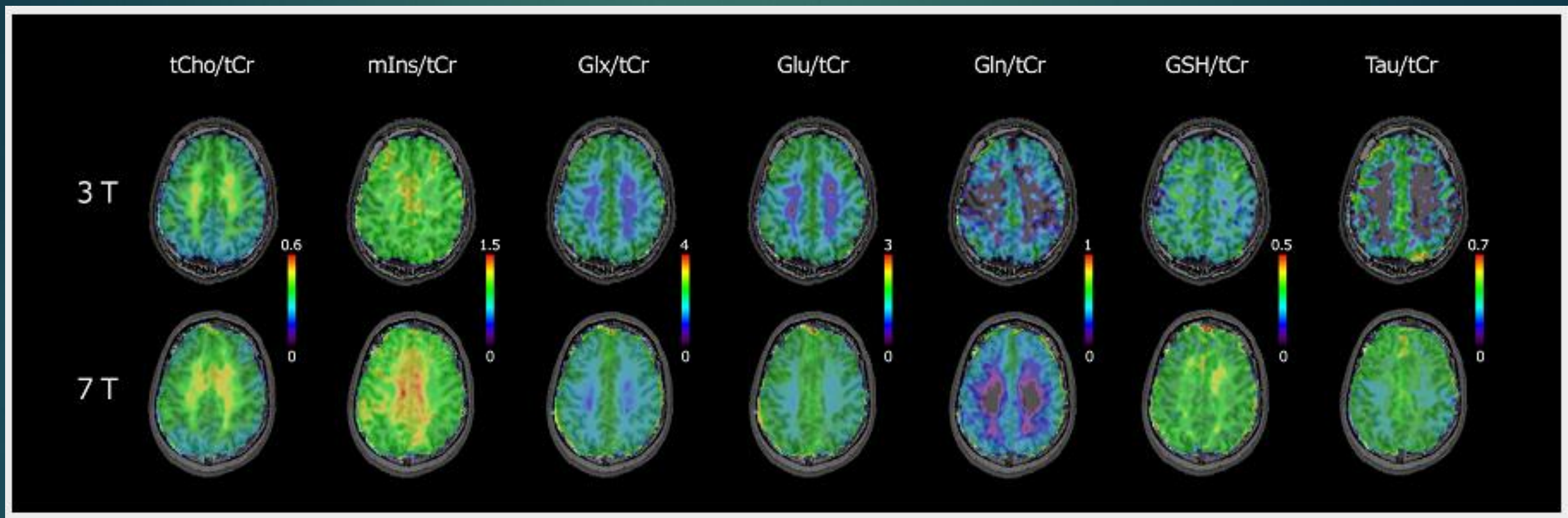


Figure 4. Metabolic maps acquired with FID-MRSI at 3T and 7T. Reliable quantification over the whole slice was possible for Glu/tCr, Gln/tCr, GSH/tCr and Tau/tCr at 7T but not at 3T. Values are displayed in a.u.

Heckova et al. ISMRM 2016

How long does it take to perform a multi- voxel 2D/3D MRSI?

2D MRSI (2 spatial+1spectral):
Total duration = TR*NEX*Nx*Ny
=1s*1*32*32= 17 minutes

3D MRSI (3 spatial+1spectral):
Total duration = TR*NEX*Nx*Ny*Nz
=1s*1*32*32*16= 4.53 hours
=1s*1*16*16*8= 34 minutes

Acceleration Techniques

The goal

- to reduce the number of excitations in order to reduce the total scan time (1-10 minutes)

The strategies

- Selective Averaging
- Parallel Imaging
- Turbo Spin Echo (TSE) techniques
- Echo-Planar (EP) techniques
- Concentric Ring Trajectories (SI-CONCEPT)
- Radial (Golden Angle View Ordering) and TV regularizer

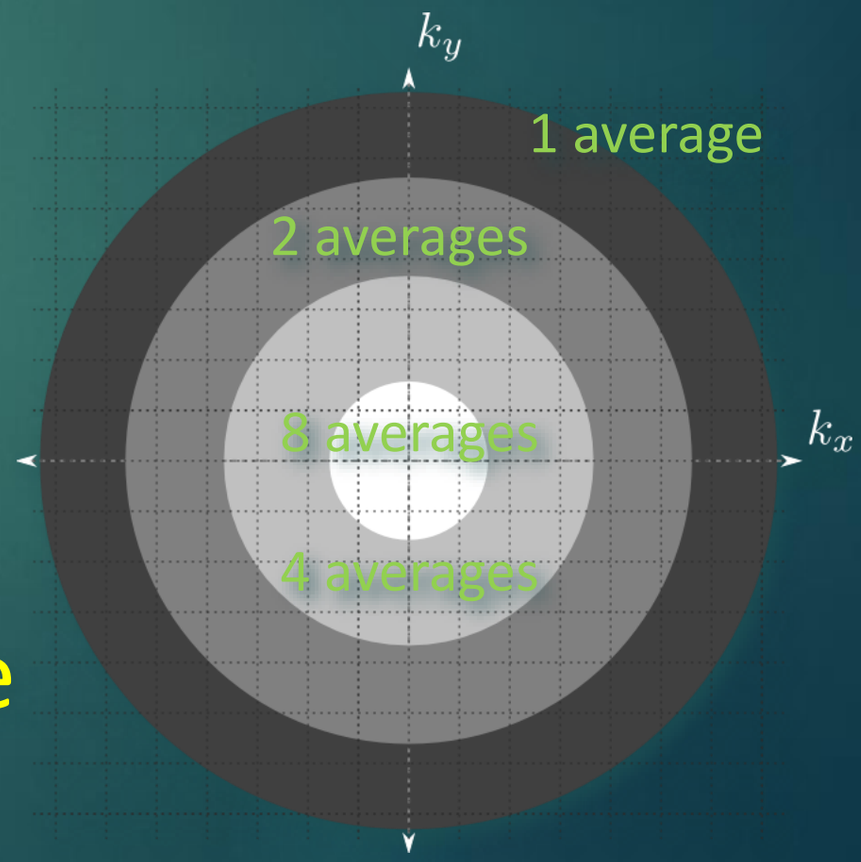
Fast MRSI

- Elliptical weighting
  - Reduced spatial sampling of k-space with only the central ellipsoid being acquired (reduction factor typically = 2)
- Parallel Imaging Reconstruction
  - Reduced acquisitions of k-space by increasing the spacing between k-space samples. Additional spatial information from multiple receiver coils is then used to increase the spatial FOV to the original size.

Selective Averaging

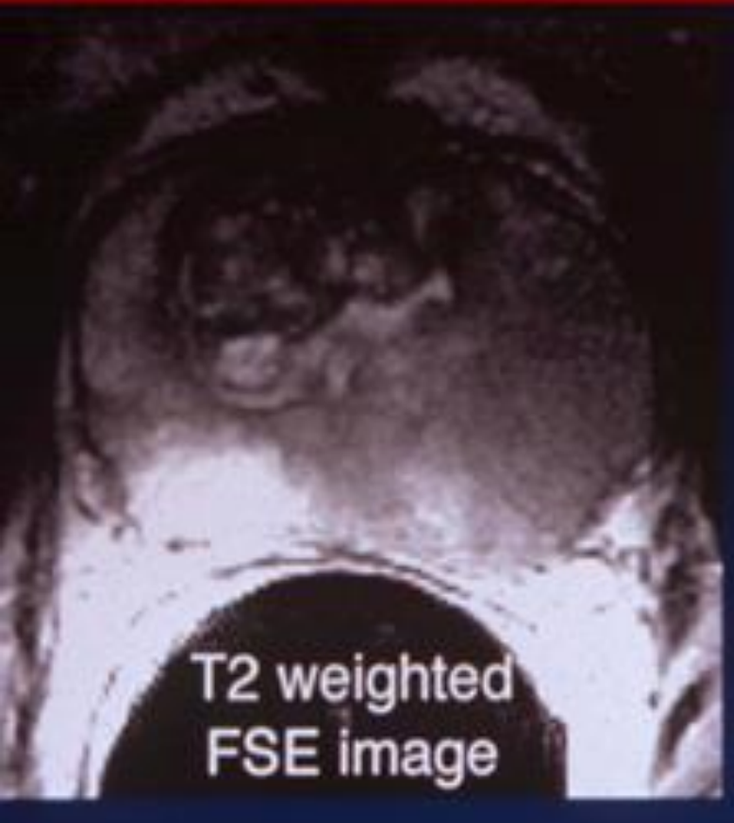
Average the parts of k-space with greater intensity

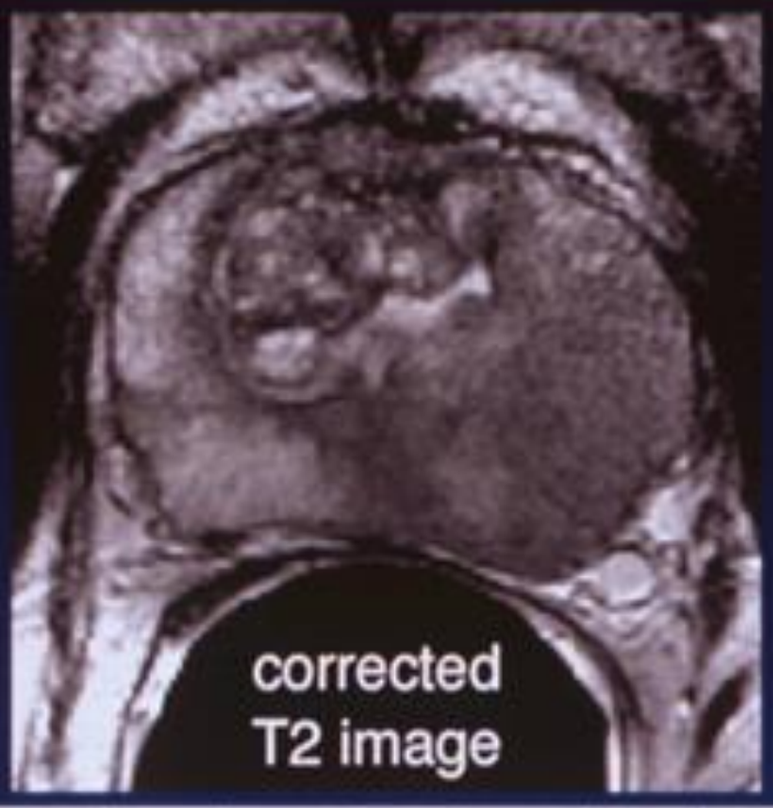
Significantly reduces total scan time

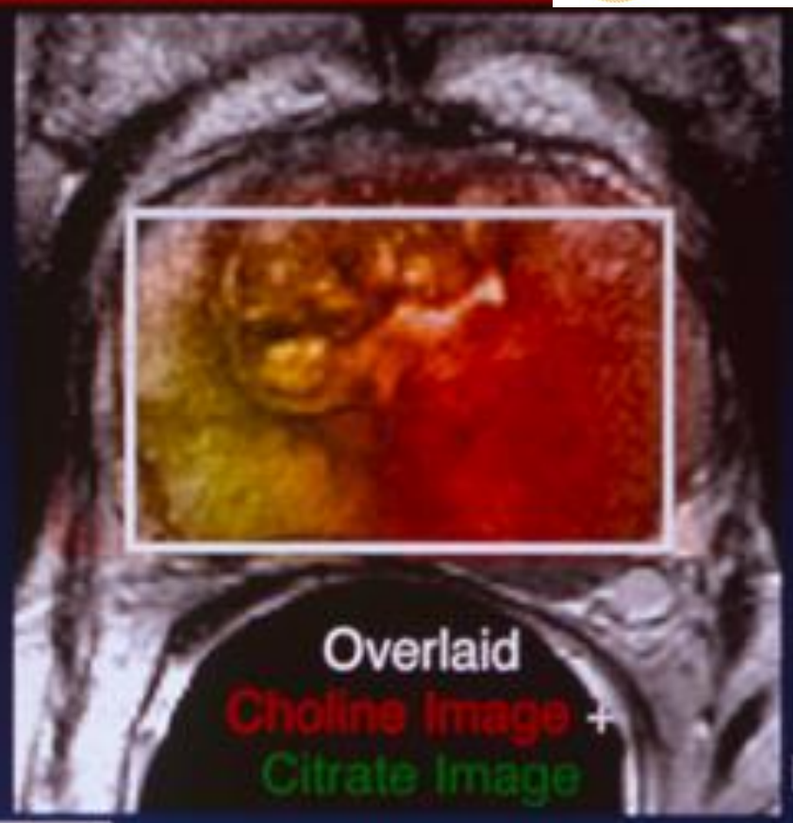


MRI/MRSI Data Display

David Geffen School of Medicine

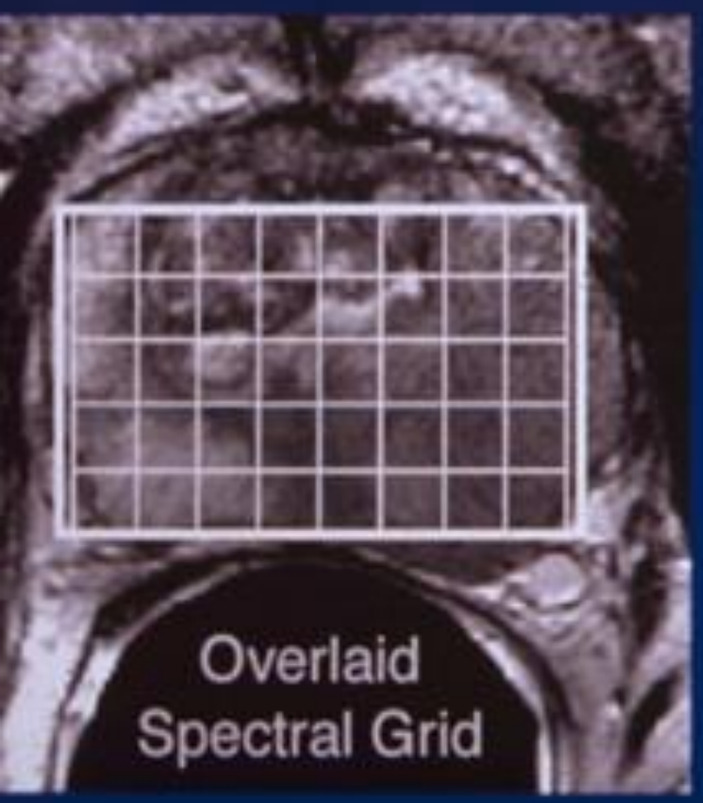


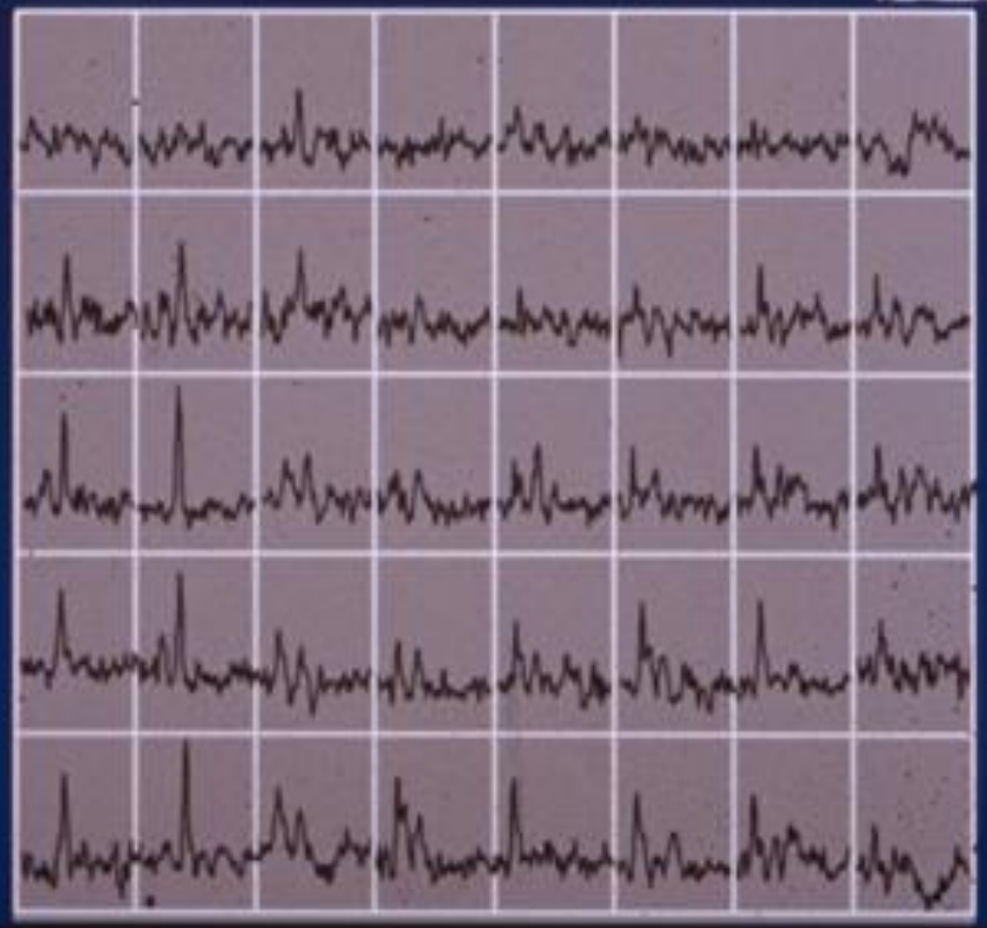


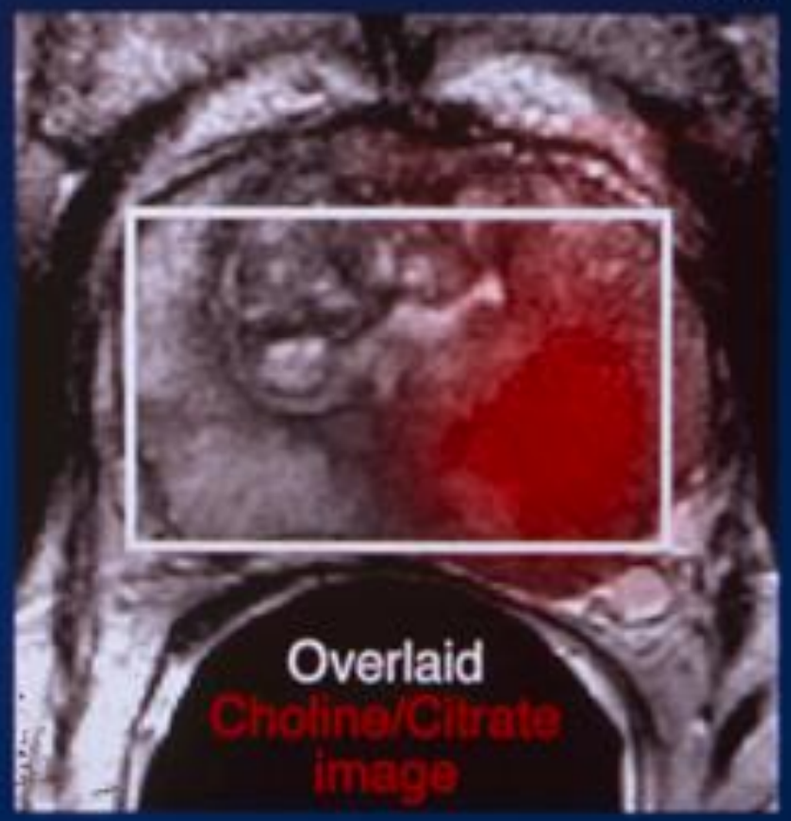


Cancer

Normal/
BPH







Parallel Imaging

Multi-coil reconstruction (SENSE/GRAPPA)

advantages

- reduced scan time

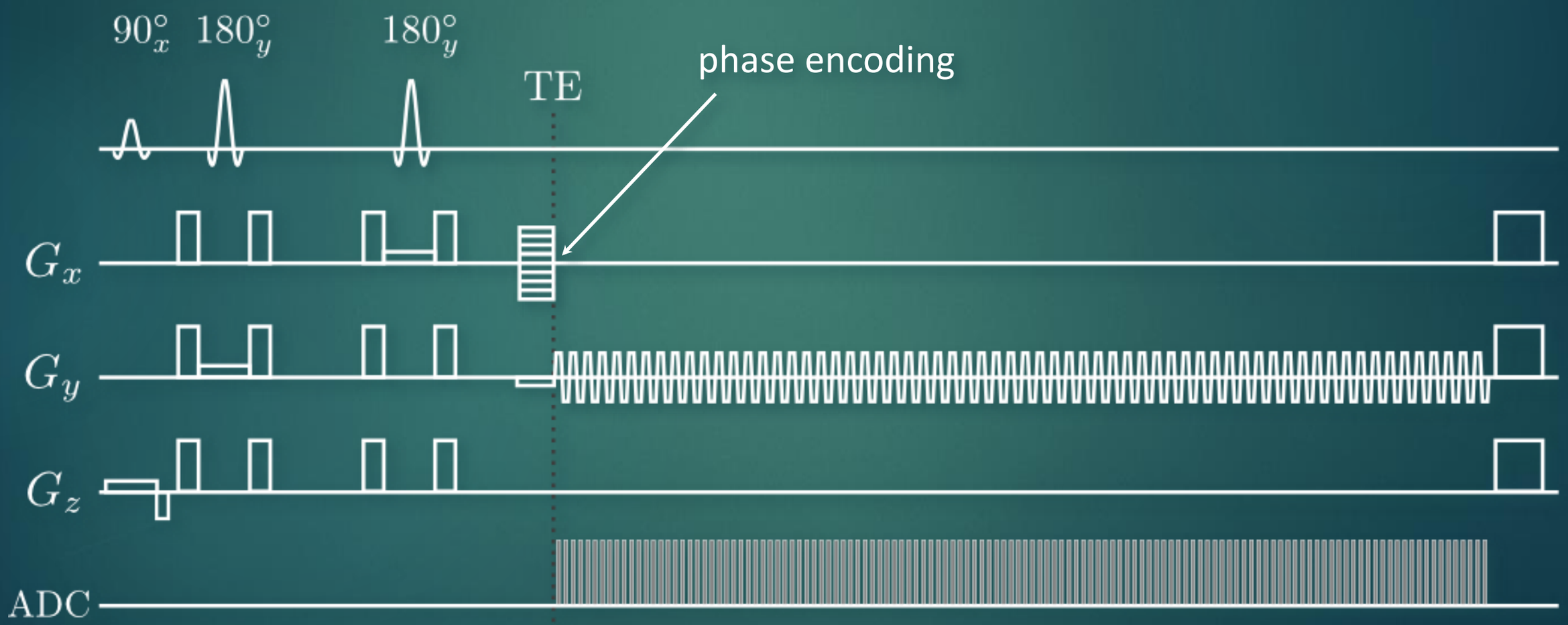
disadvantages

- reduced SNR from reduced number of excitations

Larkman and Nunes PBM 2007; Preussmann 1999; Sodickson 1999

Echo-Planar Spectroscopic Imaging (EPSI)

echo-planar spectroscopic imaging uses a repeated time-varying readout gradient to collect the same spatially encoded information as a function of time



what is the effect of the repeated bipolar gradient readout?

Mansfield 1984, Posse 1994, Lipnick 2008

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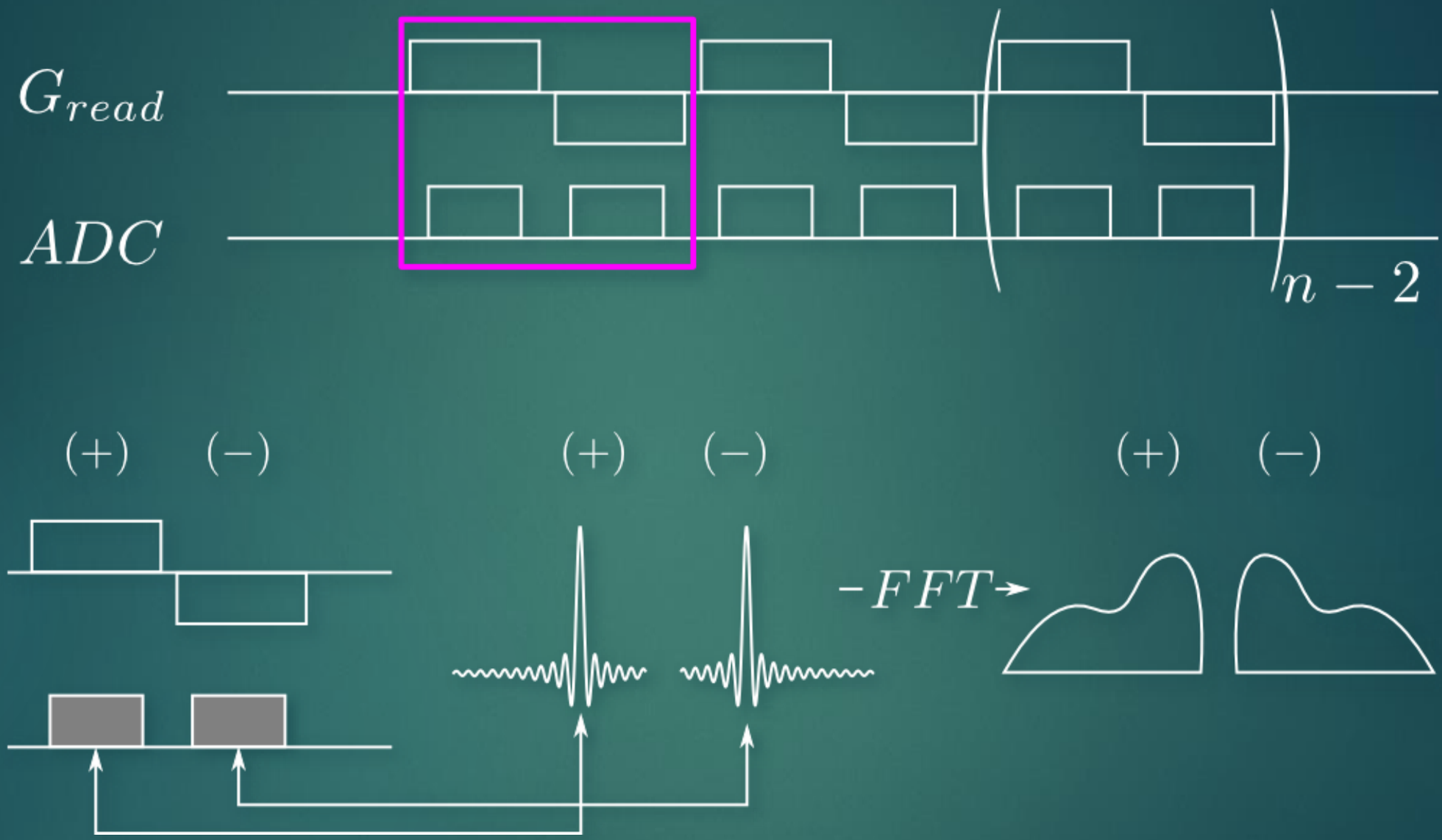
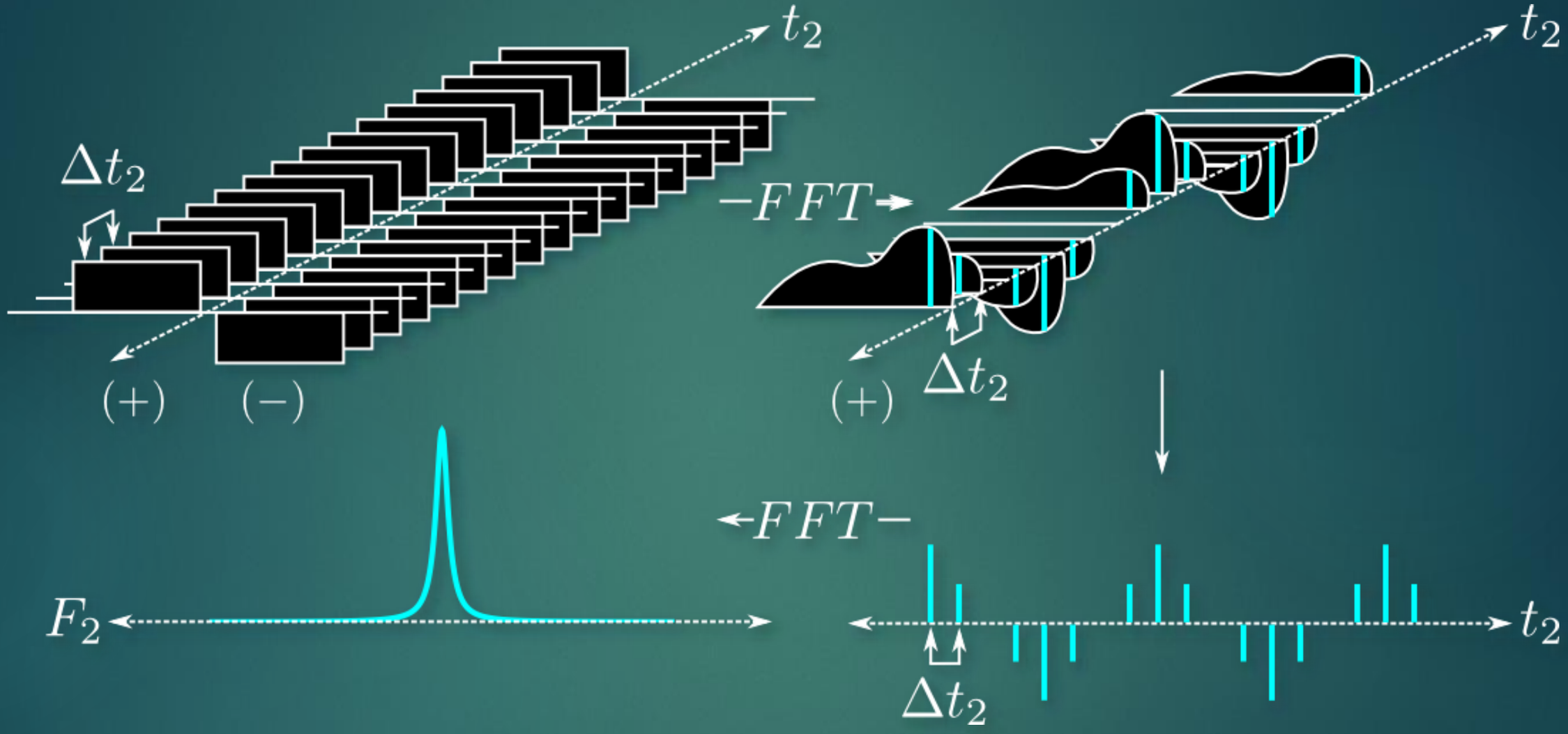
Echo-Planar SI



two sets of echoes (odd and even) form which are mirror images of each other

Mansfield 1984, Posse 1994, Lipnick 2008

Echo-Planar SI



the repeated nature of the readout gradients spatially encodes as a function of time

Mansfield 1984, Posse 1994, Lipnick 2008

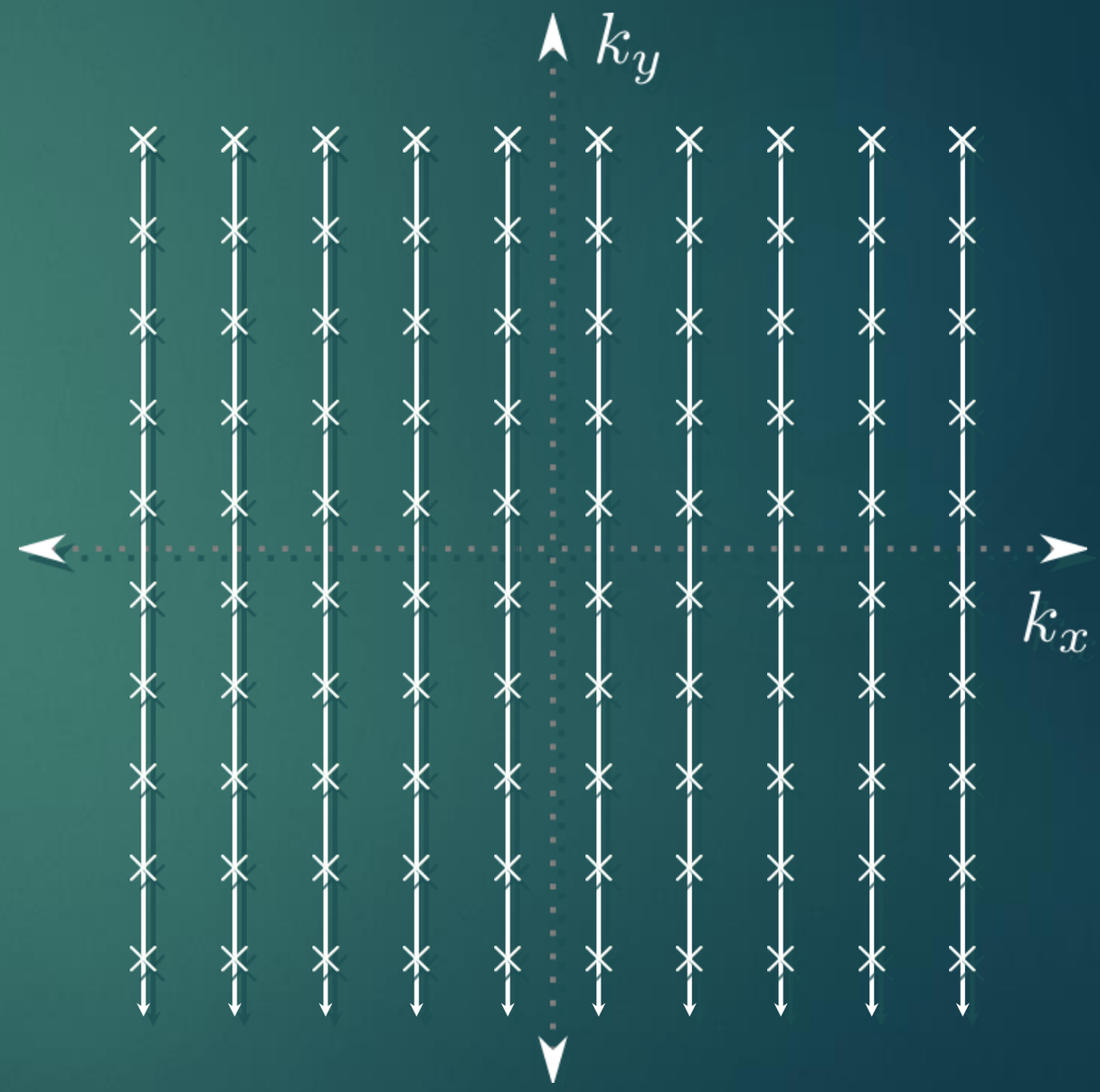
EPSI

A single line in k-space is collected in a single excitation

Image data is obtained by applying 2D FFT along spatial dimensions

Total acquisition time is thus

$$N_x \times \mathrm{TR} \times \mathrm{NEX}$$



Mansfield 1984, Posse 1994, Mulkern 2000, Lipnick 2008

Echo-Planar SI

the amount of time for a MRSI scan is thus

$$N_x \times N_z \times \mathrm{TR} \times \mathrm{NEX}$$

for a 32x32x16 scan (3D) with a TR = 1s and 1 average, the scan time is 8.5 minutes

Using all 3 phase-encoding,
3D MRSI (3 spatial+1spectral):
Total duration = TR*NEX*Nx*Ny*Nz
=1s*32*32*16= 4.5 hours

significant reduction in scan time!

Echo-Planar SI

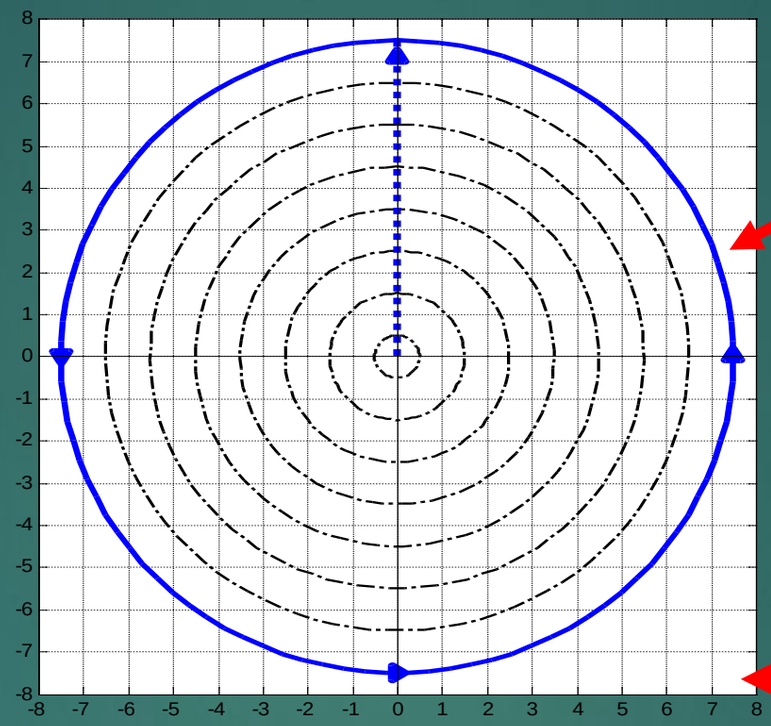
advantages

- significantly reduced scan time

disadvantages

- echo-planar readout creates undesired eddy currents which can distort spectra
- reduced SNR from reduced number of excitations
- very demanding on the hardware (reduced spectral bandwidth)

Why Concentric Circular sampling?



- More efficient k-space sampling due to symmetry of concentric circles → half the number of excitations required for similar k-space coverage

- Outer corners of k-space contain little signal and are usually filtered away anyway

Matsui 1986, Zhou 998, Wu 2009; Furuyama 2012

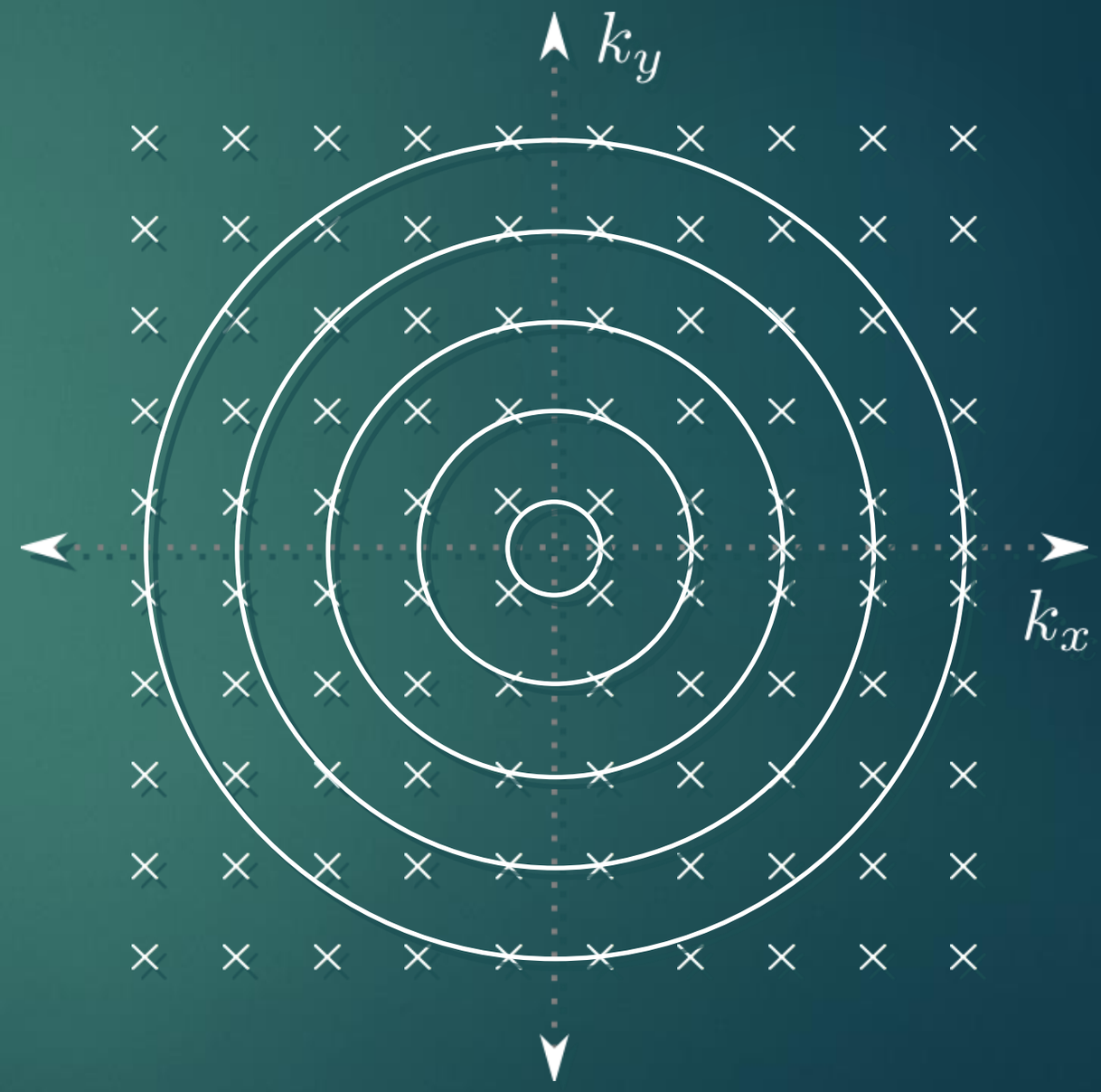
Concentric Circles (SI-CONCEPT)

A single ring in k-space is collected in a single excitation

Image data **cannot** be processed by 2D FFT since it is not cartesian

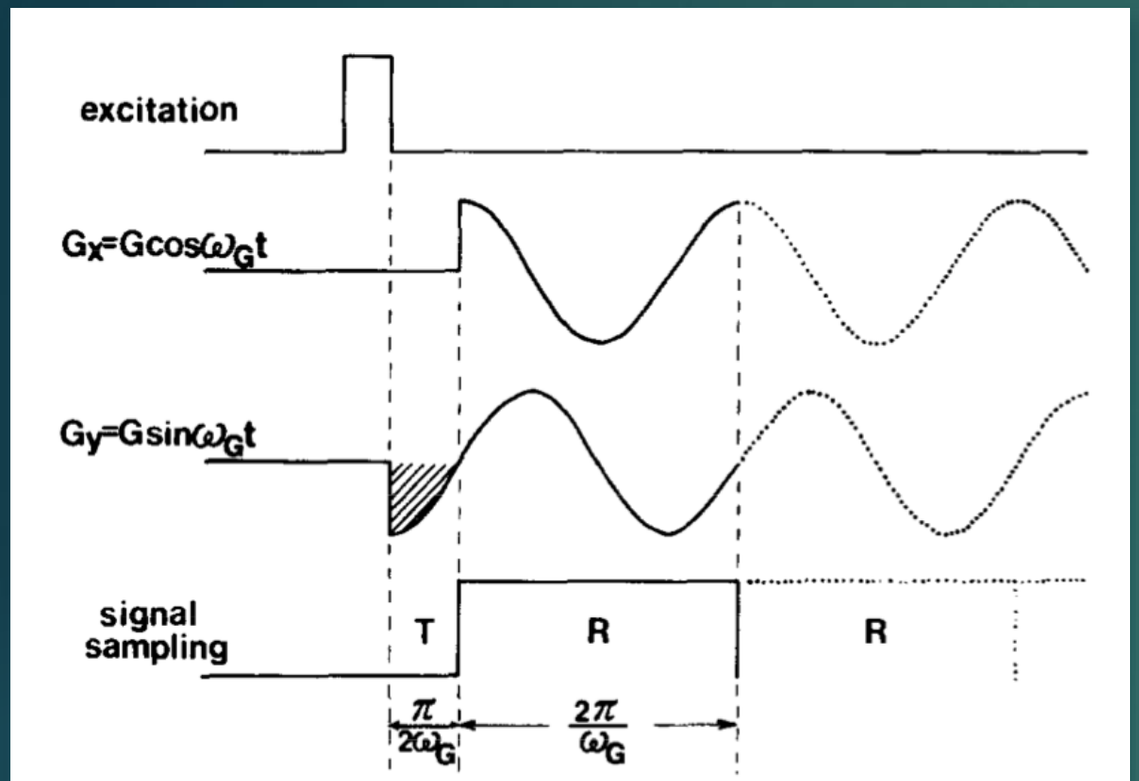
Total acquisition time is thus

$$\frac{1}{2} N_x \times \mathrm{TR} \times \mathrm{NEX}$$



JK Furuyama, NE Wislon, MA Thomas MRM 2012

What is Concentric Circular ?



Circular k-space trajectory defined by

$$k_x(t) = -k_n \sin\left(\frac{2\pi}{T}(t - TE)\right)$$

$$k_y(t) = +k_n \cos\left(\frac{2\pi}{T}(t - TE)\right)$$

where k_n is the radius of the *n*th ring and T is the spectral dwell time in the direct dimension

Gradient waveforms are thus given by

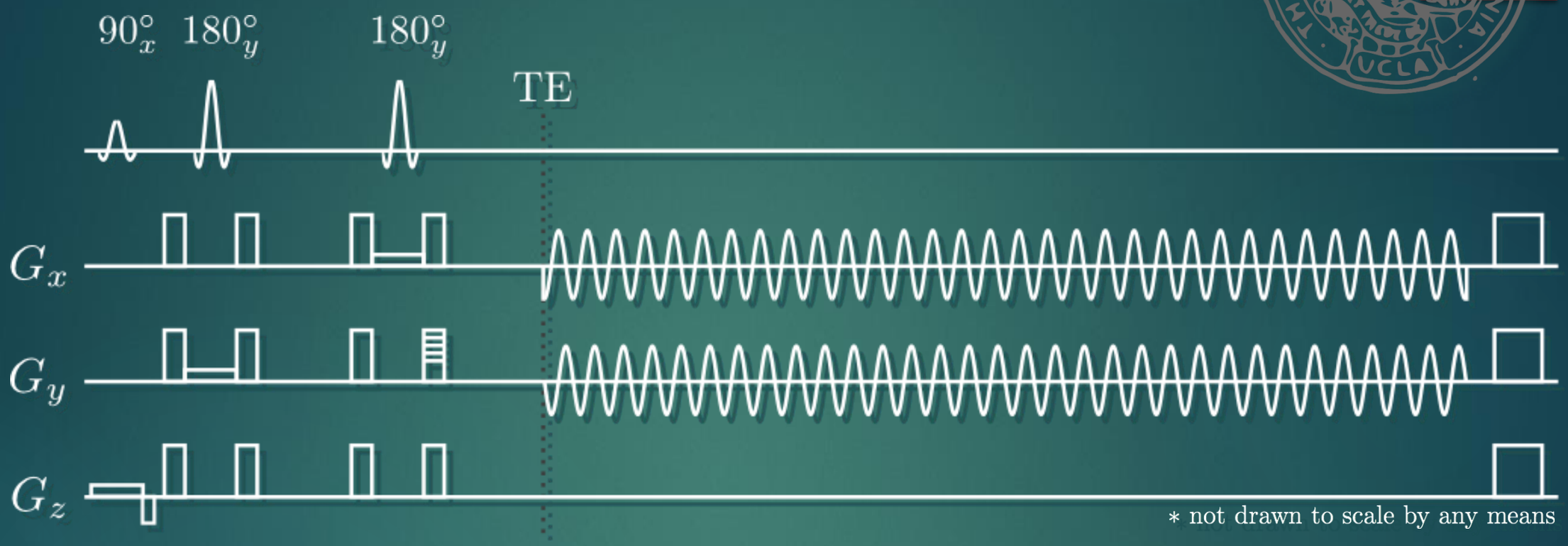
$$G_x(t) = -\frac{4\pi^2 k_n}{\gamma T}\cos\left(\frac{2\pi}{T}(t - TE)\right)$$

$$G_y(t) = -\frac{4\pi^2 k_n}{\gamma T}\sin\left(\frac{2\pi}{T}(t - TE)\right)$$

Matsui S, Kohno H, *J. Magn. Reson.* **70**, 157-162 (1986)

JK Furuyama, NE Wilson, MA Thomas MRM 2012

SI-CONCEPT Pulse Sequence



The use of a concentric k-space trajectory
is readily applied to ordinary CSI sequences

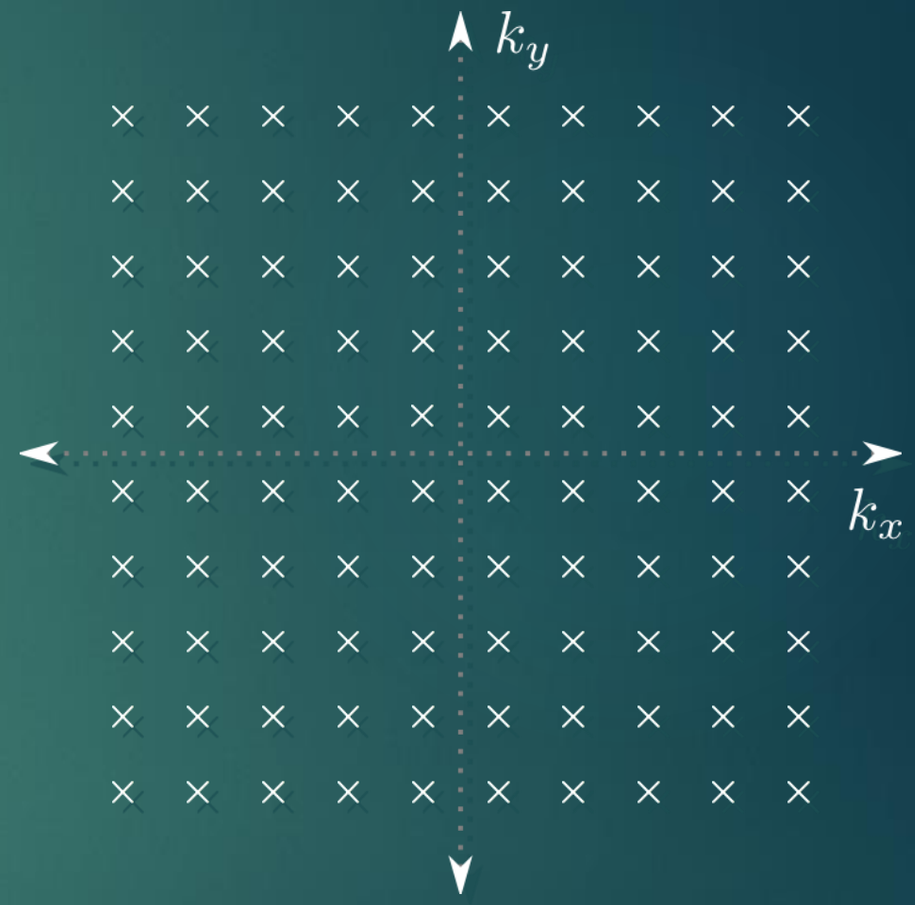
Repeatedly tracing the same circle in k-space
encodes both spatial and spectral information

JK Furuyama, NE Wislon, MA Thomas MRM 2012

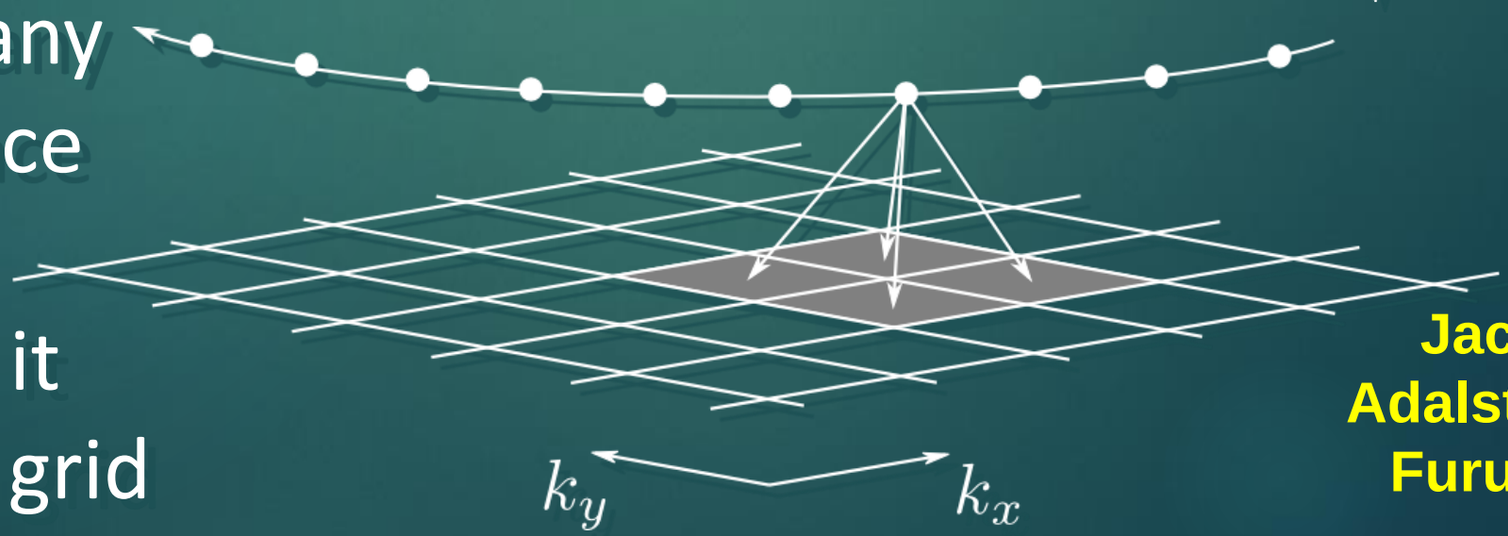
Concentric Circles

Convert polar data to cartesian?

how?

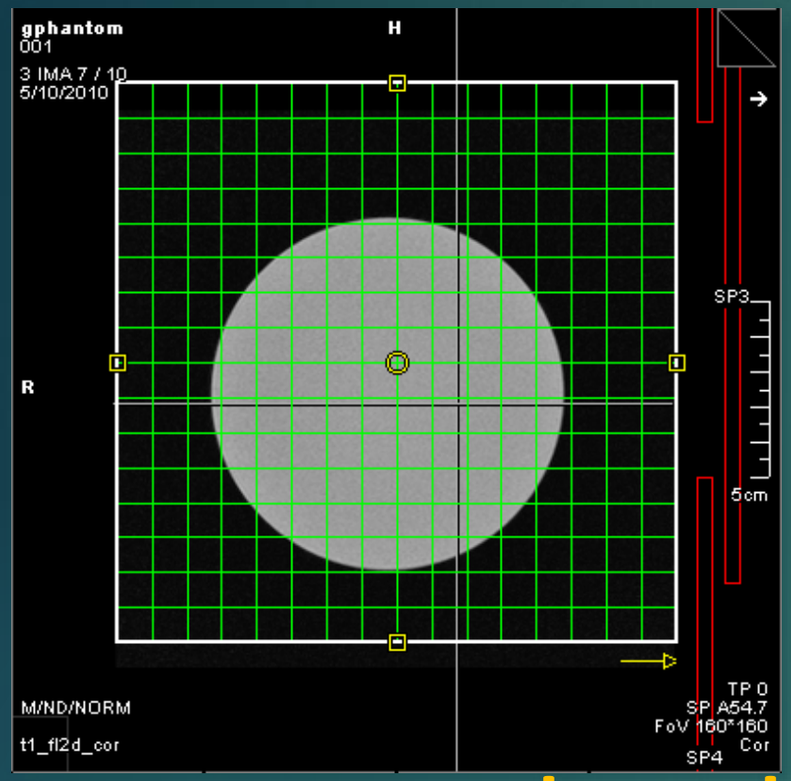


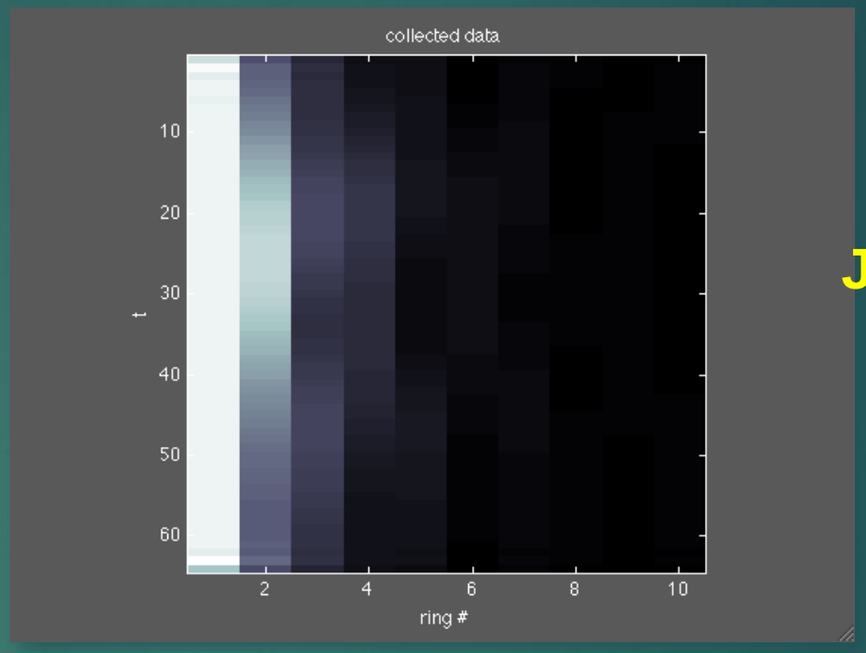
Gridding takes any arbitrary k-space trajectory and convolves it onto a cartesian grid



Jackson 1991
Adalsteinsson 1998
Furuyama, 2012

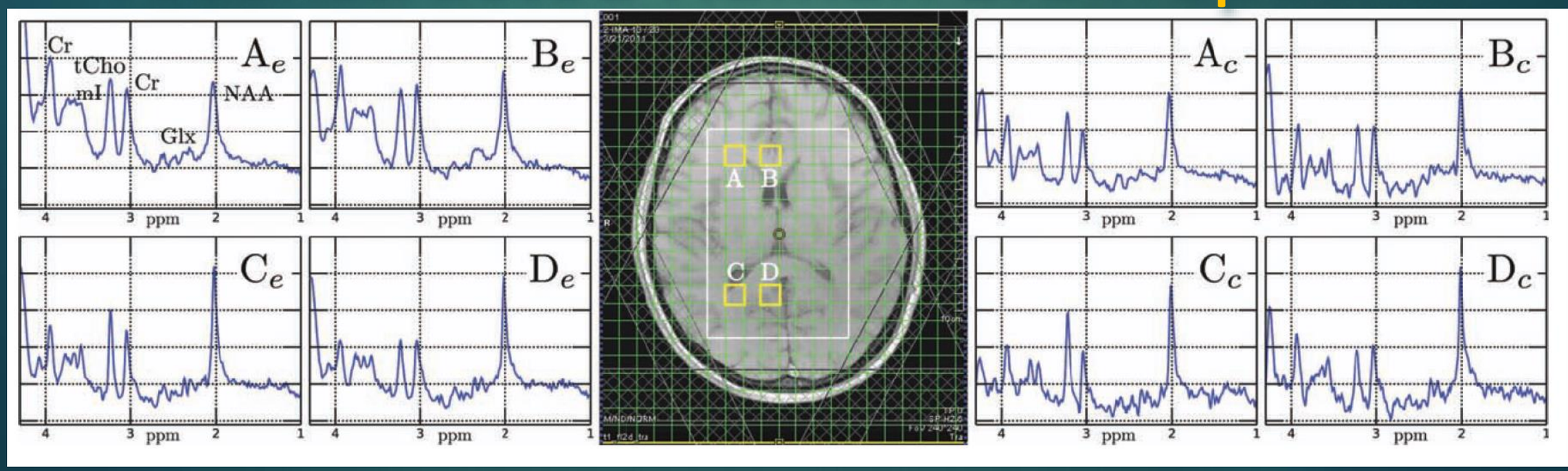
Concentric Circular Imaging
- Polar Data





JK Furuyama, NE Wislon, MA Thomas MRM 2012

- reconstructed Human Brain Spectra



MRSI vs. EPSI vs. SI-CONCEPT

$N_x \times N_y \times \mathrm{TR} \times \mathrm{NEX}$ -MRSI

$N_x \times \mathrm{TR} \times \mathrm{NEX}$ -EPSI

$\frac{1}{2} N_x \times \mathrm{TR} \times \mathrm{NEX}$ -SI-CONCEPT

Faster!

Hingerl et al. Inv Rad. 2020
*…..Brain coverage among all measured matrix sizes ranging from a 32 × 32 × 31 matrix with 6.9 × 6.9 × 4.2 mm nominal voxel size acquired in ~**3 minutes** to an 80 × 80 × 47 matrix with 2.7 × 2.7 × 2.7 mm nominal voxel size in ~**15 minutes** for different brain regions.*

Emir and coworkers, MRM 2020
*"A density-weighted concentric-ring trajectory metabolite-cycling MRSI technique was implemented to collect data with a nominal resolution of 0.25 mL within **3 minutes and 16 seconds**."*

Advantages of Concentric Circular Trajectories

- Less demanding on gradient hardware → higher spectral BW achievable (required at higher field strengths to prevent spectral aliasing)
- Eddy currents not as severe especially for *inner* k-space data
- Continuous readout during acquisition (EP-COSI without ramp sampling only samples during ~75% readout)
- Inherently less sensitive to motion artifacts
- **Lower maximum slew rates for equal resolutions and spectral BW**
- (> 50% less for actual scan parameters used)

Drawbacks

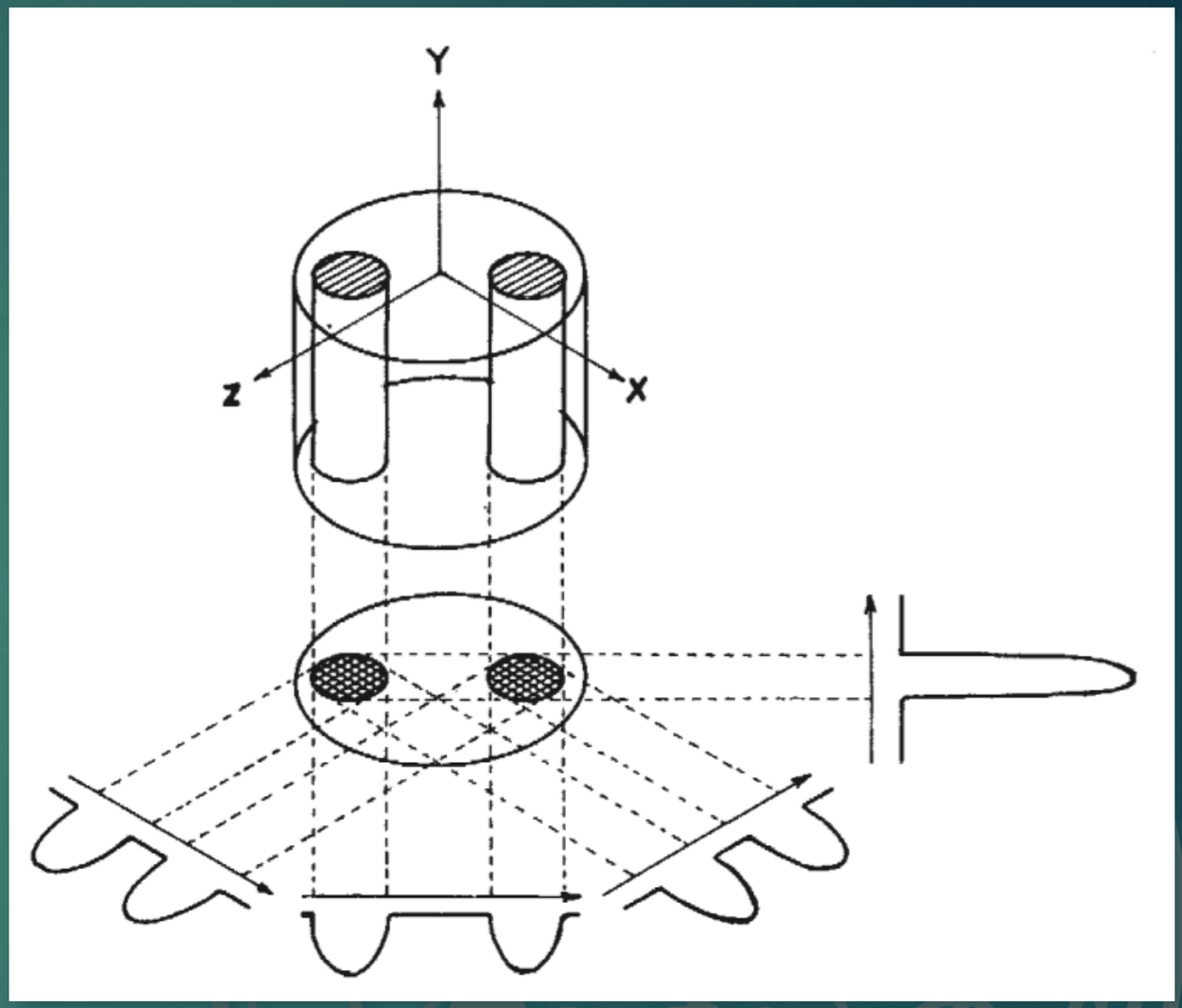
- Sampling during time-varying readout gradients leads to increased noise variance[1]
  - SNR gains from averaging compensate so that sensitivity per time in both sequences is similar

- More complicated post-processing
  - Data must be regridded in order to apply FFT
  - Alternatively, projection-reconstruction (PR) algorithms can be applied using inverse radon transform

Emir 2017; Chew 2018;
Steel 2018;; Kodibagkar 2019; Hingerl 2020

1) Pipe J and Duerk J, MRM 1995

Further Acceleration???
Projection Reconstruction/Radial

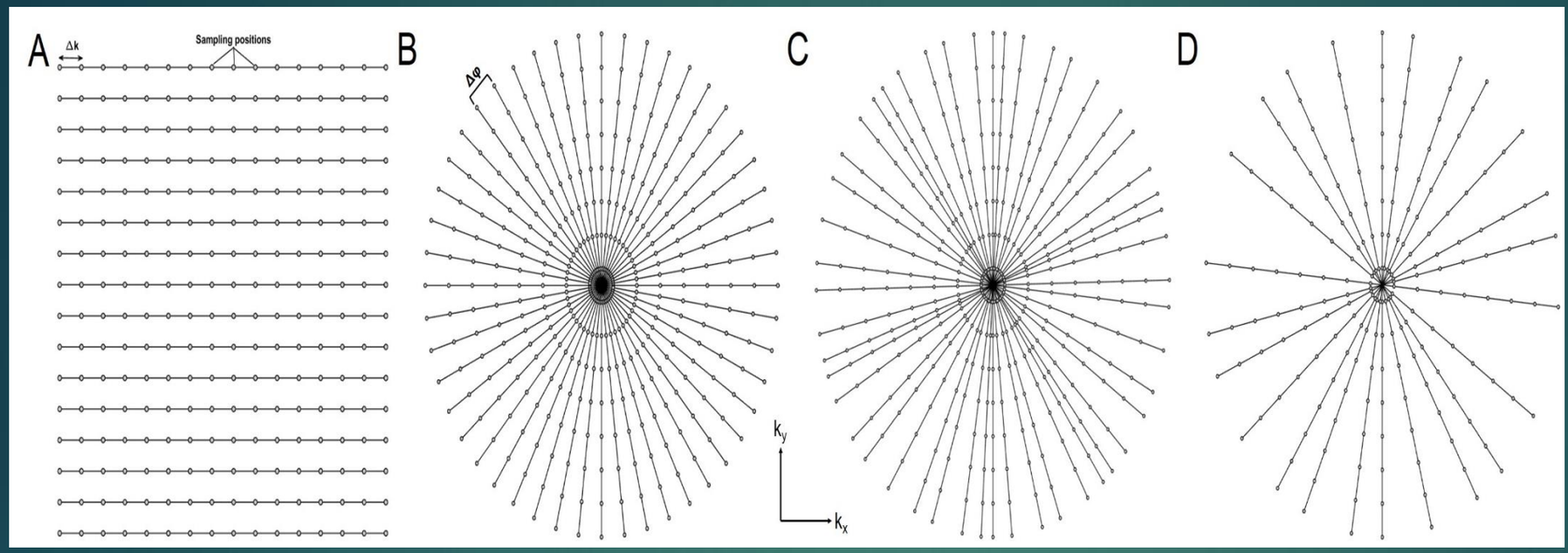


Lauterber P, *Nature* **242**, 190-191 (1973)

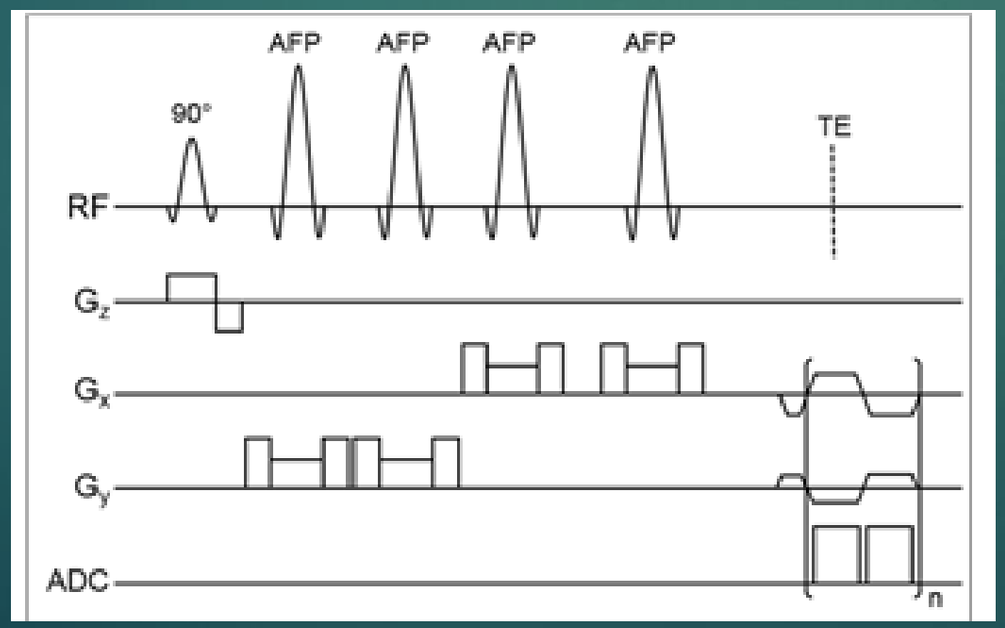
Original MRI Sequence

Series of projections taken at different angles

Radial Spectroscopic Imaging

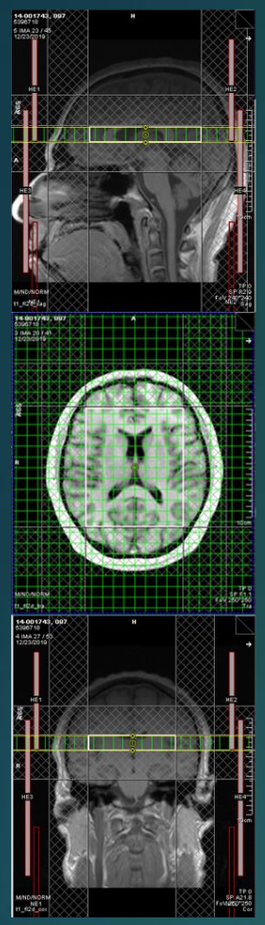


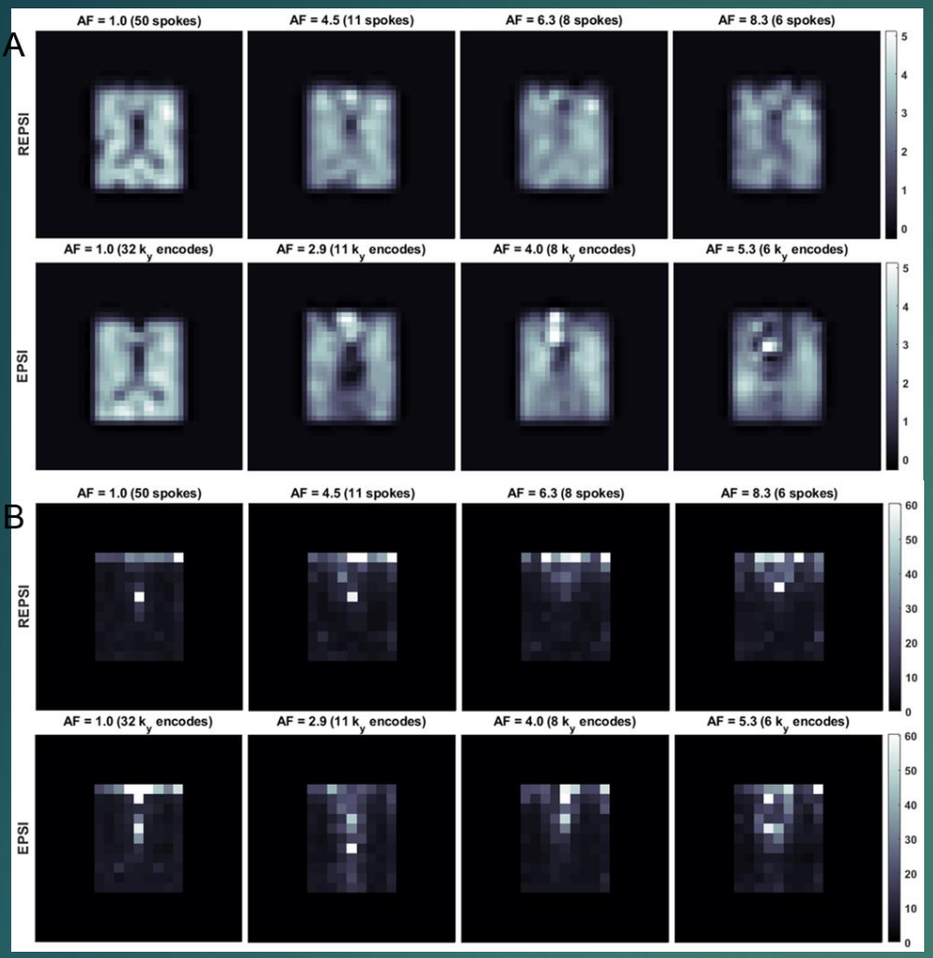
Sampling schemes using (**A**) Cartesian encoding; (**B**) radial encoding; (**C**) Golden angle radial projections successively incremented by 111.25^0, Δk=FOV. No of spokes, $n_s=(\pi/2)*n$, where n = base resolution, distance between spokes < Δk. (D) Undersampled radial acquisition (2X) compared to (C).



Saucedo, M. Sarma, MA Thomas
ISMRM 2020
MRM 2021

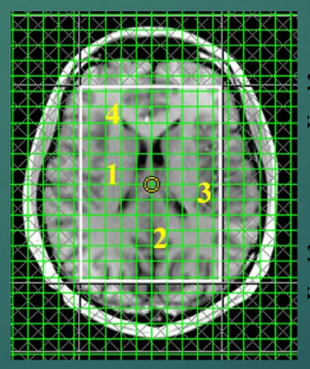
Radial Spectroscopic Imaging

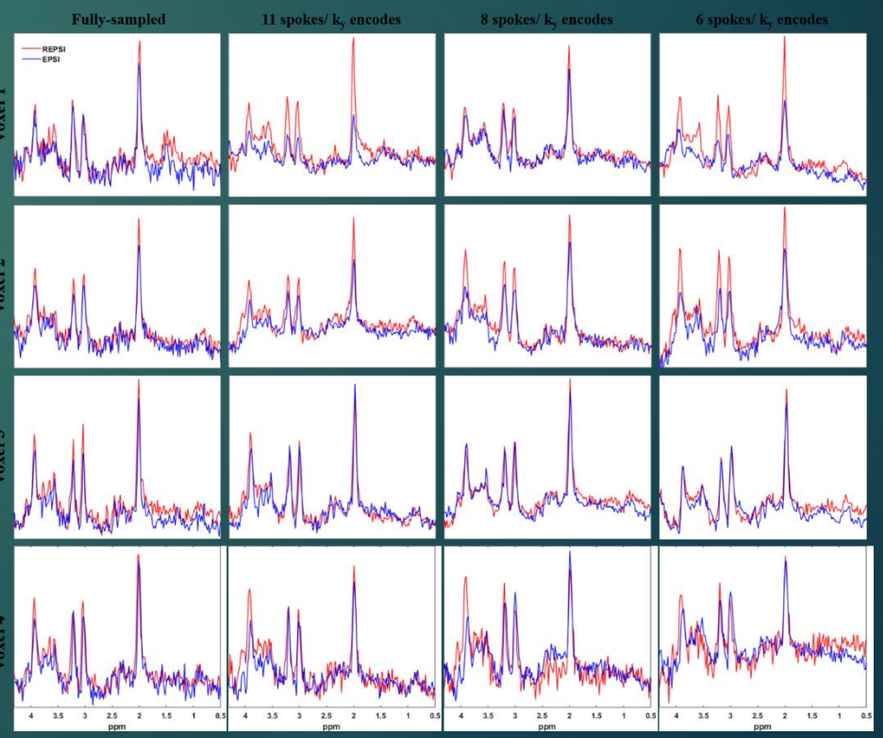




(Left) VOI localization in a 33 year-old healthy male volunteer. (A) tNAA maps from fully-sampled (AF = 1.0) REPSI and EPSI brain data (leftmost column), and tNAA maps from CS reconstructions of prospectively undersampled brain data acquired with 11, 8, and 6 radial spokes or k_y-lines. These maps are interpolated by a factor of two. (B) CRLB maps for the tNAA maps shown in (A).

Representative and CS reconstructions of prospectively undersampled *in vivo* brain data from a 32 year-old healthy male volunteer. Spectra extracted : 1 – right putamen to corona radiata, 2 - occipital gray matter, 3 – left posterior insular cortex, and 4 – frontal white matter. Both the REPSI and EPSI data were prospectively undersampled with 11, 8, and 6 acquired radial spokes or k_y-lines, respectively.



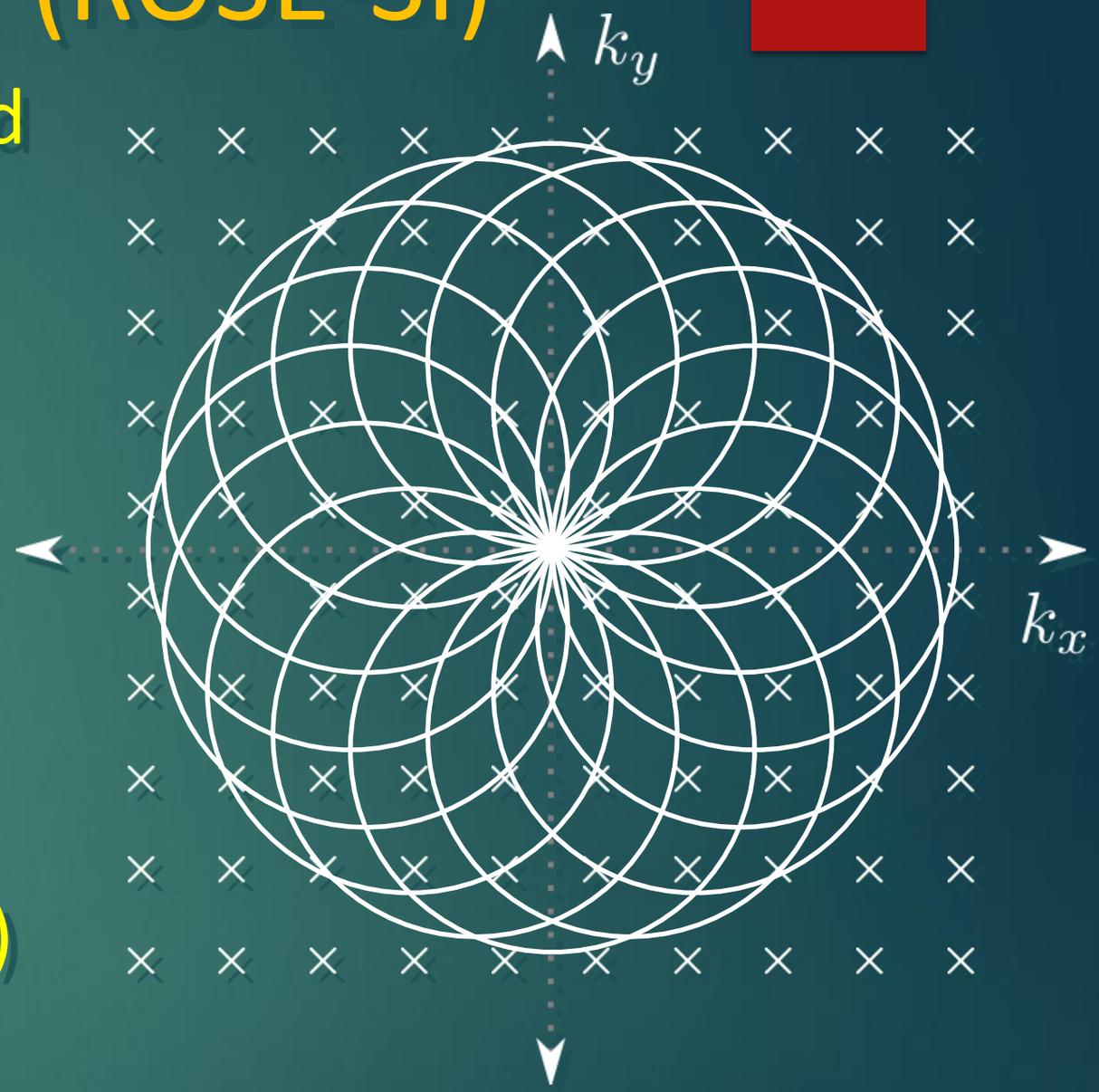


A. Saucedo, M. Sarma, MA Thomas
MRM 2021

Rosette-Trajectories-based Spectroscopic Imaging (ROSE-SI)

David Geffen School of Medicine

A single petal in k-space is collected in a single excitation



Rosette trajectory is defined as

$$k(t) = k_{max}\sin(\omega_1 t)e^{i\omega_2 t}$$

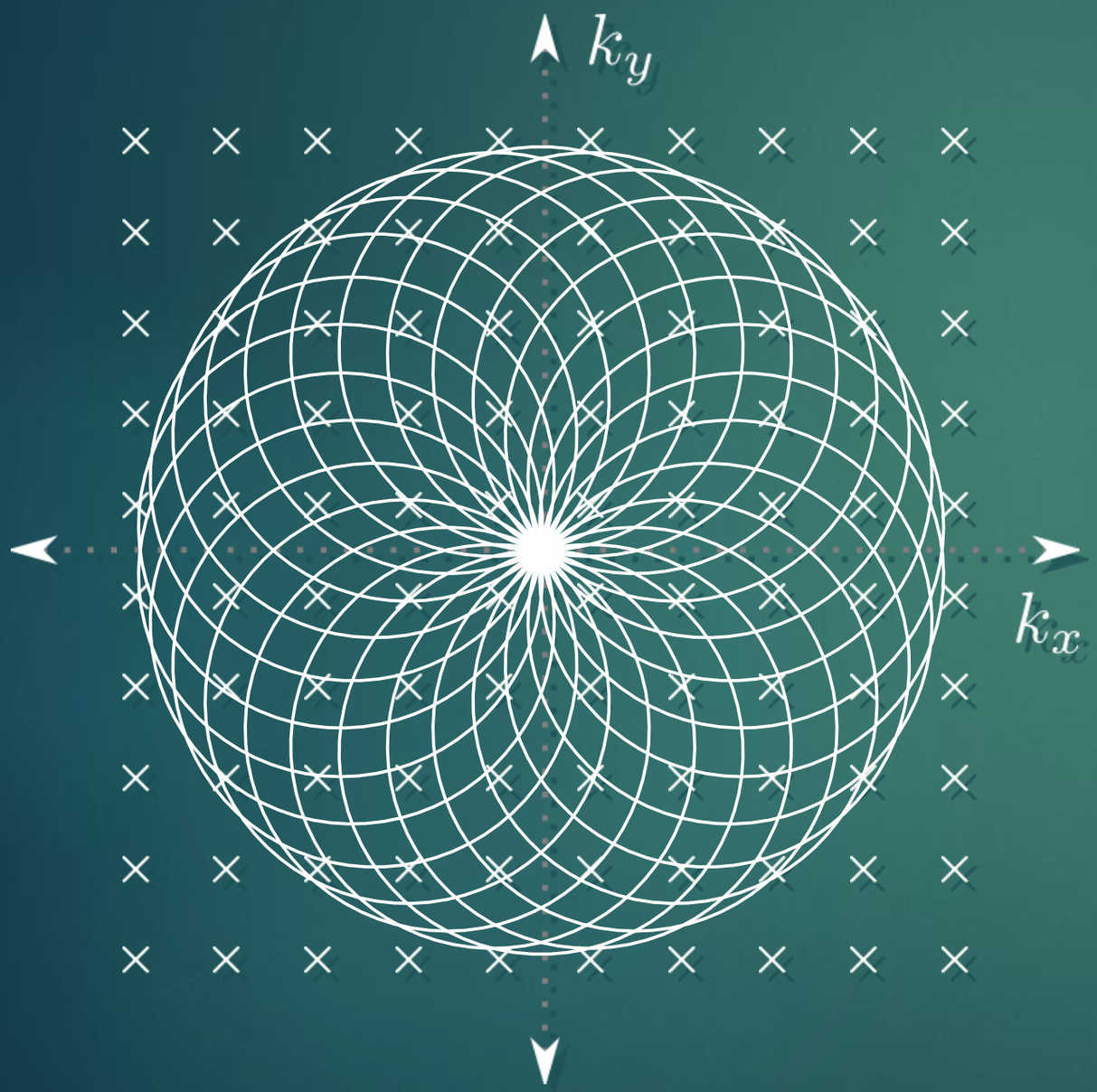
Total acquisition time depends on the radial oscillation frequency (ω_1) and the rotational frequency (ω_2)

$$\frac{\omega_2}{\omega_1} \leq 1,\ N_{sh} \cong \frac{\pi \times N_x}{\sqrt{1 + 3 \times (\omega_2/\omega_1)^2}}$$

$$\frac{\omega_2}{\omega_1} > 1,\ N_{sh} \cong \frac{\pi \times N_x}{\sqrt{3 + (\omega_2/\omega_1)^2}}$$

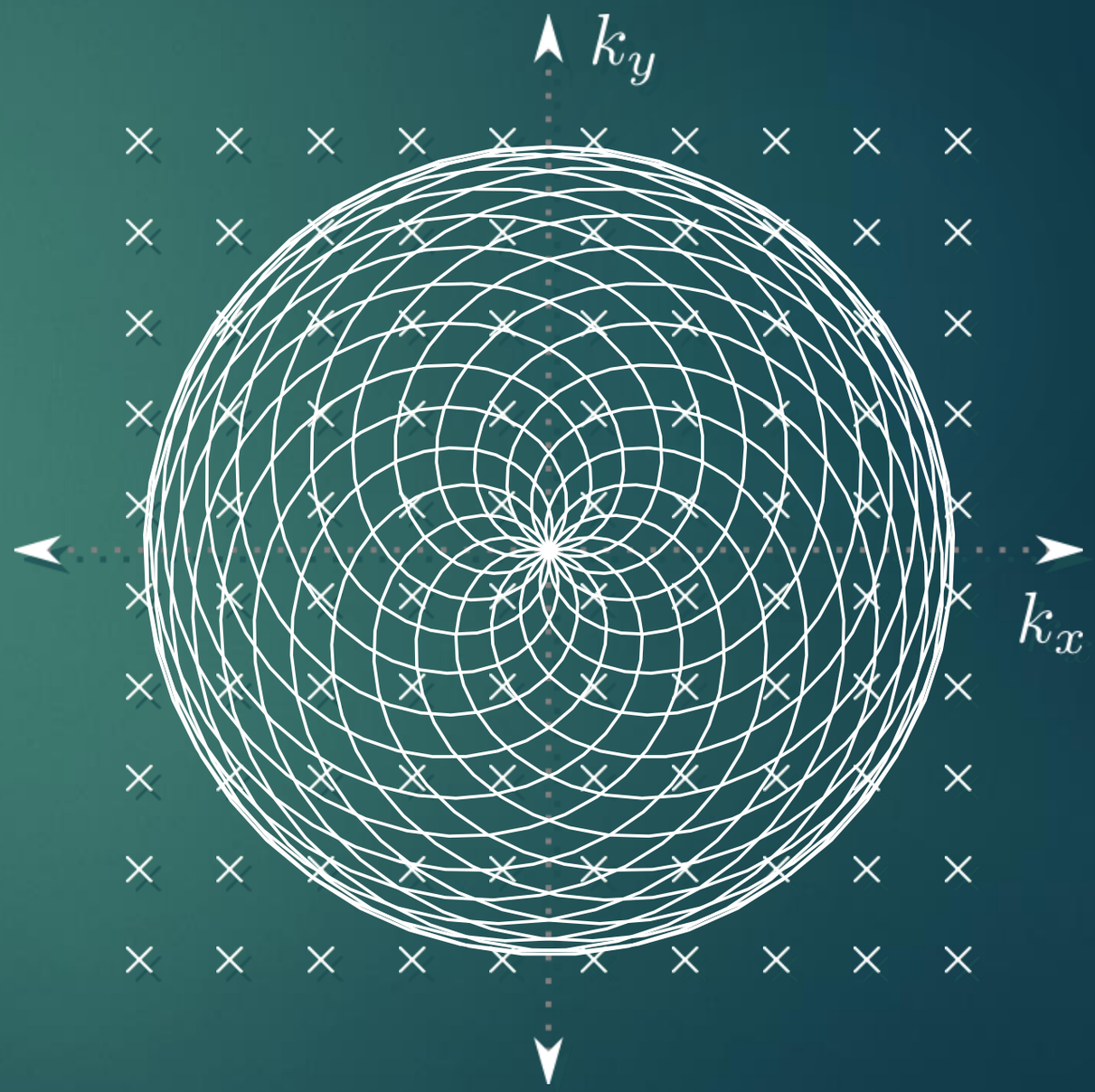
Rosette Petals (ROSE-SI)

$N_x = 16, \frac{\omega_2}{\omega_1} = 1, N_{sh} = 26$



Needs 26 petals for N_x=16

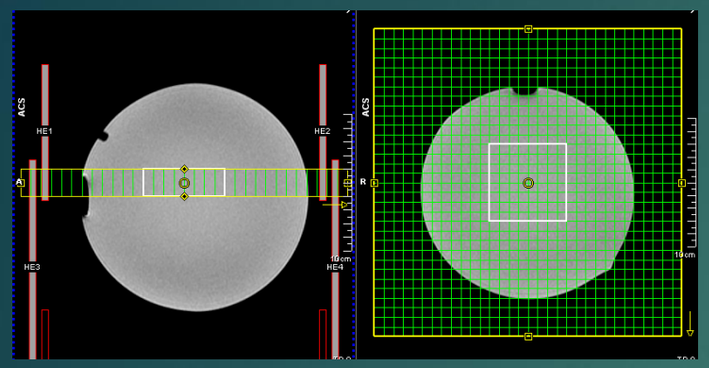
$N_x = 16, \frac{\omega_2}{\omega_1} = 3, N_{sh} = 15$



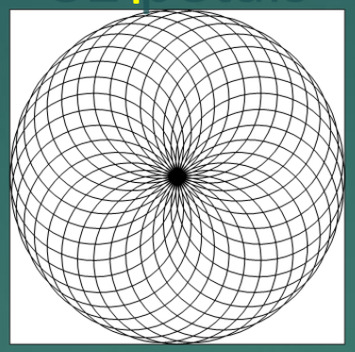
Needs only 15 petals for N_x=16

Rosette Petals (ROSE-SI)

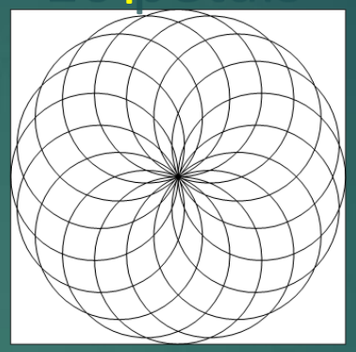
$N_x = 32, \frac{\omega_2}{\omega_1} = 1$



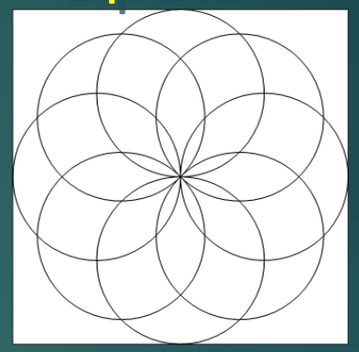
32 petals



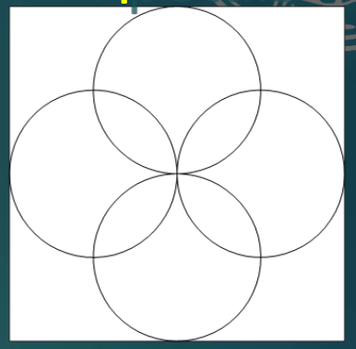
16 petals

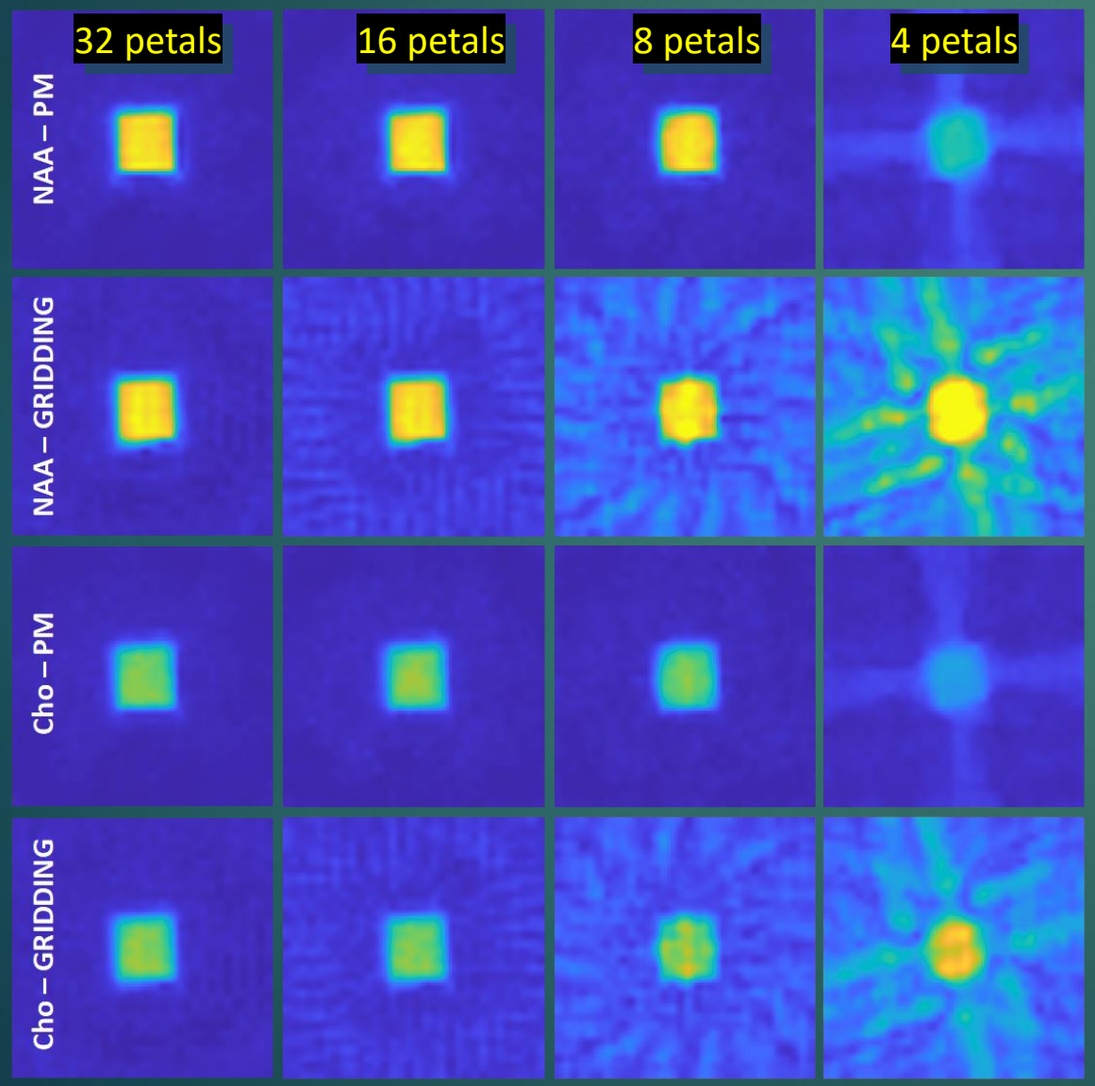


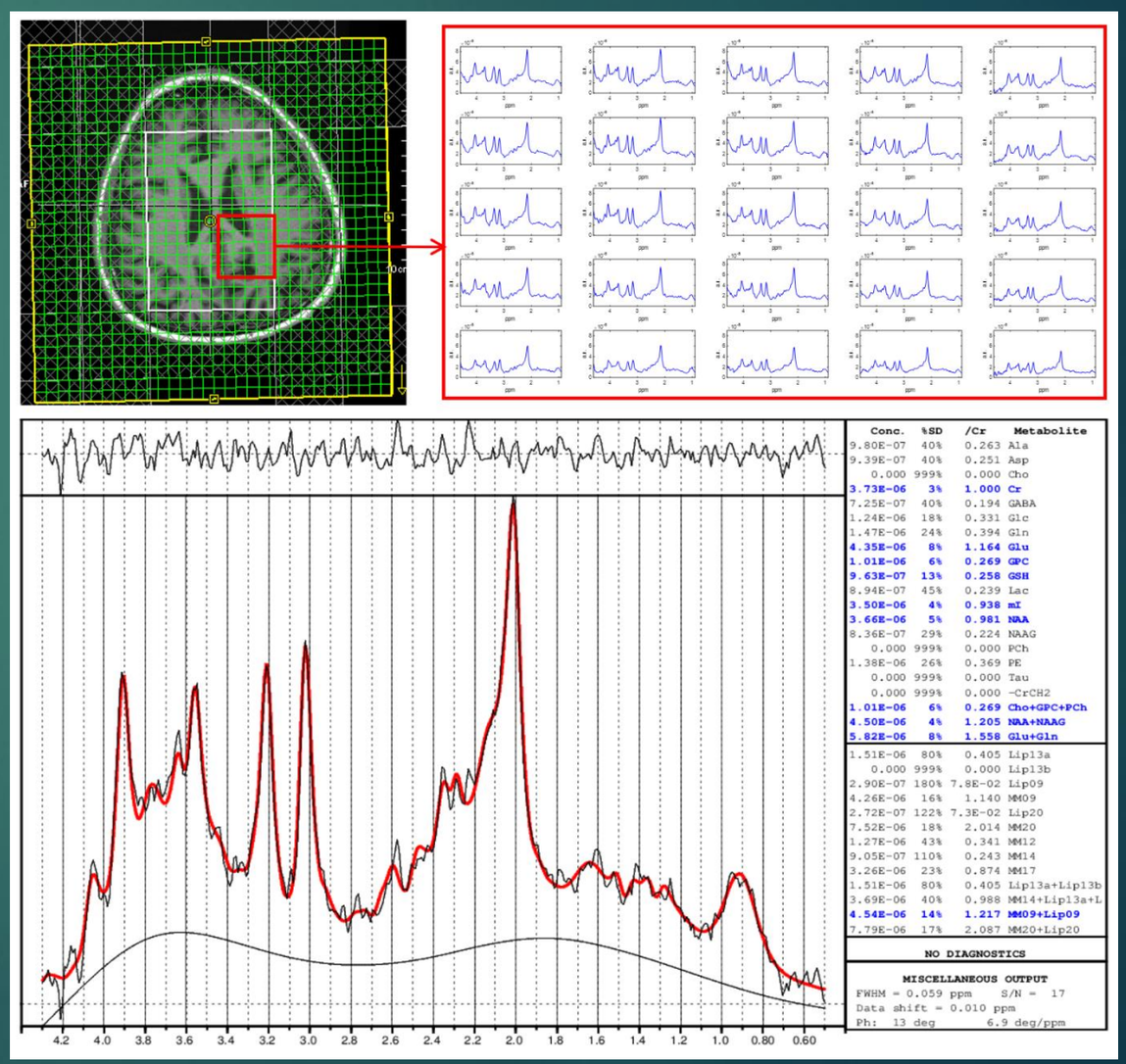
8 petals



4 petals

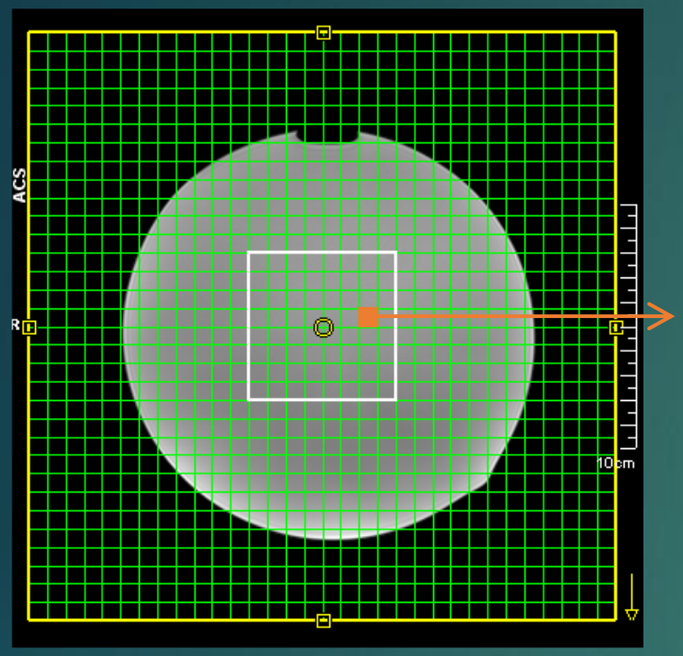


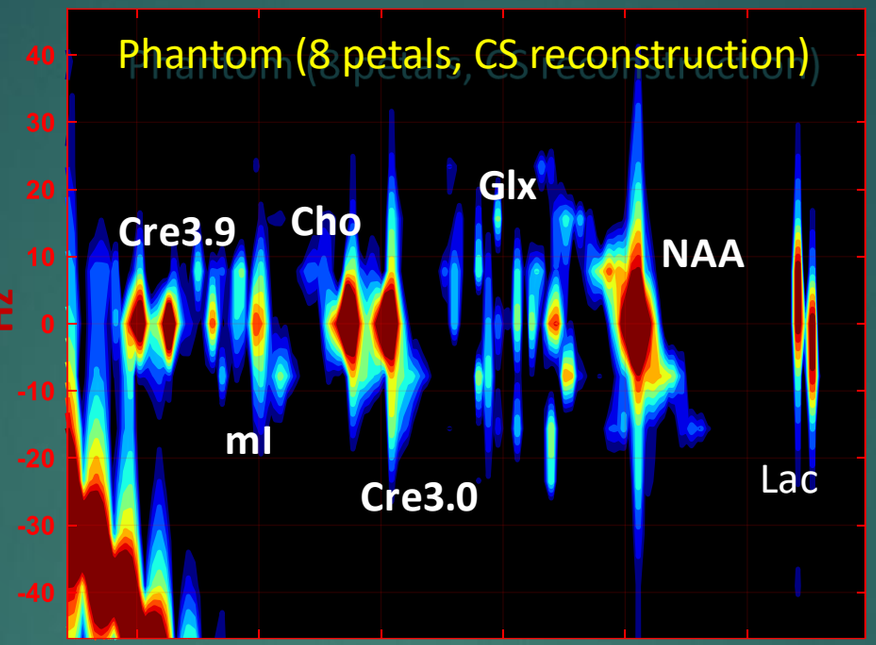




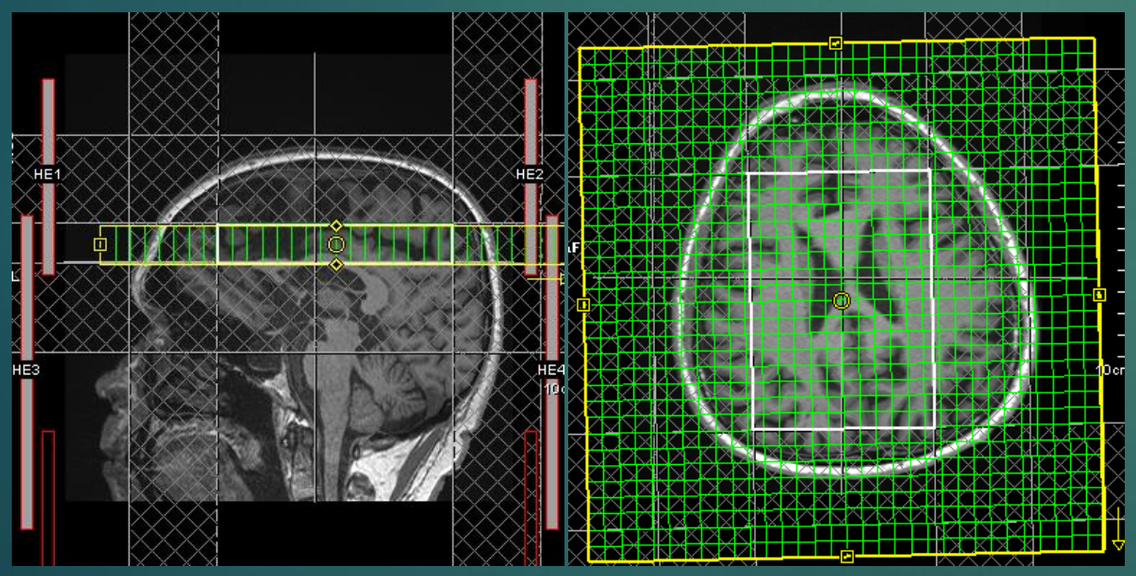
A Joy, U Emir, PM Macey, MA Thomas, ISMRM 2023

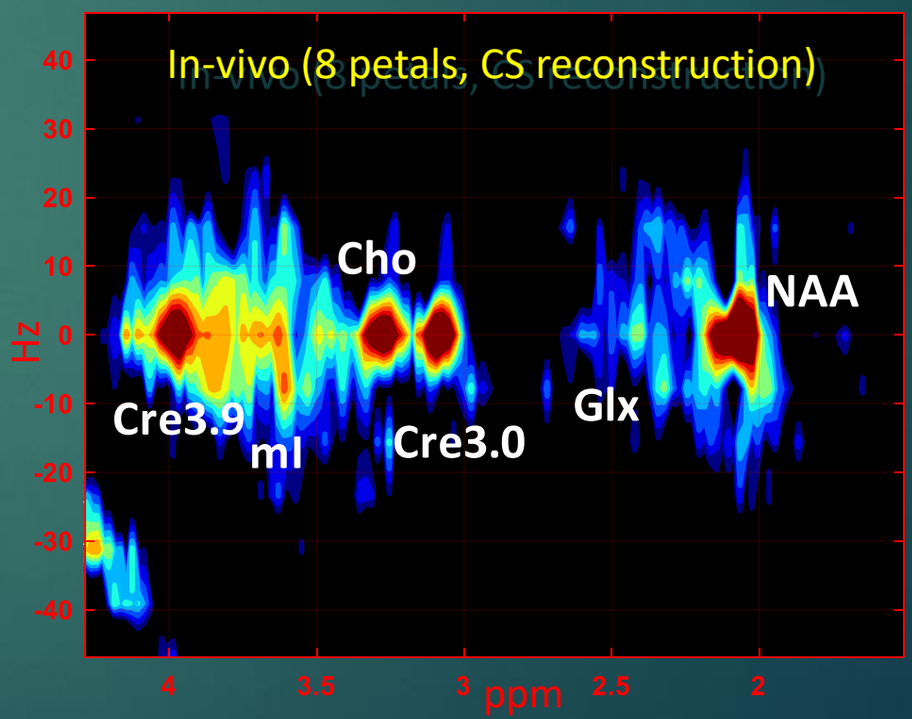
Rosette Petals (ROSE-SI)





$$N_x = 32, \frac{\omega_2}{\omega_1} = 1$$





A Joy, U Emir, PM Macey, MA Thomas, ISMRM 2023

Advantages of Rosette Trajectories

- Continuous readout during acquisition (EP-COSI without ramp sampling only samples during ~75% readout)
- Inherently less sensitive to motion artifacts (due to oversampling of center of k-space)
- Less demanding on gradient hardware (especially for lower rotational frequencies)
- Higher sensitivity than the standard CSI acquisition with square k-space support.
- Freedom in trajectory design to optimize for the available hardware by adjusting ω_2/ω_1
- Encoding speed of rosette can be used to accelerate the data acquisition process.
- Higher sampling density in central and peripheral k-space allows undersampling by reduced number of petals for accelerated acquisition and CS reconstruction

Drawbacks

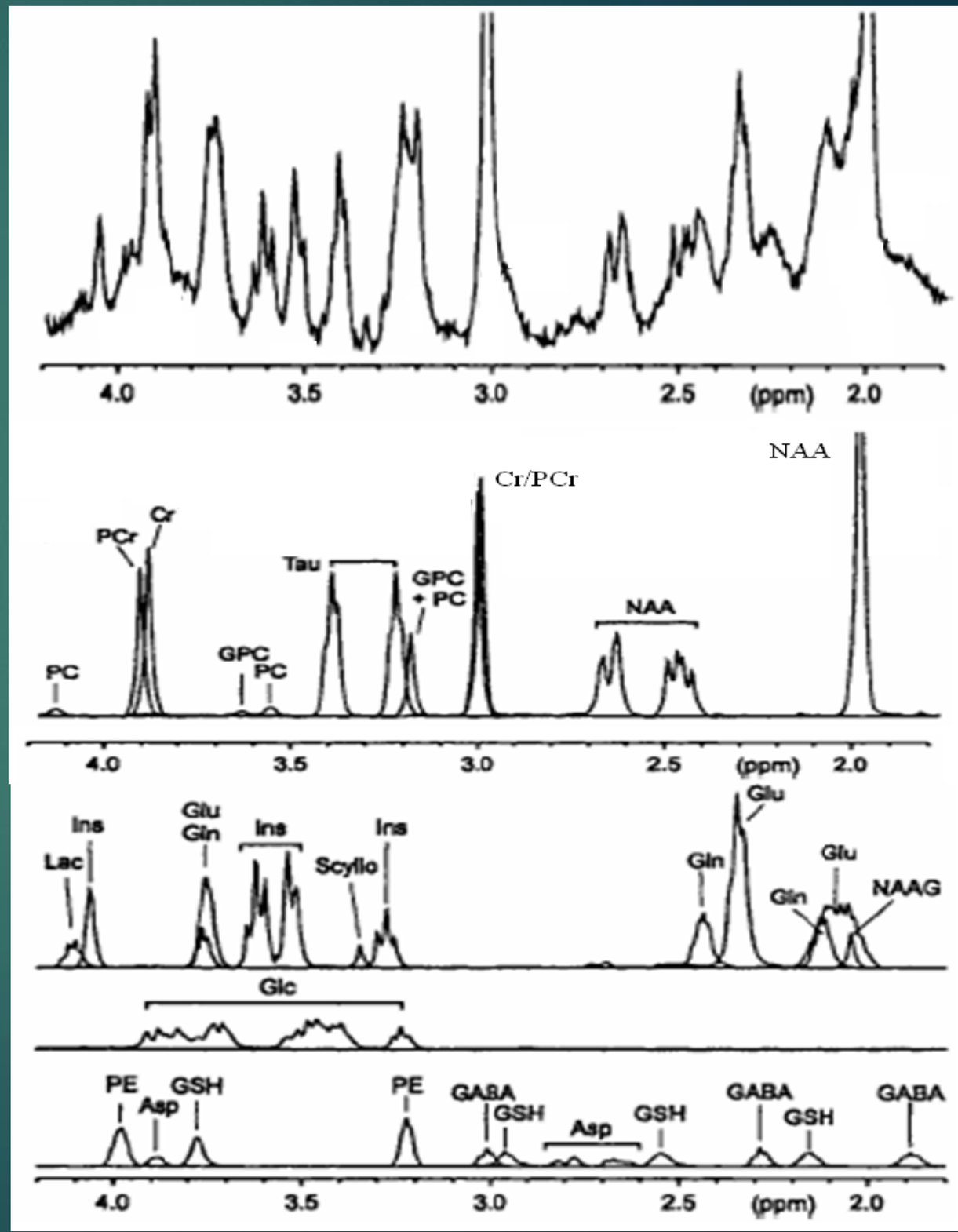
- Regular patterns of phase accrual in k-space can cause artifacts
  - More complicated post-processing
    - Data must be regridded in order to apply FFT
    - Alternatively, non-uniform fast Fourier transform (NUFFT) can be used

Noll, TMI 1997; Schirda et al., JMRI 2009;
Shen et al., MRM 2018; Joy et al., ISMRM 2023

Single-voxel localized 2D MRS : L-COSY and JPRESS

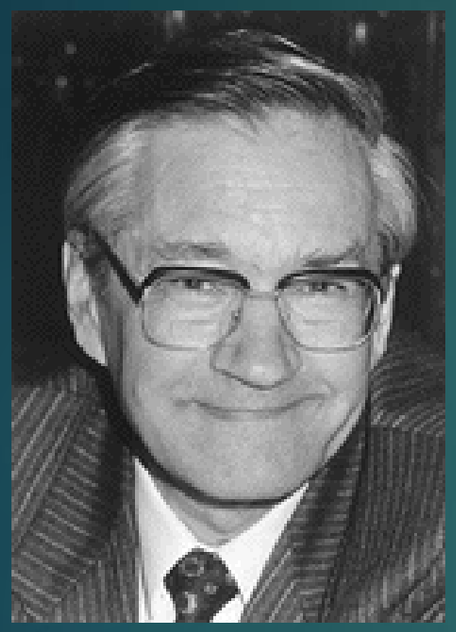
1D MRS Quantitation

- LC-Model for 1D MRS quantitation.
- Works in frequency domain using prior knowledge



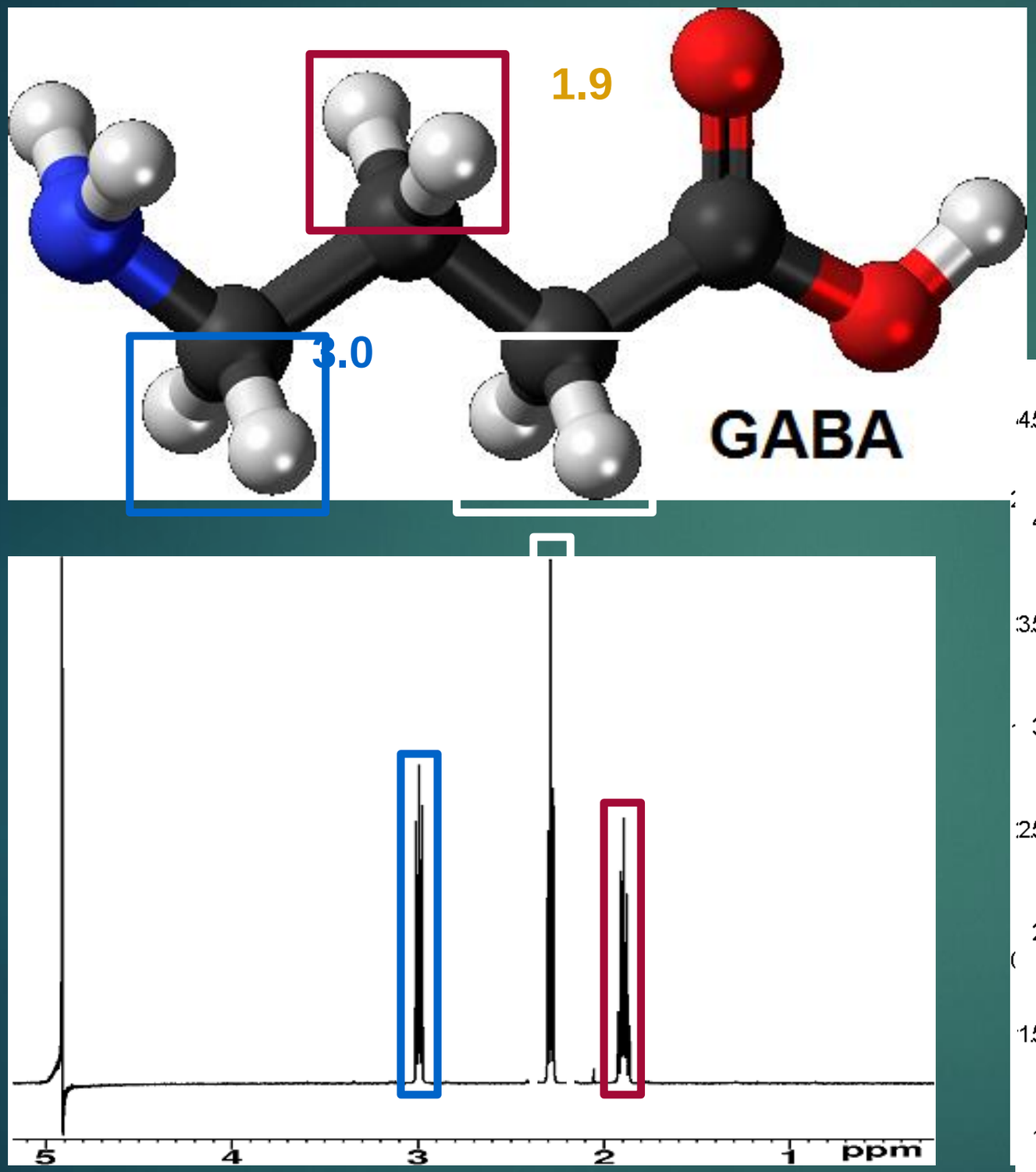
Provencher (2001)

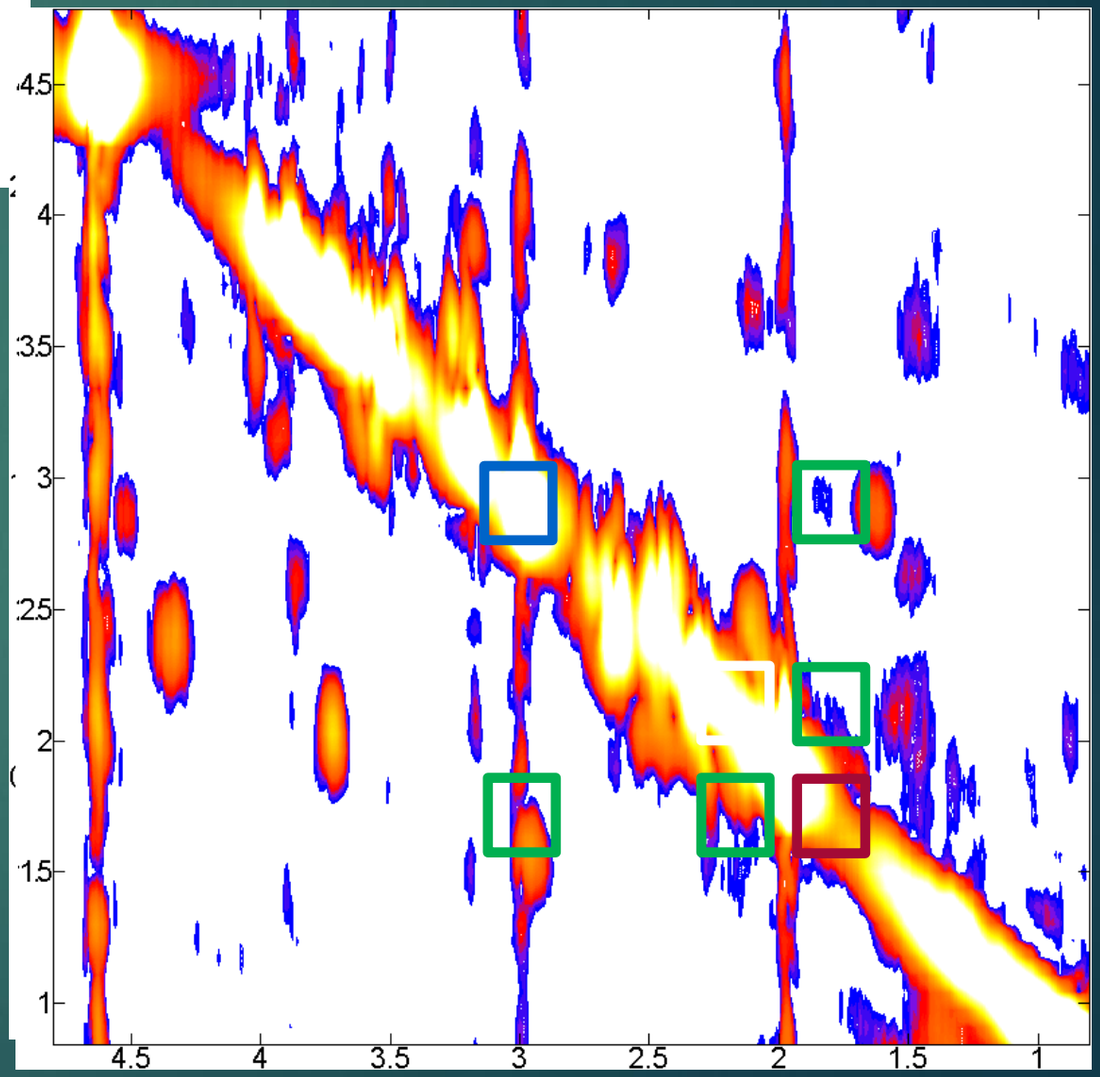
A quote from 1991 Nobel laureate Richard Ernst



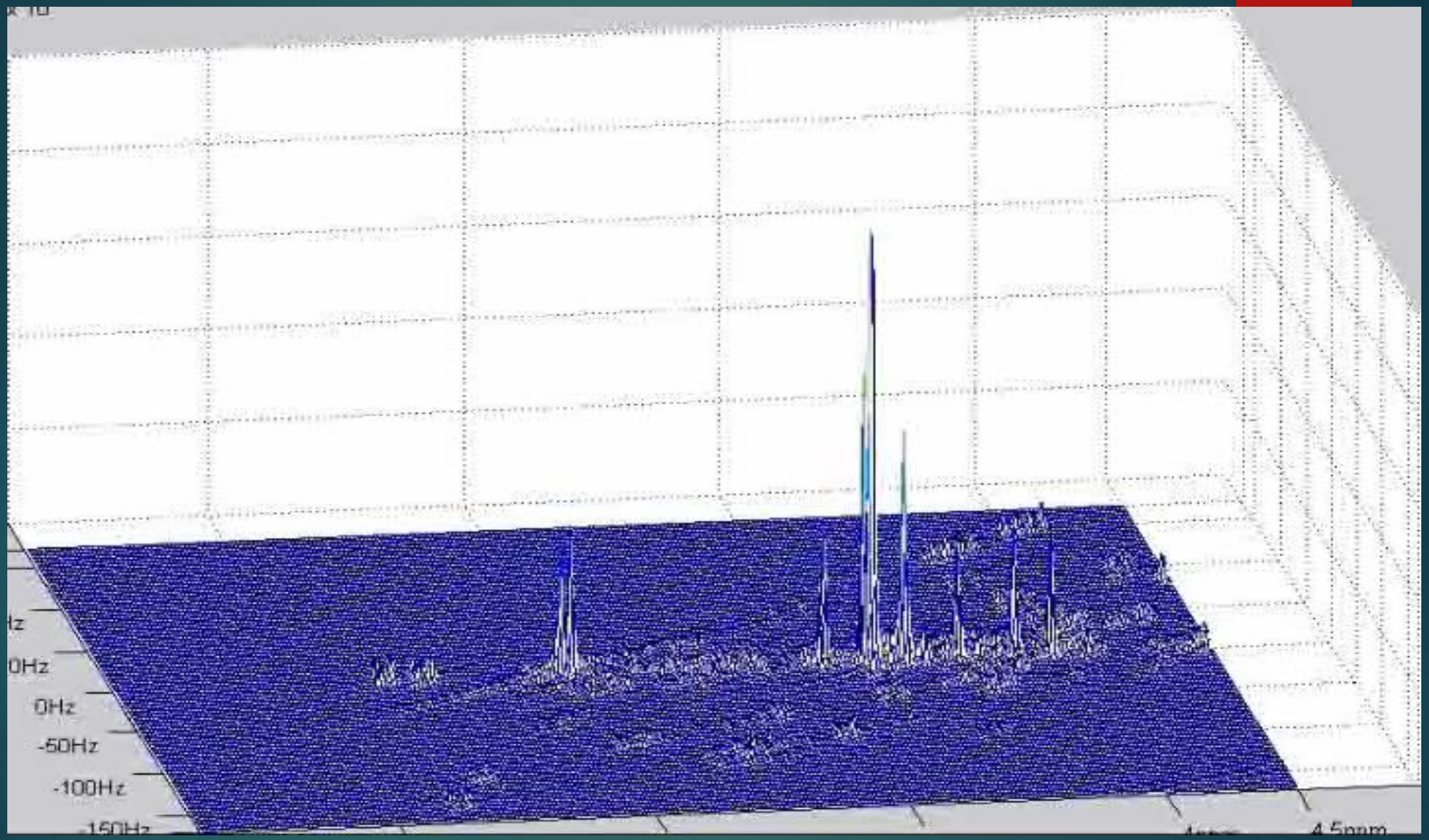
"One-dimensional spectra that are rendered inscrutable because of severe overlap may be unravelled by separating interactions of different physical origins, e.g. chemical shift and couplings, thus making it possible to spread the signals in a second frequency dimension much like opening a Venetian blind."

Why 2D Spectroscopy?

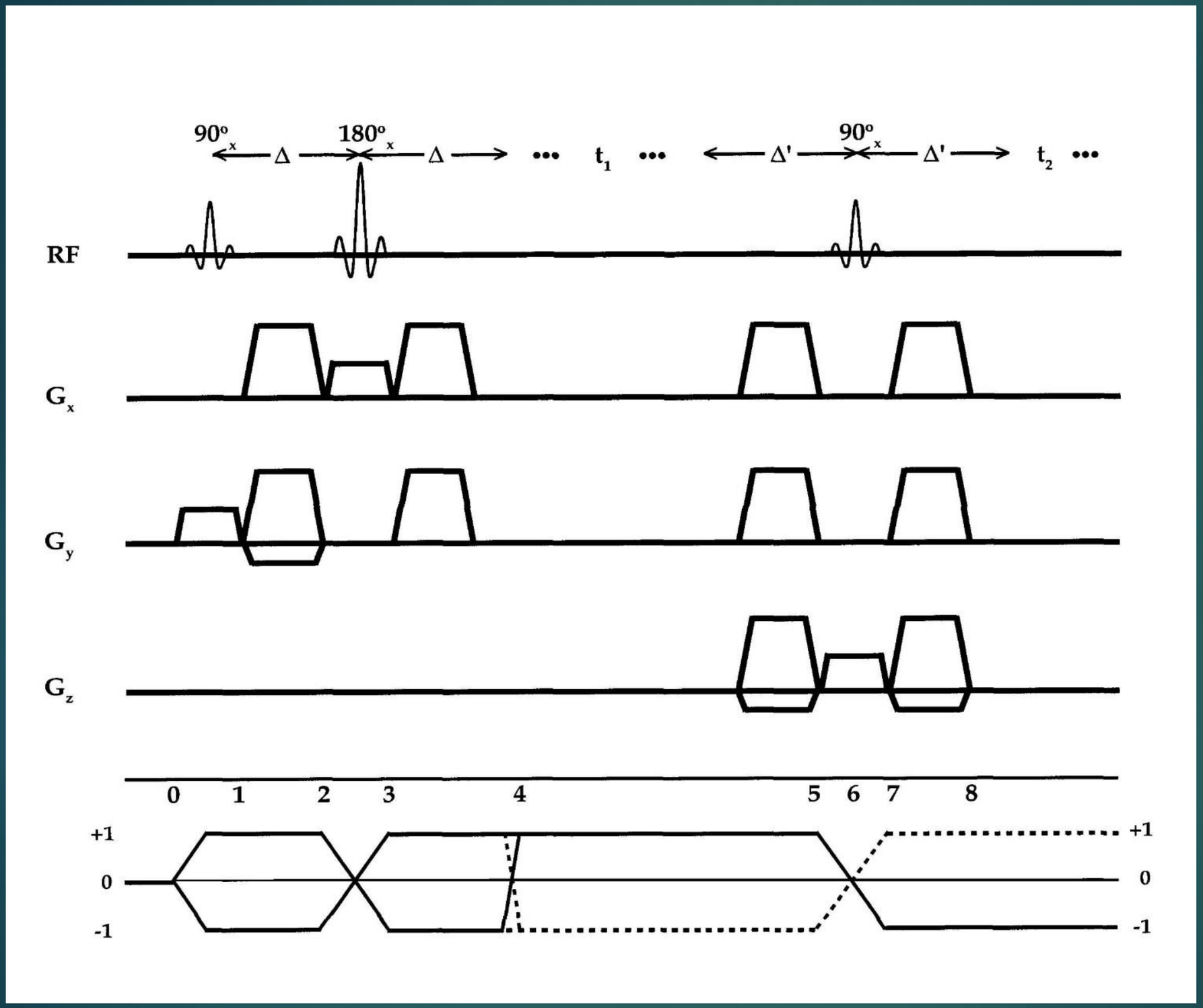




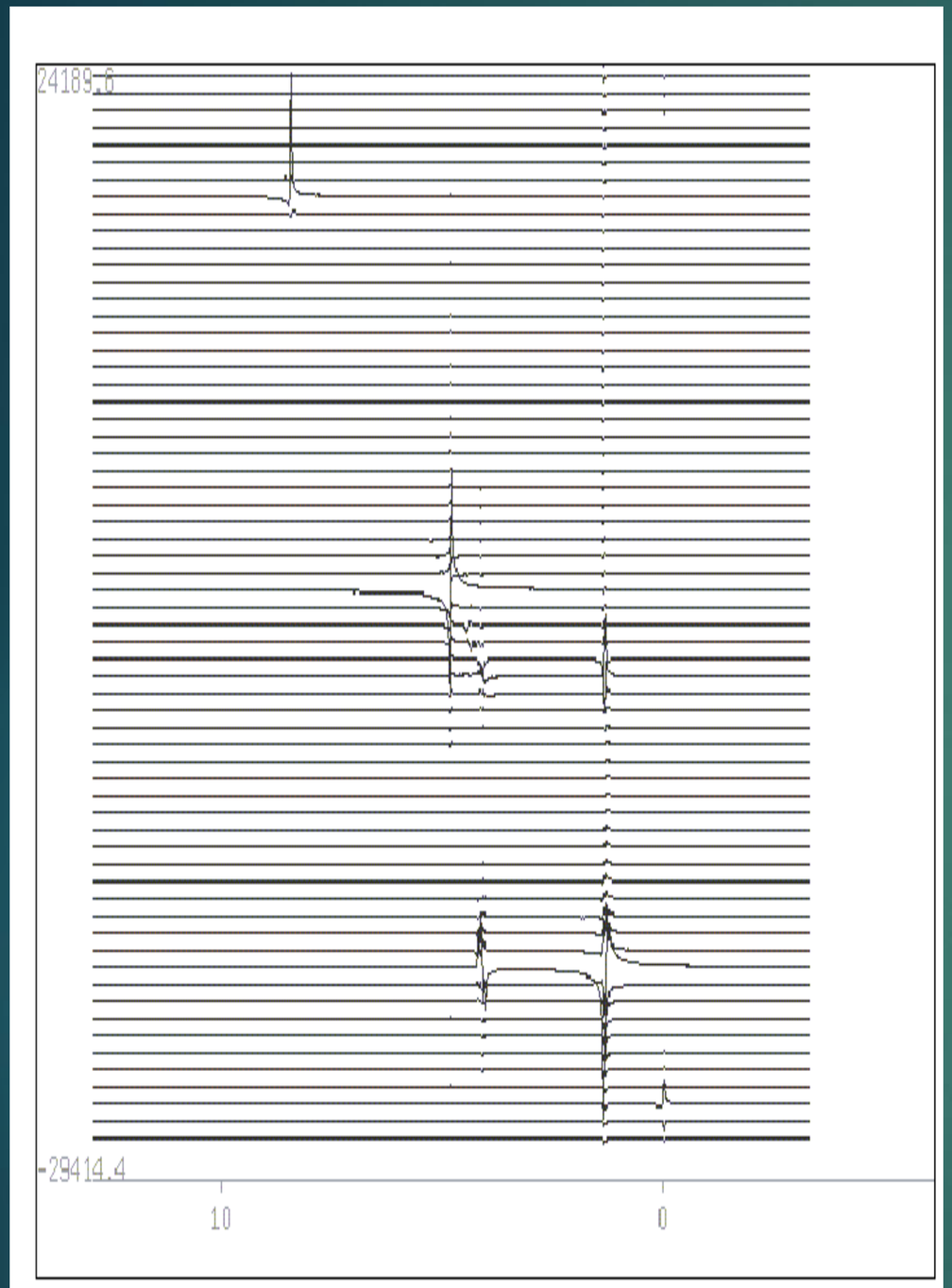
Verma et al. ISMRM 2014


-50Hz
-100Hz
4.5ppm

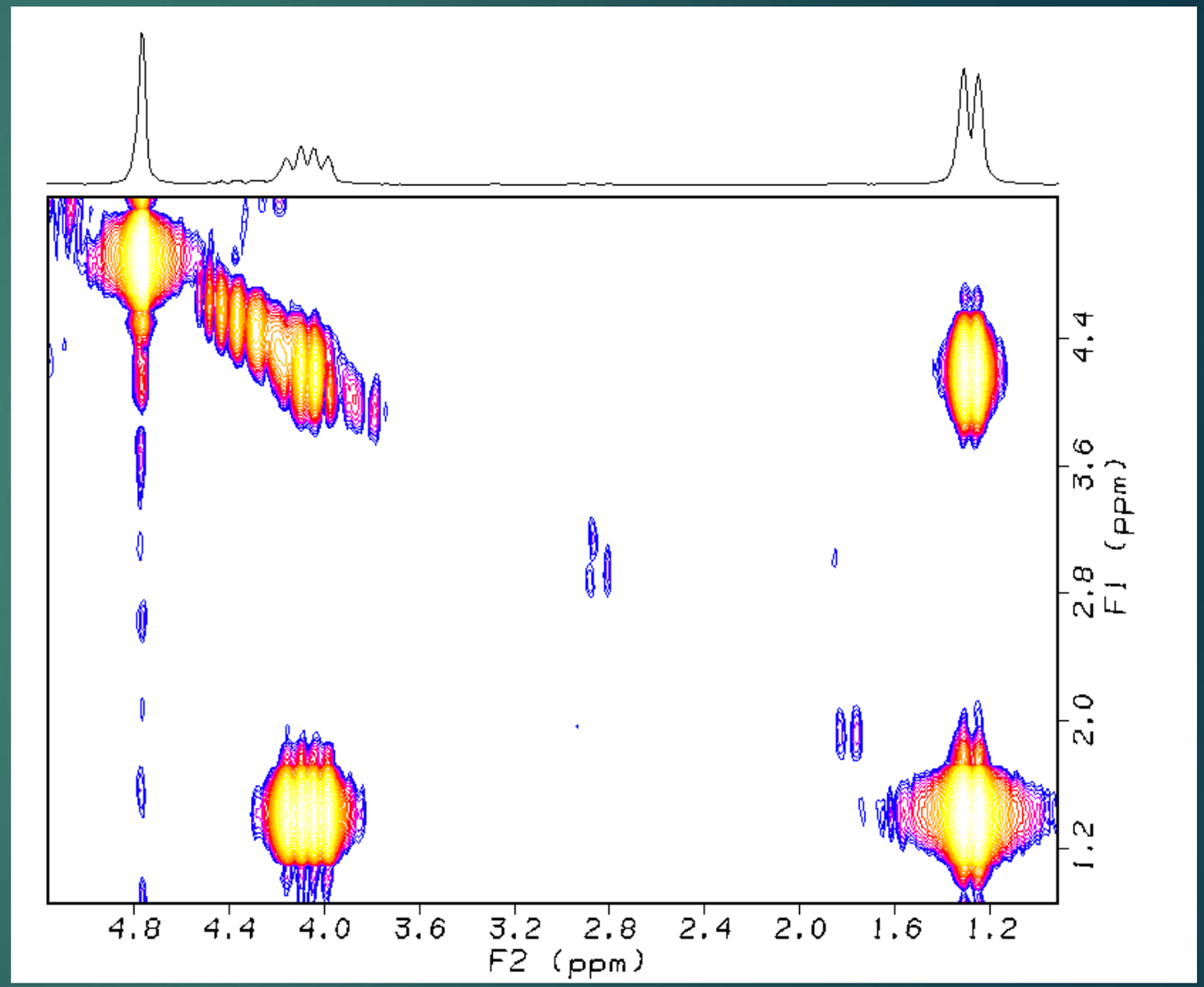
Localized 2D Correlated Spectroscopy (L- COSY)



- *Based on a spin-echo and a coherence-transfer-echo*
  Hahn (1952) / Maudsley, Wokaun and Ernst (1978) *Thomas et al. (MRM2001)*

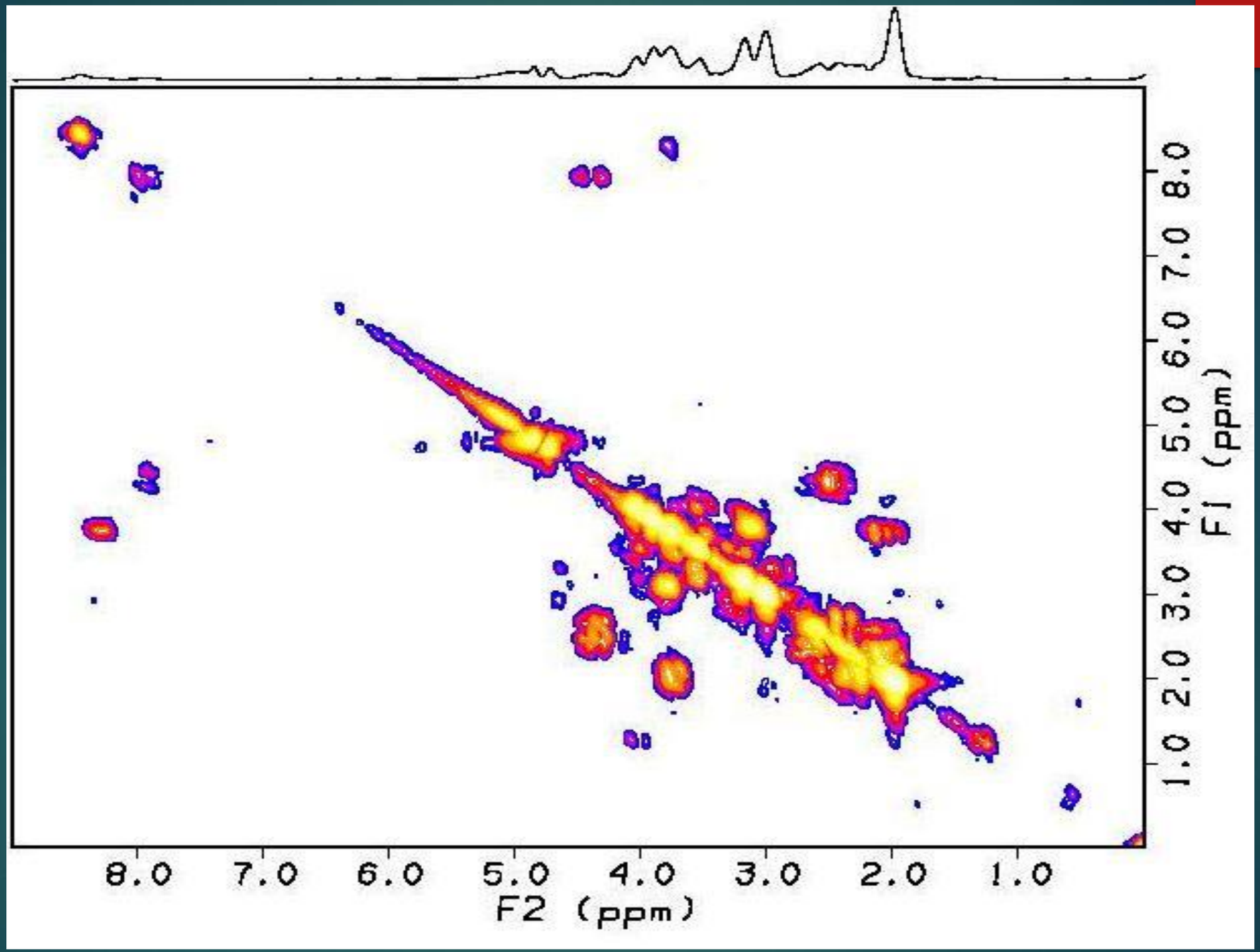

24189.6
-29414.4
10
0

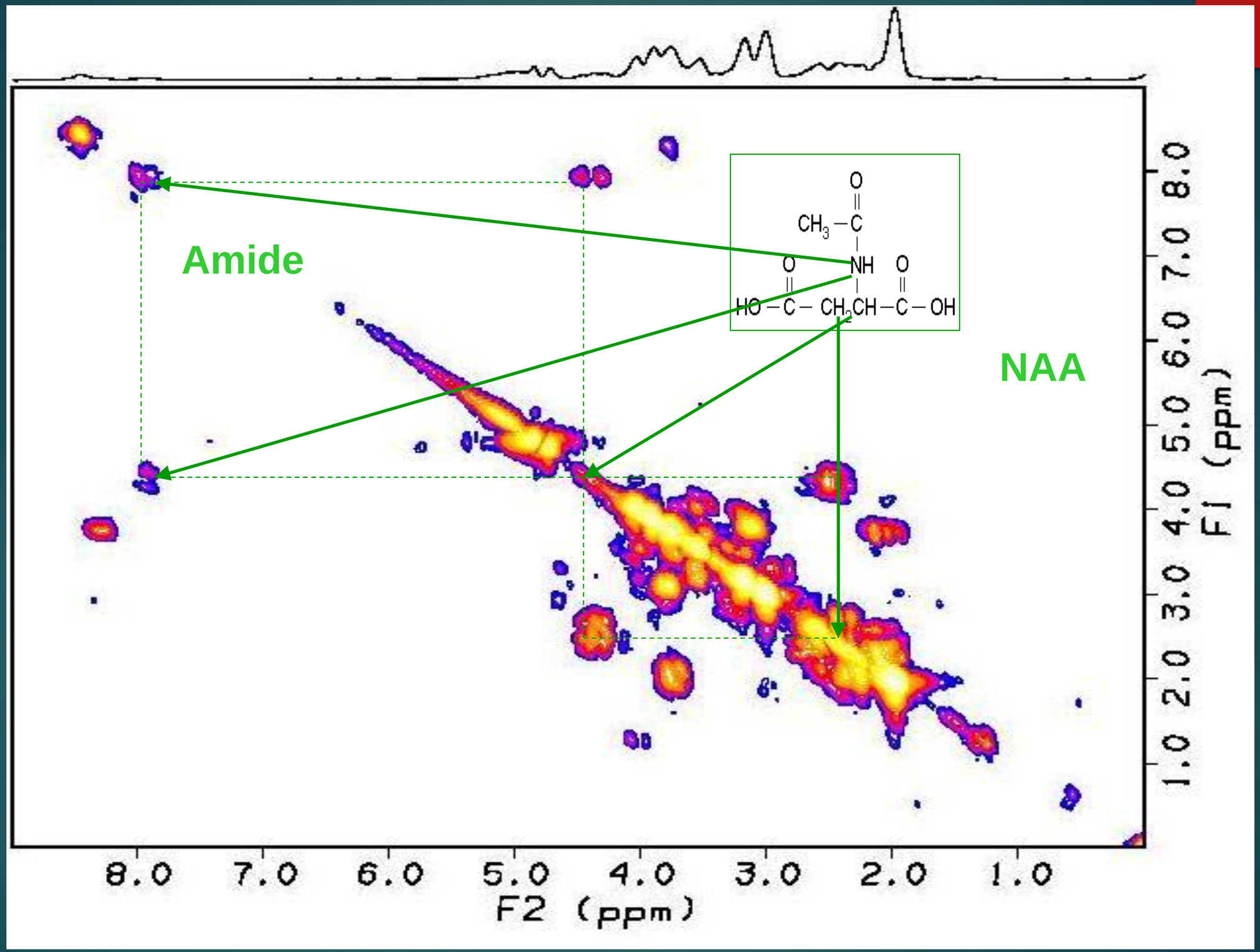
F1

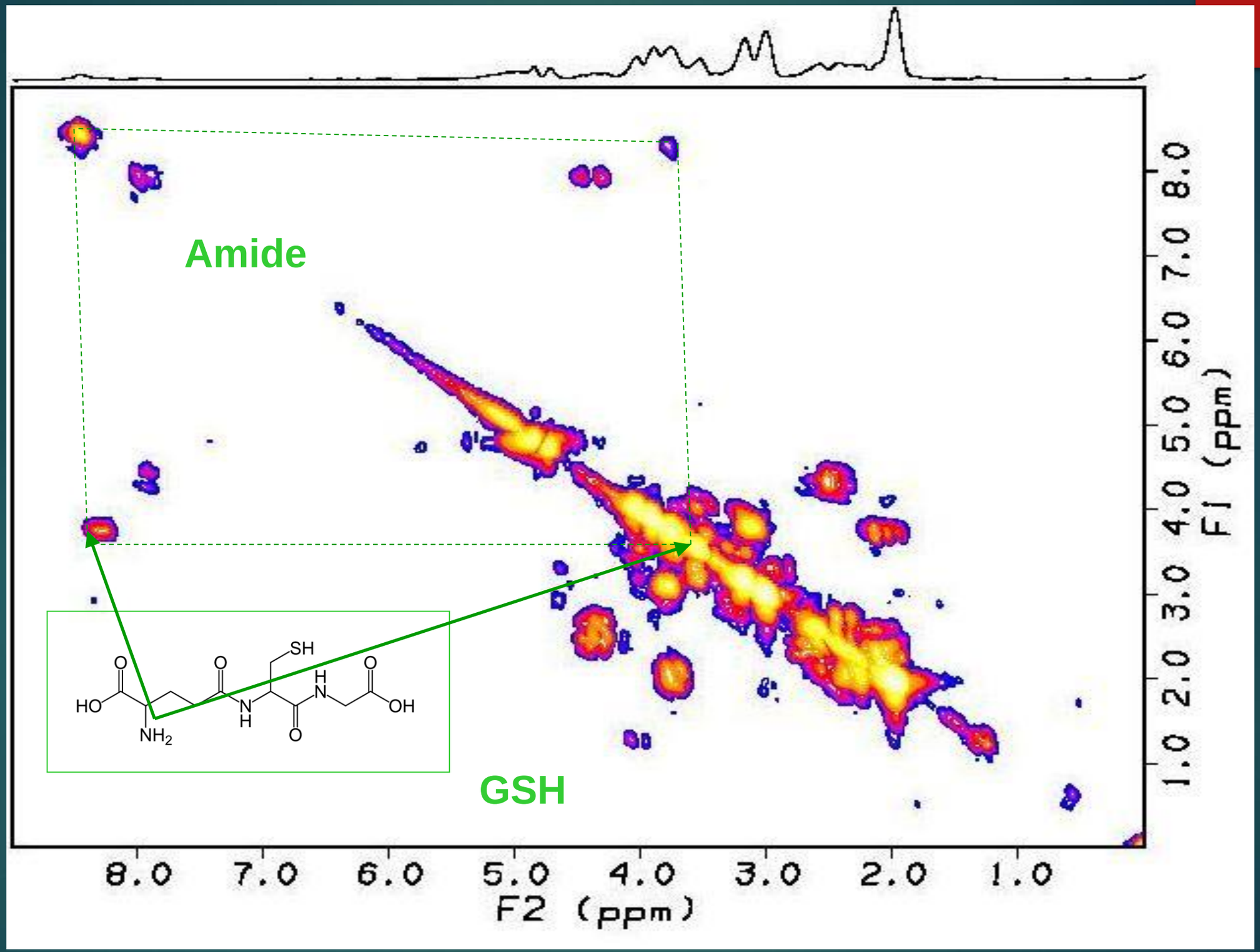

4.4
3.6
2.8
2.0
1.2
F1 (ppm)
4.8
4.4
4.0
3.6
3.2
2.8
2.4
2.0
1.6
1.2
F2 (ppm)

F2

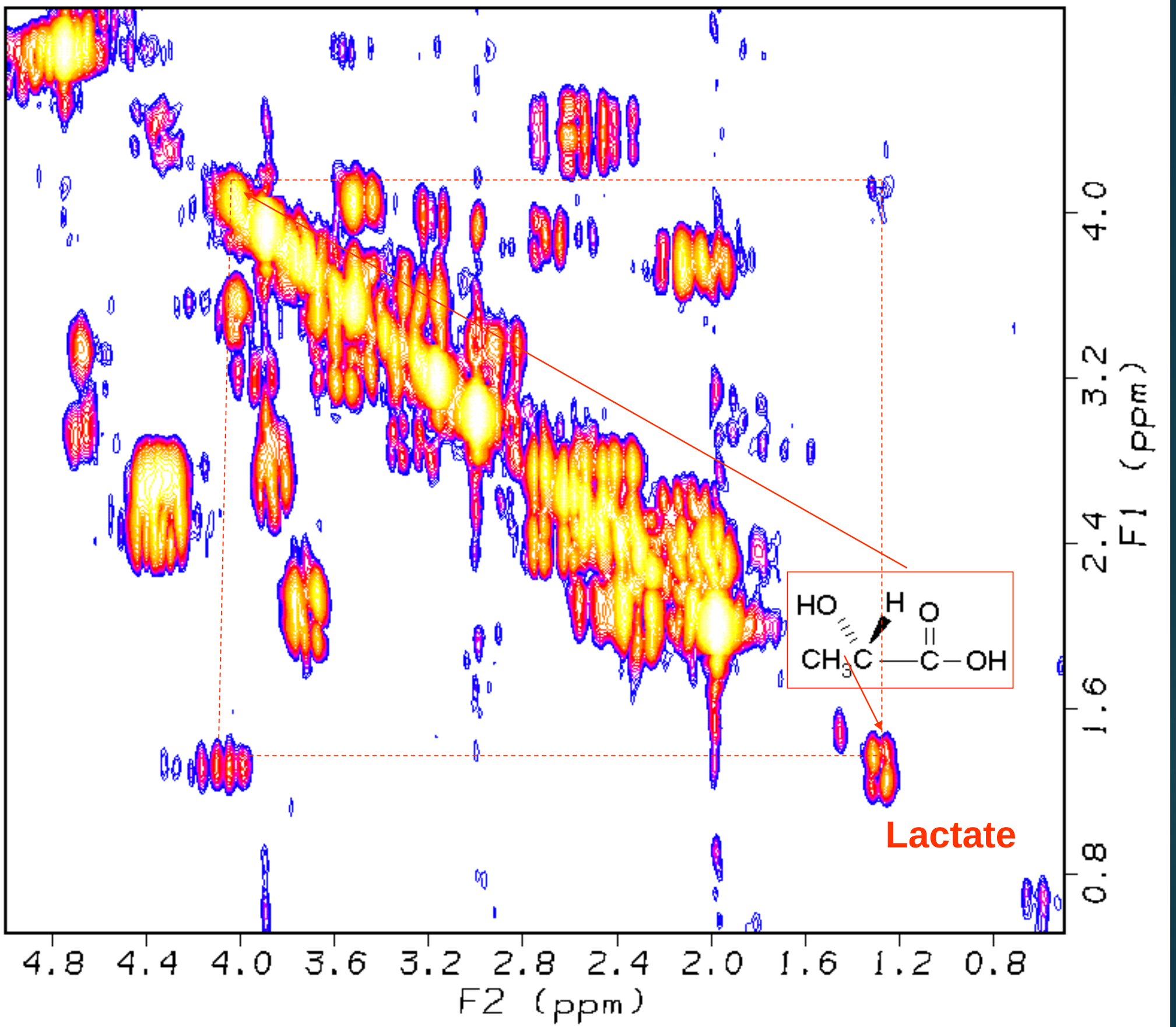
Brain Phantom 3T MRI/MRS Scanner



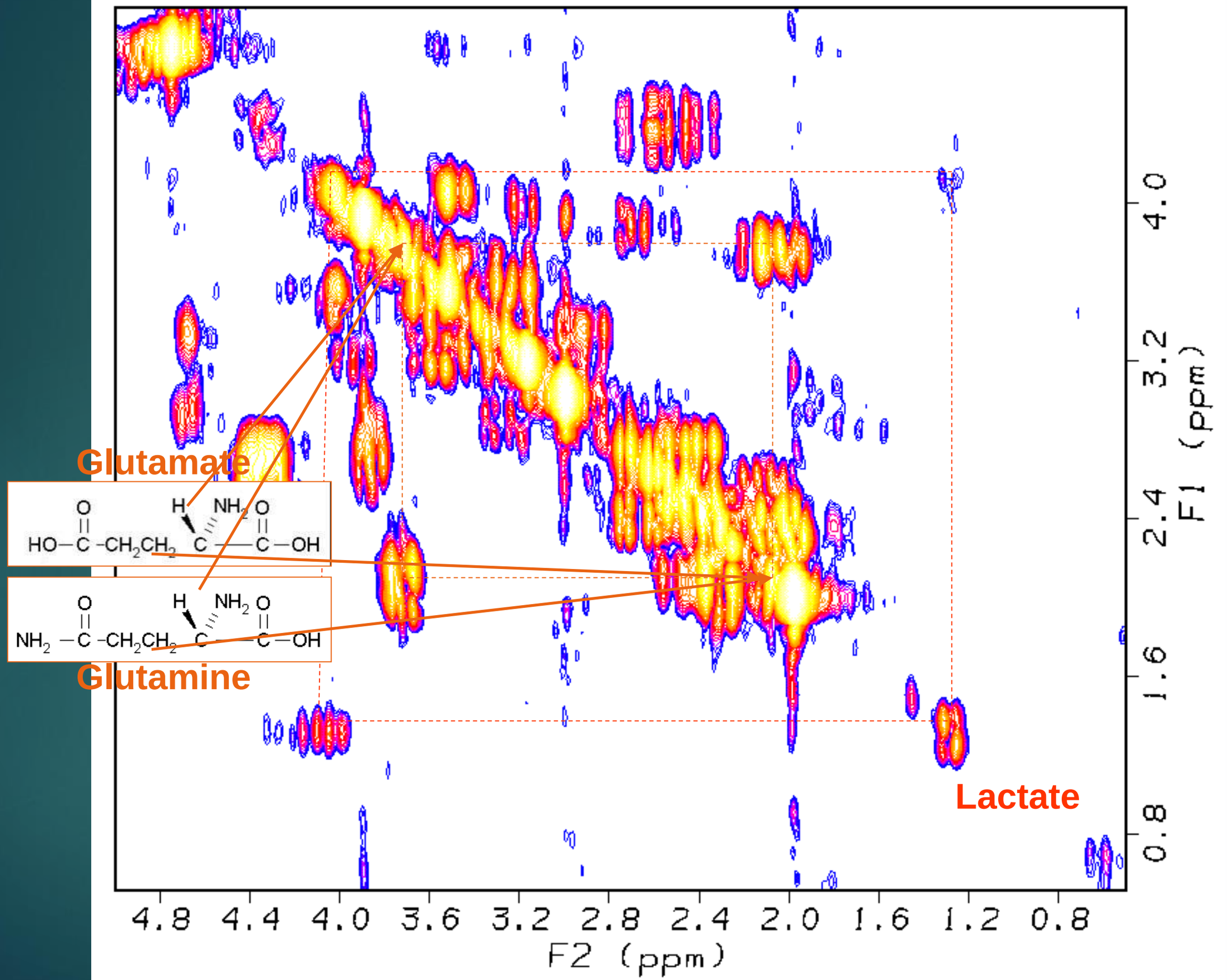

Amide
NAA
O
CH3 – C
NH
O
HO – C – CH2CH – C – OH
O
8.0
7.0
6.0
5.0
4.0
3.0
2.0
1.0
F2 (ppm)
F1 (ppm)


Amide
GSH
HO
O
NH2
O
N
H
SH
H
N
O
O
OH
8.0
7.0
6.0
5.0
4.0
3.0
2.0
1.0
F2 (ppm)
F1 (ppm)

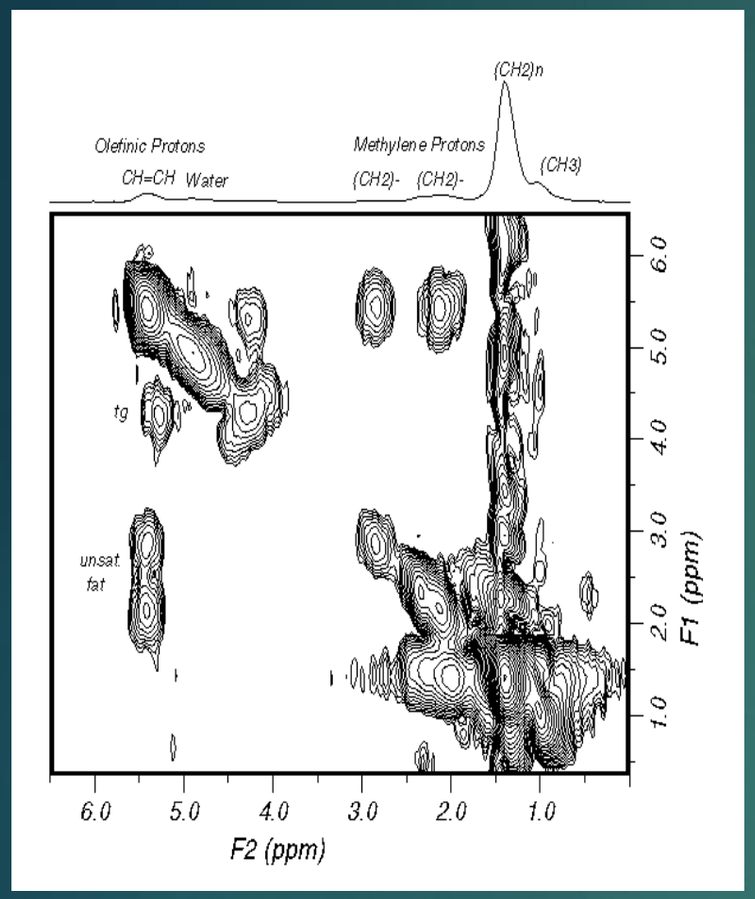
2D MRS – Gray Matter Metabolites

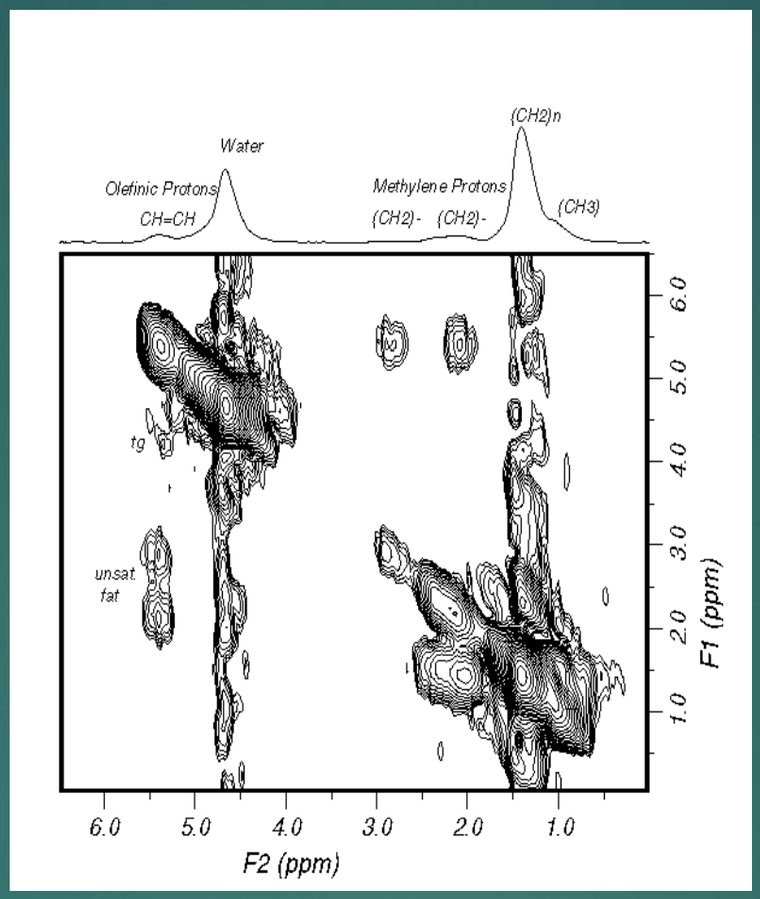


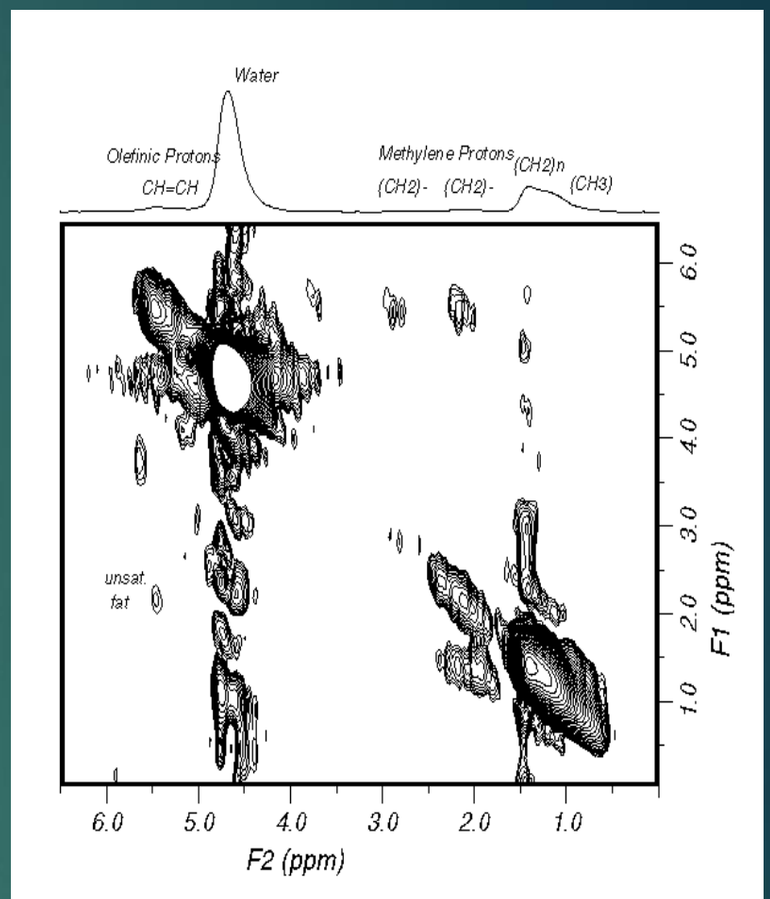
2D MRS – Gray Matter Metabolites



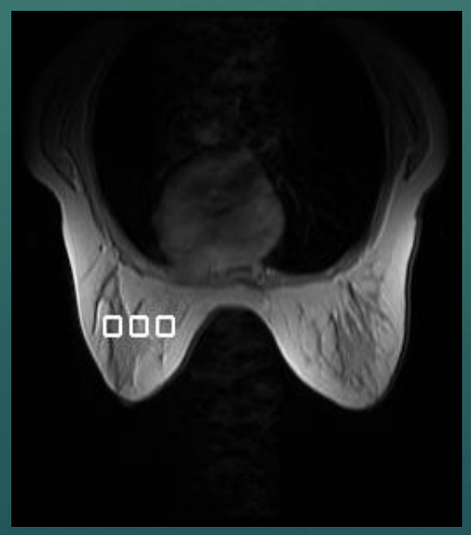
Localized 2D COSY Spectra of a 27yo healthy breast







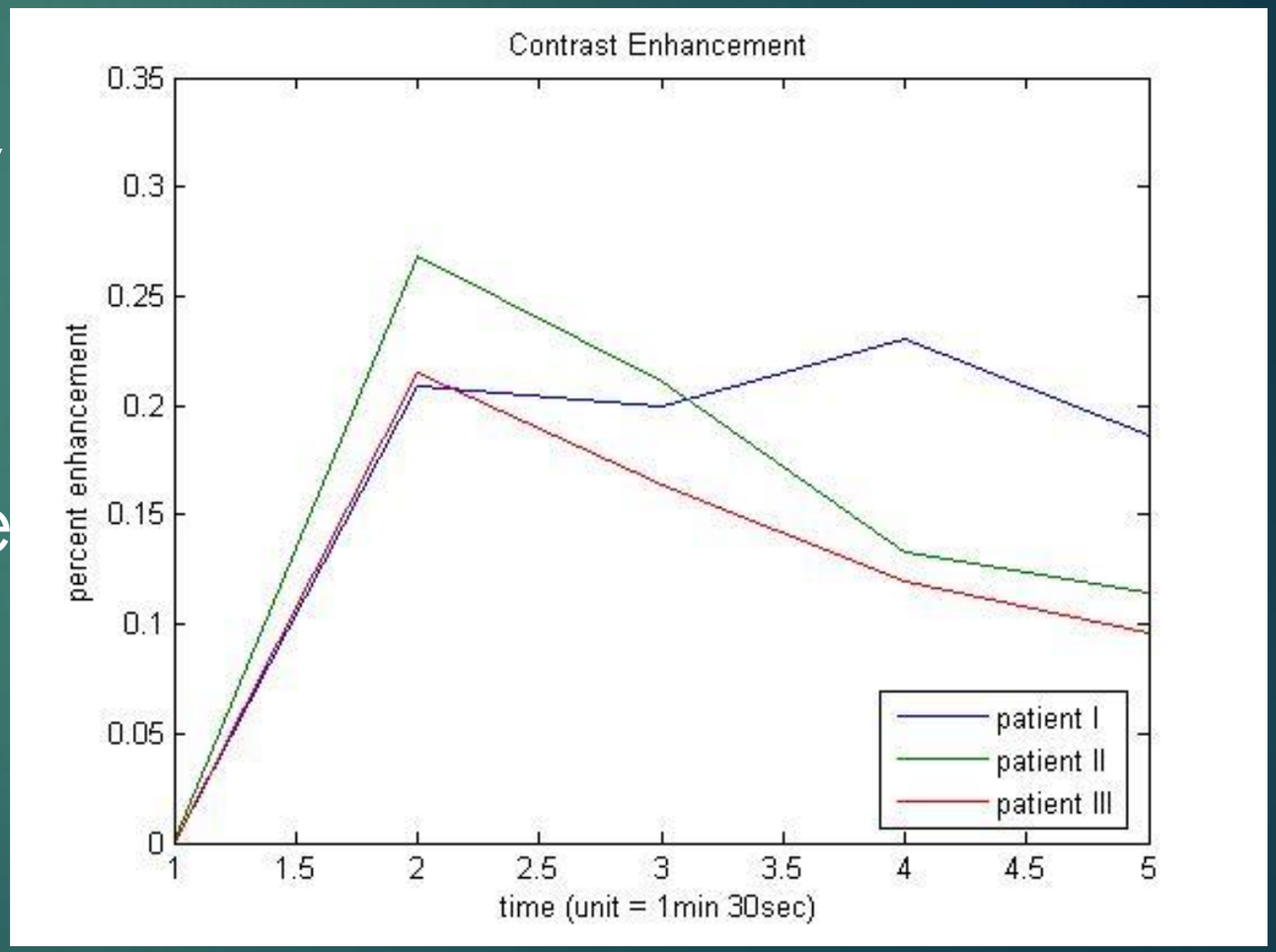
- 1x1x1 cm^3
- 40 t_1 incr.
- 8NEX/ Δt_1
- ~10 minutes
- 1.5T
- ~30 minutes for 3 locations



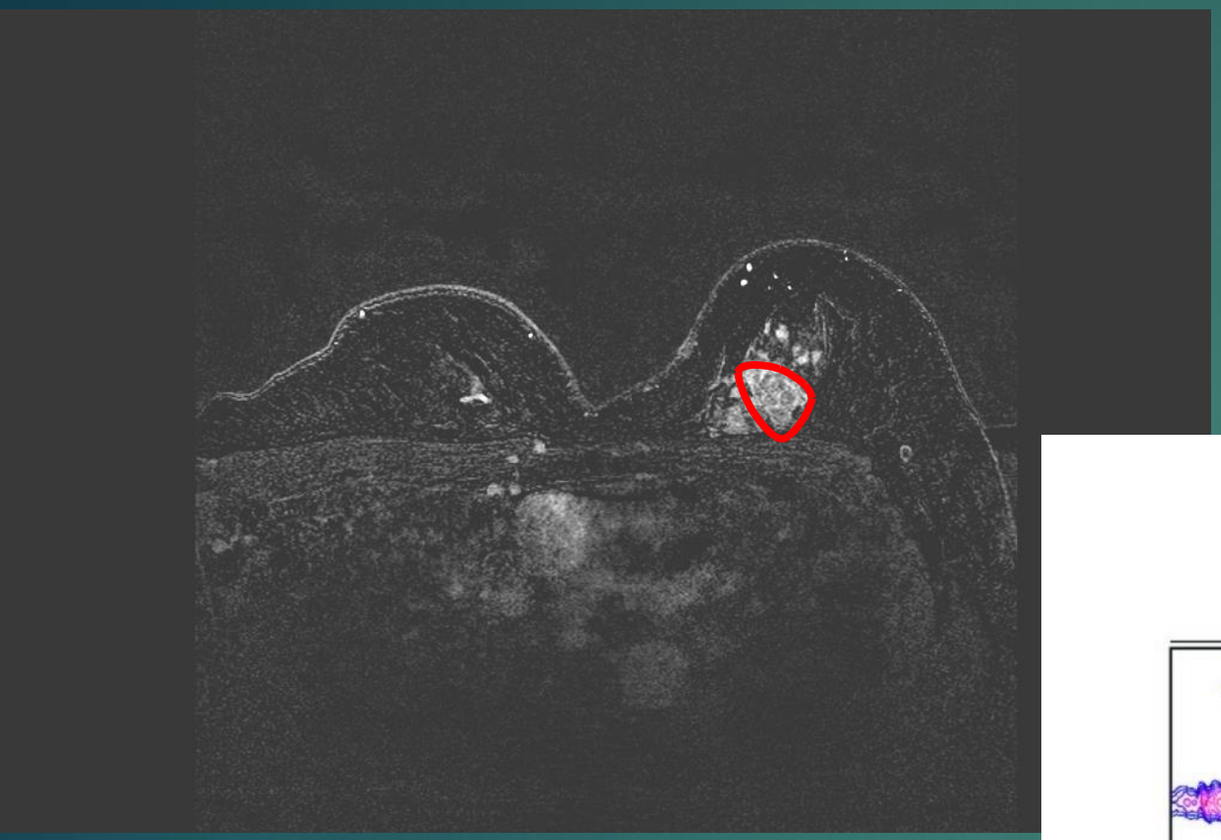
(Thomas JMRI2001)

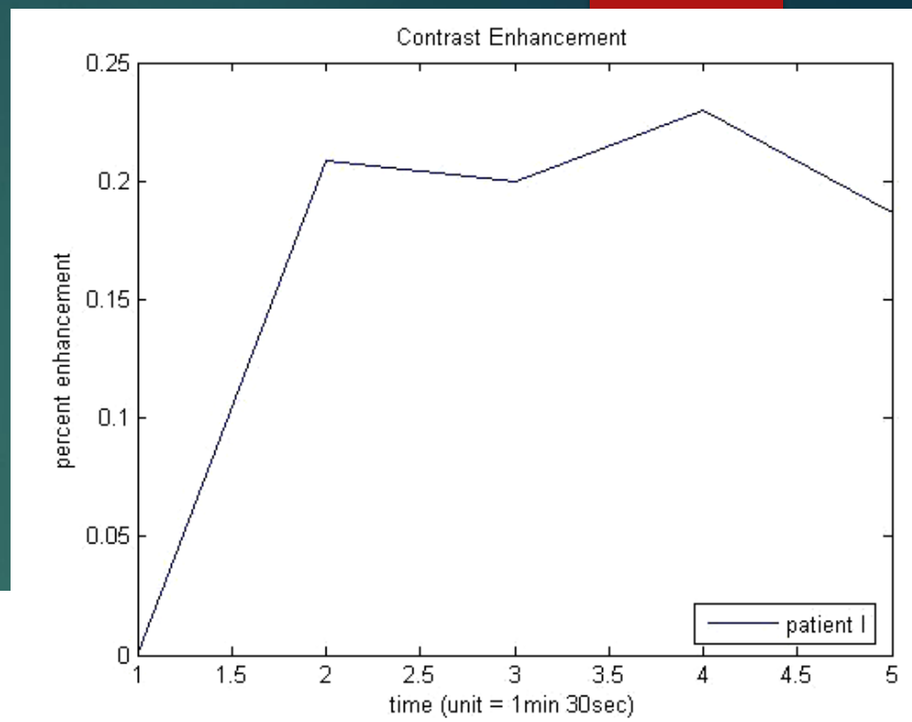
DCE-MRI

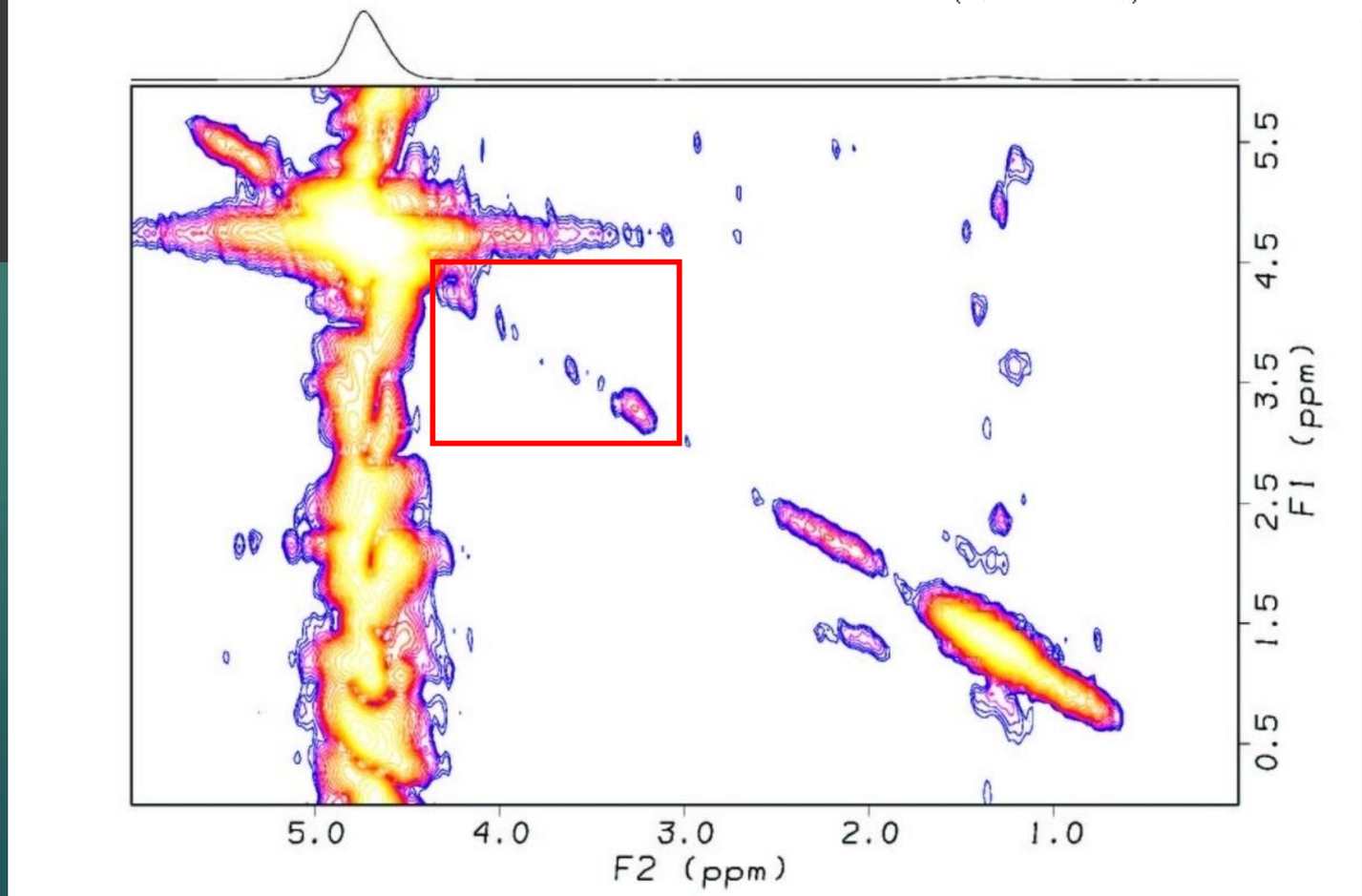
- Graphs of DCE-MRI curves for malignant patients
- Plots show uncertainty in enhancement curves
- Patient I shows a plateau shaped curve which cannot differentiate malignant from benign lesion



Malignant Patient with Type II enhancement



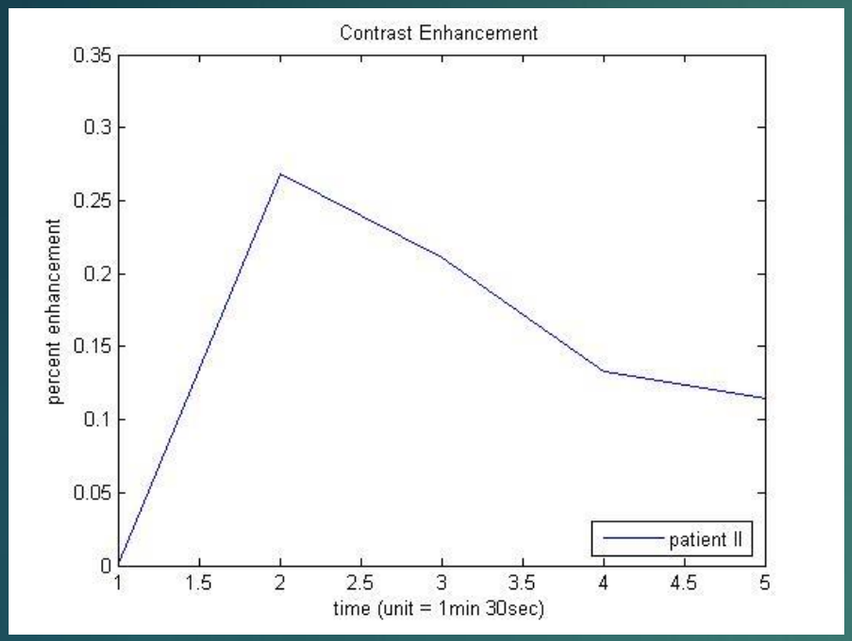




-56 yo malignant patient
-1x1x1 cm3
-45 t1 incr.
-8NEX/ Δ t1
-12 minutes

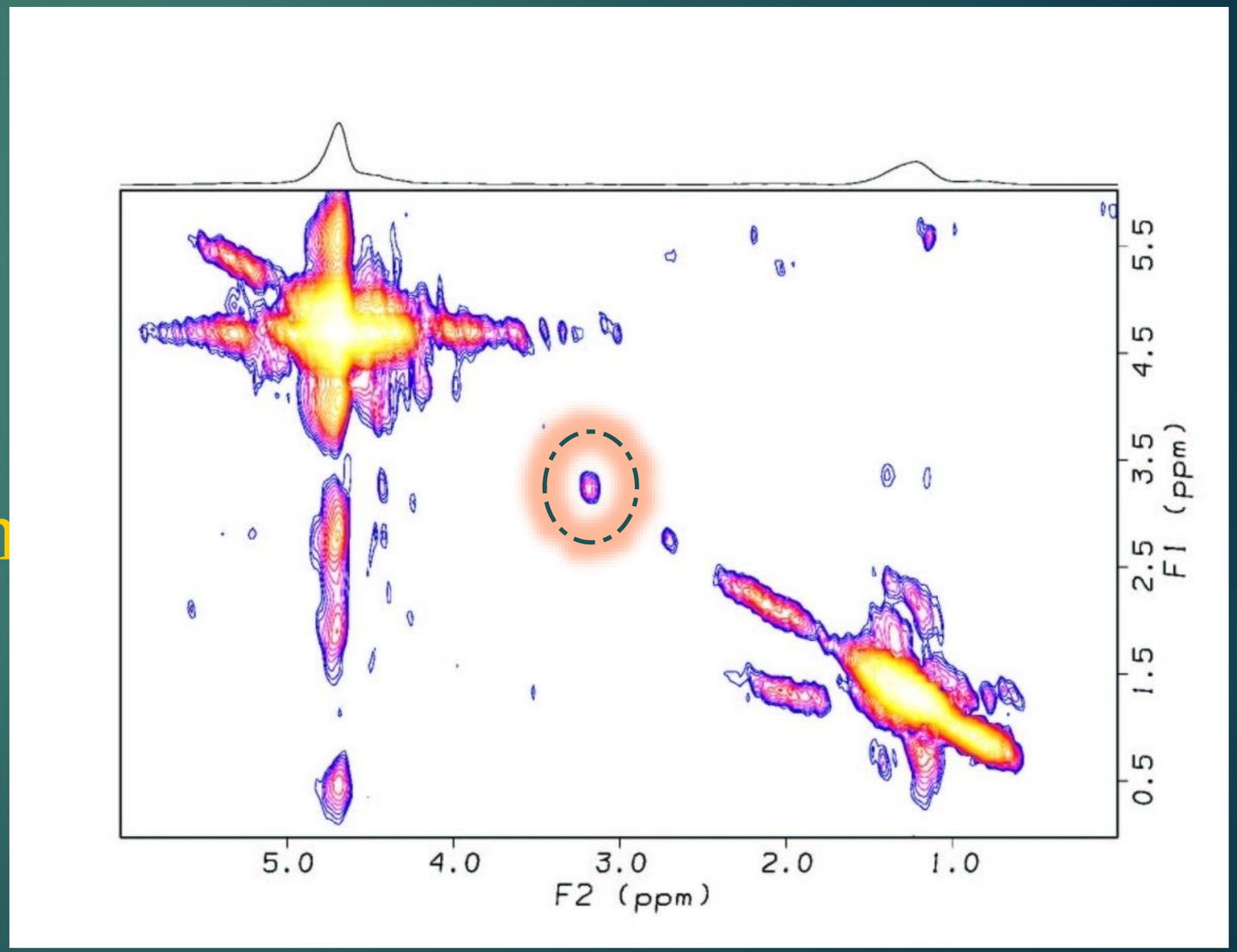
Lipnick 2010

2D L-COSY of Breast Cancer



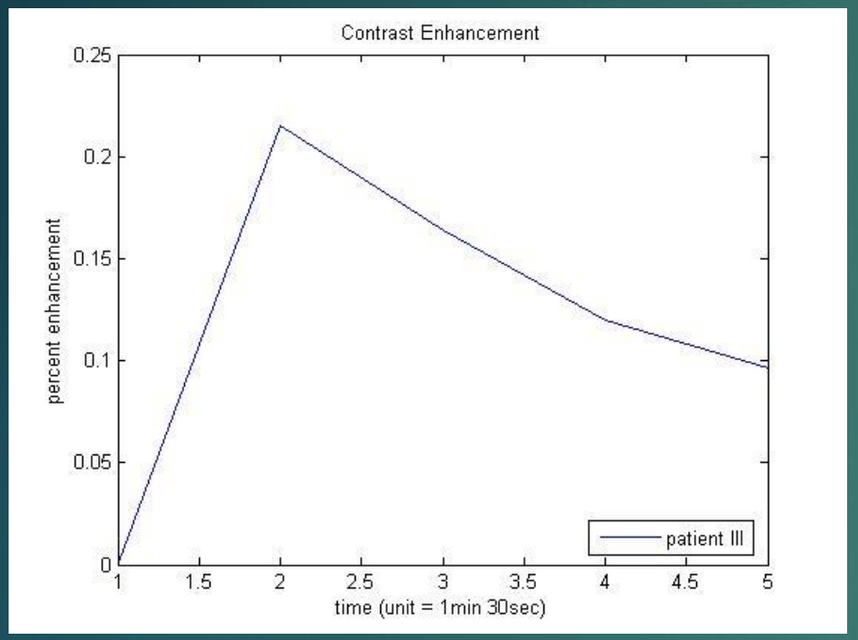
A 55 yo malignant patien

- **1x1x1** cm^3
- **45 t_1 incr.**
- **8NEX/ Δt_1**
- **12 minutes**



Lipnick 2010

2D L-COSY of Breast Cancer



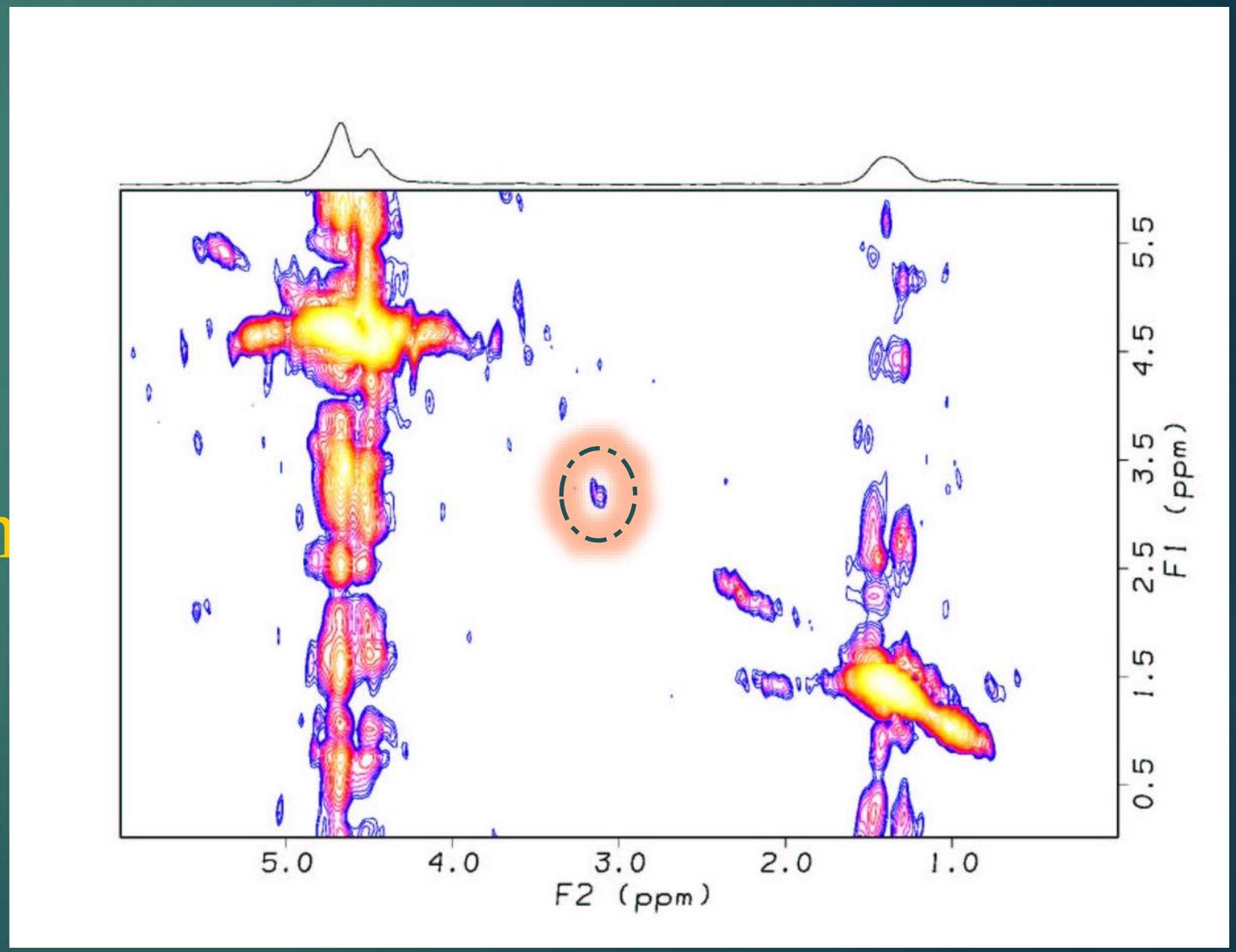
A 55 yo malignant patien

-1x1x1 cm^3

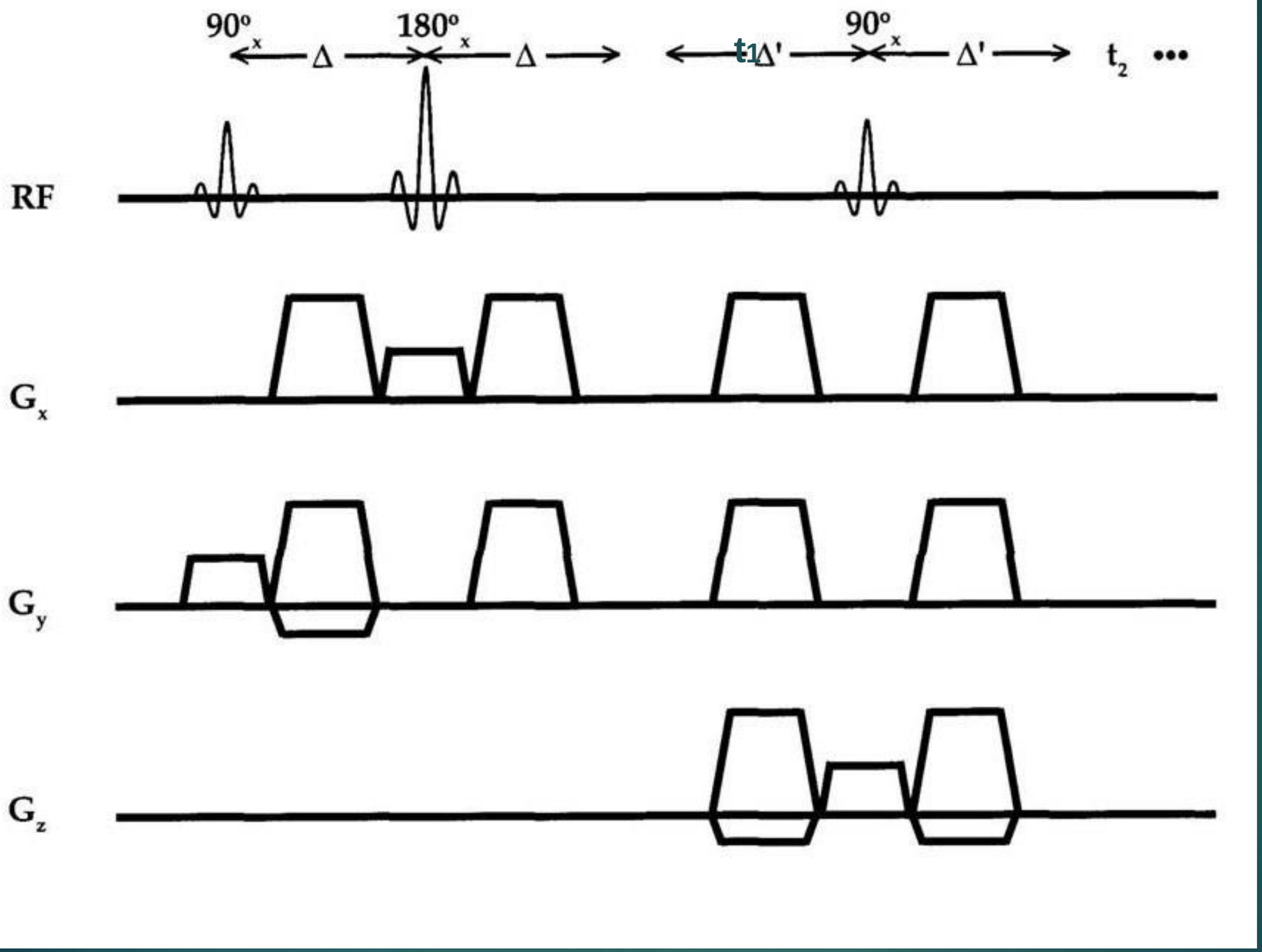
-45 t_1 incr.

-8NEX/ Δt_1

-12 minutes



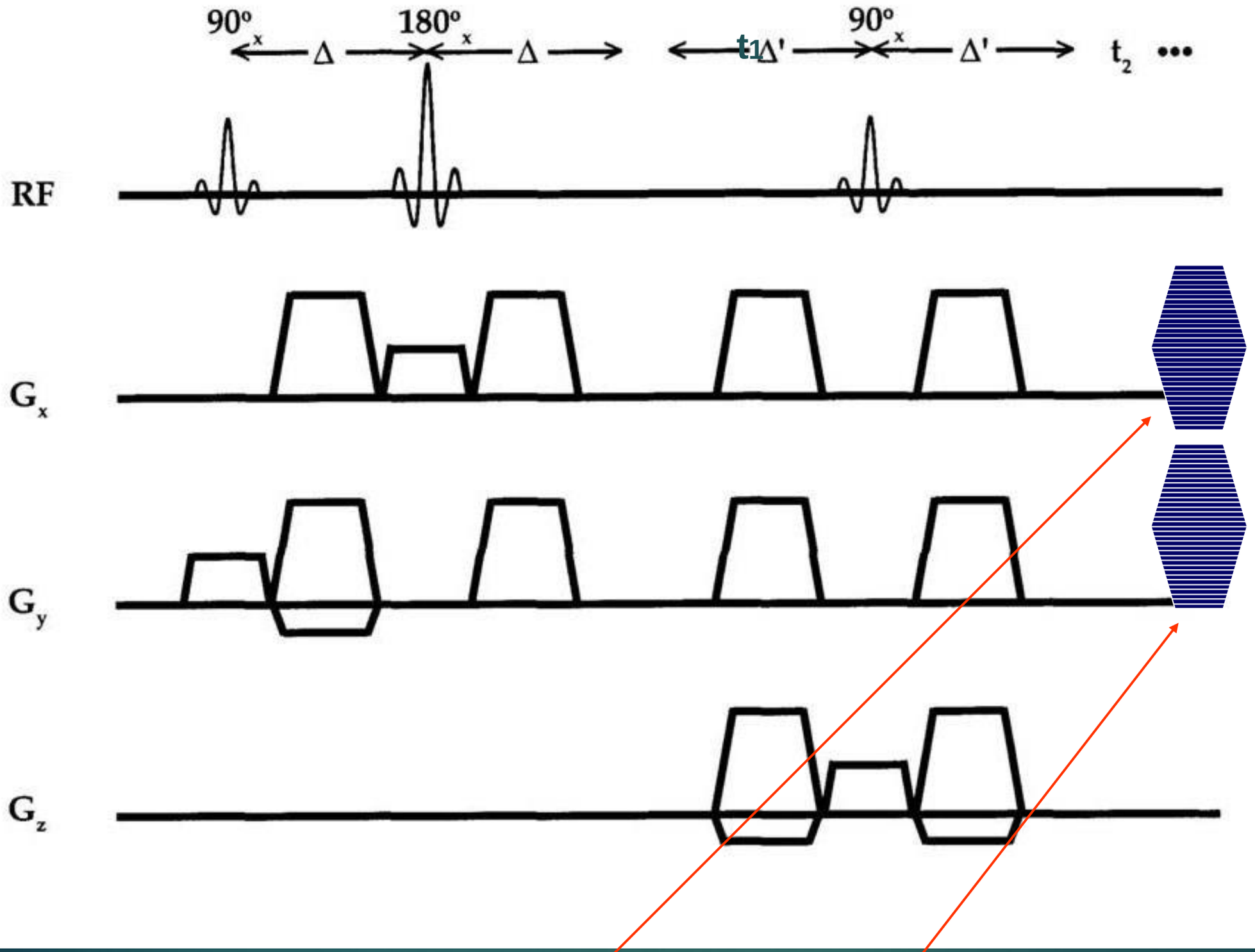
4D ^{1}H MR Spectroscopic Imaging: 2 Spectral + 2 Spatial Dimensions



Total Scan time

TR * N (t1 Encodings) * Averages

= 2s*128*8= 34minutes

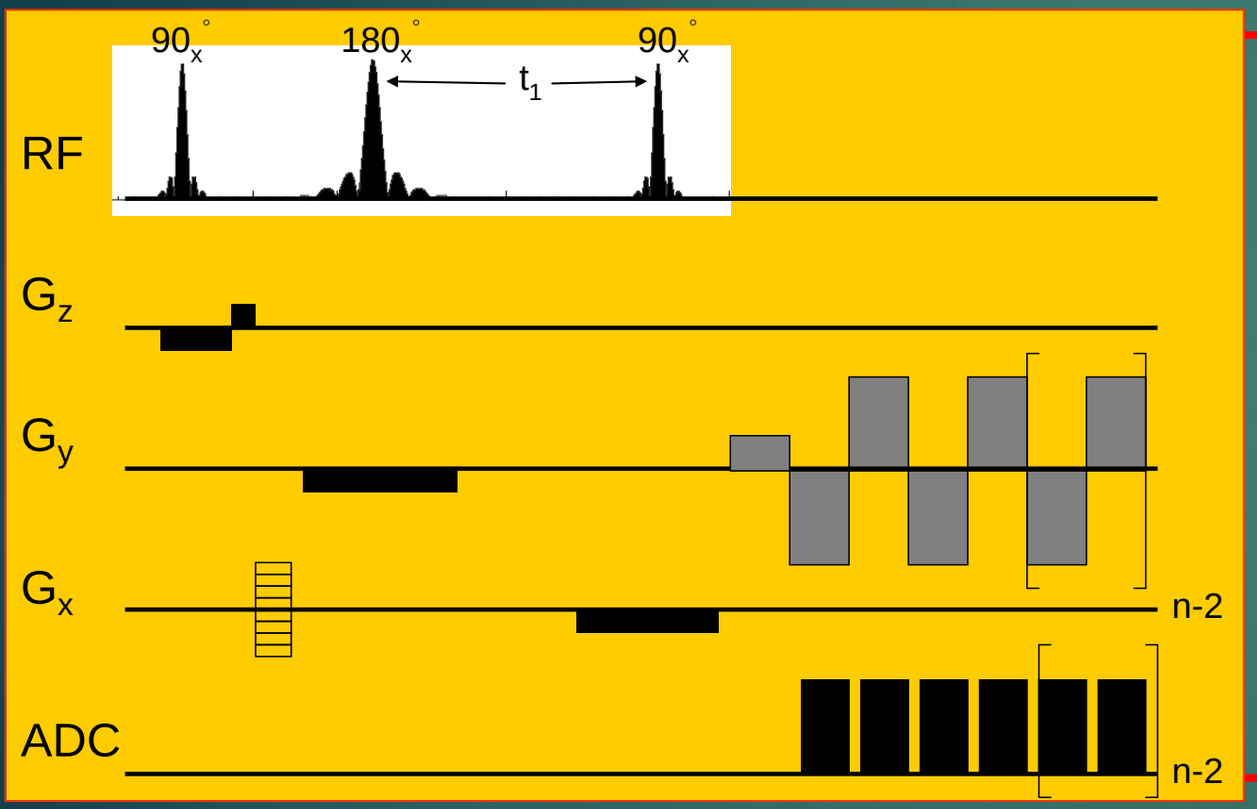


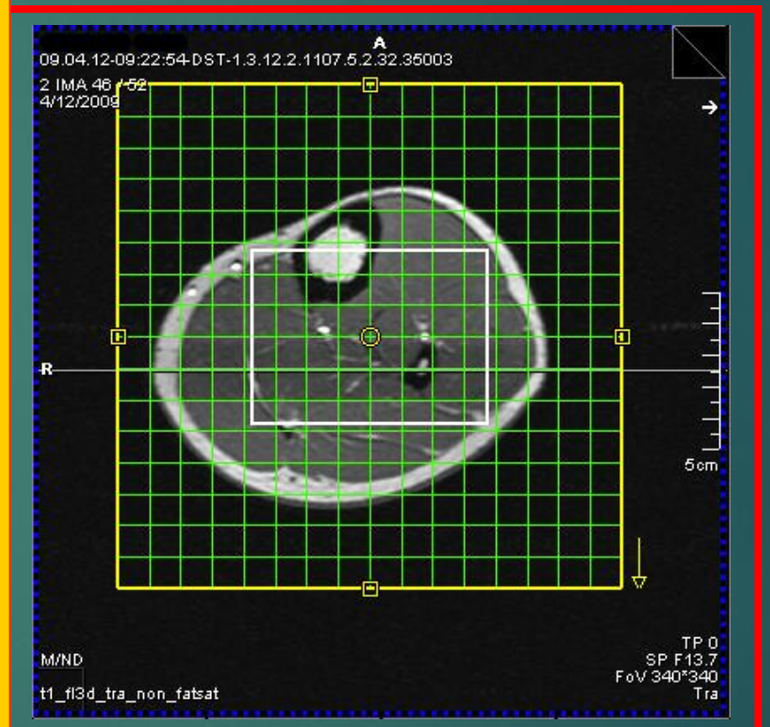
=
2s*128*1*16*1
6
= 546minutes
Or
=2s*128*1*16
= 18.2 hours

Total Scan time

TR* N (t1 Encodings) *N (y-Phase Encodings) N (X-Phase Encodings) * N (t1 Encodings) * Averages

Echo-Planar Correlated Spectroscopic Imaging (EP-COSI)



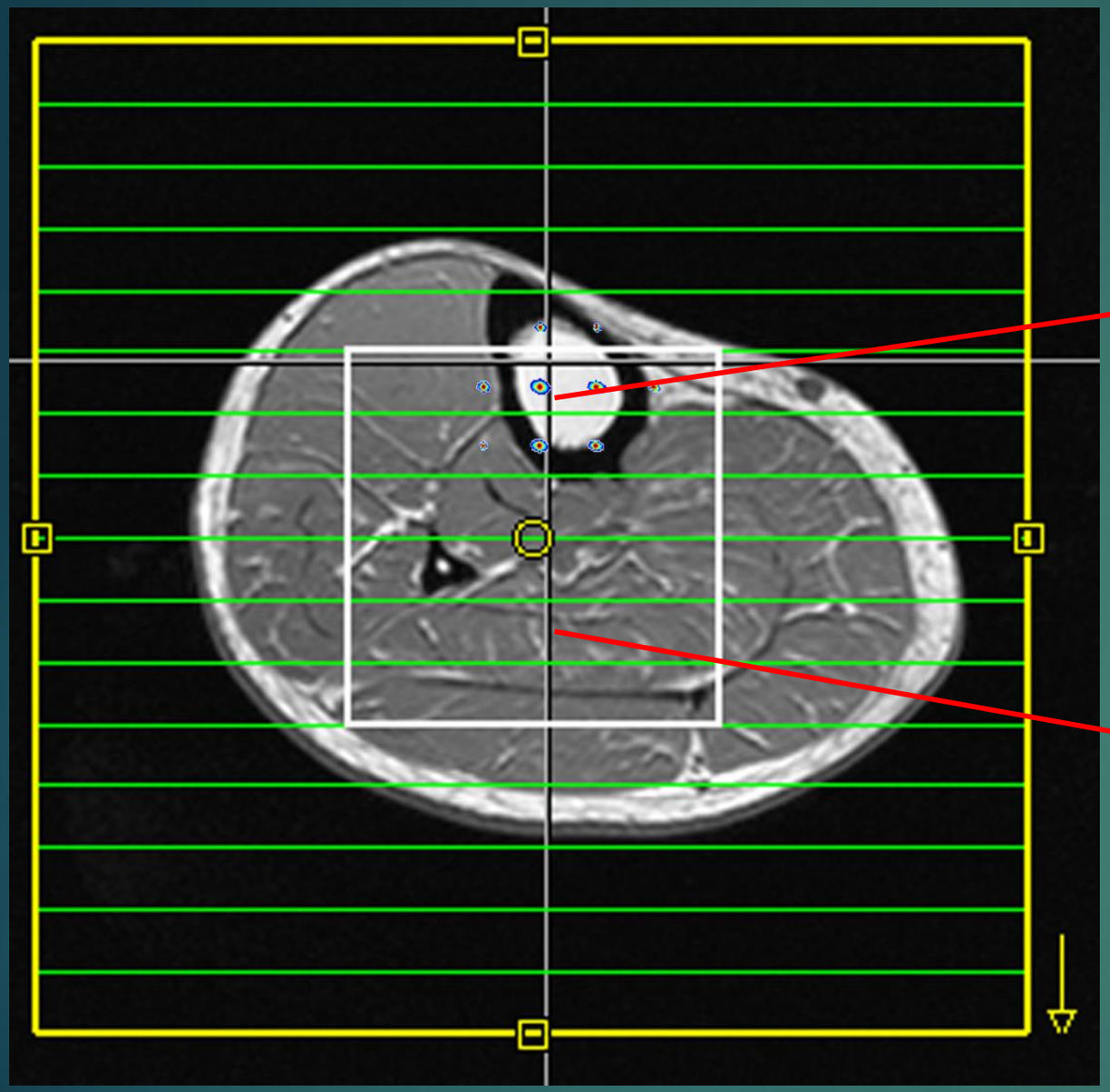


Scan time = N (X-Phase Encodings) * N (t1 Encodings) * TR

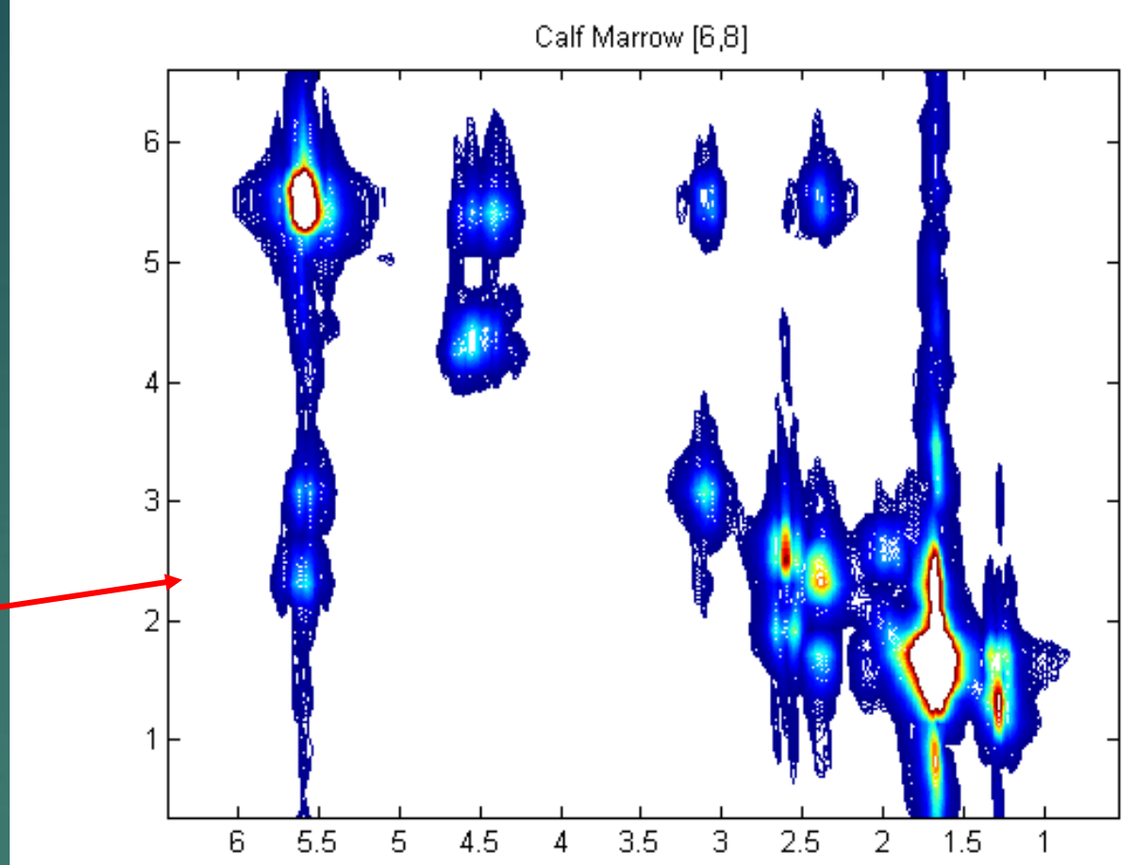
= 2s*128*1*16 = 68minutes

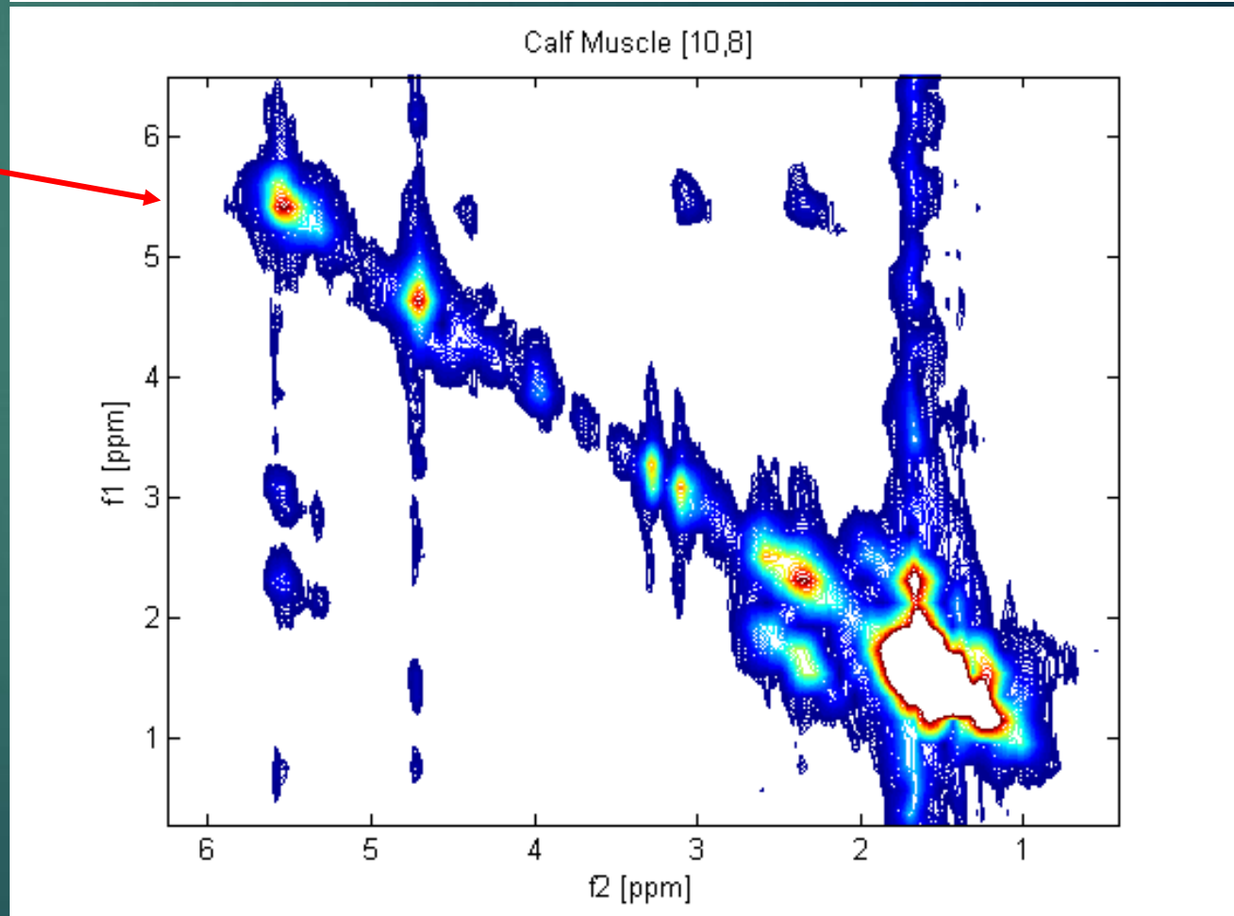
Lipnick et al, MRM 2010

EP-COSI of Human Calf *in vivo*



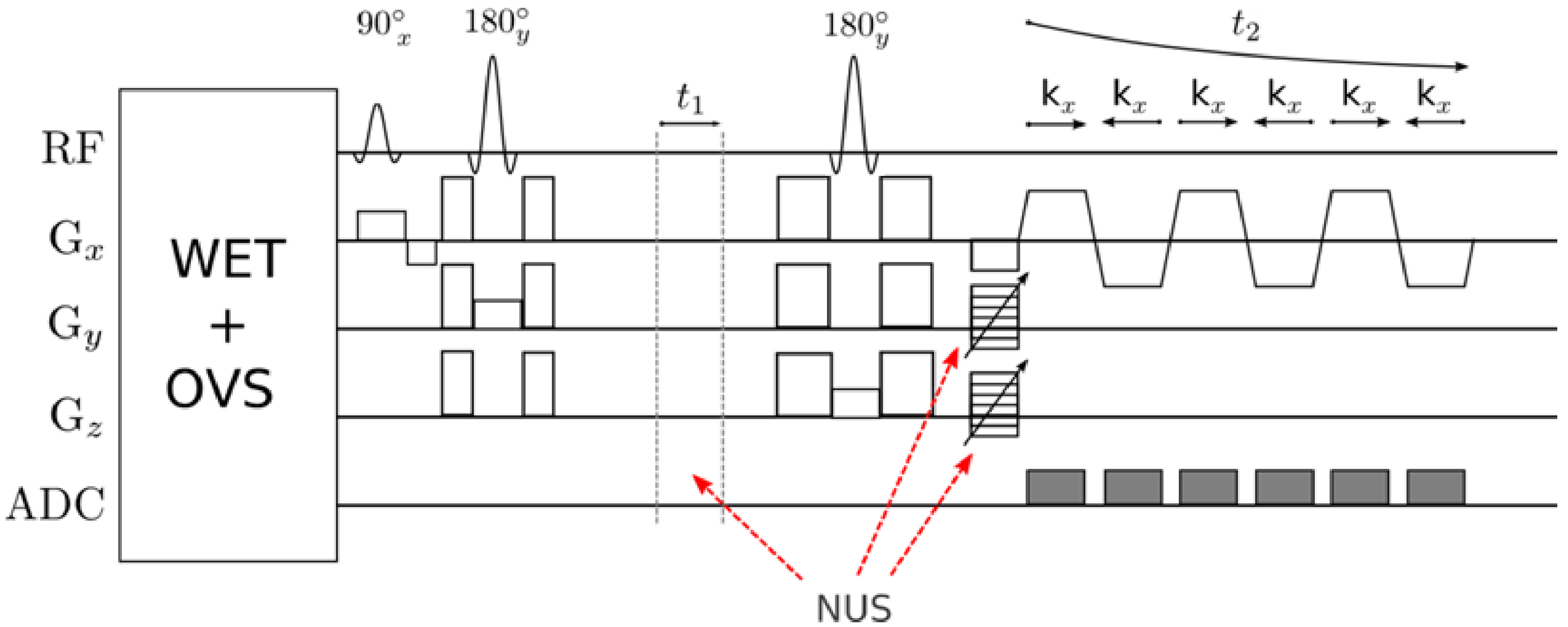
3T MRI, TR/TE=1.5s/30ms,
CP-Ext (T/R), 16x16 (x,y), FOV 16cm,
Extracted VOI of 1x1x2cm^3 and
Total Duration of 20 minutes





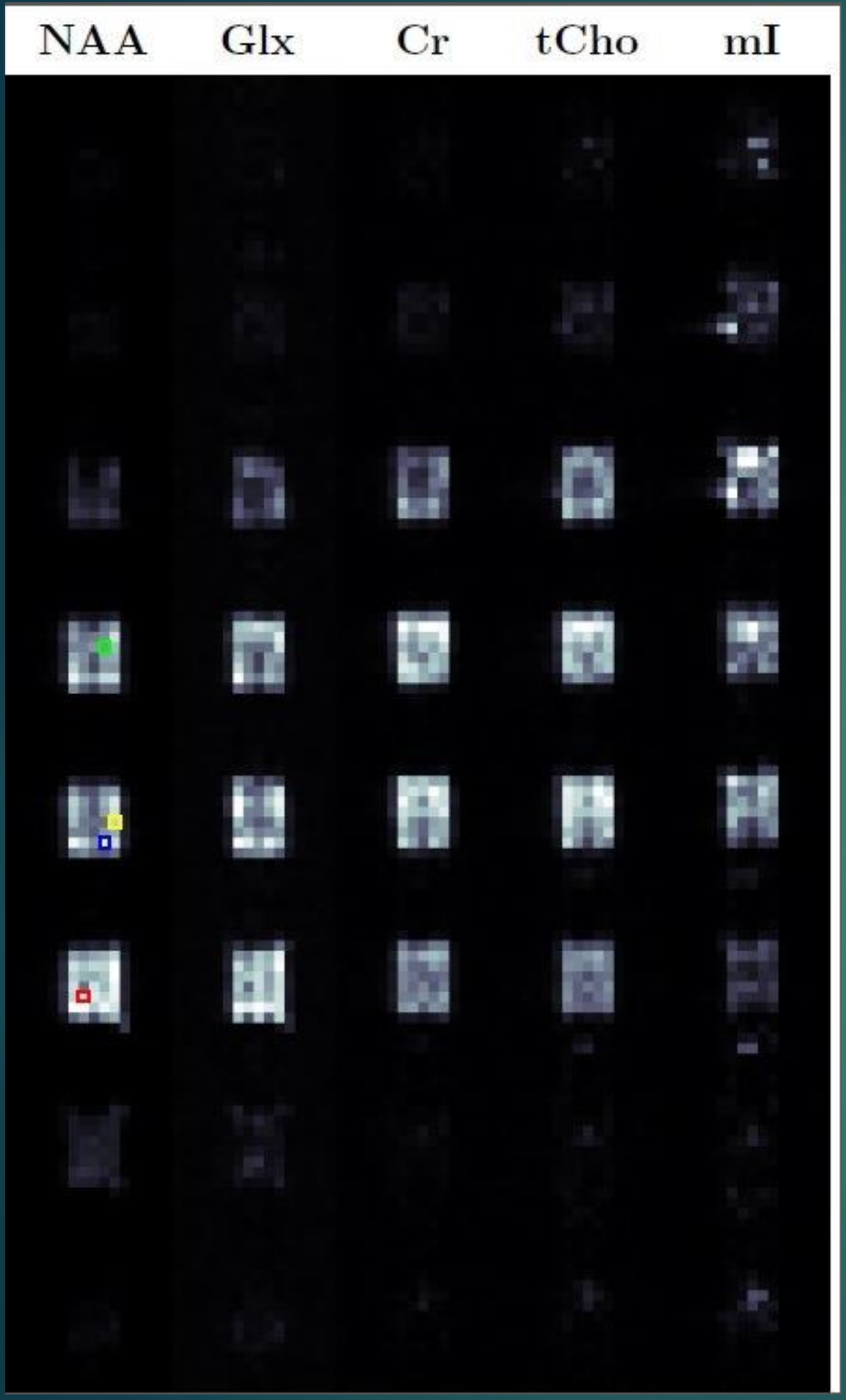
5. 2D spectral+ 3D Spatial Encoding

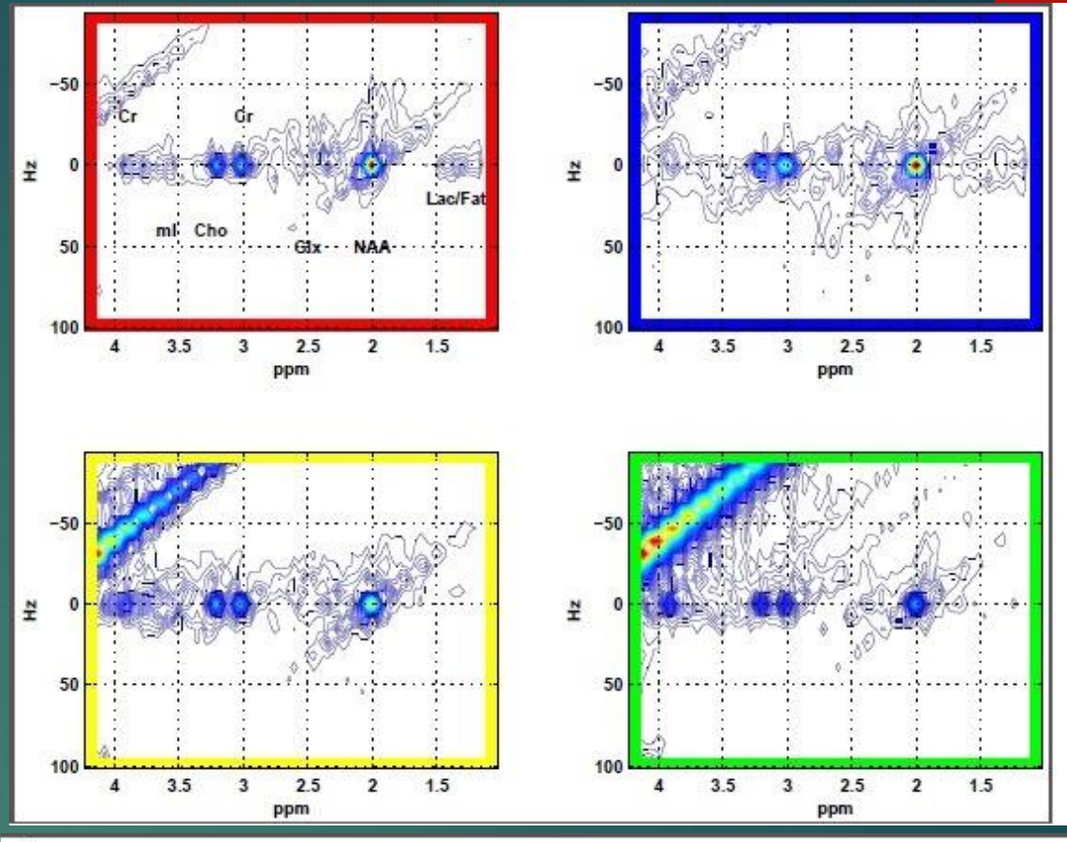
5D Echo-Planar J-resolved Spectroscopic Imaging *with NUS*

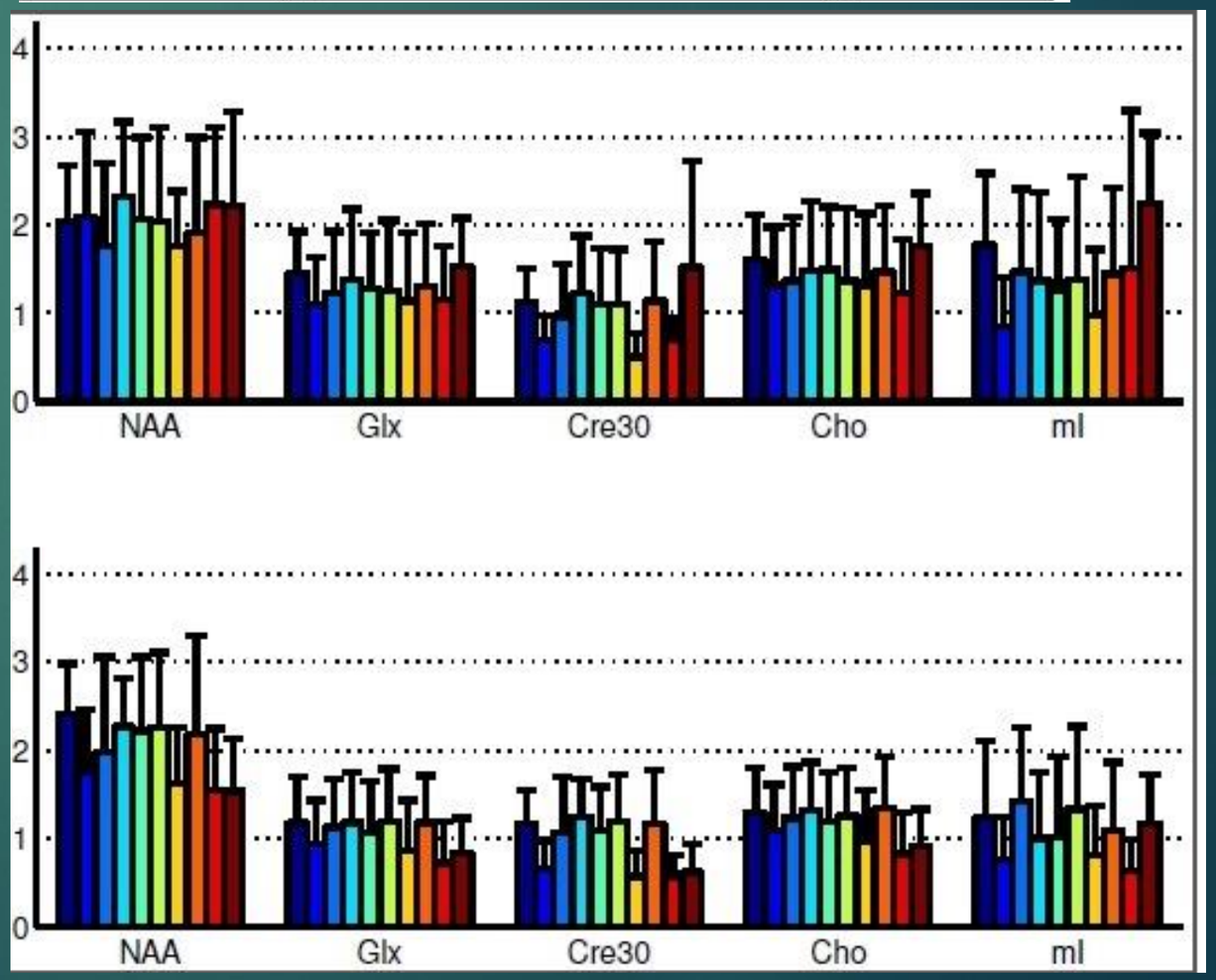


- 3D CSI/MRSI (32x32x16) -410 minutes
  - 3D EPSI (32x16) – 12.8 minutes
- 3D EPSI+2DJRES (32x16x64)- 819 minutes
- 5D EPJRESI (16x8x64) 8X NUS- 21 minutes

Wilson et al, MRM 2016





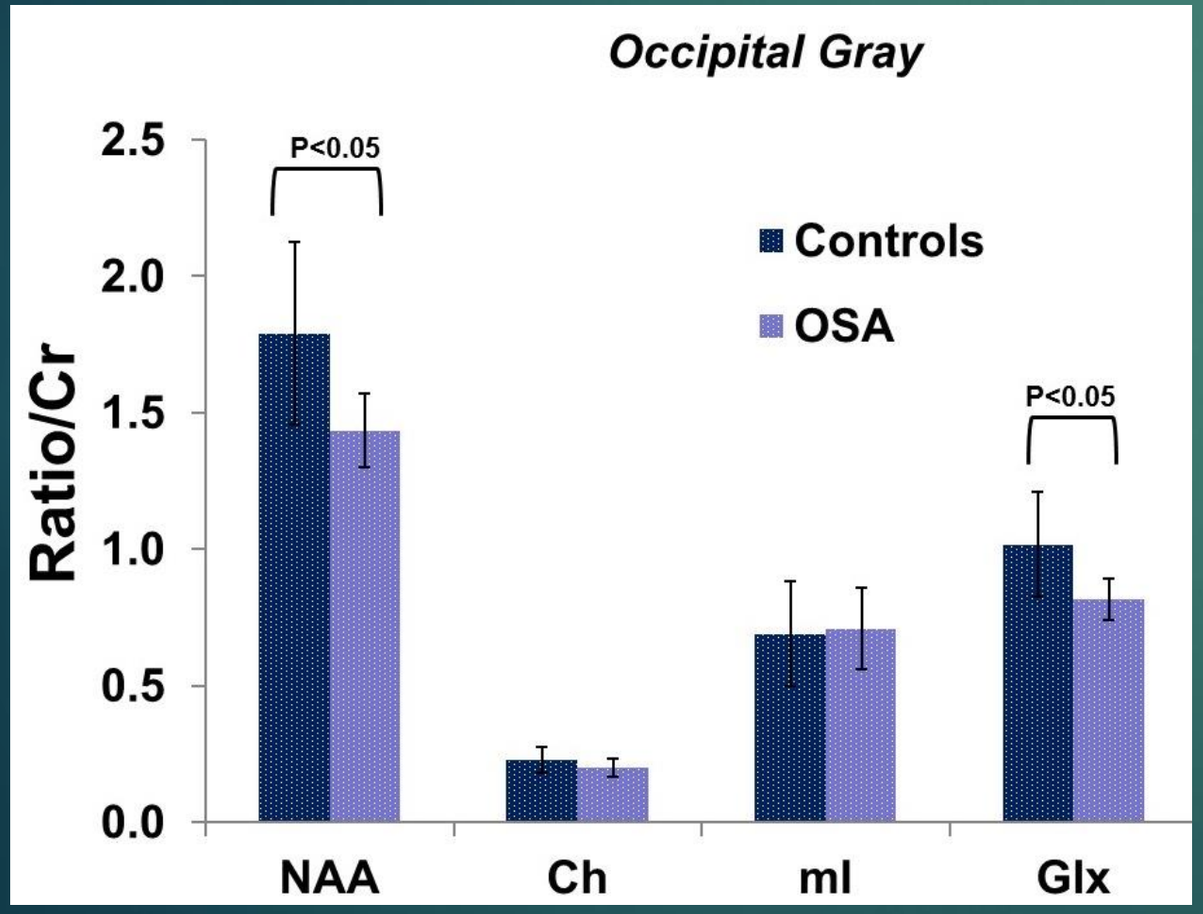


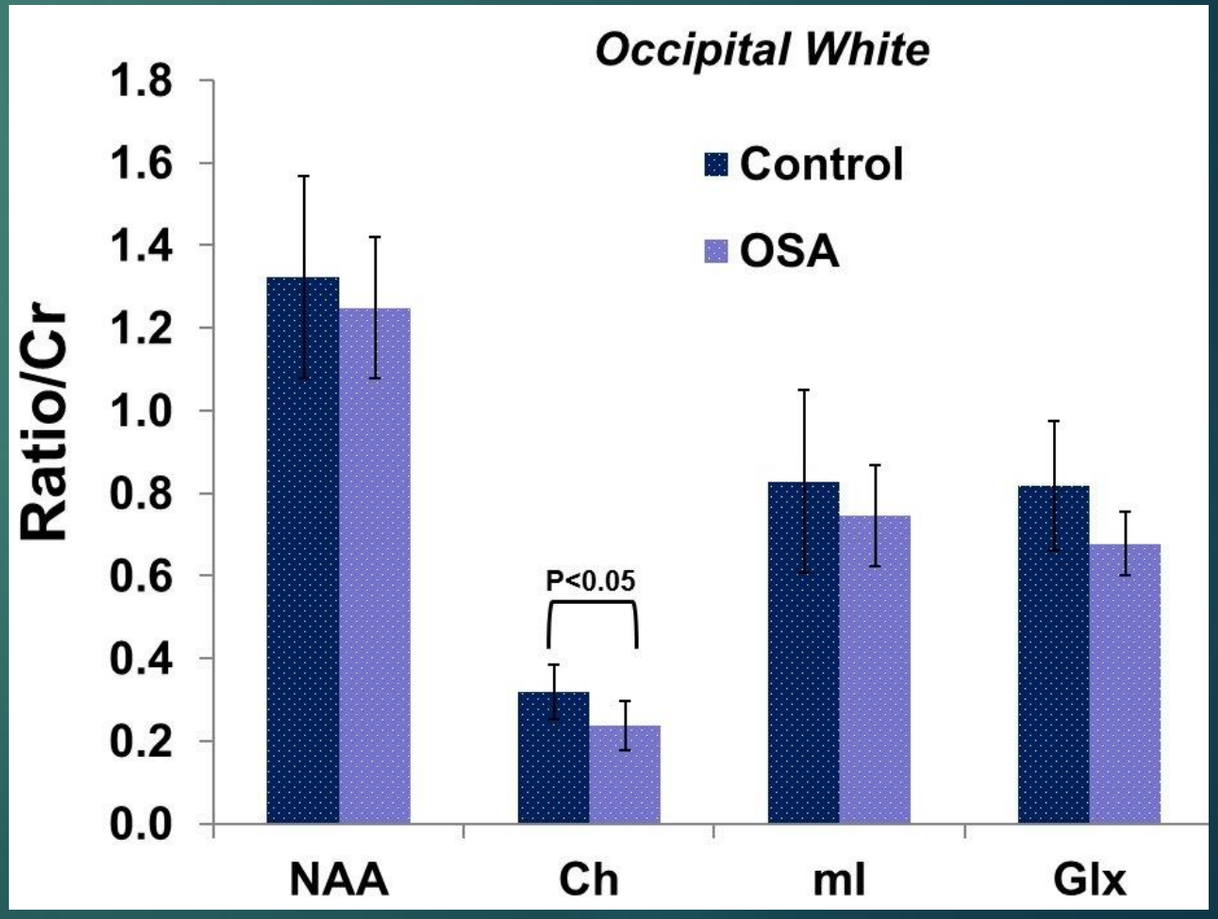
5D EP-JRESI 8X NUS-21 min

Wilson et al, MRM 2016

5D EP-JRESI 8X- OSA

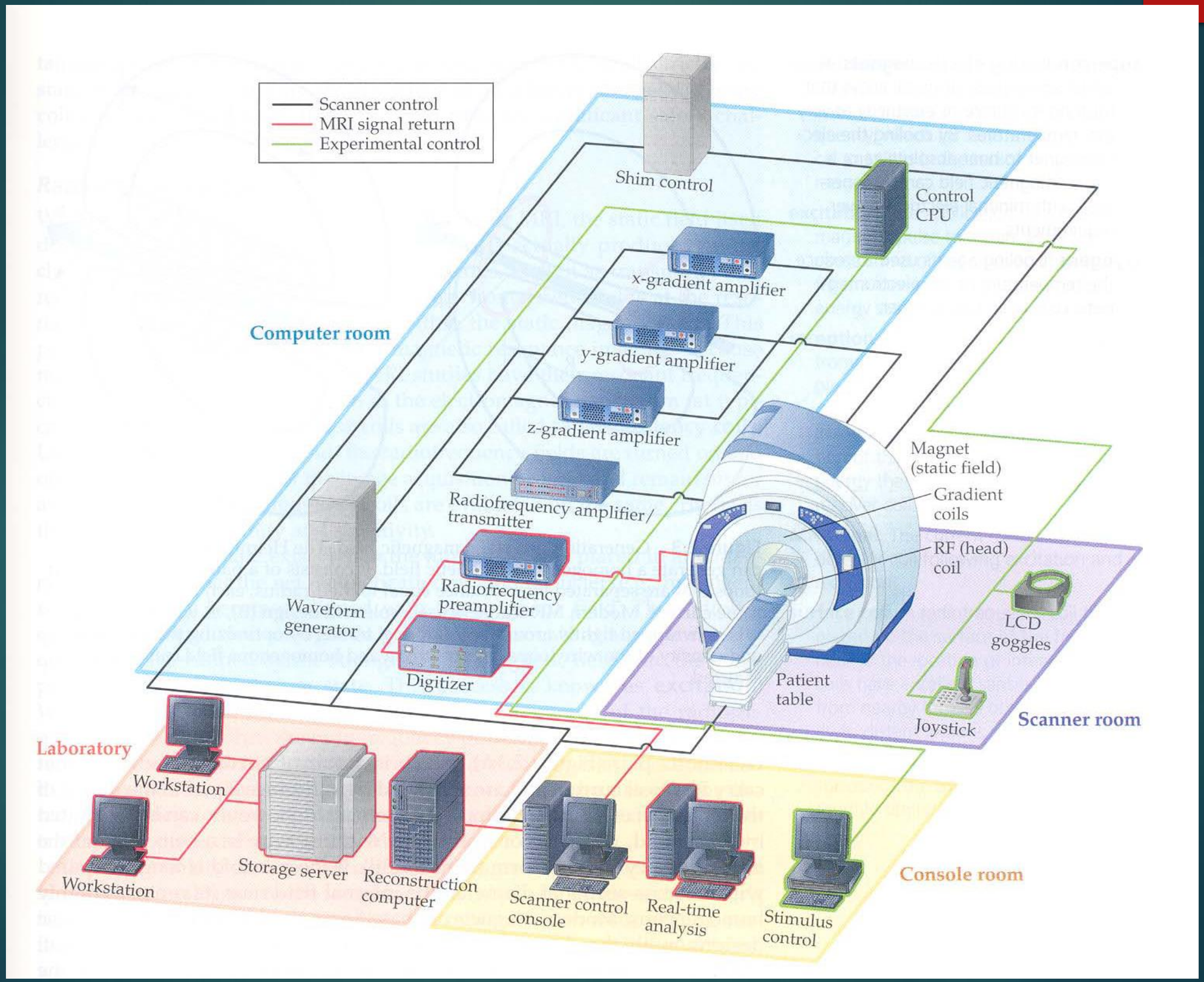
Metabolites →	NAA	Ch	mI	Glx
Healthy Controls				
Occipital White	16.94	8.52	14.33	6.08
Occipetal Gray	1.62	1.56	2.79	10.82
Left Insular Cortex	7.30	6.53	5.19	2.90
Left Parietal Insular Cortex	1.17	8.82	10.31	6.58
OSA patients post CPAP				
Occipital White	9.20	4.92	8.70	14.50
Occipetal Gray	6.03	3.20	4.00	10.00
Left Insular Cortex	2.70	1.22	18.69	1.40
Left Parietal Insular Cortex	2.70	1.97	14.81	9.56





Thomas ISMRM 2016

Overview of an MRI scanner



Conclusions

- MRI has become a revolution in Medicine during our time, thanks to NMR!
- MRI sequences can be easily translated into MR Spectroscopic Imaging

- EPSI, Spiral, SI-CONCEPT and Radial EPSI have been implemented on MRI scanners on 3T, 7T and 9.4T MRI scanners

- Accelerated Polar and Radial MRSI data need gridding to Cartesian; acquisition less than 5 minutes may facilitate functional MRSI

- 3spatial+2 spectral accelerated acquisition & the MRSI data can be post-processed using linear and non-linear reconstruction (Compressed Sensing)

- 6D MRSI (3spatial+3 spectral) and more…….

THANK YOU

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