MEDICIS-Promed Leman School on Preclinical and Clinical Imaging with Radioisotopes

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Medical imaging techniques: CT and MRI

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Medical physicist

Presentation

- Nick Ryckx, PhD
- SSRMP medical physicist
  - Institute of radiation physics, CHUV
  - Specialty in radiodiagnostics
    - Computed tomography (CT)
    - Radioscopy
    - Nuclear medicine

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- Computed tomography (CT)
  - Image acquisition
  - Dosimetry
- Magnetic resonance imaging
  - Background physics
  - Image acquisition

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- Computed tomography (CT)
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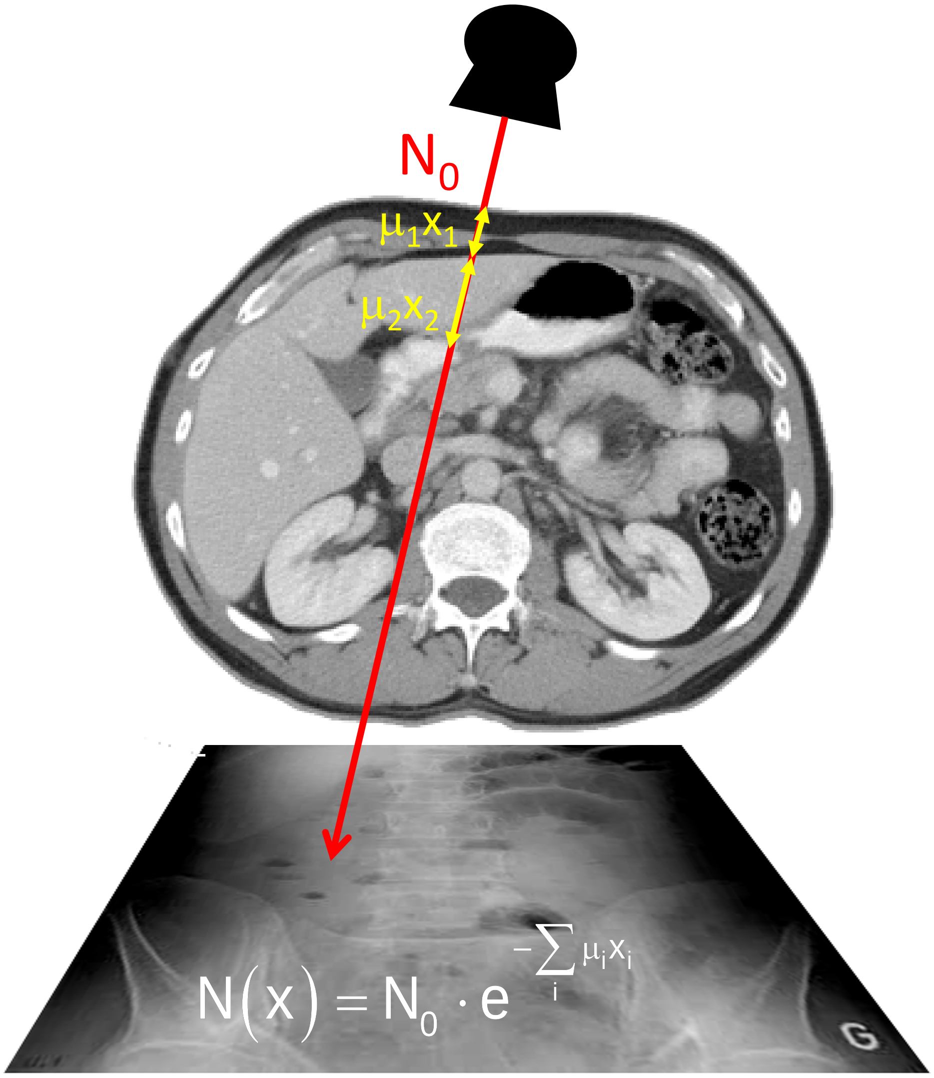
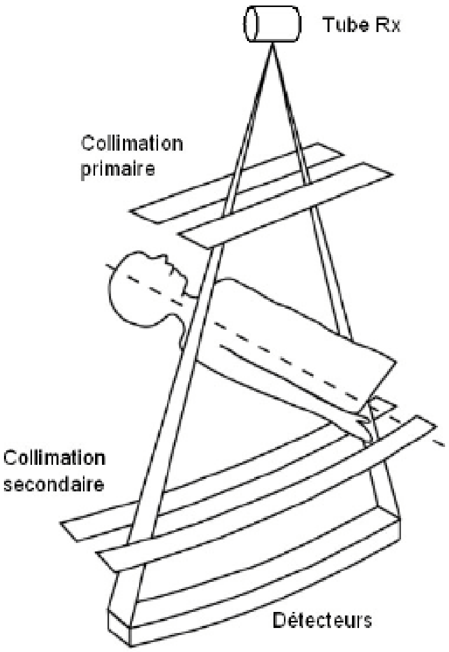
Conventional X-ray imaging

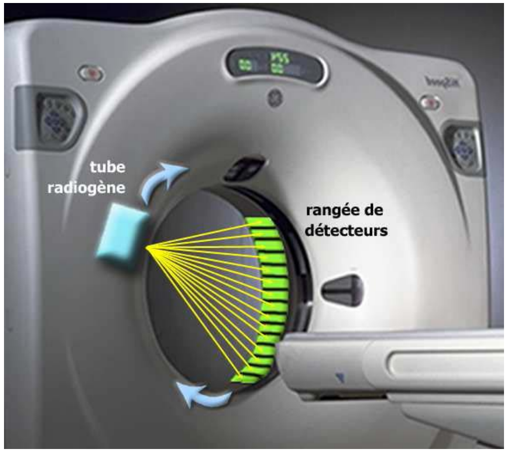


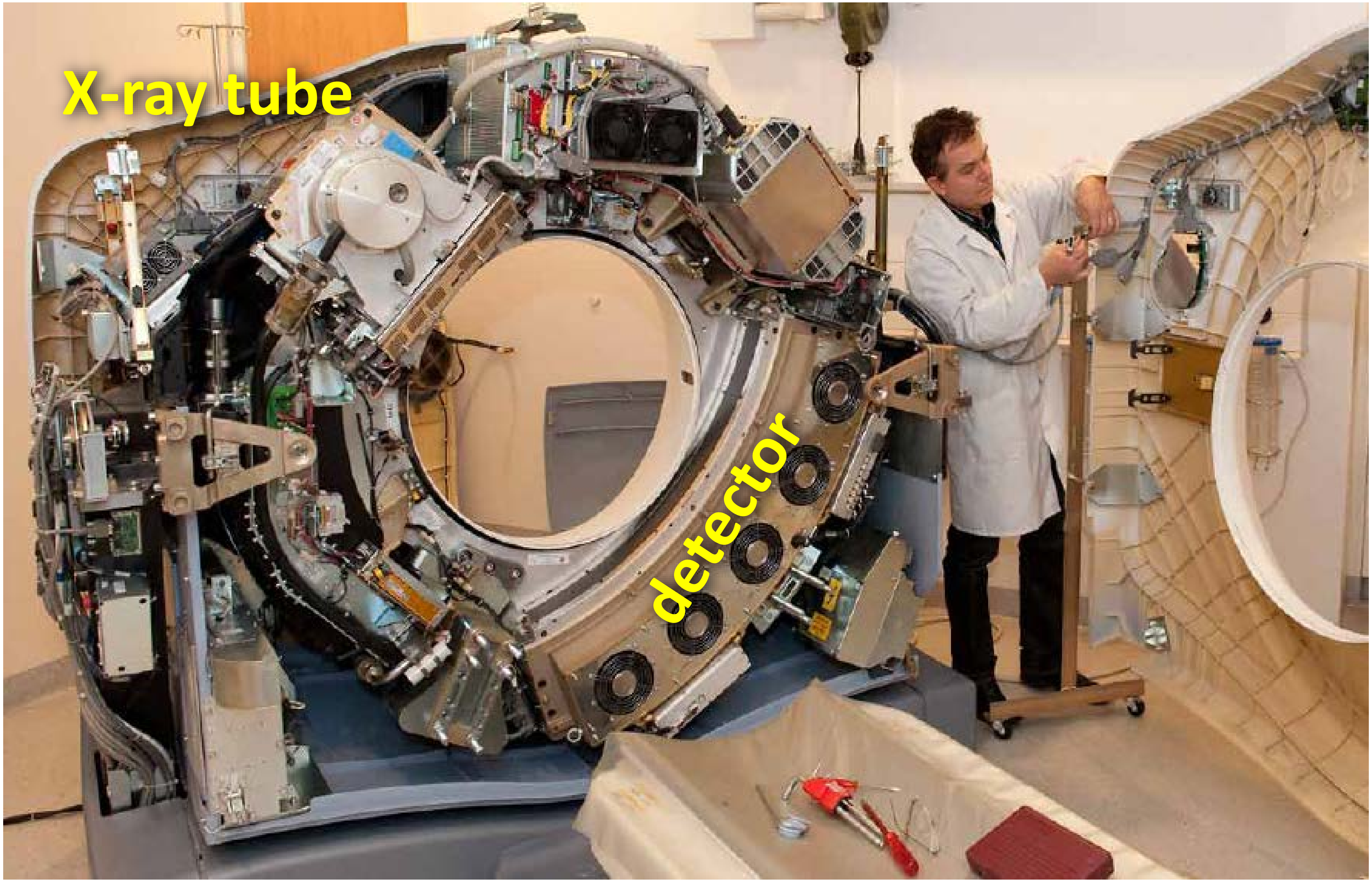
- All anatomical structures projected on a single plane
- Impossible to calculate linear attenuation coefficients (μ_t) of all body tissues

Apparatus

- Generator
- X-ray tube
- Collimation
- Detector






X-ray tube
detector

2-step procedure

- Step 1: **acquisition of X-ray attenuation profiles** under different angles angles (projections)
- Step 2: **image reconstruction** starting from the attenuation profiles

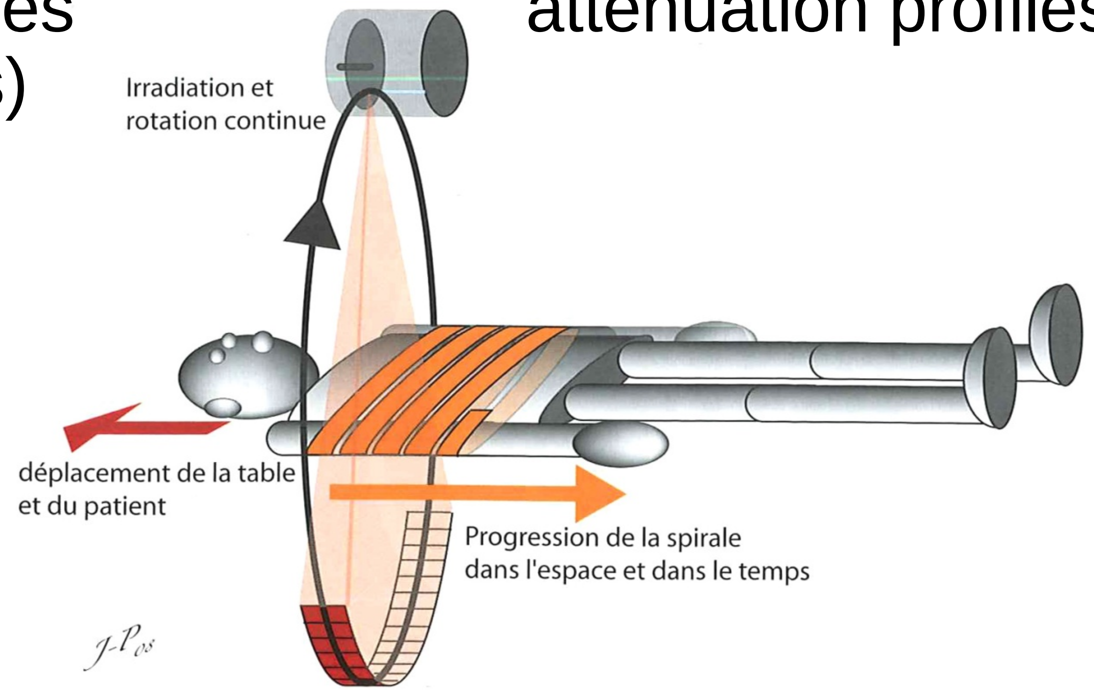
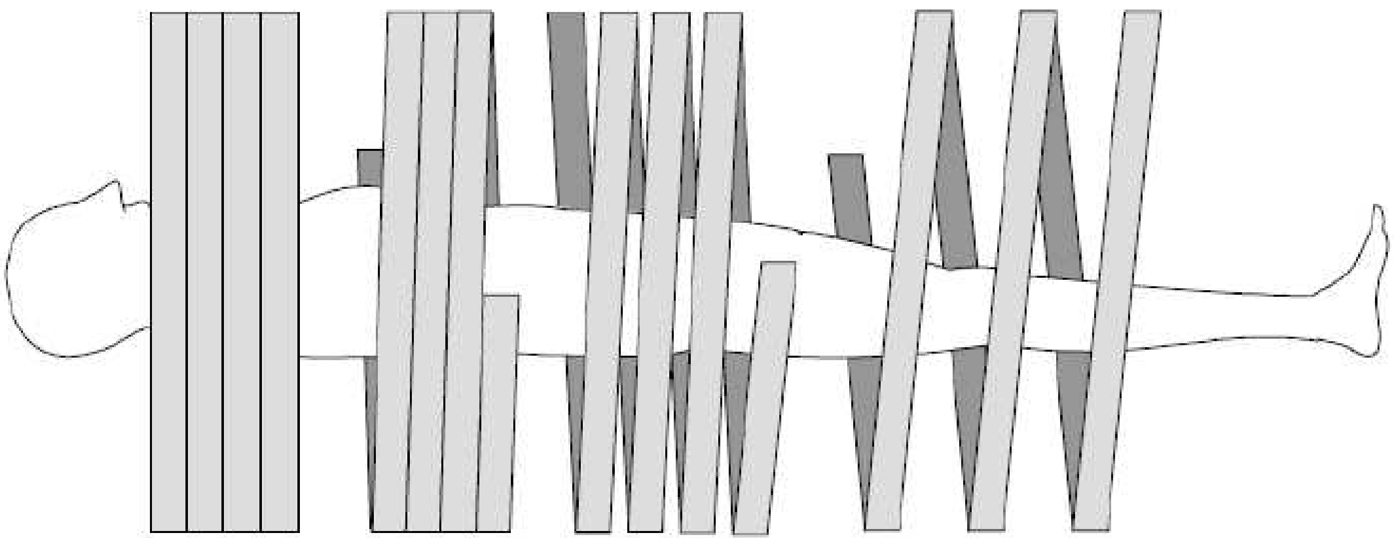


Figure 3.25. Principe du mode spiralé ou hélicoïdal.

Sequential or helical acquisition



Sequential (step-and-shoot)

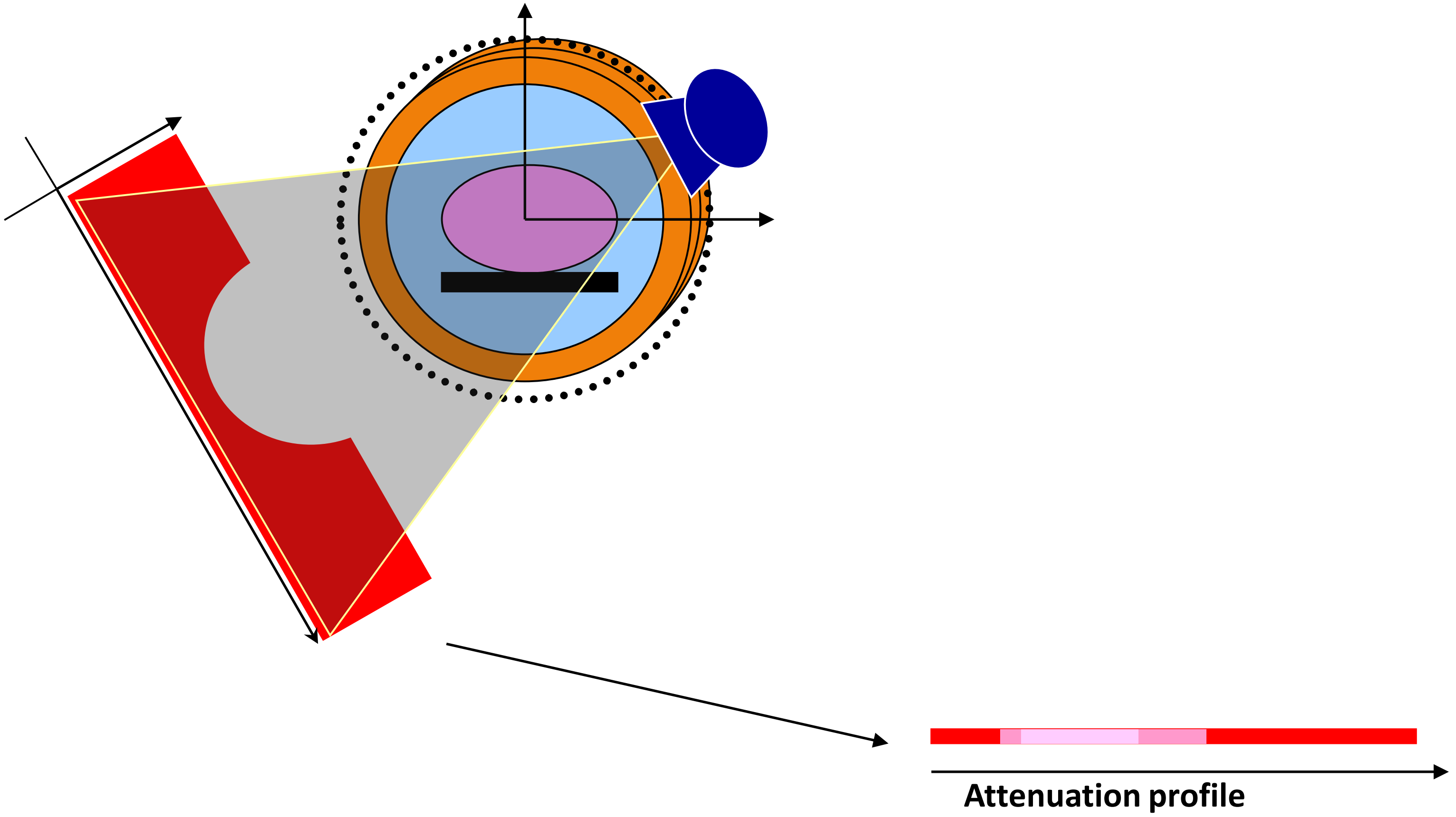
Pitch = 1
d = 1 x L

pitch = 1.5
d = 1,5 x L

pitch = 2
d = 2 x L

$$\text{Pitch} = \frac{\text{Table motion during a 360° rotation}}{\text{X-ray beam width}}$$

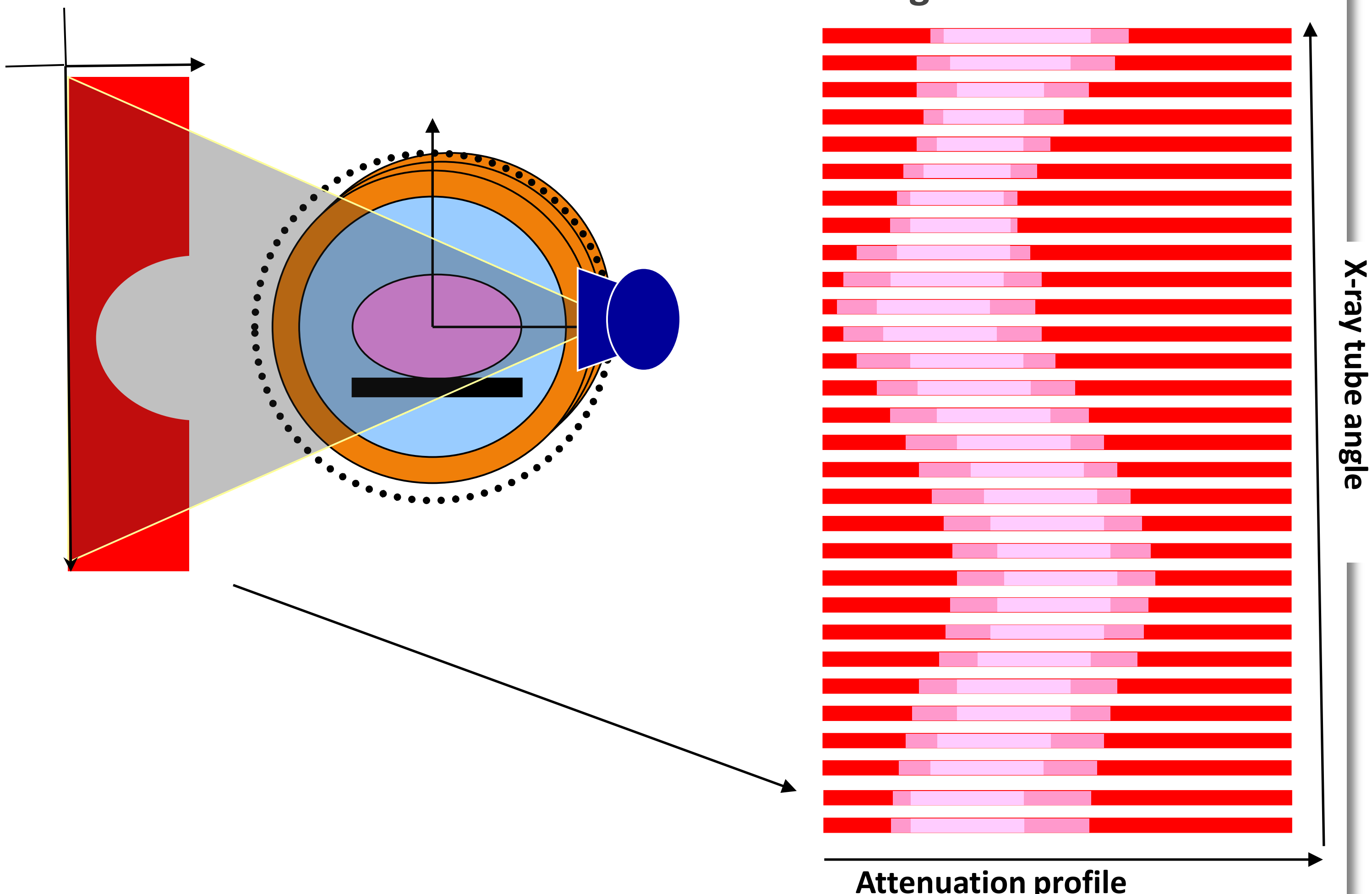
Acquisition



Acquisition

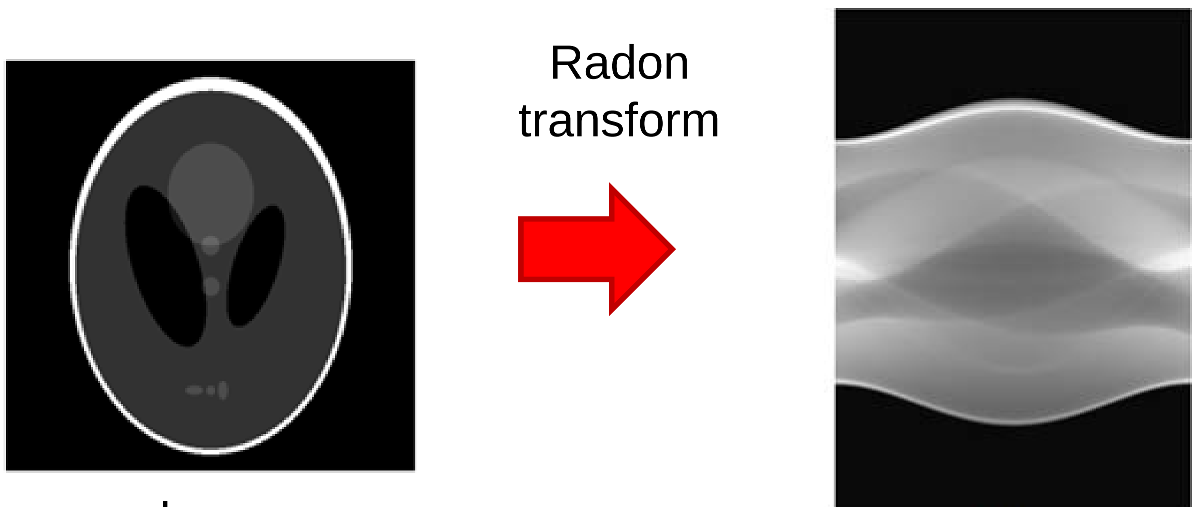
Piling of attenuation profiles as a function of tube angle:

Sinogram



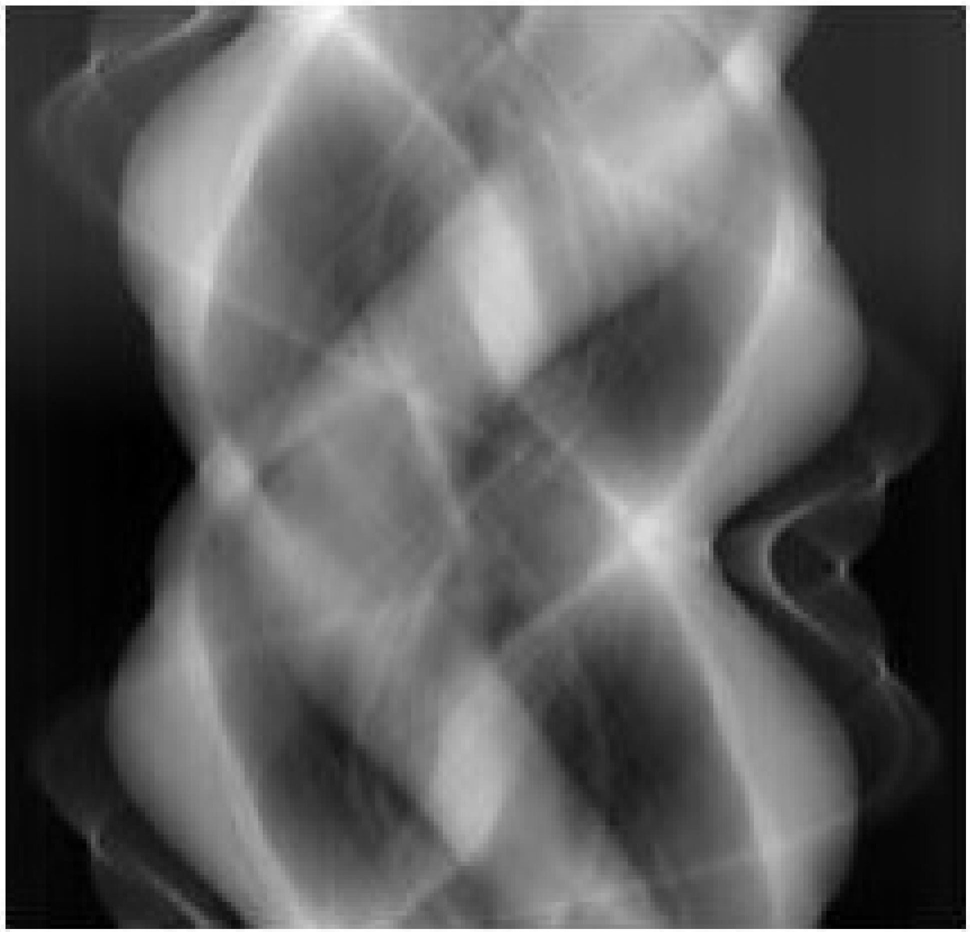
Acquisition

- Mathematical transformation of an image to a sinogram = Radon transform



Reconstruction

Starting from this acquisition …

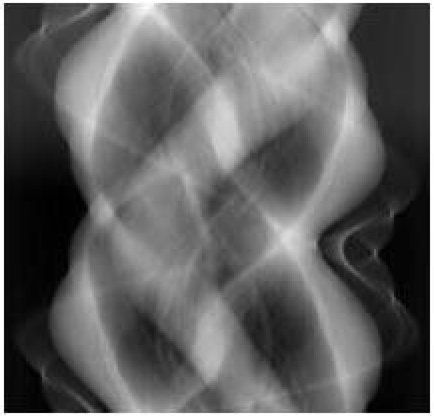


… let’s try do perform a diagnosis.

Reconstruction

- In CT, we acquire the sinogram

  ➔ Inverse problem
- To reconstruct the image from the sinogram
  - Inverse Radon transform



Sinogram

➔ Inverse Radon transform ➔

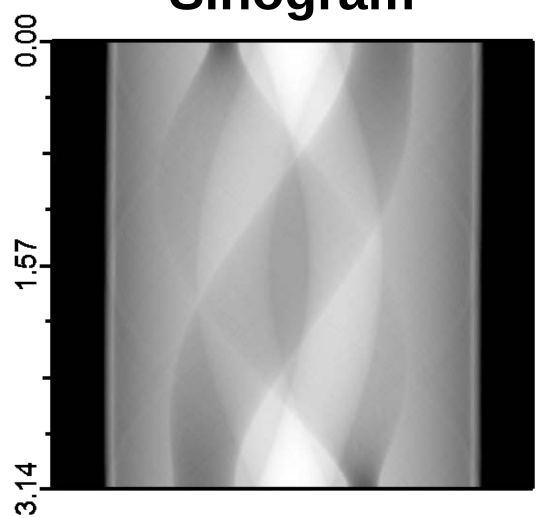
Image

Reconstruction

Original

Filtered back-projection (FBP)

Sinogram



Back-projection

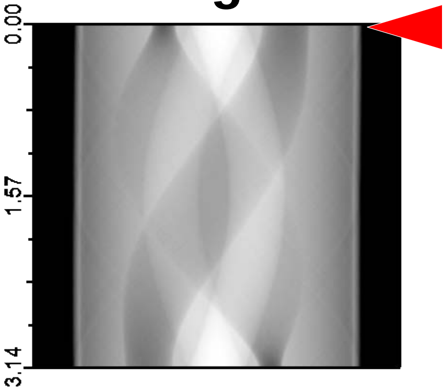
Basic principle of back-projection

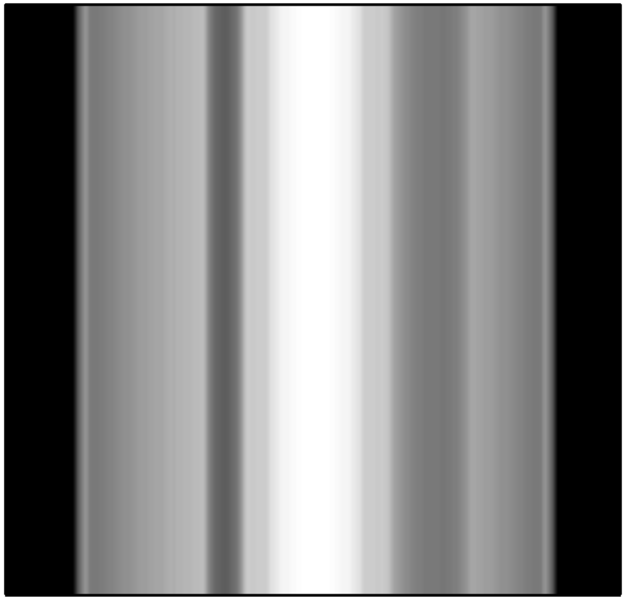
FBP

Each line in the sinogram is an attenation profile at a given tube angle

Original

Sinogram



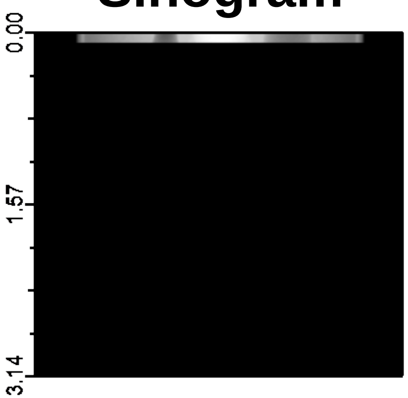


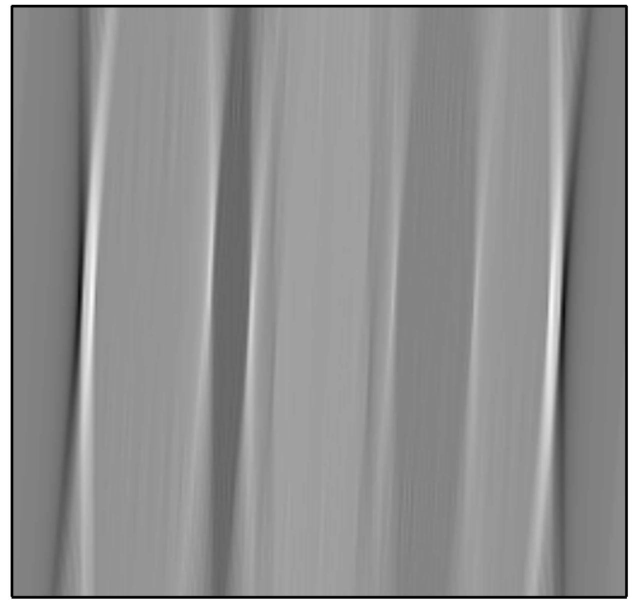
Only one attenuation profile used

FBP

Original

Sinogram



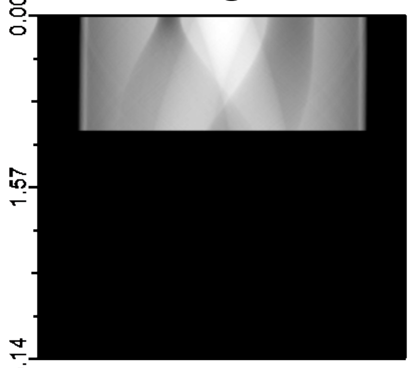


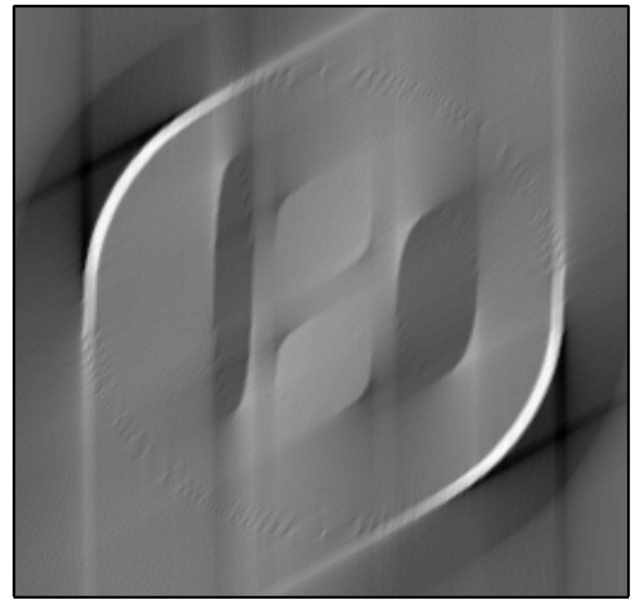
5° used

FBP

Original

Sinogram



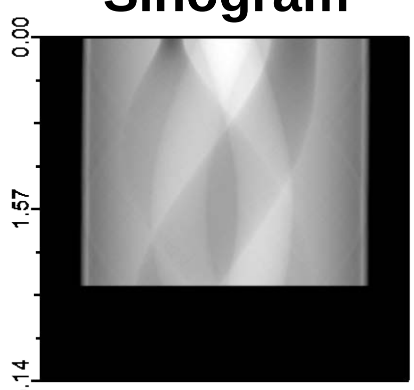


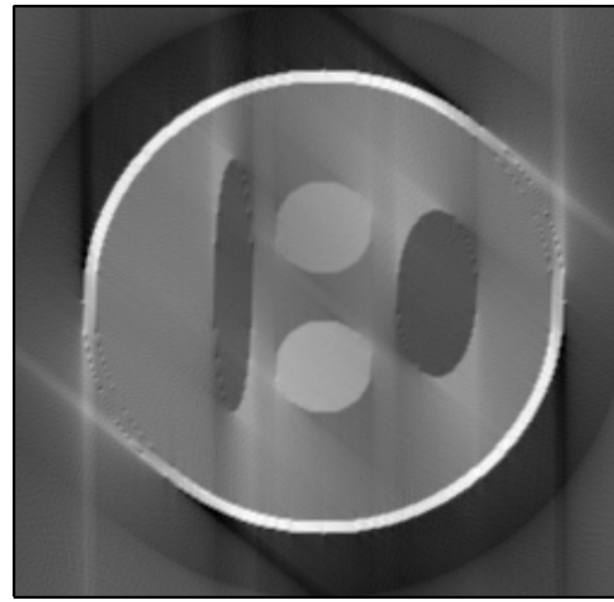
60° used

FBP

Original

Sinogram



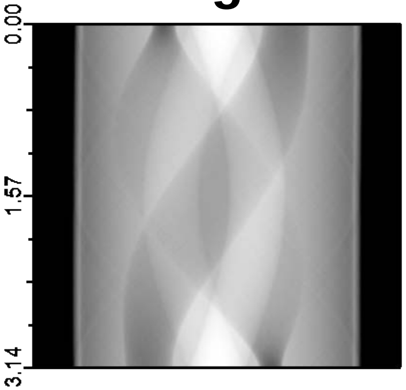


130° used

FBP

Original

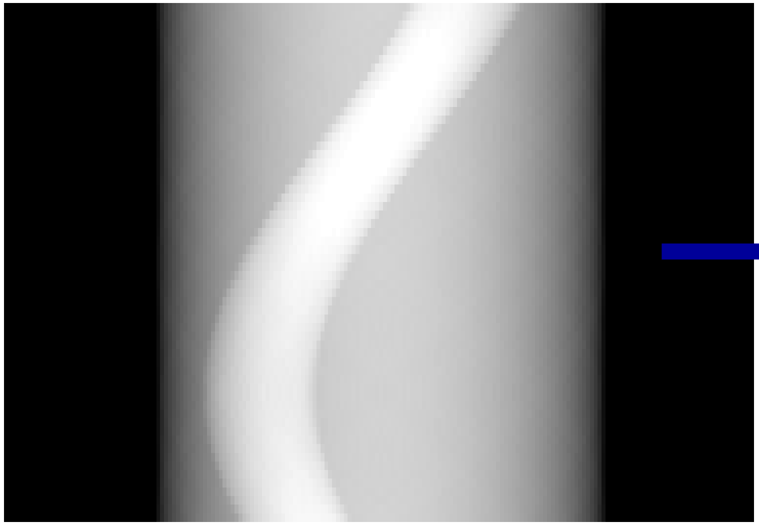
Sinogram



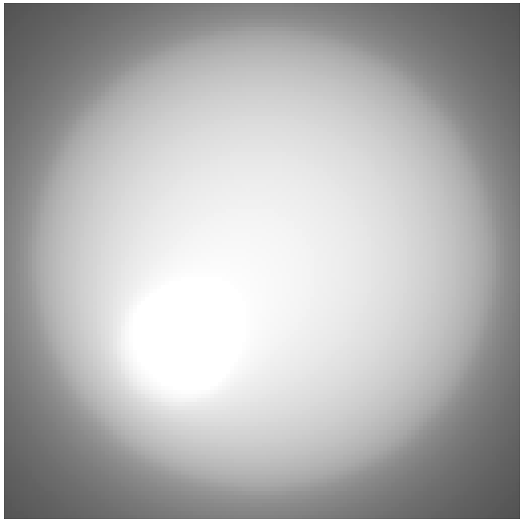
180° used

Reconstruction

Simple back-projection (BP): Problem



Sinogram

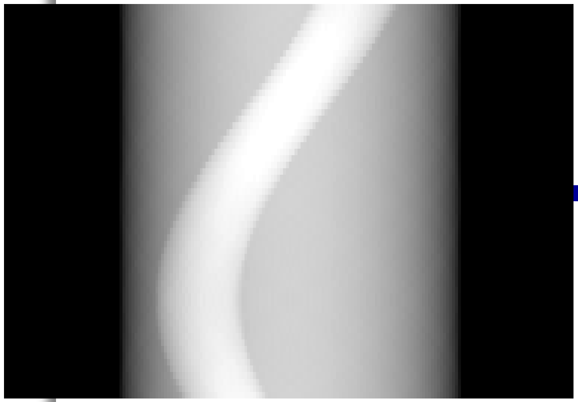


Image

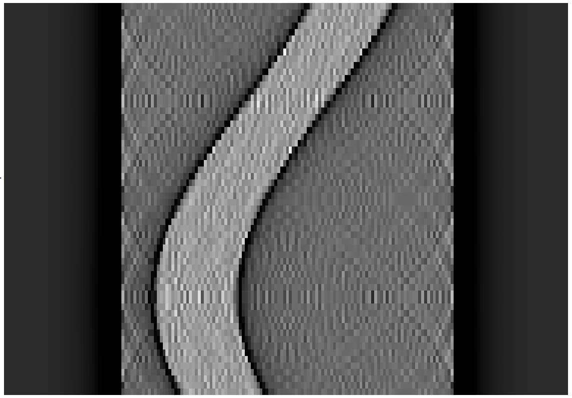
Reconstruction

Filtered back-projection:

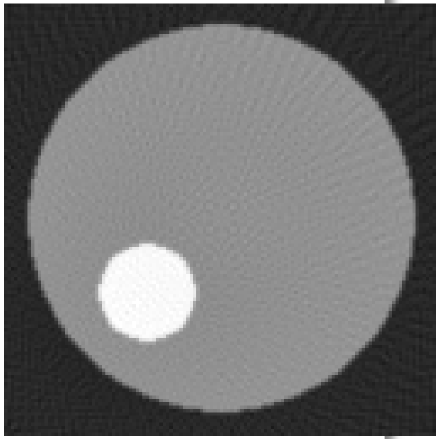
- Selection of relevant spatial frequencies



Sinogram



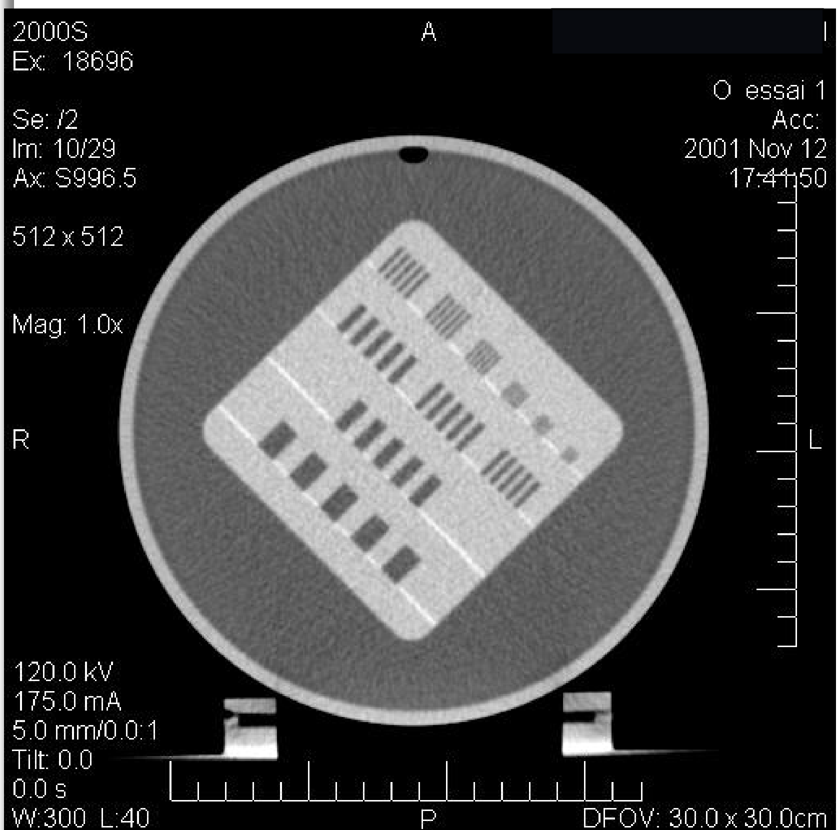
Filtered sinogram



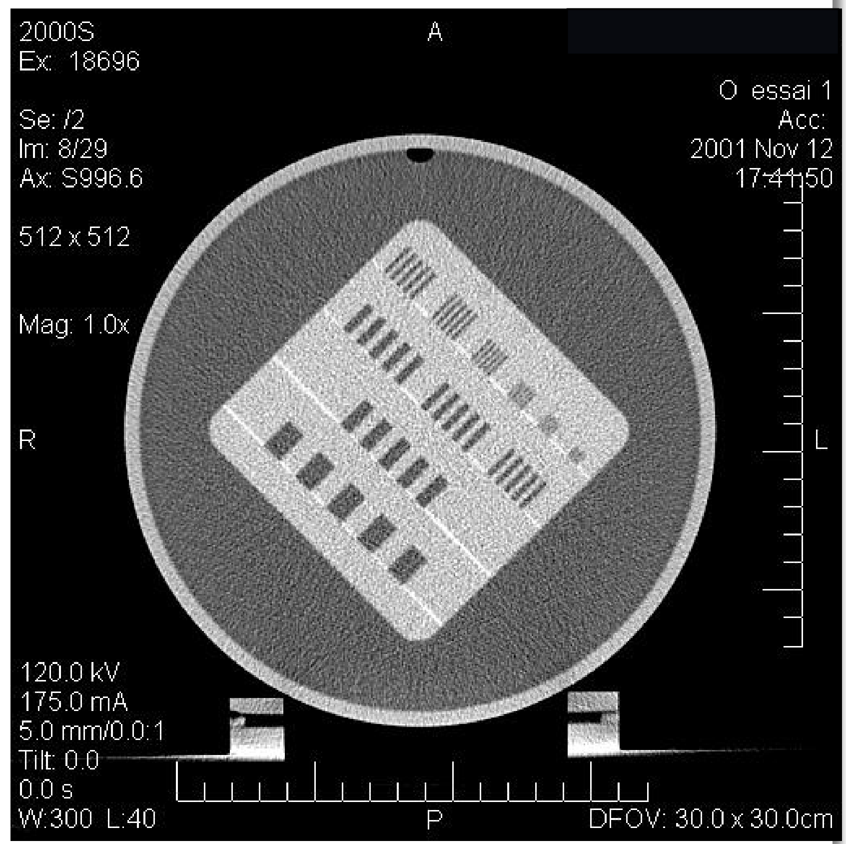
Image

Reconstruction: Filter

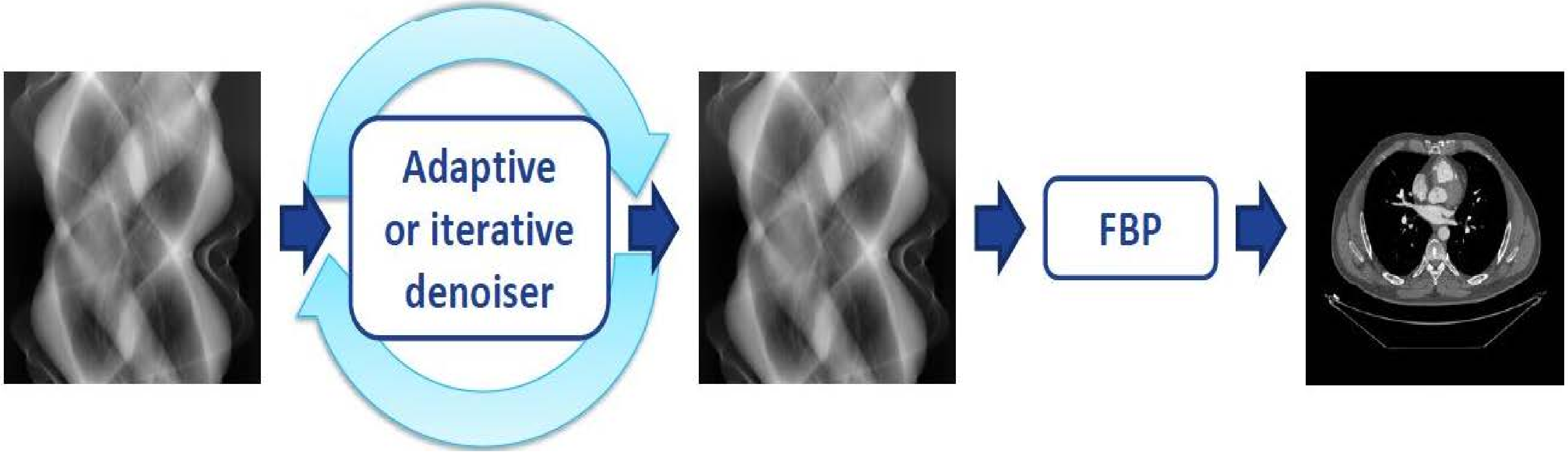
Smooth



Bone

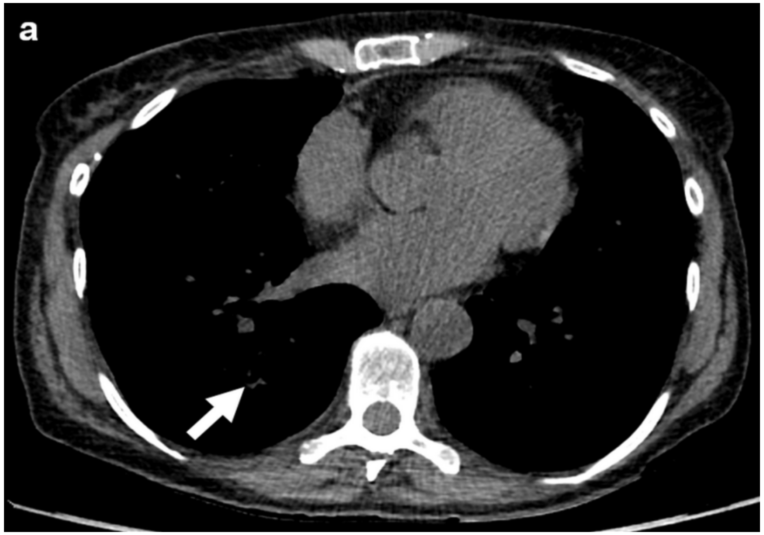


Iterative reconstruction (1st generation)

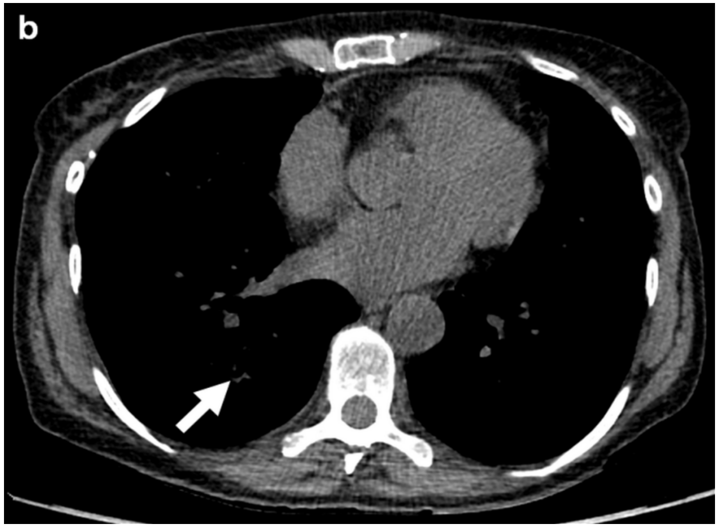


- Sinogram denoising
- Image reconstructed with modified sinogram
- Possibility of reducing patient doise due to lower noise levels for identical doses

Iterative reconstruction (1st generation)



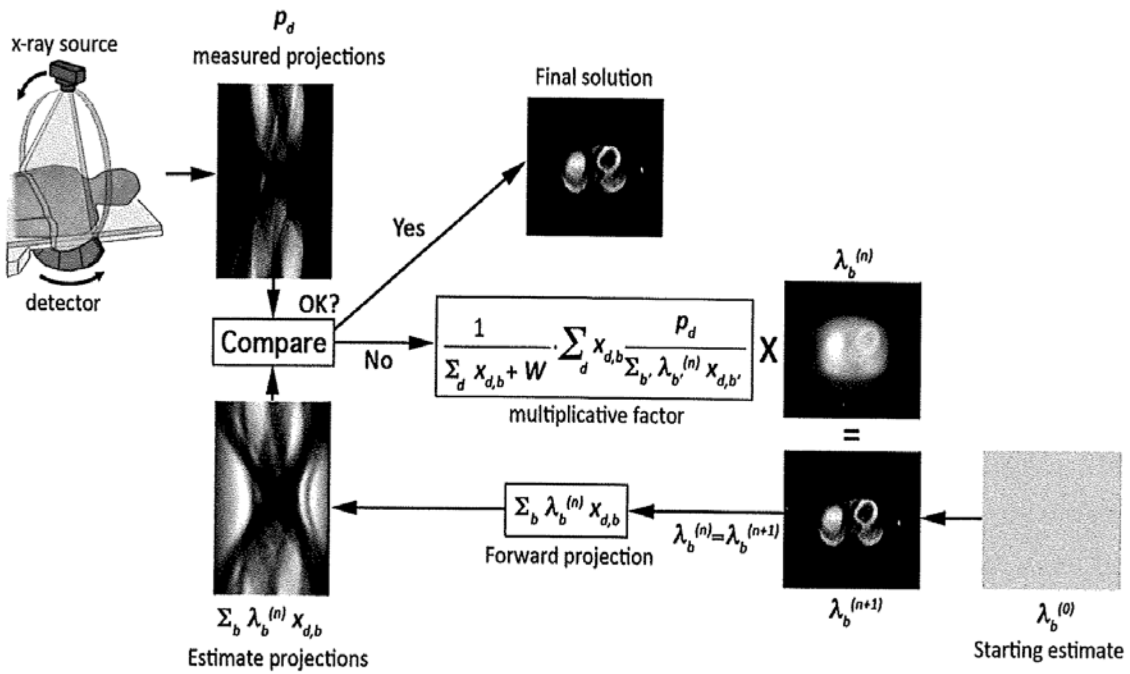
FBP



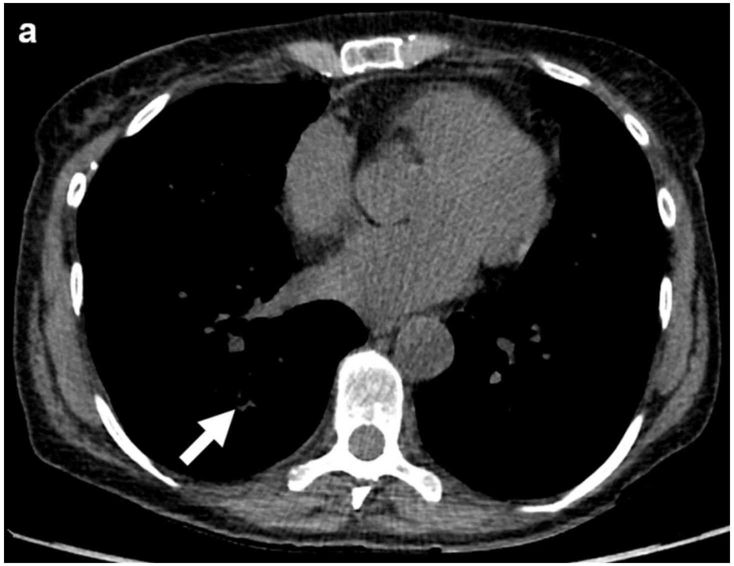
ASIR (40%)

Iterative reconstruction (2nd generation)

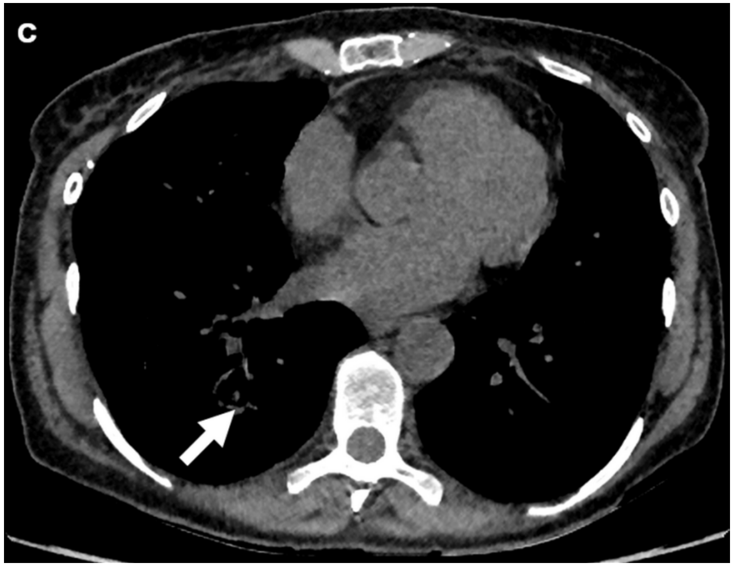
- Optimisation problem
  - Back-and forth between sinogram space and image space
    - CT optics are modeled
    - Noise statistics are modeled



Iterative reconstruction (2nd generation)



FBP



MBIR

Hounsfield scale: $HU = 1000 \times \frac{\mu - \mu_{water}}{\mu_{water} - \mu_{air}}$

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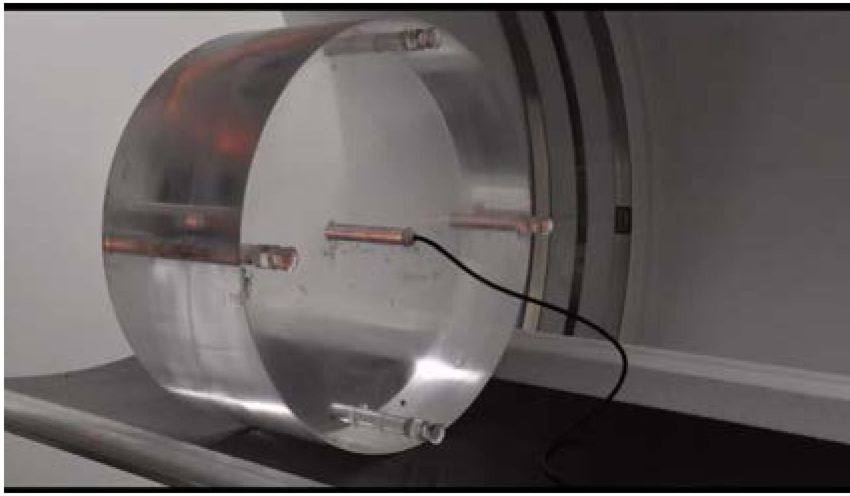
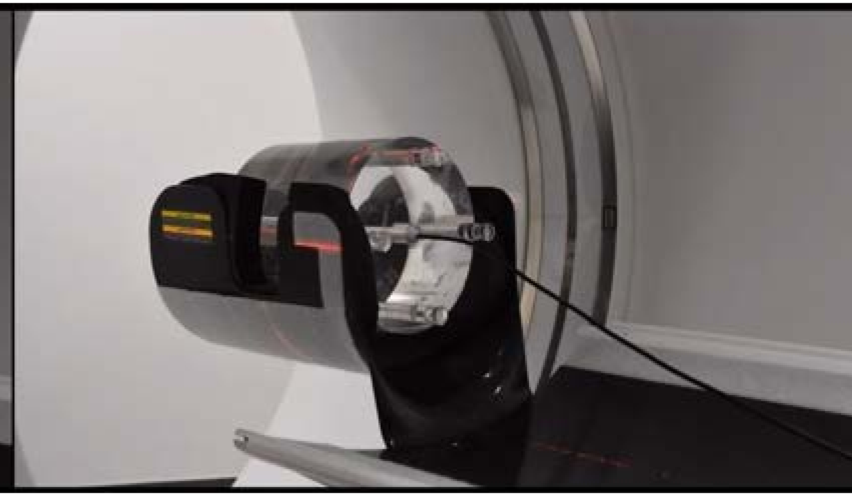
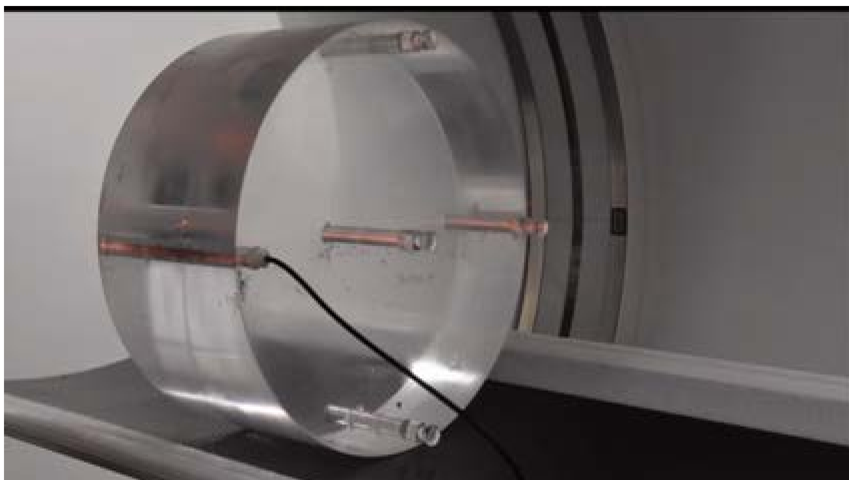
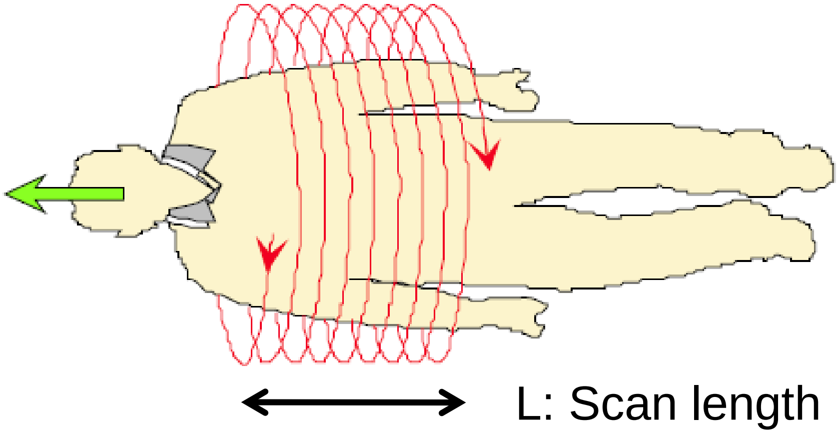




$$CTDI_{vol} = 100\,mm \cdot \frac{\left(\frac{1}{3} \cdot center + \frac{2}{3} \cdot < periphery >\right)}{collimation\ [mm]}$$

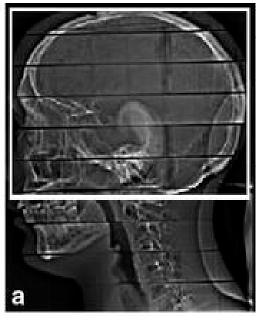


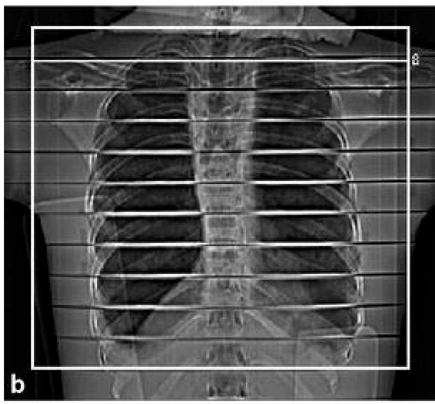
Dose-length product

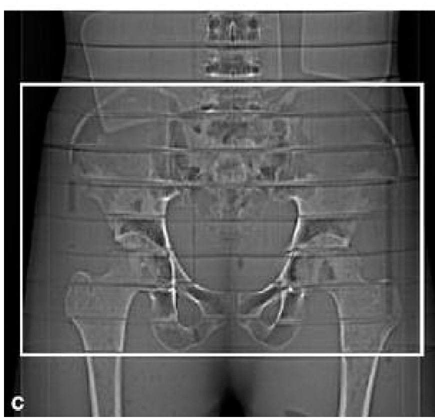


DLP: $CTDI_{vol}$ x L

From DLP to effective dose





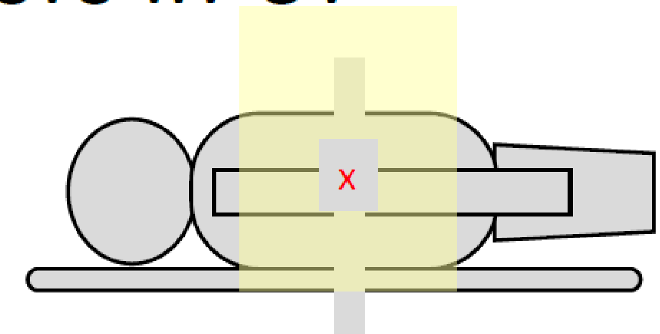


	DLP x [mGy cm]	Conversion factor = [mSv / (mGy cm)]	Effective dose [mSv]
L_1	$DLP_1 = CTDI_{vol} \times L_1$	x 0.0019 (head)	$= E_1$
		x 0.0052 (neck)	
L_2	$DLP_2 = CTDI_{vol} \times L_2$	x 0.0146 (chest)	$= E_2$
		x 0.0153 (abdomen)	
L_3	$DLP_3 = CTDI_{vol} \times L_3$	x 0.0129 (pelvis)	$= E_3$

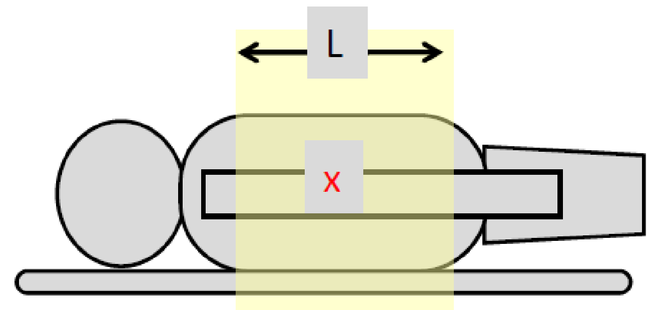
$$E = \Sigma E_i$$

Dose Descriptors in CT

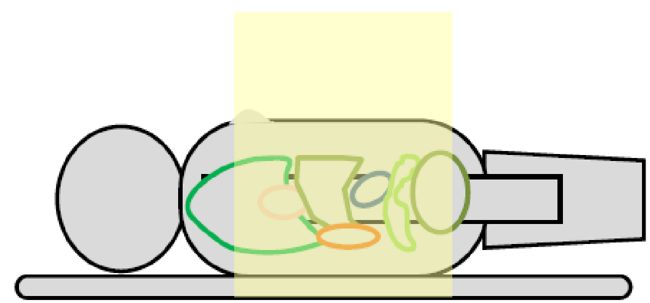
- $CTDI_{vol}$ (mGy)
  - ~ represents local absorbed dose
  - Good for protocol comparison



- $DLP = CTDI_{vol} \times L$ (mGy.cm)
  - represents total absorbed dose
  - ~ represents relative risk

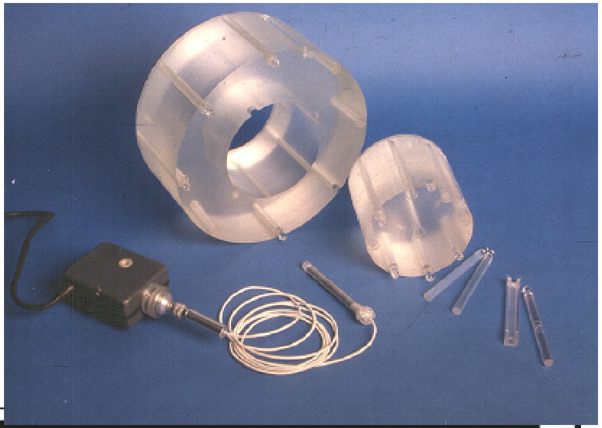


- Effective dose (mSv)
  - Sensitivity of organs accounted for
  - E = DLP x k
  - k values region specific



CTDI is not patient dose

- Patients are not cylinders of Perspex
- The integration length (≡ scan length) is not 100 mm
- Patients come in different sizes



Radiology

CT Dose Index and Patient Dose:
They Are *Not* the Same Thing[1]

canton de vaud

CHUV

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CHUV

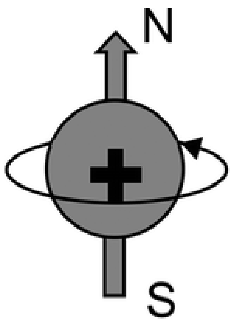
Radiography

MR imaging

- Limitations of X-ray imaging
  - Mainly anatomical imaging
  - Limited soft tissue contrast

Origin of the signal in MRI: Nucleus

Mass: spin (S)



Charges: magnetic momentum (μ)

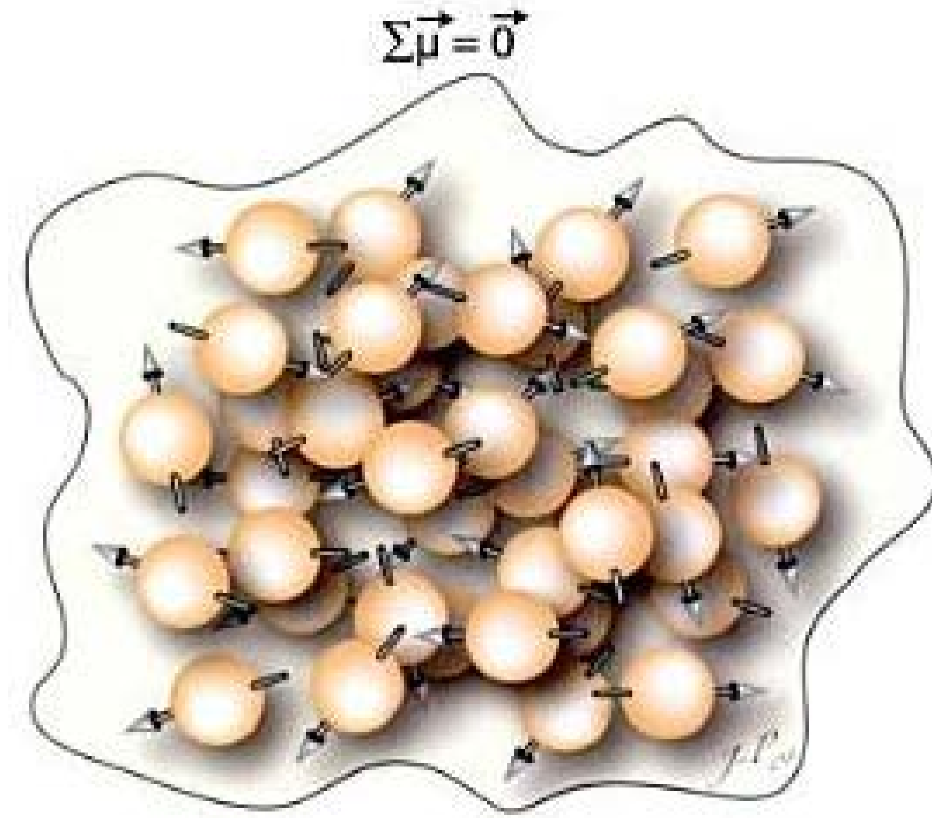
- Rotation of a mass leads to a kinetic momentum
  - Quantified spin
  - Proton, neutron and electron have a spin equal to 1/2
- Rotation of charges leads to a magnetic momentum (small magnet)

$$\mu = \gamma \cdot S$$

γ : Gyromagnetic ratio

MRI : imaging of hydrogen nuclei → proton imaging

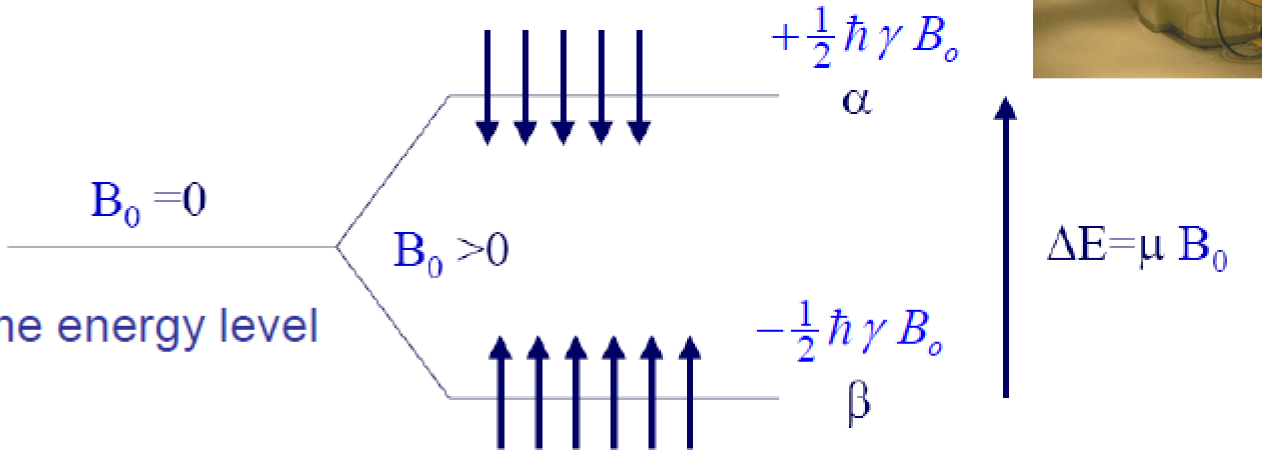
In the absence of a magnetic field $\mathbf{B_o}$, the microscopic magnetisations $\boldsymbol{\mu}$ are oriented randomly



Resulting macroscopic magnetisation is zero.

Effect of a static magnetic field on a magnetic momentum

Splitting of energy levels



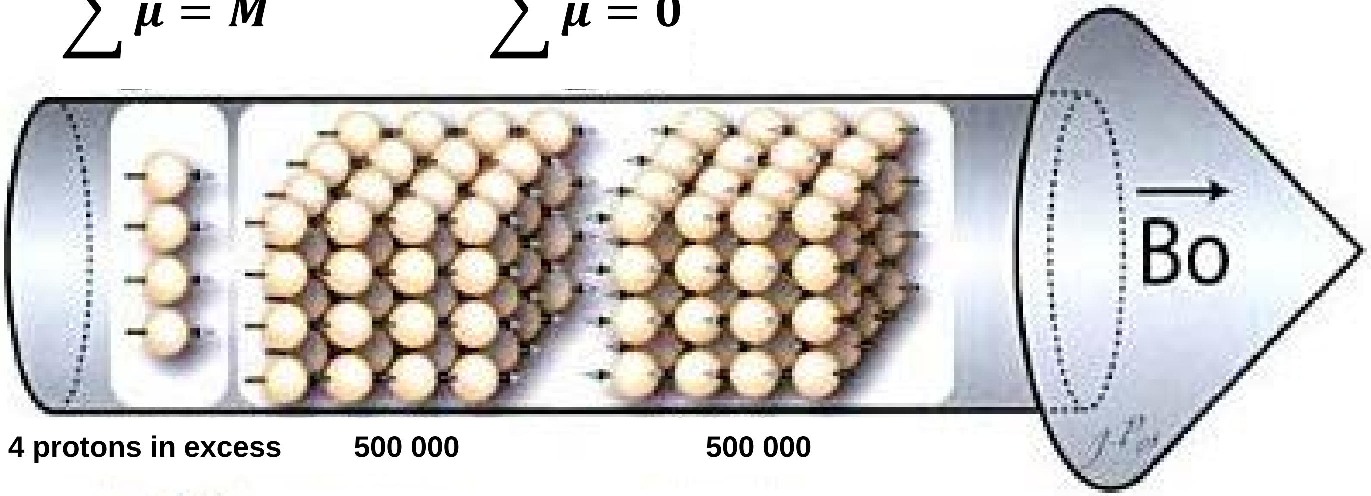
Number of levels available :
2 I + 1

At equilibrium:
$N\beta/N\alpha = \exp - \Delta E/kT$

If B_o = 1.5 T
$N\beta/N\alpha$ =0.001 %

Magnetic field $\mathbf{B_o}$ induces a macroscopic magnetisation **M**

$\sum \vec{\mu} = \vec{M}$ $\sum \vec{\mu} = \vec{0}$

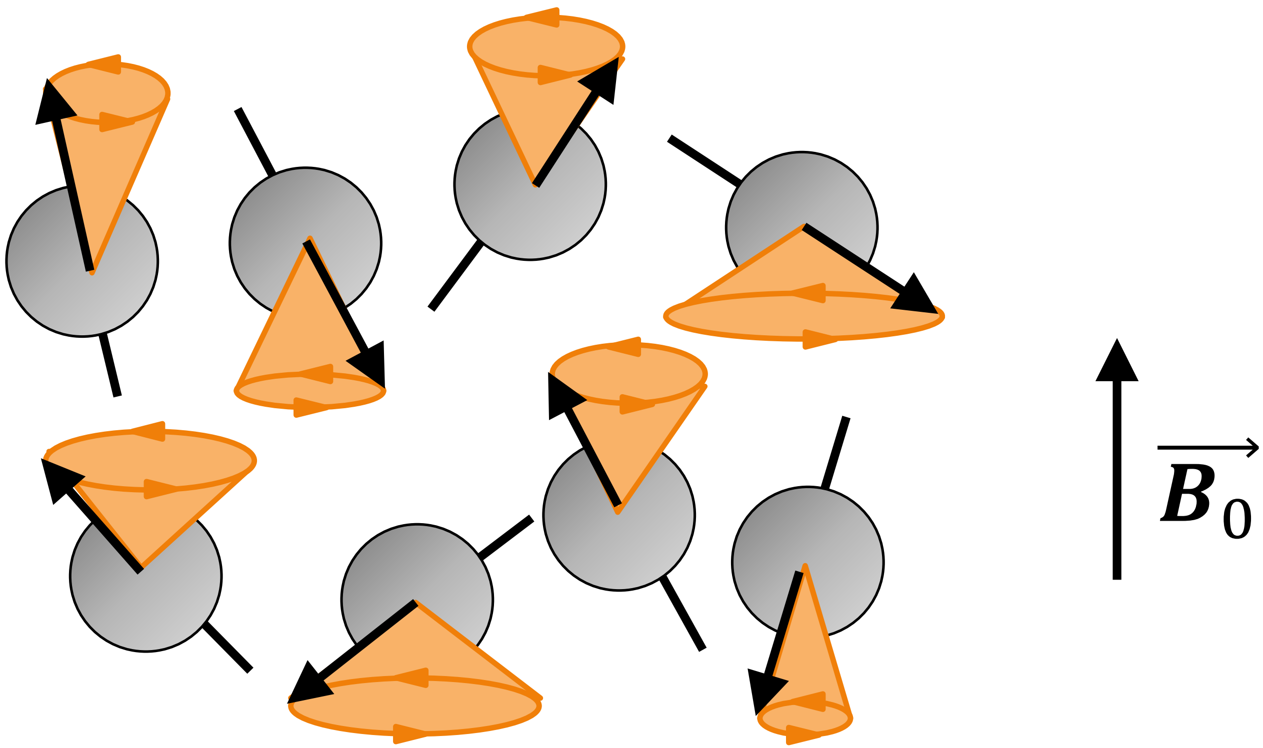


4 protons in excess 500 000 500 000

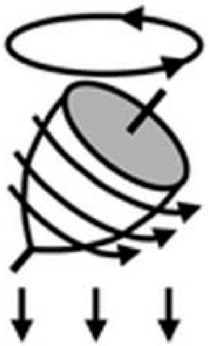
$\sum \vec{\mu} = \vec{M}$

$\vec{M}$ is parallel to $\vec{B_0}$ → **along the patient main axis (head-foot)**

In the presence of a magnetic field **B**, the microscopic magnetisations μ precess (turn) around **B**

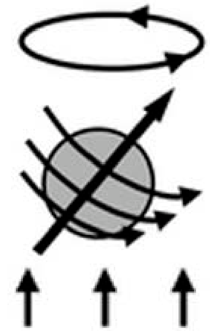


Spinning top in gravitation field



Magnetic moment in a magnetic field (proton)

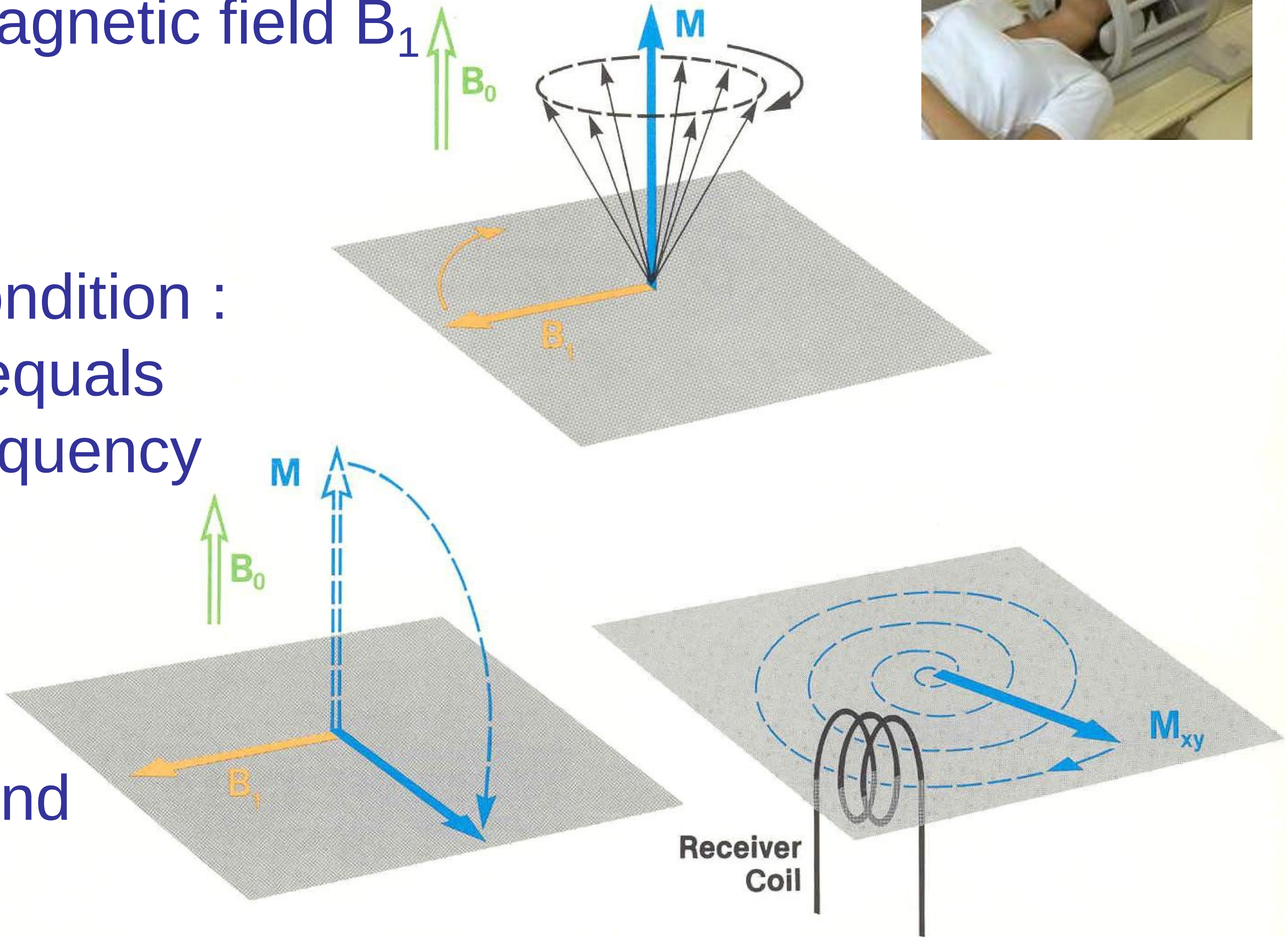
$$\nu = \frac{\gamma}{2\pi} \cdot \mathbf{B_o}$$



- The spinning frequency is proportional to the static magnetic field B_o (Larmor frequency ν)
- For protons: ν = 42.58 MHz @ 1 T

Magnetic resonance experiment

A radiofrequency corresponds to a rotating magnetic field B_1



Resonance condition : B_1 frequency equals the Larmor frequency

M rotates around B_0 and B_1

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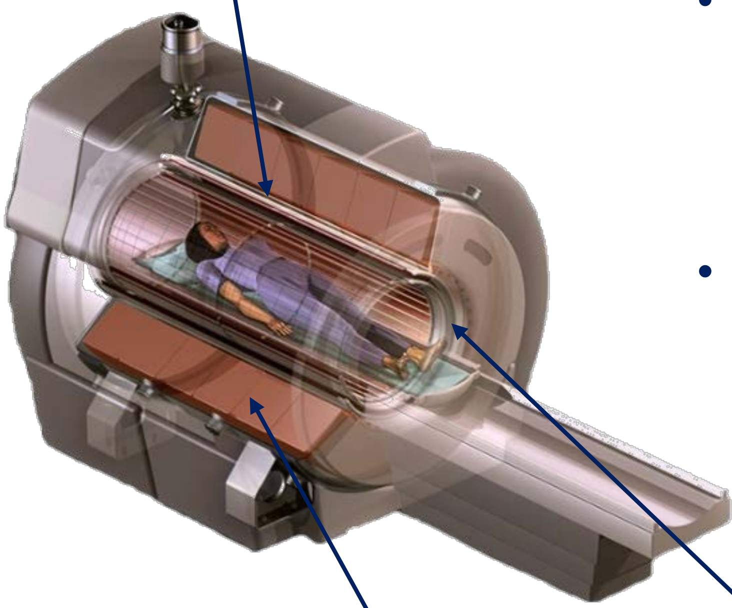
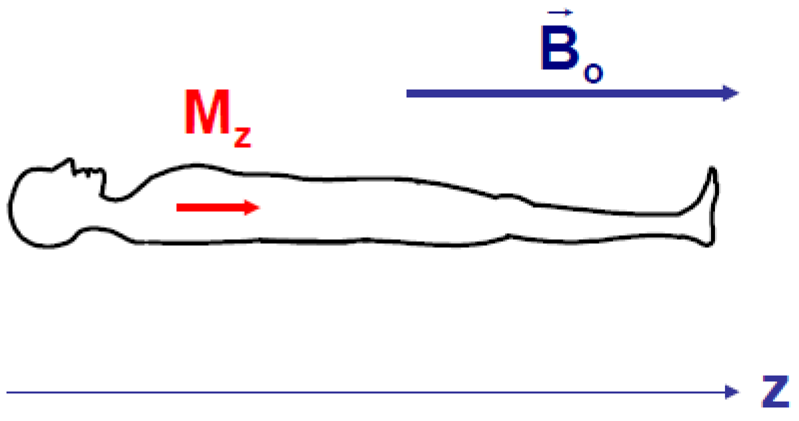
Apparatus

- B_o: Very high magnetic field (tesla, T)
  - Constant in space and time
  - Creation of the signal
- B_1: Excitation magnetic field (mT)
  - Electromagnetic wave (radiofrequency, rf)
  - Magnetic field varying across time
  - Perturbation of the signal
- $G_{x,y,z}$: magnetic field gradients ($mT.m^{-1}$)
  - Magnetic field that varies in a given direction in space
  - Adds up for several seconds to B_o
  - Allows for signal localisation



Phase 1

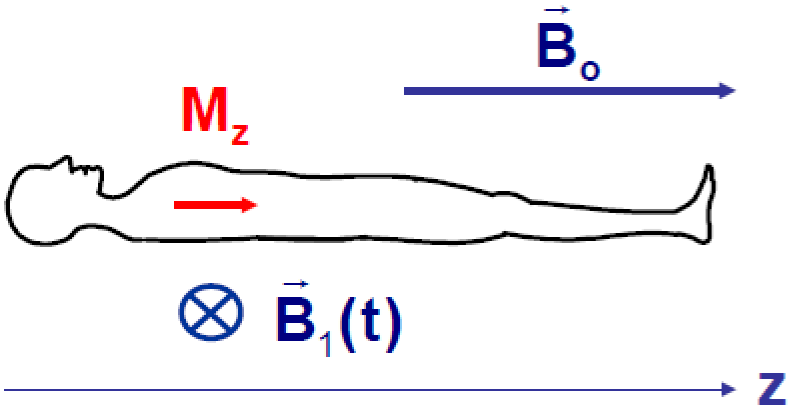
- The patient is placed in a static magnetic field (B_o)



B_o creates $\vec{M}$ which projection along z is M_z

Phase 2

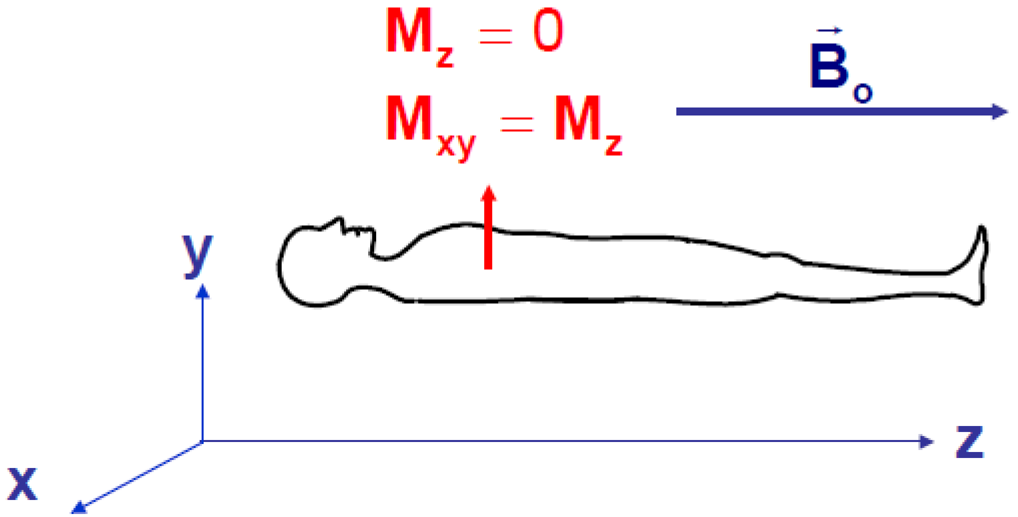
- The direction of M_z is modified by applying a radio frequency, $B_1(t)$, at Larmor frequency (42 MHz/T)
- M_z rotates around B_1 and B_o



Use of an excitation antenna: tilt of M_z in the transverse plane x,y. $M_z \rightarrow M_{xy}$

Phase 3

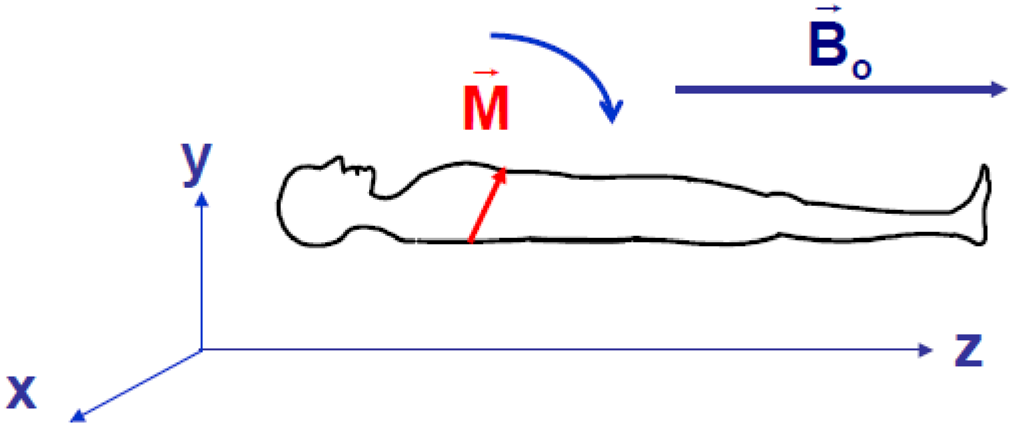
- For a tilt of 90° M_z is now equal to 0



The magnetization vector is now in the transverse plane and rotates around B_o (B_1 is switched off). One can detect M_{xy} in that plane with a reception antenna.

Phase 4

- Magnetization returns to its equilibrium position
→ energy relaxation



The amplitude of M_{xy} decreases as a function of time

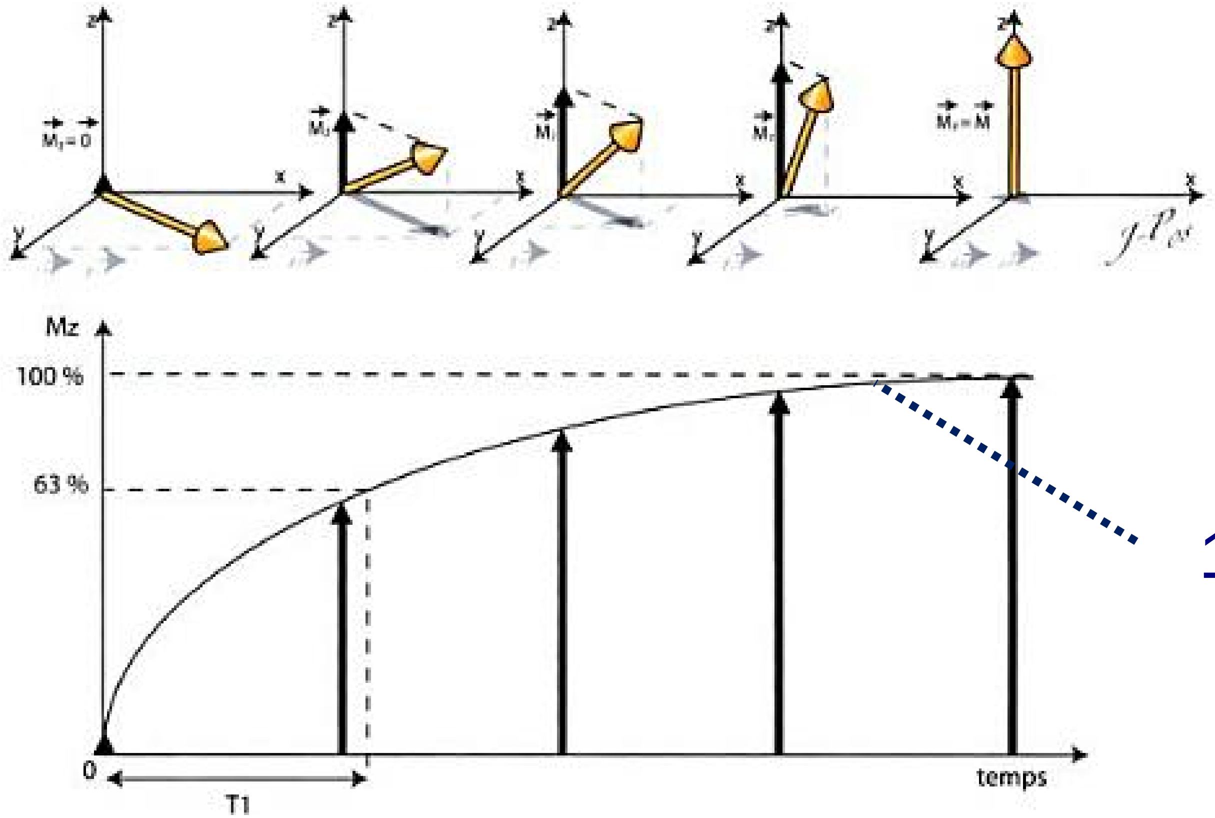
The amplitude of M_z increases as a function of time

MRI sequences

- To obtain an image
  - $\vec{M}$ is flipped multiple times in the transverse plane and the signal is read out
    - Typically 256 times for a 256x256 pixel image
    - MR sequences are usually long (several minutes)
  - Time interval between RF excitations (flipping of $\vec{M}$ in the transvers plane) : **TR**
    - Repetition time
  - Time interval between the flipping of $\vec{M}$ in the transverse plane and signal readout: **TE**
    - Echo time (time between RF excitation and middle of echo)

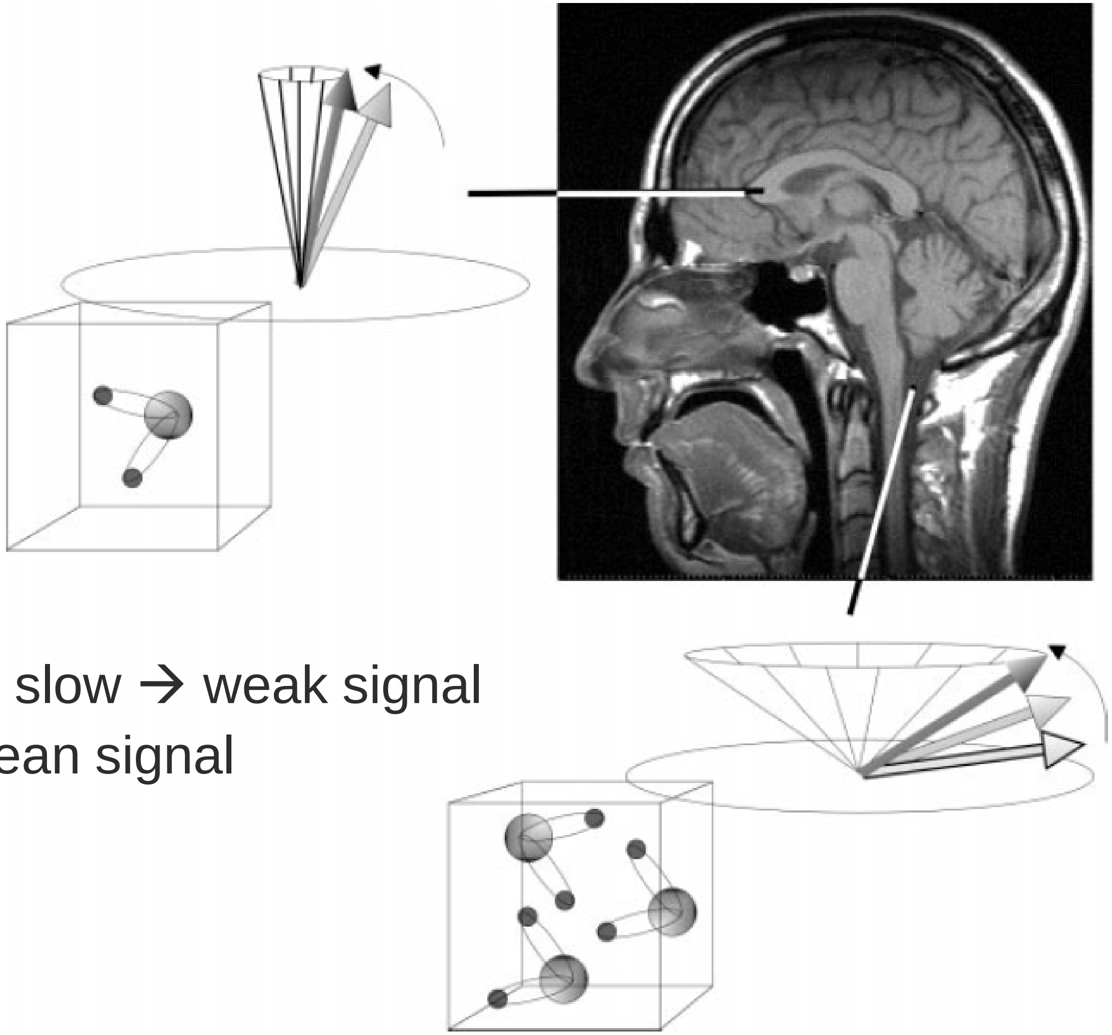
Longitudinal relaxation T_1

The return of the longitudinal magnetisation $\mathbf{M_z}$ (along $\overrightarrow{B_0}$) to equilibrium occurs exponentially (T_1 = 63% of regrowth).
The spins dissipate their energy (from the RF excitation) to their environment.
Spin-lattice relaxation or T_1



$$1 - \exp\left(-\frac{t}{T_1}\right)$$

T_1 effect on the image



Regrowth of M_z:

Solids and liquids: very slow → weak signal

Soft tissue: mean → mean signal

Fat: fast → high signal

Transverse relaxation (T_2)

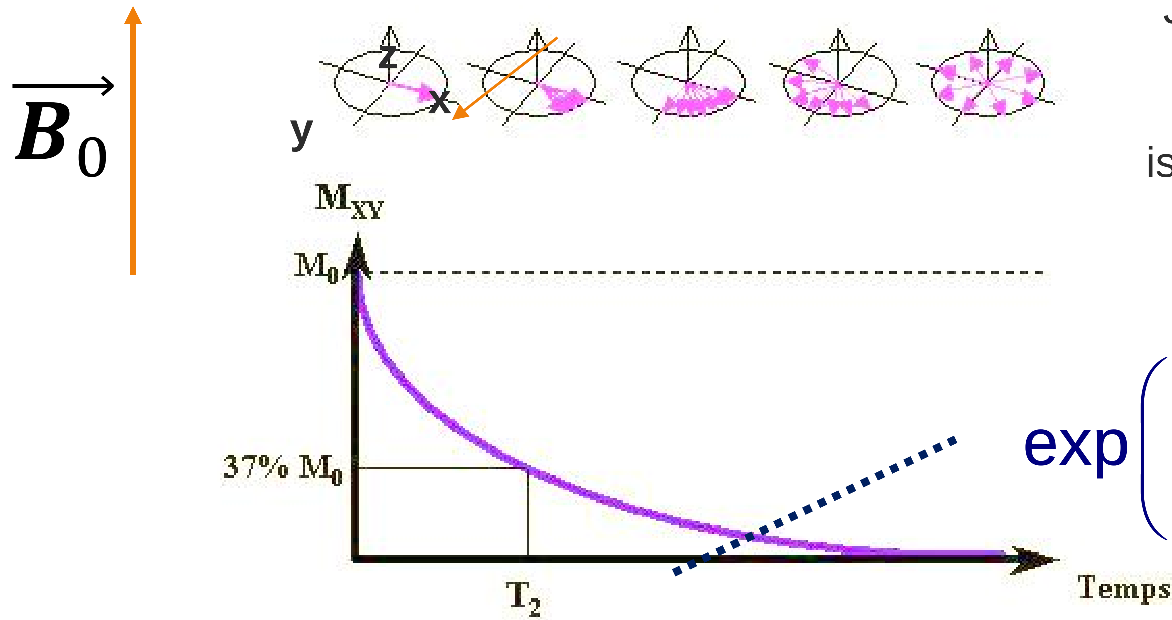
Mechanism simultaneous to T_1 and much faster

Dephasing of individual spins

Diminution of M_{xy} (signal)

No energy exchange with the lattice (medium): **spin-spin relaxation or T_2**

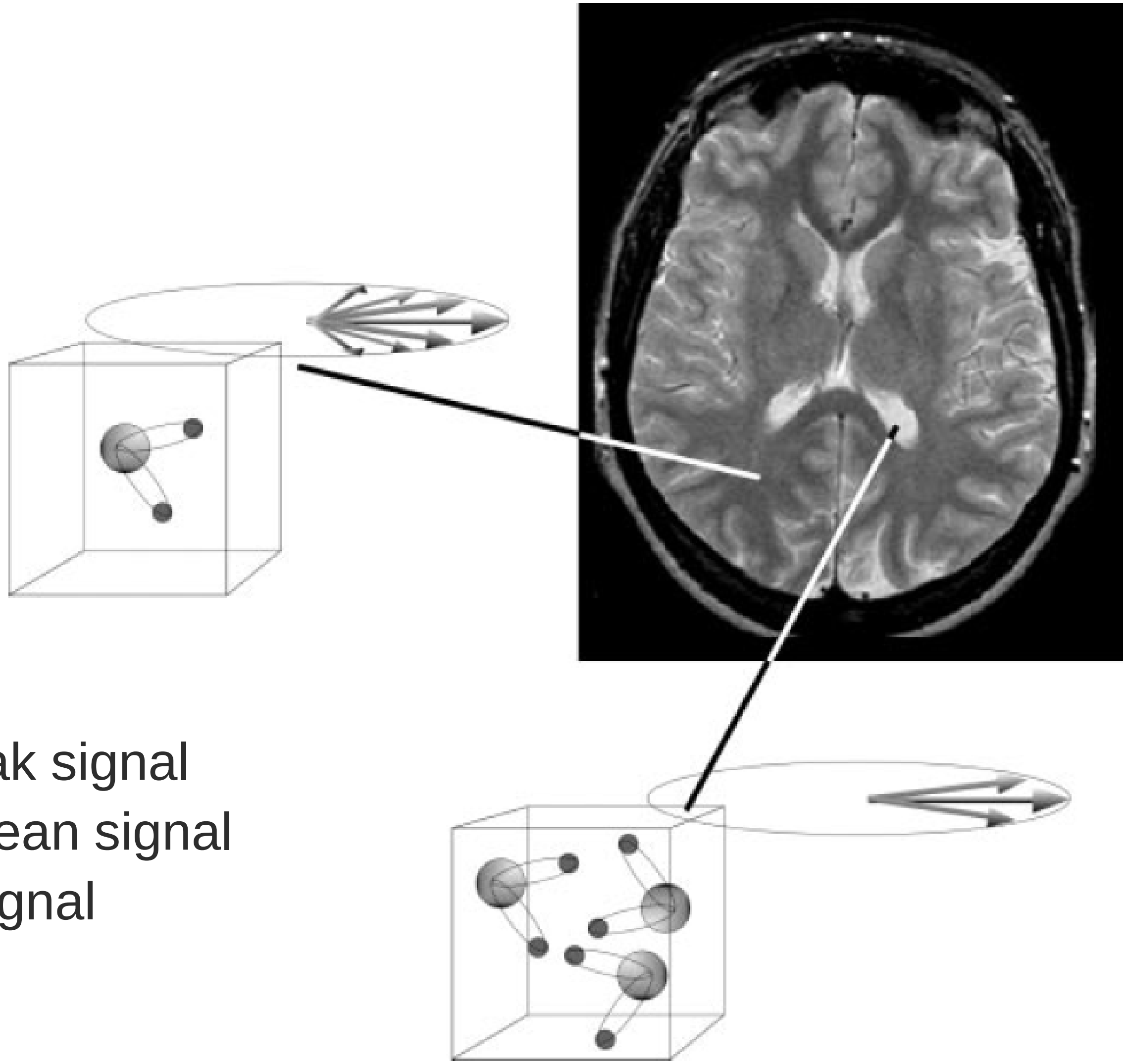
Measures the magnetic homogeneity of the environment



Just after the 90° pulse, all the magnetisation is in the transverse plane M_{xy}

$$\exp\left(-\frac{t}{T_2}\right)$$

T2 effect on the image



Decay of M_{xy}:
Solids: very fast → weak signal
Soft tissue: mean → mean signal
Liquids: slow → high signal

Temps de relaxation T_1 et T_2

- Indicative values of tissue relaxation times (usually $\mathbf{T_1 >> T_2}$)

Tissue	T1 @1,5 T (msec)	T2 (msec)
Fat	260	80
Liver	500	40
Muscle	870	45
White matter	780	90
Grey matter	900	100
Cerebrospinal liq.	2'400	160

Classical contrasts in MRI

- Proton density (number of H atoms in water and fat molecules) per unit volume
  - No T_1 relaxation effect
    - RF pulses are far from one another
      - **Long TR**
  - No T_2 relaxation effect
    - The signal is recorded as fast as possible after the RF pulse
      - **Short TE**

- T_1 contrast (proton density modified by T_1)
    - **Short TR**
    - **Short TE**

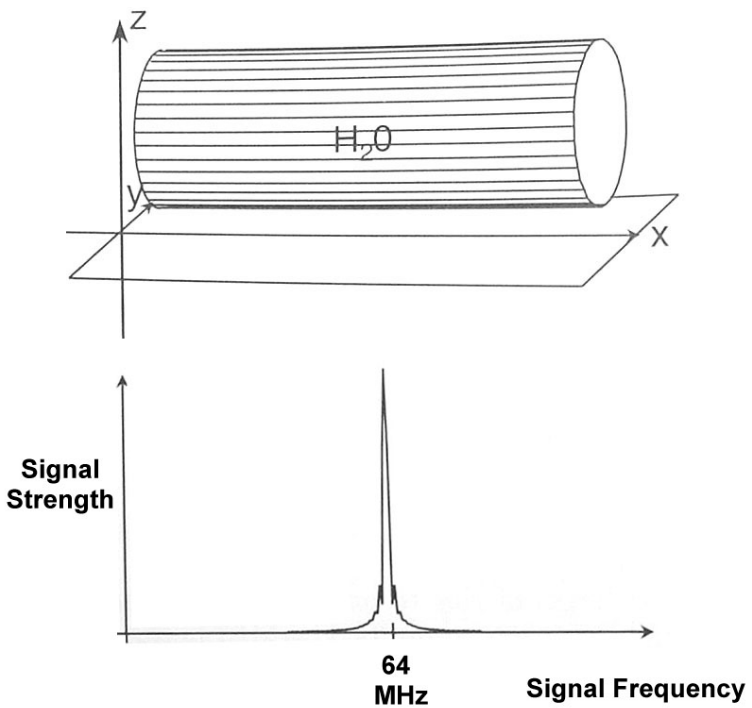
- T_2 contrast (proton density modified by T_2)
    - **Long TR**
    - **Long TE**

canton de vaud

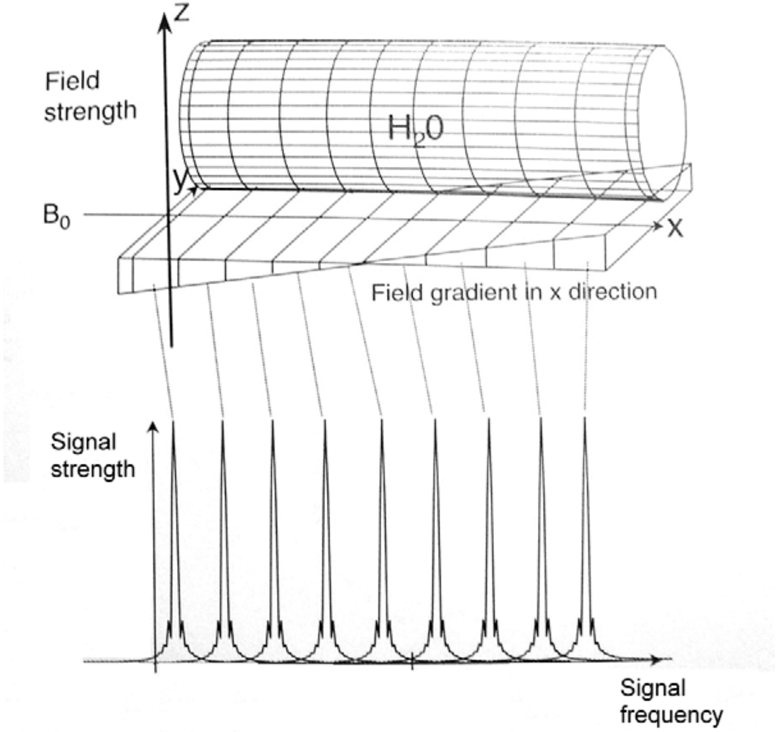
CHUV

Localisation of the signal

- Slice selection: activation of a gradient coil



B_o constant: only one Larmor frequency

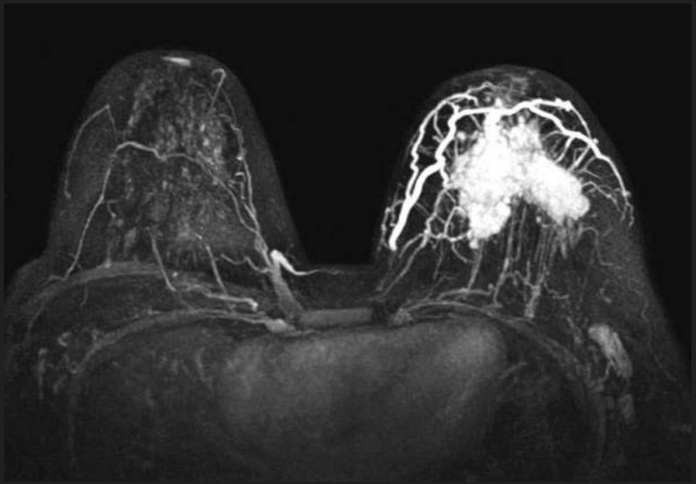


B_o + gradient: Larmor frequency depends on the position

Particular applications of MRI

Cardiac MRI

Applications in oncology



T_1 image (Gd enhanced)
(mammary carcinoma)

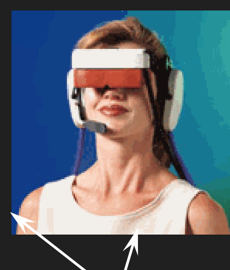
T_2 image
(bone tumour)

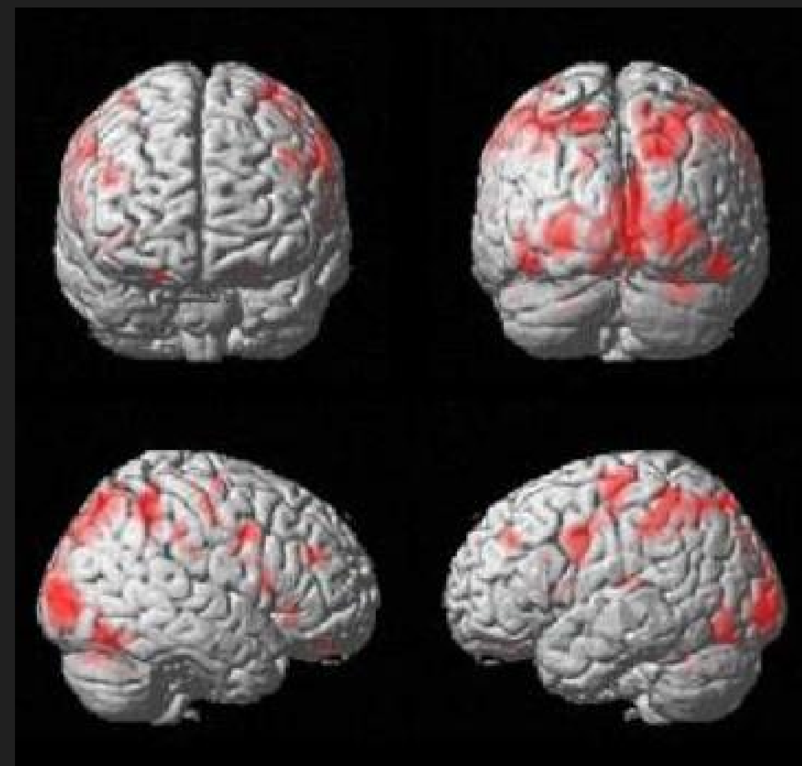
T_2 image
(cyst)

T_1 image
(cerebral metastasis)

Functional imaging (fMRI)

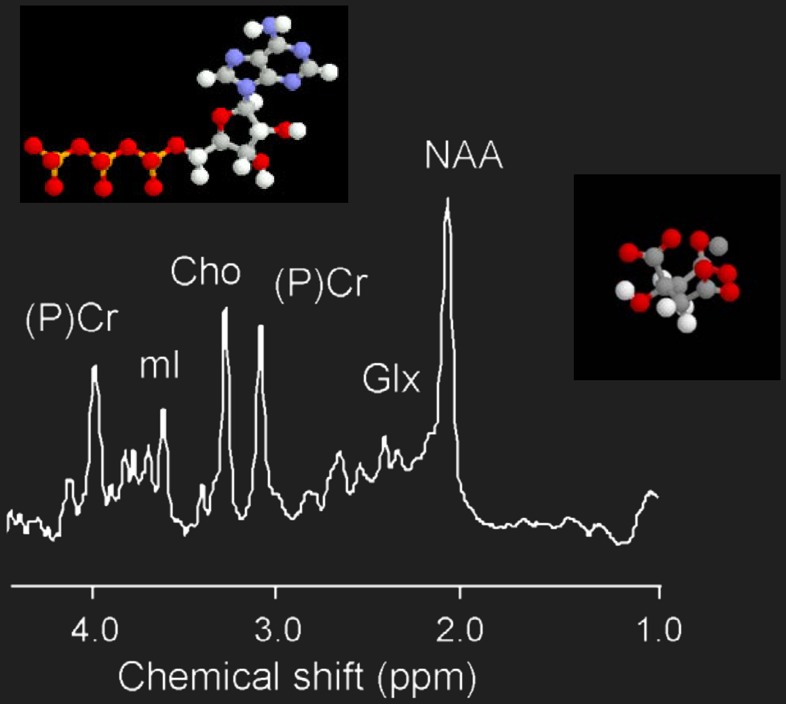
Different possible stimuli

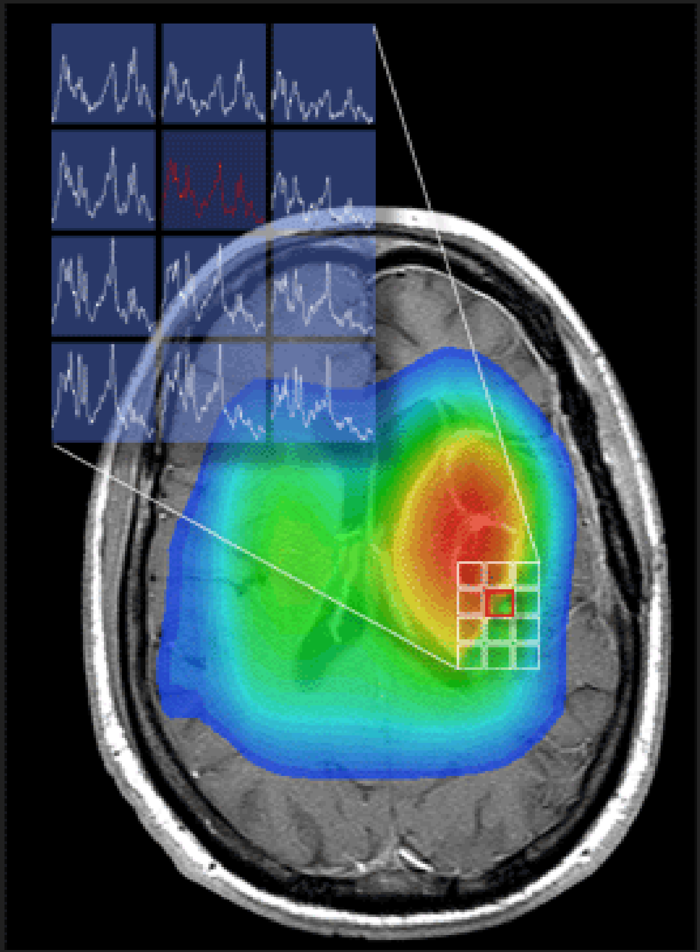




Functional imaging of a blind person reading a text written in Braille.

MR spectroscopy (MRS)





Reference

- Excellent course on youtube
  - Paul Callaghan
  - 10 videos between 6 and 10 minutes

https://www.youtube.com/user/magritek/videos

Thank you for your attention

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