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Joint CI-JAI advanced accelerator lecture series

Imaging and detectors for medical physics

Lecture 3: X-ray imaging

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Course layout

Day	AM 09.30 – 11.00	PM 15.30 – 17.00
Week 1		
6th June	Lecture 1: Introduction to medical imaging	Lecture 2: Detectors for medical imaging
7th June	Lecture 3: X-ray imaging	
8th June		Tutorial
Week 2		
13th June	Lecture 4: Radionuclides	
14th June	Lecture 5: Gamma cameras	Lecture 6: SPECT
16th June	Lecture 7: PET	
Week 3		
22nd June	Tutorial	

Books

1. N Barrie Smith & A Webb
   Introduction to Medical Imaging
   Cambridge University Press

2. Edited by M A Flower
   Webb's Physics of Medical Imaging
   CRC Press

3. A Del Guerra
   Ionizing Radiation Detectors for Medical Imaging
   World Scientific

4. W R Leo
   Techniques for Nuclear and Particle Physics Experiments
   Springer-Verlag

X-ray in the body

Ref. 1 – Chapter 2, Ref. 2 – Chapter 2

- X-rays going through patient's body get attenuated:

$$I(x) = I_0 e^{-\mu(E)x}$$

I_0 = X-ray fluence in entrance

$I(x)$ = X-ray fluence at position x = fluence in exit

$\mu(E)$ = X-ray linear attenuation coefficient

- X-ray linear attenuation coefficient $\mu[cm^{-1}]$ depends on X-ray energy
- In tissue mass attenuation coefficient often used $\mu/\rho[cm^2 g^{-1}]$, with $\rho[g/cm^3]$ = tissue density

X-ray transmission imaging

- Basis = differential absorption of X-rays by tissues = for ex. bone absorbs X-ray more than soft tissue

Tissue	$\mu(cm^{-1})$	$I(x)/I_0(x = 1\ cm)$	**Difference to muscle (%)**
Air	0.000	1.0	+20
Blood	0.178	0.837	+0.2
Muscle	0.180	0.835	0
Bone	0.480	0.619	-26

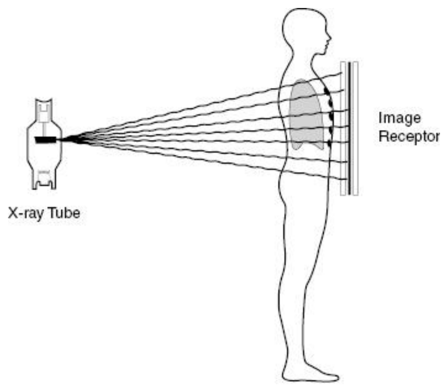
- Contrast agents = chemicals introduced in patient's body to enhance contrast between tissues

X-ray transmission image formation

- Image formation:
  1. X-rays from source directed toward patient → some X-rays absorbed + some X-rays transmitted
  2. X-rays transmitted detected in exit from patient
  3. Measured in exit from patient = fluence distribution = linear attenuation coefficient distribution
- Some X-rays scattered inside patient = image noise / background

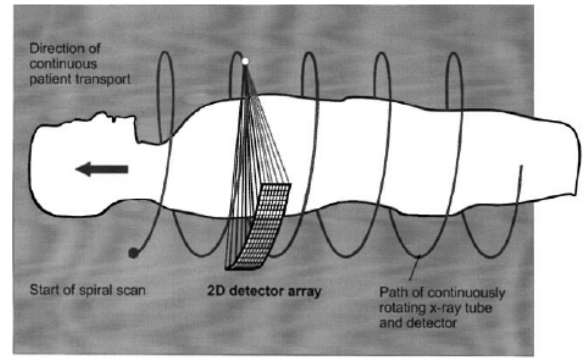
X-ray imaging techniques

Planar radiography



- Image = 2D projection of all tissues between X-ray source and detector
- X-ray source and detector fixed

Computed Tomography



- Image = 3D image of body region
- X-ray source and detector rotate at high speed around patient + patient moved in third direction
- Disadvantage respect to planar radiography = much higher dose

Other X-ray imaging techniques

- X-ray fluoroscopy

  Images are acquired continuously → to study passage of X-ray contrast agent through GI tract

- Digital mammography

  Images are acquired with lower X-ray energies than standard X-ray scans → to obtain images with much finer resolution

- Digital subtraction angiography

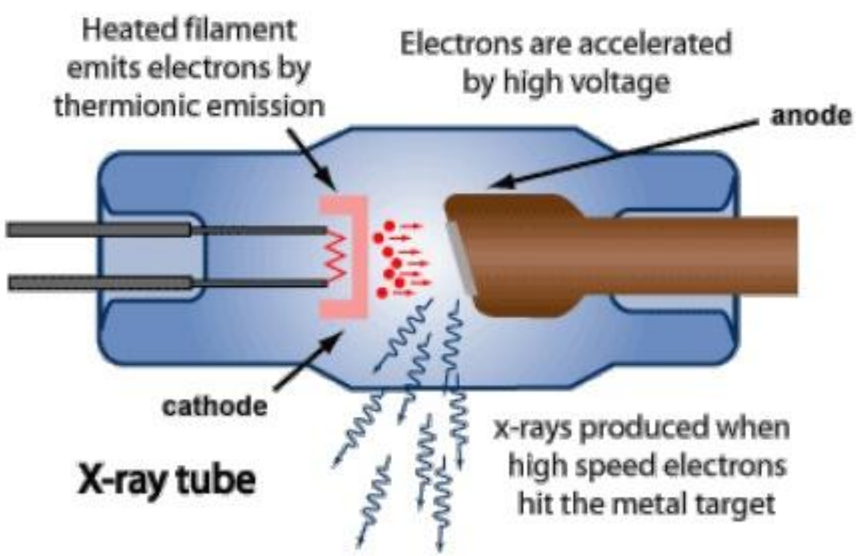
  Images are acquired at extremely high resolution → to image vasculature

- Digital X-ray tomosynthesis

  Hybrid planar radiography – CT: fixed screen + rotating source

X-ray tube

X-ray source for transmission imaging = X-ray tube



- Cathode = filament + focusing cup
- Anode = target that rotates at high speed to reduce localised heat
- Filament and target usually tungsten
- Efficiency for e^{-} conversion in X-rays $\sim 1\%$, rest dissipated in heath
- Strong vacuum inside tube $\rightarrow$ unimpeded path between cathode and anode
- Oil surrounding the envelope = dissipates heat from anode

Materials for the filament and target

- Tungsten: most commonly used

Characteristics	Advantages
Emission at $\sim 2000\ ^{o}$C	High and stable e^{-} thermionic emission
Melting point $3370\ ^{o}$C	Can withstand very high temperatures generated in the anode
High $Z = 74$	High X-ray production efficiency[1]
Good thermal conductivity + Low vapour pressure	Can operate in very high vacuum

[1]Bremsstrahlung yield increases with Z

- Molybdenum: used in digital mammography that requires very low energy X-rays = less heat generated

X-ray tube parameters

Tube parameters	Values
Accelerating voltage ΔV_{C-A}, kVp	25÷140 kV[1]
Tube current I from the cathode to the anode	50÷400 mA for 2D radiography Up to 1000 mA for CT
Exposure time	Limited by anode heating

[1] 25 kV for mammography, 140 kV for bone and chest

- These parameters are chosen by the operator according to the specific application
- 2D radiography and CT scanners = different set-up $\rightarrow$ same X-ray tube cannot be used for both

Power rating

- Power rating – Definition

  Maximum power dissipated in an exposure time of 0.1 s

- Exercise

  Q = What is the maximum exposure time of a tube with a power rating of 10 kW, when operated at 125 kV with 1 A of current? What modality is this?

  A =

$$Power\ dissipated = kVp * I = 125\ kV * 1\ A = 125\ kW$$

$$Exposure\ time * Power\ dissipated = Power\ rating \rightarrow$$

$$Exposure\ time = \frac{Power\ rating}{Power\ dissipated} = \frac{10}{125} = \boxed{80\ ms}$$

$\boxed{\text{Modality is CT}}$

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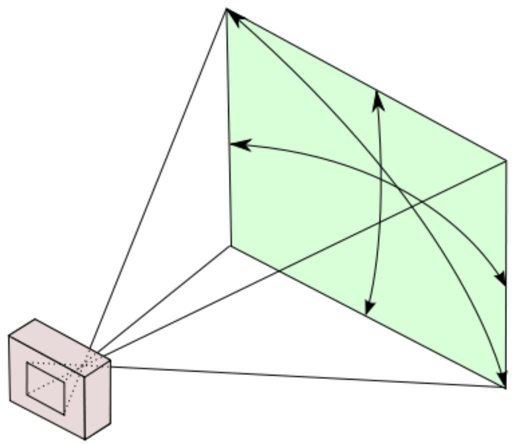
A couple of definitions

Field-of-view (FOV)

https://en.wikipedia.org/wiki/Field_of_view

- FOV of optical instruments or sensors = solid angle through which detector is sensitive to radiation = solid angle imaged by the detector

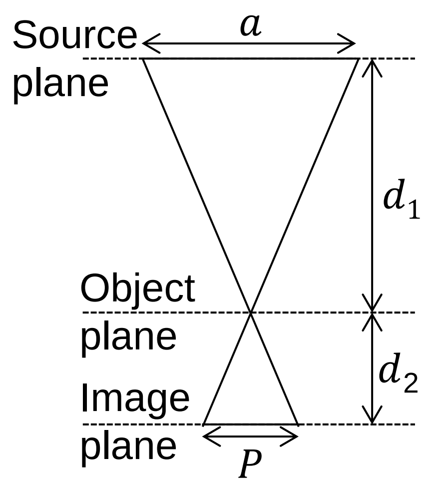


Penumbra

Ref. 2 – Chapter 2.5.5

- Penumbra P = unsharpness / blurring in the image due to finite size of X-ray source:

$$P = a\frac{d_1}{d_2}$$



Instrumentation for 2D X-ray radiography

- X-ray tube: generates the X-ray beam
- Collimator: reduces patient's dose and amount of Compton scattered X-rays
- Anti-scatter grid: reduces amount of Compton scattered X-rays = background/noise $\rightarrow$ increases image contrast
- Digital detector: converts transmitted X-rays into light and then into electric signal
- Read-out electronics: digitises and reads the signal from the detector

Collimators and grids

Collimators

- Sheets of lead placed between X-ray source and the patient
- Restrict dimension of the beam to the FOV in 1D or 2D → reduce amount of X-rays reaching the patient = only X-rays inside FOV reach the tissue → dose reduced + scattered reduced

Anti-scatter grids

- Parallel or slightly divergent strips of lead foil with aluminium spacers
- Amount of scattered X-rays absorbed depends on length, thickness and separation of lead strips
- Some non-scattered X-rays are absorbed → increase in dose to get same image intensity of one without grid

Detectors and electronics

Ref. 1 – Chapter 2.7

- Computed radiography

  Instrumentation = detector plate + separate reader

- Digital radiography

  Instrumentation = detector and reader are one unit

  1. Indirect = X-ray converted into light by scintillator → light converted into electric signal by photon detector
  2. Direct = X-ray converted into electric signal by materials such a:Se.

     Less efficient than indirect conversion device

Signal-to-noise ratio (SNR)

Ref. 1 – Chapter 2.8.1

- Signal = N of X-rays arriving on detector
- Statistical fluctuations in number of X-rays detected per unit area → noise
- Statistical fluctuation follow Poisson distribution → $\sigma_{noise} = \sqrt{\mu}$ with μ mean value

$$SNR = \frac{N}{\sigma_{noise}} = \frac{N}{\sqrt{\mu}} \propto \sqrt{N}$$

- Exercise: What is the dose increase if doubling SNR?

A: $2 \times SNR = 2 \times \sqrt{N} = \sqrt{4 \times N} \rightarrow 4 \times N = \boxed{4 \times Dose}$

Factors affecting SNR

1. X-ray tube current I and exposure time t_e:
$$SNR = \sqrt{I \times t_e}$$
2. X-ray tube kVp: the higher kVp the higher the X-ray energy $\rightarrow$ greater penetration in tissue $\rightarrow$ signal increases $\rightarrow$ SNR increases in a non-linear way
3. Detector efficiency: the higher the efficiency the more X-rays are detected $\rightarrow$ signal increases $\rightarrow$ SNR increases
4. Patient size and body part to be imaged: the greater the tissue thickness the higher the X-ray attenuation $\rightarrow$ signal decreases $\rightarrow$ SNR decreases
5. Anti-scatter grid: attenuates Compton scattered X-rays $\rightarrow$ reduces signal $\rightarrow$ SNR decreases

Spatial resolution

Ref. 1 – Chapter 2.8.2

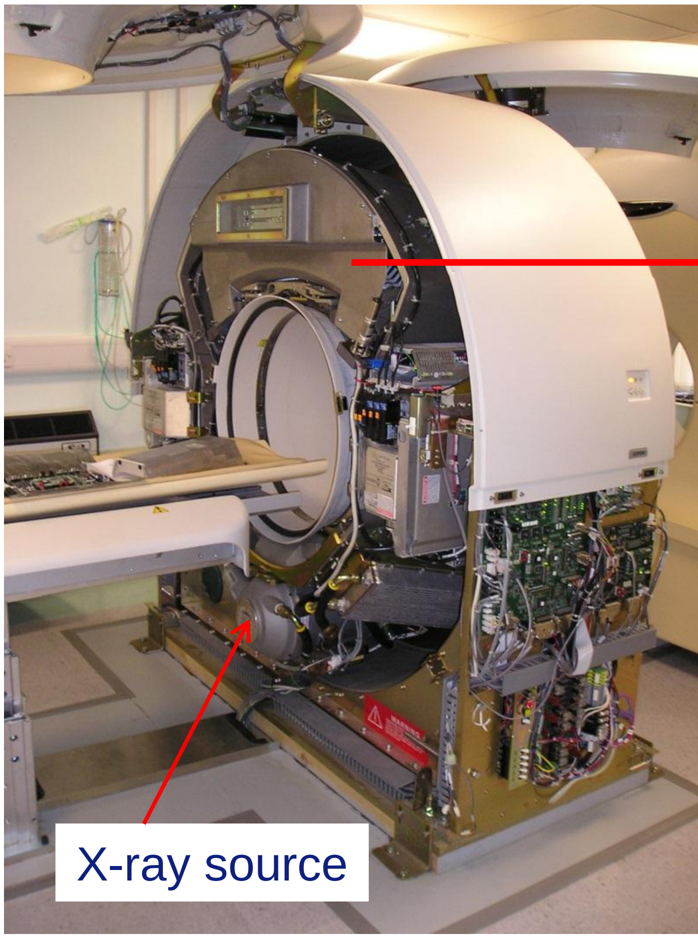
- Factors affecting spatial resolution:
  1. Set-up geometry: penumbra P = unsharpness / blurring in the image due to finite size of X-ray source generates → ideal set-up:
     a. Smallest possible X-ray spot size
     b. Patient on top of detector
     c. Large distance between source and patient
  2. Detector's properties: detector's intrinsic spatial resolution

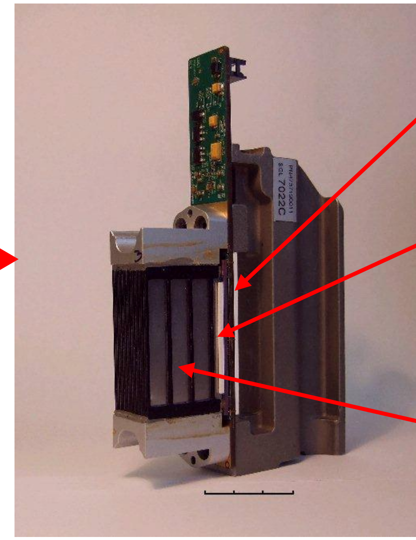
Contrast-to-noise ratio (CNR)

Ref. 1 – Chapter 2.8.3

- Factors affecting CNR:
  1. X-ray energy: the higher the energy the more X-rays undergo Compton scattering → CNR decreases
  2. FOV: up to 30 cm the larger the FOV the higher the number of Compton scattered X-rays reaching the detector → CNR decreases; above 30 cm there is no change
  3. Thickness of body part being imaged: the thicker the section the more X-rays undergo Compton scattering + the more X-rays get absorbed → CNR decreases
  4. Anti-scatter grid: reduces the Compton scattered X-rays reaching the detector → CNR increases

Computed Tomography (CT) scanners





CT scanner:
- X-rays rotating source
- Diametrically opposite detector unit

Market:
~30,000 scanners worldwide, 60 millions CT scans performed annually in USA

Courtesy Mike Partridge (Oxford)

Computed Tomography (CT)

Ref. 1 – Chapter 2.12

- Basic principle:
  - Conventional CT:
    1. Series of 1D projections at different angles is acquired continuously by synchronously rotating the X-ray source and detectors through one complete revolution around the patient
    2. The 1D projections are combined by the process of filtered backprojection to form the 2D CT image, also called slice
  - Spiral / helical and multi-slice helical CT
    1. 2D slices are acquired as in conventional CT
    2. Multiple adjacent slices are acquired by moving the patient's couch along the direction perpendicular to the slices' plane to give 3D images

Instrumentation for conventional CT

Instrumentation	Notes
X-ray tube	Same as in planar radiography kVp = $80, 100, 120, 140$ kV
Collimators	Same as in planar radiography
Anti-scatter grids	Same as in planar radiography but usually integrated in the detector array
Detectors	Only one detector unit = 1D array of several hundred 15×1 mm^2 detectors[1] along circumference
Heavy gantry	Has fixed to it X-ray tube and detector unit and rotates at high speed

[1]Detector = scintillator (converts X-rays into light) + photodiode (converts light into electric signal)

- Note: detector's orientation = wider side (15 mm) along couch axis $\rightarrow$ slice thickness determined by width of collimated beam that is < 15 mm

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Instrumentation for helical CT

- X-ray source and couch moved at the same time $\rightarrow$ X-ray path = helical
- Conventional CT set-up modified as follows:

1. Power supply and signal transmission cables are substituted by multiple slip-rings

   Reason: impossible to have fixed cables for power supply and signal transmission to read-out system

2. X-ray tube: specially designed to withstand very high temperatures in anode

   Reason: X-rays produced (almost) continuously $\rightarrow$ no cooling period $\rightarrow$ anode reaches very high temperatures = higher than in conventional CT

Instrumentation for multi-slice helical CT

- Same operation as helical CT but bigger detector unit $\rightarrow$ larger volumes can be imaged in a given time
- Same set-up as of the helical CT but with different geometry of the detector unit = 2D array of smaller detectors
  1. Along couch axis = detector size is much smaller (can be ~ 0.5 mm) but there are multiple rows that cover up to 16 cm $\rightarrow$ slice thickness determined by detector width = smaller than in helical CT
  2. Along circumference = detector size (1 mm) and number of detectors per row are the same as for helical CT 1D array

Dual-source CT

- Dual-source CT = 2 X-ray tube + multi-slice detector chains

  Reason: increases temporal resolution = 2 x temporal resolution of single-source CT

  Gantry's rotation = gravitational forces on scanner → rotation speed limited ($< 100 \div 160$ ms for 180^o) → temporal resolution limited

- Features:

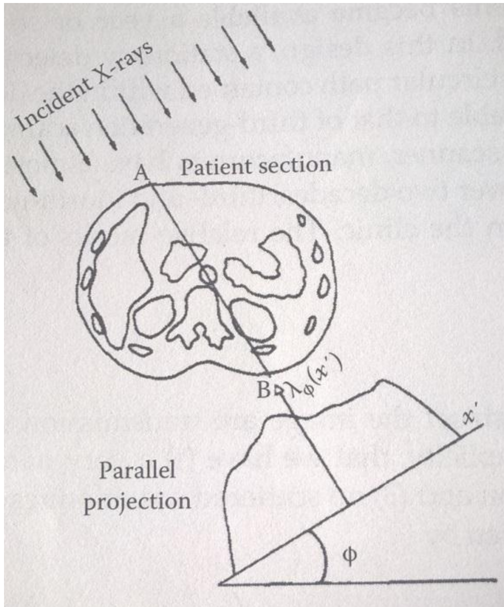
  1. Set-up: 1 standard chain (can be used alone) + 1 chain with narrow-arc detector = smaller FOV (~2/3) (only used with other)
  2. Data acquisition modes:
     a. Single energy = both tubes operated at same kVp
     b. Dual energy = tubes operated at different kVp = 140 keV and 80 keV → better contrast between different tissues

2D image reconstruction in CT

Ref. 2 – Chapter 3.2

- Data acquired and used to reconstruct image = transmission measurements:
  1. Exit (attenuated) X-ray beam intensity detected
  2. Ratio attenuated (exit) / unattenuated (entry) X-ray beam intensity → projections
  3. Reconstruction = extract linear attenuation coefficients from projections
  4. Image = display of linear attenuation coefficients' distribution

Transmitted intensity



Taken from Ref. 2 pg. 104

- xy frame = centred on body
- $x'y'$ rotating frame = centred on scanner
- X-ray source on y' axis

- Transmitted intensity $I_\phi(x')$:

$$I_\phi(x') = I_\phi^0(x') exp\left(-\int_{AB} \mu[x,y]\,dy'\right)$$

I_ϕ^0 = unattenuated, entry intensity

$\mu[x,y]$ = 2D distribution of linear attenuation coefficients

- Assumptions:
  1. Very narrow pencil X-ray beam
  2. Monochromatic radiation
  3. No scatter radiation reaching the detector

Projections

- A single projection $\lambda_\phi(x')$ is defined as:

$$\lambda_\phi(x') = -\ln\left[\frac{I_\phi(x')}{I_\phi^0(x')}\right]$$

$$= \iint_{-\infty}^{\infty} \mu[x,y]\,\delta(x\cos\phi + y\sin\phi - x')\,dx\,dy$$

$\delta =$ Dirac delta function $\rightarrow$ picks out AB path

- Reconstruction = to invert equation above = recover $\mu[x,y]$ from set of measured projections $\lambda_\phi(x')$

Image reconstruction

Ref. 2 – Chapter 3.6

- Mathematics of transmission CT and theory of image reconstruction from projections = research field on its own
- Reconstruction techniques:
  1. Convolution and backprojection methods also called filtered backprojection methods
  2. Iterative methods
  3. Cone-Beam reconstruction

  → extract spatial (2D) distribution of linear attenuation coefficients

Filtered backprojection reconstruction algorithms

Ref. 2 – Chapter 3.6

- Two steps to extract $\mu[x,y]$:
  1. Filtered / Convolution step: measured projection $\lambda_\phi(x')$ is filtered to give a filtered projection $\lambda_\phi^\dagger(x')$ = measured projection $\lambda_\phi(x')$ convolved with filtering operator $p(x')$:
  $$\lambda_\phi^\dagger(x') = \lambda_\phi(x') * p(x')$$
  2. Backprojection step: filtered projection $\lambda_\phi^\dagger(x')$ is backprojected = distributed over the $[x,y]$ space to give $\mu[x,y]$:
  $$\mu[x,y] = \int_0^\pi \lambda_\phi^\dagger(x')\, d\phi \,|_{x'=x\cos\phi+y\sin\phi}$$

Iterative reconstruction algorithms

Ref. 2 – Chapter 3.7

- Developed in early days, abandoned, now back in use
- Basic principle:
  1. Computed backprojections $\lambda'(\phi, x)$ at position (ϕ, x):

$$\lambda'(\phi, x) = \sum_{i=1}^{N} \alpha_i(\phi, x)\mu_i$$

     N = number of 2D pixels in the image

     α_i = average path length of projection through i pixel

     μ_i = linear attenuation coefficient density in i pixel

  2. α_i calculated once at start
  3. μ_i calculated <u>iteratively</u> until λ' closely resemble measured backprojections $\rightarrow$ image created from μ_i

Cone-Beam reconstruction algorithms

Ref. 2 – Chapter 3.8

- Two main categories:
  1. Exact Cone-Beam reconstruction algorithms

     Convert measured 1D projection data into plane integrals + use backprojection → complex and require high dose → considered impractical for medical applications
  2. Approximate Cone-Beam reconstruction algorithms

     Do not calculate full set of plane integrals → simpler and require less dose → widely used

Data interpolation in helical CT

Ref. 2 – Chapter 3.5

- Data acquired along helix and not within 2D plane → one (single-slice scanner) or few (multi-slice scanner) projections available in given plane → interpolation to get full set of projections for image reconstruction

1. Interpolation techniques for single-slice scanners:
   - 360º LI (Linear Interpolation): See for ex. W A Kalender et al., *Radiology* **176**, pg. 181-183 (1990)
   - 180º LI (Linear Interpolation): See for ex. C R Crawford & K F King, *Med. Phys*. **17**, pg. 967-982 (1990)
   - Other techniques: J. Hsieh, *Med. Phys.* **23**, pg. 221-229 (1996)
2. Interpolation techniques for multi-slice scanners:
   - See for ex. H. Hu, *Med. Phys.* **26**, pg. 5-18 (1999)

CT number

- $CT\ number$ of tissue = fractional difference of tissue linear attenuation coefficient μ_{tissue} relative to water μ_{water} measured in units of 0.001 = Hounsfield units (HU):

$$CT\ number = \frac{(\mu_{tissue} - \mu_{water})}{\mu_{water}} \times 1000$$

- Data acquired are rescaled in terms of $CT\ number$

2D image display

- Image formation steps:
  1. Backprojections are measured
  2. μ_i are calculated from backprojections for each i pixel
  3. $CT\ numbers$ are calculated and displayed
- "Display" = 512 × 512 matrix of 2D 12 bits pixels $\rightarrow$ $CT\ number$ range = $-1000 \div 3095$ HU. Some manufacturers offer increased range to $\sim 20{,}000$ HU (useful for areas with metal implants)
- Display monitor = typically 256 grey levels $\rightarrow$ windowing techniques = map selected range of $CT\ numbers$ (window width) onto grey scale

CT numbers of some tissues

Tissue	Density and μ_{tissue}	$CT\ number$ (HU)[1]
Bone	High $\rightarrow \mu_{bone} \gg \mu_{water}$	$1000 \div 3000$
Blood	Low $\rightarrow \mu_{blood} > \mu_{water}$	40
Muscle	Low $\rightarrow \mu_{muscle} > \mu_{water}$	$10 \div 40$
Brain (grey matter)	Low $\rightarrow \mu_{brain,g.m.} > \mu_{water}$	$35 \div 45$
Brain (white matter)	Low $\rightarrow \mu_{brain,w.m.} > \mu_{water}$	$20 \div 30$
Water		0
Lipid	Very low $\rightarrow \mu_{lipid} < \mu_{water}$	$-50 \div -100$
Air	Very low $\rightarrow \mu_{air} \ll \mu_{water}$	-1000

[1]At 70 keV

- Soft tissues = low density = $CT\ numbers$ very close to each other and to zero. Can still be resolved and reconstructed in CT

Signal-to-noise ratio (SNR)

- Sources of image noise:
  1. Poisson fluctuations
  2. Reconstruction algorithm
  3. Electronic noise = small contribution
- Poisson fluctuations propagates through reconstruction algorithm $\rightarrow$ object of uniform density μ appears mottled:

$$SNR = \frac{\mu}{\Delta\mu}$$

  $\Delta\mu =$ RMS fluctuation in μ reconstructed around mean
- Contrary to other imaging modalities, CT image noise not affected by pixel size

Spatial resolution

Ref. 2 – Chapter 3.9.1

- Spatial resolution = two terms:
  1. In the scan plane
  2. Perpendicular to the scan plane
- Factors affecting the spatial resolution:
  1. Spatial resolution in the scan plane: acquisition parameters (sampling frequency and bandwidth) and reconstruction algorithm
  2. Spatial resolution perpendicular to the scan plane: collimation

Low-contrast resolution

Ref. 2 – Chapter 3.9.1

- The smaller are the details with low-contrast that can be resolved the higher is the imaging efficacy
- Low-contrast resolution = diameter of the smallest low-contrast detail visible on the image
- Factors affecting low-contrast resolution:
  1. SNR
  2. Spatial resolution
  3. Reconstruction algorithm

Artefacts

1. Partial-volume artefacts

   Due to X-ray beam divergence or anatomical structures not perpendicular to slice → regions with density not corresponding to any real tissue

2. Beam-hardening artefacts

   Due to faster absorption of low-energy X-ray beam components → beam hardens → false reduction in density + false details = ex. dark bands

3. Aliasing artefacts

   Due to wrong sampling

4. Motion artefacts

   Due to patient movement during scan = inconsistencies in the projections → “artificial” sudden changes in attenuation

5. Equipment-related artefacts

   Due to changes in performance → artefacts depend on faulty components = ex. rings due to drifts in detector performance

Effects of reconstruction algorithms on image quality

Ref. 2 – Chapters 3.9.9, 3.9.10 and 3.9.11

- Effect of spiral interpolation algorithms

  Some degree of blurring of the image is introduced

- Effect of iterative algorithms

  Noise is lower → dose could be reduced

  Noise texture is different → challenge for the radiologist as not used to it

- Effect of Cone-Beam reconstruction algorithms

  ‘Wave’ or ‘windmill’ artefacts can be introduced

Quality control of CT scanners

Ref. 2 – Chapter 3.11

- X-ray tube tests
- Scan localisation
- CT dosimetry
- Image quality
- Helical scanning

X-ray imaging dose

- X-ray imaging = ionising radiation = associated dose
- Dose = damage:
  1. Deterministic effects
  2. Stochastic effects
- Damage = side effects $\rightarrow$ concern
- Dose needs to be quantified:
  - Absorbed dose in tissue D_T
  - Equivalent dose in tissue H_T
  - Effective dose in tissue E_T

Dose quantification in CT

- X-ray beam = divergent $\rightarrow$ beam profile across slice not uniform $\rightarrow$ CT dose index $CTDI$
- $CTDI$ measured not on patients but on dosimetry phantoms
- Dose delivered to patients is complex function of:
  1. Scanner parameters = geometry, X-ray beam quality and filtering
  2. Size of patient
  3. Acquisition parameters
- Empirical relation between dose on phantom and effective dose on patient

CT dose index

- CT dose index $CTDI$:

$$CDTI_{100} = \frac{1}{NT}\int_{-50\ mm}^{+50\ mm} D(z)\, dz$$

N = number of slices

T = slice width

D = dose profile along axis of rotation z

- Dosimetry phantoms used = two cylindrical Perspex phantoms:

1. Diameter 16 cm
2. Diameter 32 cm

Other CT dose indexes

- $CTDI$ depends on where on plane → weighted $CTDI_w$:

$$CTDI_w = \frac{1}{3}CTDI_{centre,100} + \frac{2}{3}CTDI_{periphery,100}$$

$CTDI_{centre,100} = CTDI_{100}$ at centre of phantom

$CTDI_{periphery,100} = CTDI_{100}$ 1 cm under phantom surface

- Average dose in volume irradiated $CTDI_{vol}$:

$$CTDI_{vol} = \frac{CTDI_w}{p}$$

$p =$ pitch of helical scan = $\frac{couch\ increment\ in\ one\ revolution}{slice\ thickness}$

Doses associated to imaging procedures

- Approximate effective doses for common X-ray imaging procedures

Body section (Procedure)	Effective dose (mSv)	
	Planar radiography	CT scan
Chest	0.04	8.3
Abdominal	1.5	7.2
Brain		1.8
Lumbar spine	2.4	

- Exact dose depends on:
  1. Imaging system used
  2. Patient's size

CT –vs– planar radiography

- CT disadvantages

CT much more complex than planar radiography
CT much more expensive than planar radiography
CT delivers higher dose to patients

- CT advantages

CT allows contrasts down to 1% to be imaged → distinguishes soft tissue	Planar radiography allows contrasts only down to 2% to be imaged → cannot distinguish soft tissues
CT provides 3D images	Planar radiography provides only 2D images → 3D body structure collapsed on 2D film