UNIVERSITY OF OXFORD

## Joint CI-JAI advanced accelerator lecture series

# Imaging and detectors for medical physics

# Lecture 6: SPECT

Dr Barbara Camanzi

barbara.camanzi@stfc.ac.uk

# 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

# Single Photon Emission Computed Tomography (SPECT)

Ref. 1 – Chapter 3.9

- Basic principle:
  - Source:
    1. Radioisotope usually emitting a single $\gamma$-ray per nuclear disintegration $\rightarrow$ from where 'Single Photon'
    2. Radiotracer injected in patient $\rightarrow$ $\gamma$-rays are emitted from inside the body $\rightarrow$ from where 'Emission'
  - Provides 3D images using CT techniques $\rightarrow$ from where 'CT':
    1. System rotates around patient $\rightarrow$ detects $\gamma$-rays at different angles $\rightarrow$ 2D projections[1]
    2. Tomographic reconstruction algorithm applied to 2D projections $\rightarrow$ 3D images

[1]Signal at each image point calculated as sum = projection of radioactivity along line through body

# SPECT –vs– CT

- SPECT similar to CT

| | SPECT | CT |
|---|---|---|
| Radiation | $\gamma$-rays | X-rays |
| Source location | Internal to body of patient<br>Injected into patient | External to body of patient |
| Type of imaging | Emission | Transmission |

# SPECT scanner components

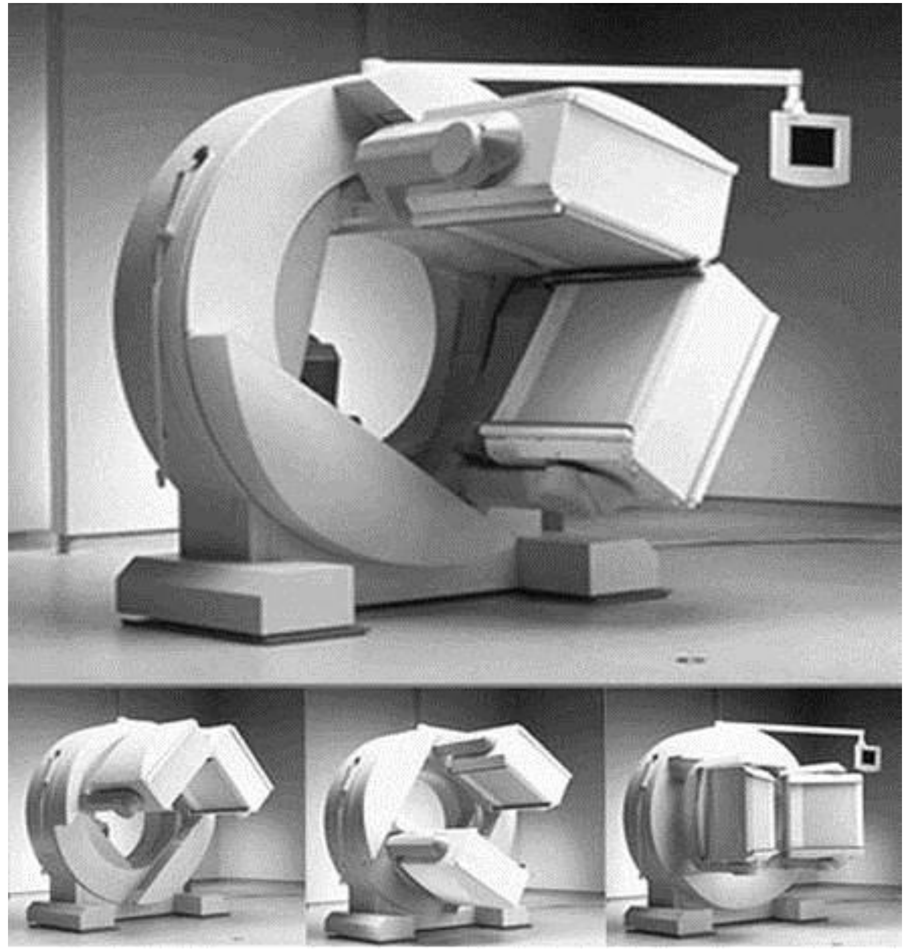
1. Two or three gamma cameras = heads =multi-headed systems → acquire multiple views simultaneously
2. Converging collimator often used inside gamma cameras → increases $SNR$
3. Moving bed often attached → allows whole body studies

Two heads

Three heads

# Heads positioning

- Heads can be positioned at different angles and locations and moved inwards and outwards to make distance to organ under study as close as possible:
  - Outwards = to accommodate body
  - Inwards = to make gamma cameras close to the brain



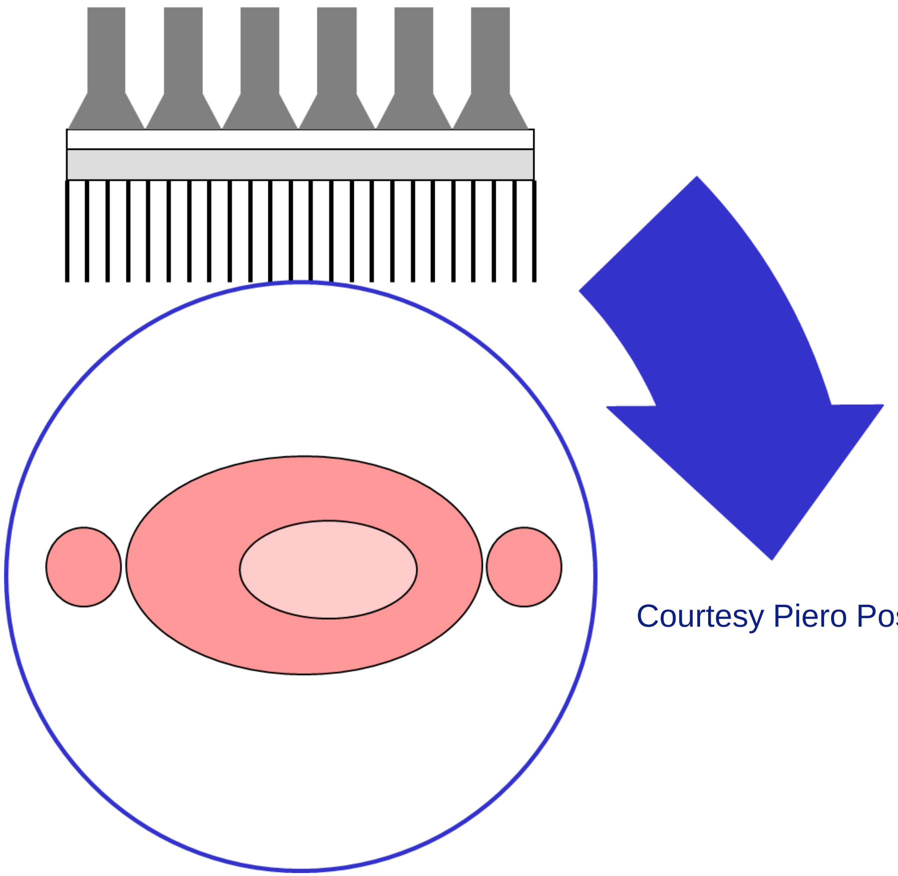
After Piero Posocco (Imperial College)

# Data acquisition

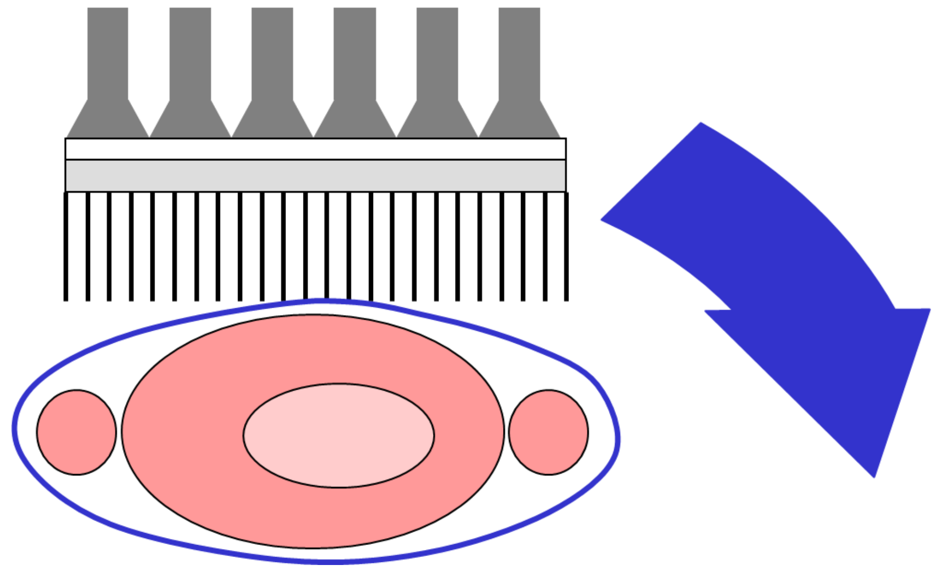
- Rotation of gamma cameras around patient $\rightarrow$ different thickness of body in FOV = different source – detector distance at different positions that affects:
  1. Degree of tissue attenuation
  2. Distribution of $\gamma$-rays in body
  3. Spatial resolution: due to diverging-cone shape of line of response defined by collimator
- As a result:
  – Projections at $180°$ to each other not identical
  – $360°$ rotation known as conjugate counting needed = combines data acquired from opposing views $\rightarrow$ reduces attenuation effects and gives more uniform resolution

# 360° rotation orbits

## Circular orbit




## Contoured orbit




Courtesy Piero Posocco (Imperial College)

- Higher spatial resolution
- Needs complex path for detector head

# Typical acquisition parameters

| Parameter | Value |
|---|---|
| Number of angular views | 64÷128 |
| Linear sampling pitch | 2÷3 mm |
| Data collection | Over 360° |
| Display matrix | 64 × 64 or 128 × 128 |
| FOV | ~40÷60 cm trans-axially |
| Axial coverage[1] | 30÷40 cm |

[1]Achieved by stacking images

# Image formation

- Two steps:
  1. Application of corrections to the data in two steps:
     a. Scatter
     b. Depth-dependent attenuation
  2. Image reconstruction

# Scatter

- Events acquired in energy window of width that depends on energy resolution of gamma camera $\rightarrow \gamma$-rays scattered at large angles can be still accepted
- Ex.: for a system with energy resolution $20\%$ and detection window = $127 \div 153$ keV $\rightarrow$ $140$ keV $\gamma$-rays from $^{99}Tc^{m}$ scattered up to $50°$ will fall in the detection window $\rightarrow$ will be accepted
- Contribution from scattered events:
  - Typically of same order as unscattered events even with collimator

# Scatter contribution

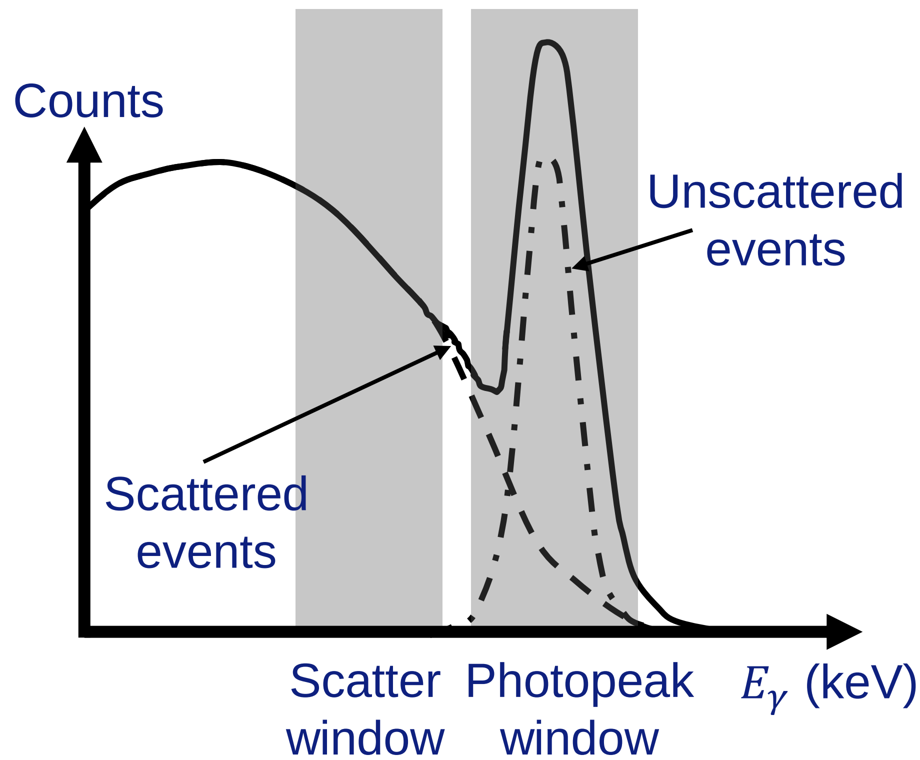
- Contribution from scattered $\gamma$-rays typically of same order as unscattered $\gamma$-rays even with lead collimator
- Number of scattered $\gamma$-rays greatest closer to areas of high concentration of radiotracer $\rightarrow$ position dependent correction
- Scattered $\gamma$-rays have lost position information $\rightarrow$ get assigned to wrong pixel $\rightarrow$ overestimation of radioactivity in that pixel $\rightarrow$ image contrast reduced $\rightarrow$ clinically relevant details may be obscured

# Scatter correction methods

- Two methods:
  - Dual-energy window approach = most common method
  - Triple-energy window approach
- Ratio of scattered to unscattered $\gamma$-rays may be as high as $40\%$ even when using narrow energy windows

# Dual-energy window approach

- Two windows:
  1. Primary photopeak window centred at photopeak
  2. Secondary scatter sub-window set to lower energy range
- Assumptions:
  1. Contributions in windows:
     a. Photopeak window: scattered and unscattered $\gamma$-rays
     b. Scatter sub-window: only scattered $\gamma$-rays
  2. Constant ratio between signal in photopeak window and signal in scatter window

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# Dual-energy window data processing

- Steps:
  1. Projections formed separately from data in photopeak and scatter windows
  2. Scatter projections scaled by scaling factor
  3. Scaled scatter projections subtracted from photopeak projections
- Limitation: multiple scatter more likely in scatter window → spatial distribution of scatter somewhat different in photopeak and scatter windows

# Dual-energy window scaling factor

- The scaling factor $SF$ is given by:

$$SF = \frac{W_P}{2W_S}$$

$W_P$ = width photopeak window

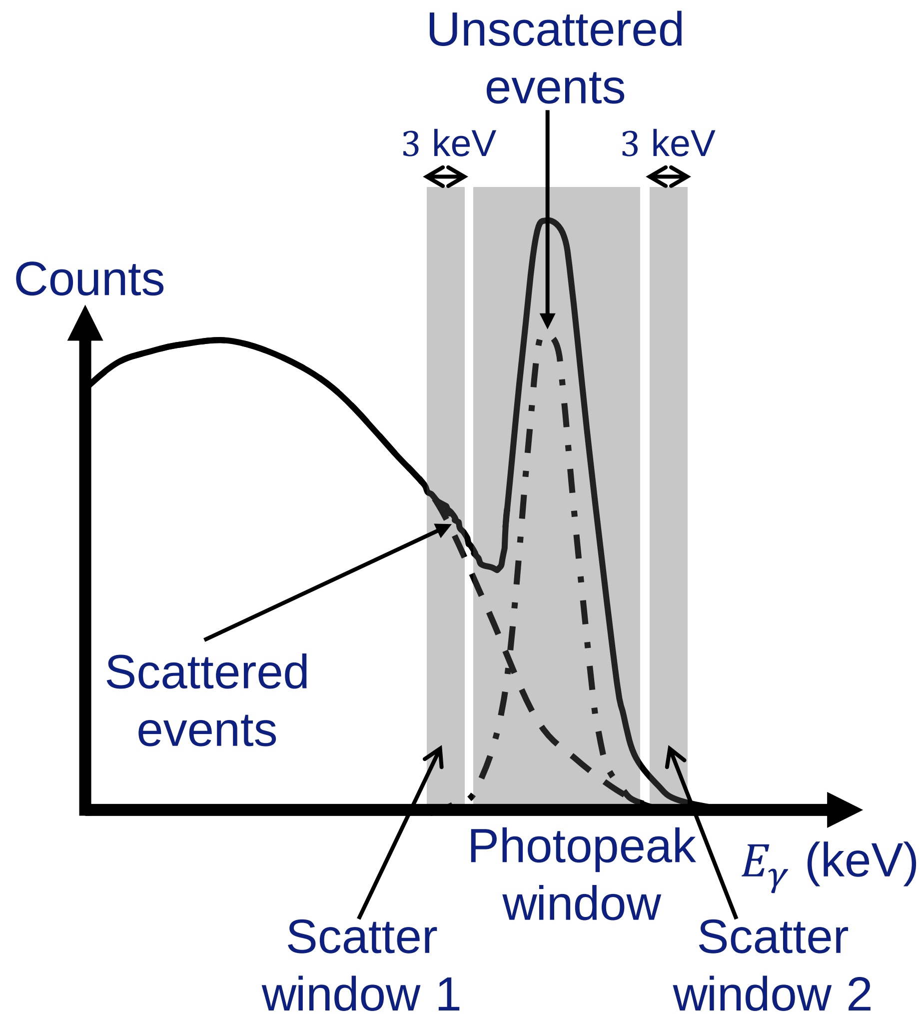
$W_S$ = width scatter window

$2$ = factor determined experimentally

- $SF$ depends on:
  1. Energy resolution of gamma camera $\rightarrow$ choice of detection window
  2. Size of object to be scanned

# Triple-energy window approach

- Three windows:
  1. Primary photopeak window centred at photopeak and with width narrower than gamma camera detection window → 15% width for a gamma camera with width of detection window 20%
  2. Two secondary very narrow scatter sub-windows on both sides of photopeak window of width 3 keV centred on the edge of gamma camera detection window sides

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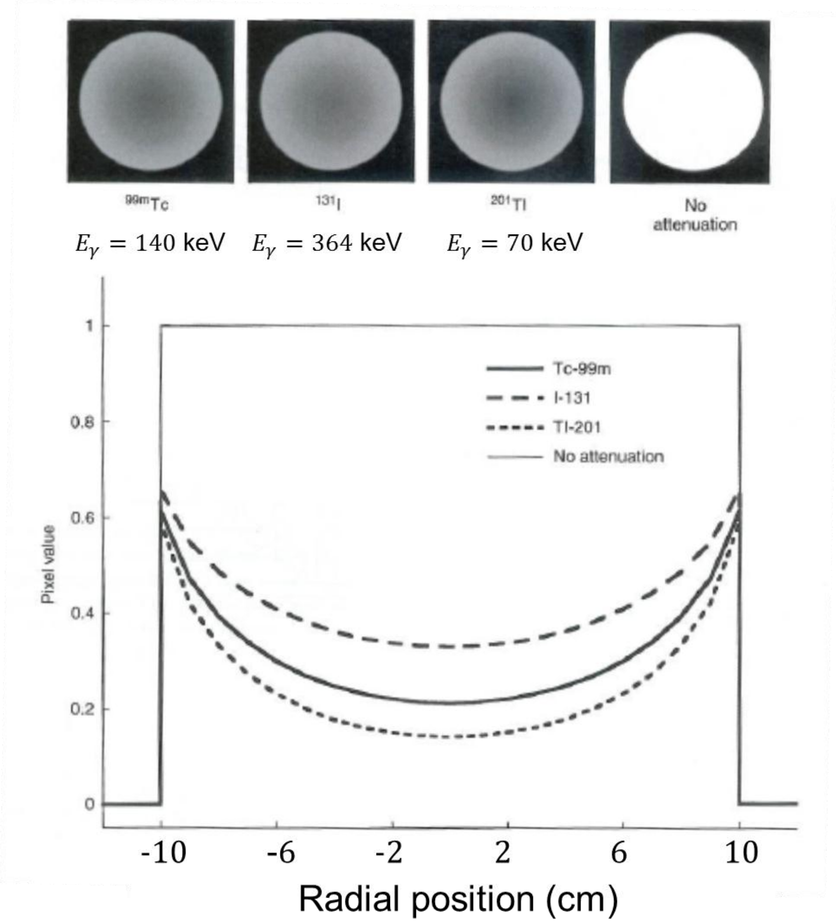
# Triple-energy window data processing

- Steps:
  1. Projections formed separately from data in photopeak and two scatter windows
  2. Scatter projections scaled by width of scatter windows
  3. Scaled scatter projections from two windows added
  4. Resultant scatter projections multiplied by half width of photopeak window
  5. Final scatter projections subtracted from photopeak projections

# Attenuation

- $\gamma$-rays from radiotracer inside body travel through patient to get to detector $\rightarrow$ attenuation
- Attenuation depends from:
  1. Amount of tissue $\gamma$-rays go through $\rightarrow$ $\gamma$-rays from deeper in the body / further away from detector more attenuated
  2. $E_\gamma$ $\rightarrow$ lower energy $\gamma$-rays more attenuated
- SPECT measures radiotracer distribution and not attenuation $\rightarrow$ attenuation creates image artefacts $\rightarrow$ correction
- Attenuation = bigger magnitude effect than scatter

# Attenuation contribution: an example

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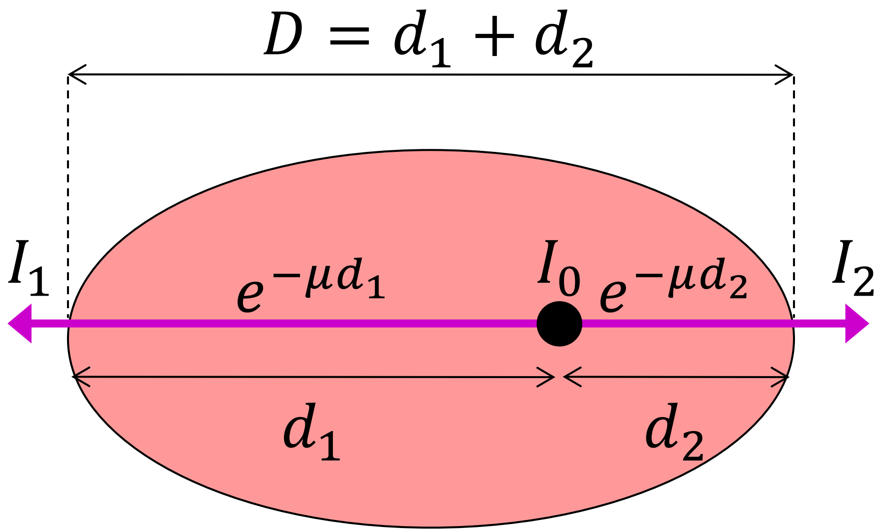
- Simulated reconstructed images:
  - Three images from 20 cm diameter cylindrical phantom loaded with three different radiotracers $^{99}Tc^{m}$, $^{131}I$, $^{201}Tl \rightarrow$ $\gamma$-rays of different energies
  - One image with no attenuation
- Reconstructed image:
  - Uniform when there is no attenuation
  - Non uniform due to attenuation

# Attenuation correction with conjugate counting

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$d_1, d_2 =$ distances source – detector along opposing directions

$I_0 =$ unattenuated signal

$I_1 = I_0 e^{-\mu d_1} =$ signal along direction 1

$I_2 = I_0 e^{-\mu d_2} =$ signal along direction 2

$\mu =$ linear attenuation coefficient

- $\mu$ assumed constant throughout patient to determine correction for attenuation = attenuation correction factor $ACF$
- Geometric mean $I_m$ of two signals $I_1$ and $I_2$ measured in opposing directions is then given by:

$$I_m = \sqrt{I_1 \cdot I_2} = \sqrt{I_0 e^{-\mu d_1} \cdot I_0 e^{-\mu d_2}}$$
$$= I_0 \sqrt{e^{-\mu(d_1+d_2)}} = I_0 e^{-\mu D/2}$$

- The unattenuated signal $I_0$ is given by the attenuated signal $I_m$ multiplied by $ACF$:

$$I_0 = I_m \times ACF$$

- $ACF$ is therefore given by:

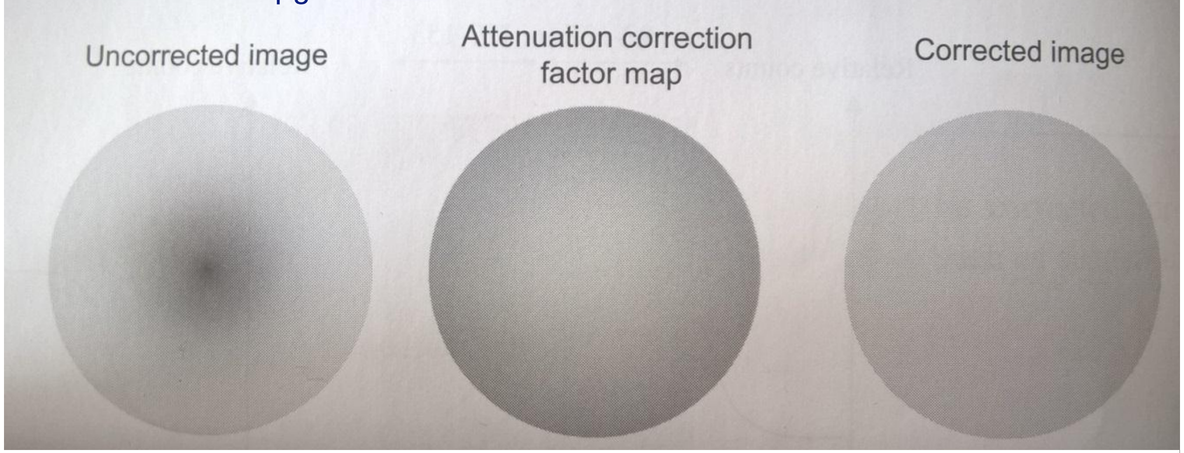
$$ACF = e^{\mu D/2}$$

- Both $\mu$ and $D$ need to be measured to determine $ACF$

# Attenuation correction methods

- Two analytical methods:
  - Chang's multiplicative method
  - Transmission scan method
- One 'empirical' method:
  - Mean shape and attenuation of an average patient used

Taken from Ref. 1 pg. 114



# Chang's multiplicative method

- Steps:
  1. Initial image of radiotracer distribution formed by filtered backprojection without corrections → attenuation ignored
  2. Initial image used to estimate outline contour of patient → distance $x$ that $\gamma$-rays have to travel through tissue
  3. Attenuation correction factor $ACF$ calculated for each pixel in reconstructed image as:

     $$ACF = e^{\mu x}$$

     $\mu =$ linear attenuation coefficient
  4. Attenuation correction factor is applied to each pixel to generate corrected reconstructed image = each projection is scaled up by the attenuation correction factor

# Chang's multiplicative method limitation

- Assumption used by the method = tissue attenuation coefficient uniform through part of body imaged
- Validity of assumption:
  – Valid for brain and abdomen
  – Not valid for thorax where tissue attenuation is highly spatially dependent
- Method can only be used for brain and abdomen imaging but not for cardiac or thoracic imaging

# Transmission scan method

- A transmission scan is acquired simultaneously with a patient scan
- Transmission scan = external $^{153}Gd$ source emitting $\sim 102$ keV $\gamma$-rays $\rightarrow$ measures tissue attenuation $\rightarrow$ equivalent to low-statistics, low-spatial-resolution CT scan
- $^{153}Gd$ half-life = $242$ days $\rightarrow$ radioactivity constant over several months $\rightarrow$ does not need replacing frequently

# Transmission scan acquisition

- Two scans performed simultaneously:
  1. Patient scan with $^{99}Tc^{m}$ = emission scan
  2. Transmission scan with external $^{153}Gd$ source
- $E_{\gamma}^{Gd} \ll E_{\gamma}^{Tc} \rightarrow$ two scans acquired at same time using dual-energy window
- Third window between other two can be used to correct scattered $\gamma$-rays leaks from $^{99}Tc^{m}$ to $^{153}Gd$ windows
- Separate transmission scan acquired without patient $\rightarrow$ no object in front of gamma camera = reference scan

# Transmission scan correction calculation

- Projections $\lambda$ calculated comparing $^{153}Gd$ signal intensities with $I_{trans}$ and without $I_{ref}$ the patient:

$$\lambda = -\ln \frac{I_{trans}}{I_{ref}}$$

- Map of attenuation coefficients $\mu$ at each position calculated from projections:

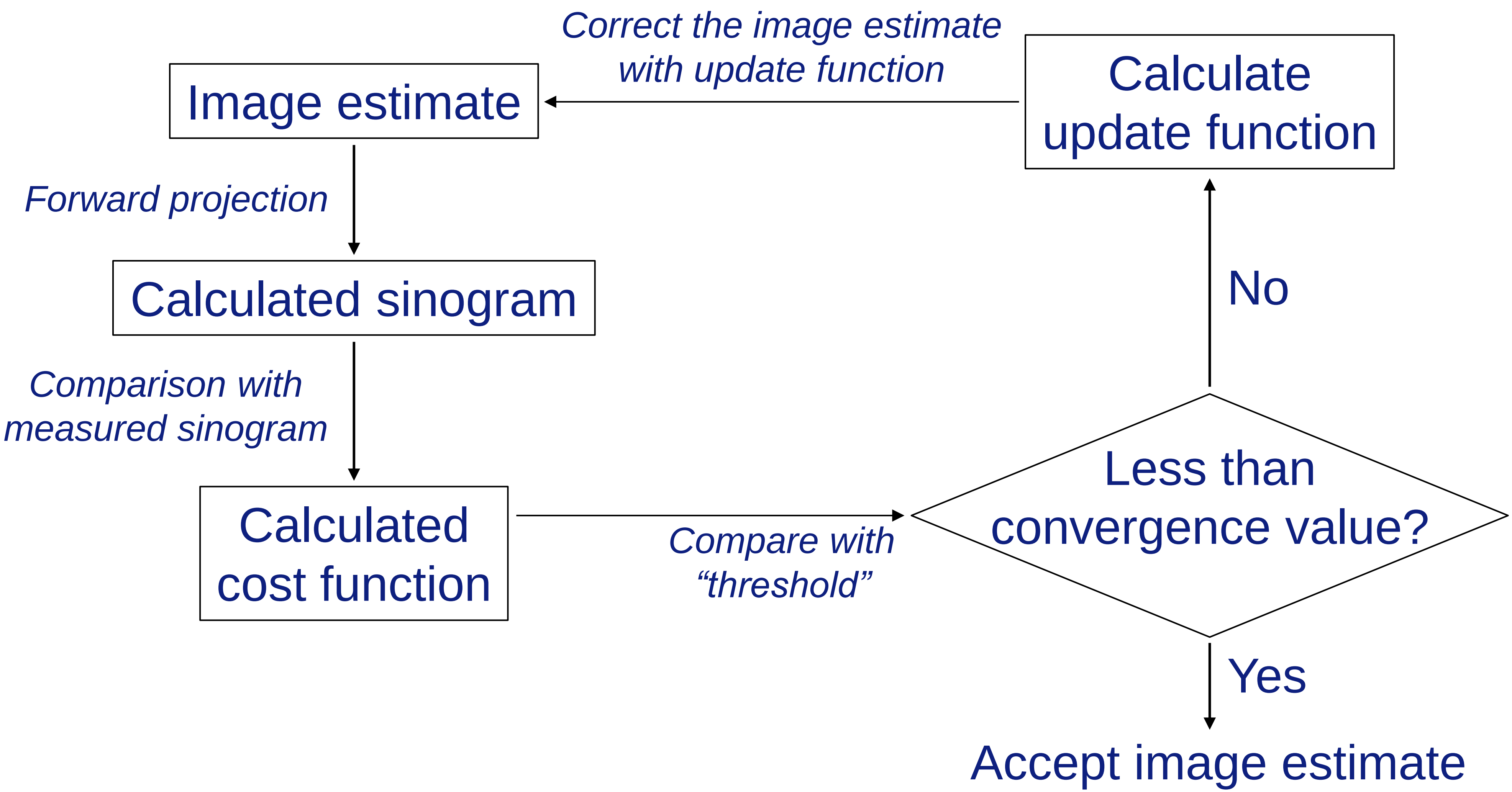
$$I_{trans} = I_{ref} e^{-\mu x}$$

- Map of attenuation coefficients used to correct image

# Image reconstruction

- SPECT uses CT techniques to reconstruct 2D axial slices from acquired projections and from there 3D distribution of radionuclides → two classes of reconstruction algorithms used:
  1. Filtered backprojection techniques = identical to CT
  2. Iterative techniques: computer intensive → methods / algorithms developed to speed it up, including using a subset of acquired profiles → most common algorithm = maximum-likelihood expectation-maximum (ML-EM) method

# Iterative reconstruction



# Common radiotracers for SPECT

| Radiotracer | $E_\gamma$ (keV) |
|---|---|
| Myocardial perfusion (rest/stress) | |
| $^{201}Tl$ | $68 \div 80$ |
| $^{99}Tc^m$-Sestamibi | $140$ |
| $^{99}Tc^m$-Tetrofosmin | $140$ |
| Cerebral perfusion | |
| $^{99}Tc^m$-HMPAO | $140$ |
| $^{99}Tc^m$-ECD | $140$ |
| Oncology | |
| $^{67}Ga$ | $93,\ 185,\ 300$ |
| $^{201}Tl$ | $68 \div 80$ |
| $^{99}Tc^m$-Sestamibi | $140$ |
| Somastatin receptors: $^{99}Tc^m$-Depreotide, $^{111}In$-Octreotide | $140\ (^{99}Tc^m),\ 171,\ 245\ (^{111}I)$ |
| Labelled antibodies, peptides | |

# Main clinical applications

- Myocardial perfusion studies = majority of SPECT scans:
  - To detect coronary artery diseases or myocardial infarction
  - Single image = multi-slice data set = typically 100,000 counts
  - Spatial resolution $\sim$14 mm
- Brain studies:
  - To detect areas of reduced blood flow associated with stroke, epilepsy or neurodegenerative diseases / conditions such Alzheimer
  - Single image = multi-slice data set = typically 500,000 counts
  - Spatial resolution $\sim$7 mm $\leftarrow$ distance source – detector smaller than in myocardial studies