Nuclear diagnostics and Magnetic Resonance Imaging

Lecture 4: Nuclear diagnostics IV: SPECT

K. Long
Imperial College London/STFC

K.Long@Imperial.ac.uk

Outline

1. **Single photon emission computed tomorgraphy; SPECT**
   - Introduction
   - Reconstruction
   - Attenuation correction strategies
   - Scatter correction; overview
   - Example images

Section 1

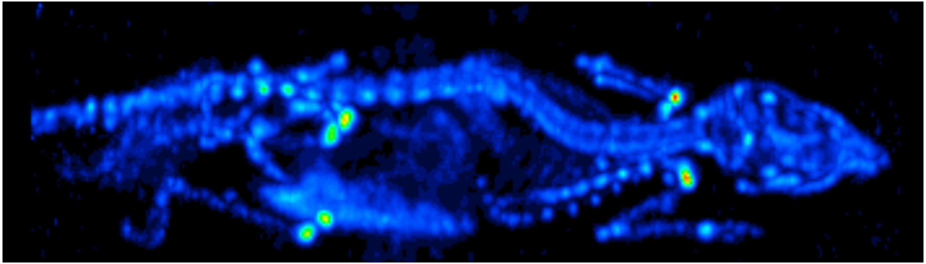
Single photon emission computed tomorgraphy; SPECT

Introduction

Gamma camera gives single projected image of object; cf conventional x-ray image

SPECT: Single Photon Emission Computed Tomography; cf X-ray CR

SPECT image of mouse with bone tracer



Click here for C. Lackas' animated gif on wikipedia

Image prepared by C. Lackas

Typical SPECT systems

Two (or more) gantry-mounted gamma cameras:

- Gamma cameras rotate around patient; 2D cross section
- Images taken from multiple angles
- Bed moves in longitudinal direction

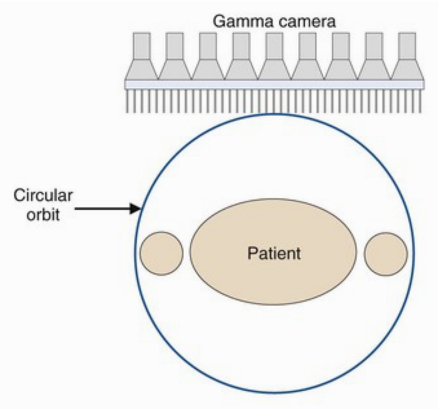
Allows 3D images to be reconstructed

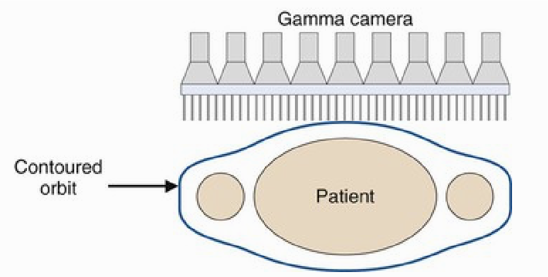
Ring of planar or pinhole gamma cameras

- 2D images obtained without rotation of detectors
- Images taken from multiple angles at the same time
- Bed moves in longitudinal direction

Allows 3D images to be reconstructed

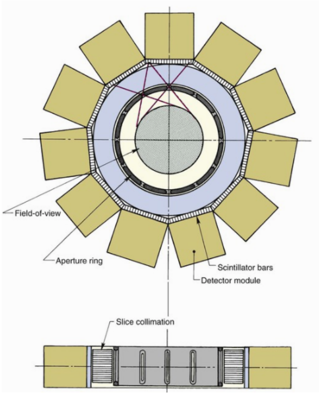
Circular and contoured orbits



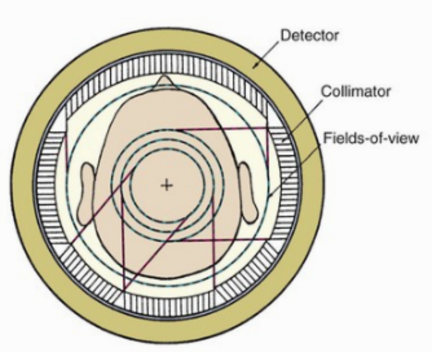


"Elliptical" orbit is more complicated but has the advantage of increased precision

Alternative configurations, in this case for brain scans



Aperture ring (12 slits) rotates

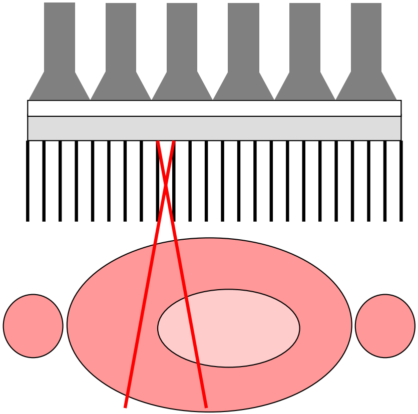


Each collimator section has its own field-of-view diameter

Typical parameters

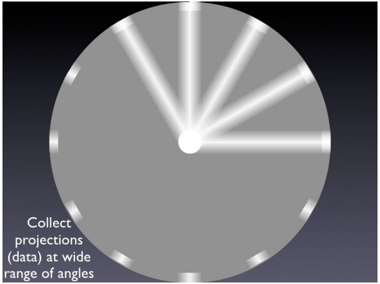
- 64 to 128 angular views
- 2-3 mm linear sampling along longitudinal axis
- 360° data collection
- Reconstructed on 64×64 or 128×128 matrix
- Field of view $\sim$ 40–60 cm transaxially
- Stack of images covering $\sim$ 30–40 cm longitudinally

Projection on image plane

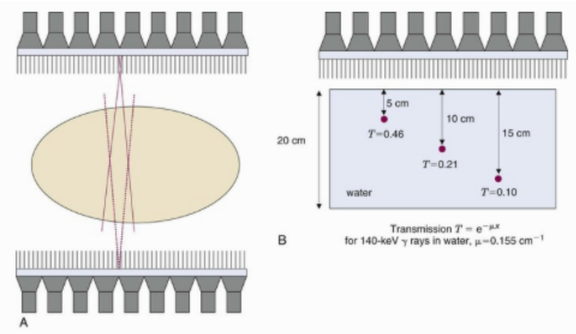


Absorptive collimator means that each hole views a pencil-like area of the object

Repeat for a wide range of angles:



Geometric response; conjugate counting



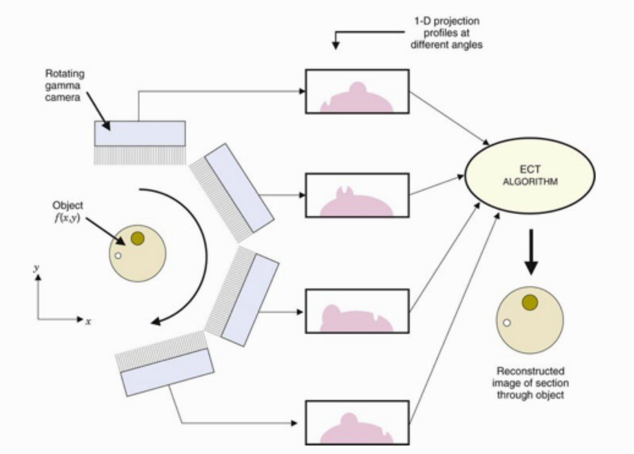
$I(r,\phi) \neq I(r,\phi+\pi)$

- Attenuation of γ intensity
- Divergence of image cone
- Measurements at ϕ and $\phi+\pi$ "conjugate"

Example of attenuation of γs from ^{99m}Tc:

- Exploit conjugate measurements to correct for lost attenuation

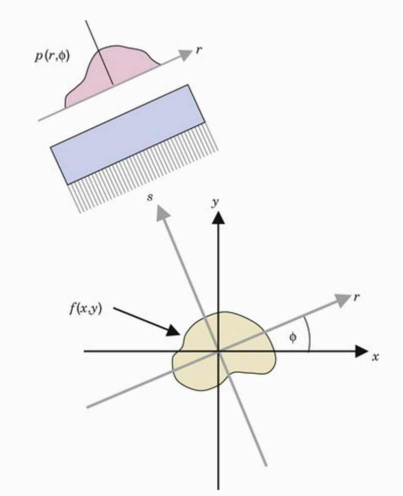
Back projection



ECT: Emission computed tomography

TCT: Transmission computed tomography

Back projection: local coordinate system



Local coordinate system r, s:

- r: coordinate along gamma camera
- s: distance camera to source

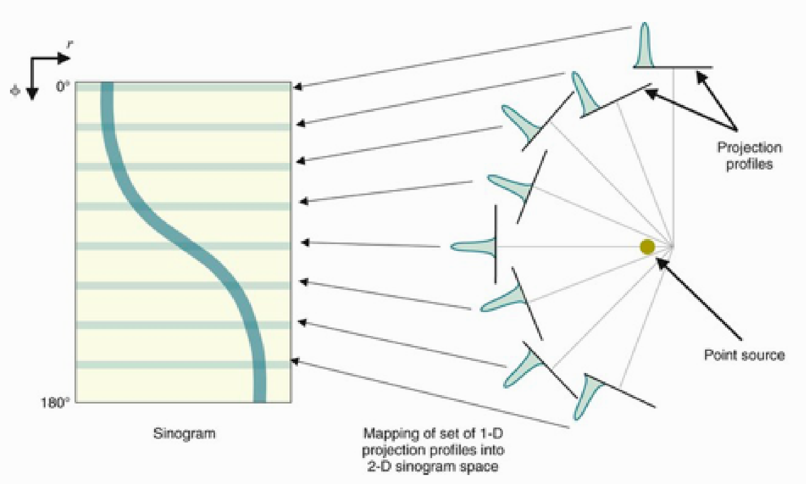
r, s coordinates related to x, y by:

$$\begin{aligned} r &= x\cos\phi + y\sin\phi \\ s &= -x\sin\phi + y\cos\phi \end{aligned}$$

x, y may be reconstructed using:

$$\begin{aligned} x &= r\cos\phi - s\sin\phi \\ y &= r\sin\phi + s\cos\phi \end{aligned}$$

Back projection: sinogram



Back projection: profile construction

Measurement at each projection measures a response "profile", $p(r, \phi)$

Want to reconstruct the activity in a particular slice, $f(r, s)$ or $f(x, y)$

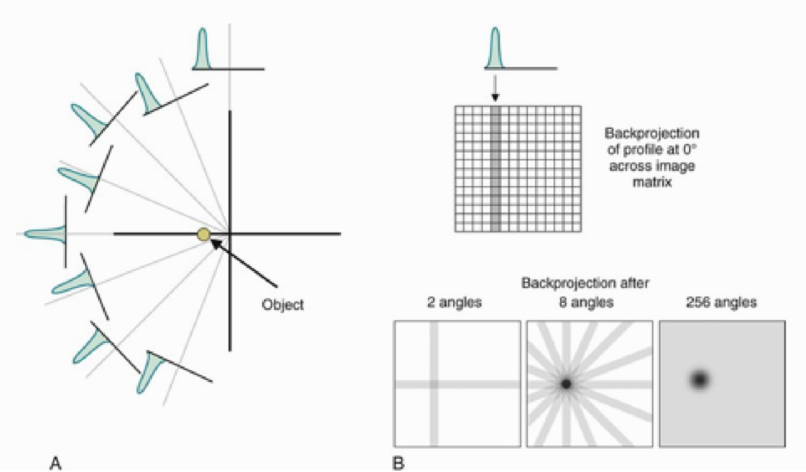
In "simple" back projection, the total response measured at a particular r_i, ϕ_i is divided between the pixels along the projected coordinate s

The total (uncorrected) activity, $f'(x, y)$ within a pixel at coordinate x, y is then given by:

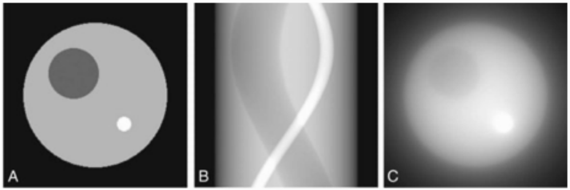
$$f'(x, y) = \frac{1}{N} \sum_{i=1}^{N} p(r_i, \phi_i)$$

where the sum runs over the N projections that illuminate the pixel at x, y

Back projection: illustration



Back projection: illustration using a simple phantom



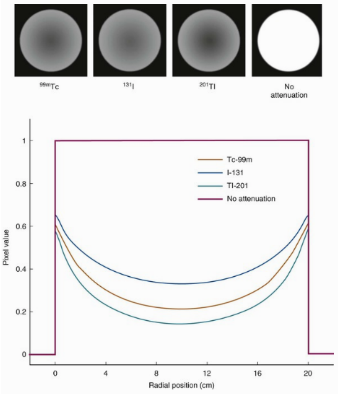
Main features of phantom appear in the image . . . but . . .

The attribution of activity to pixels along the projected coordinate s causes “spoke-like” image with few projections

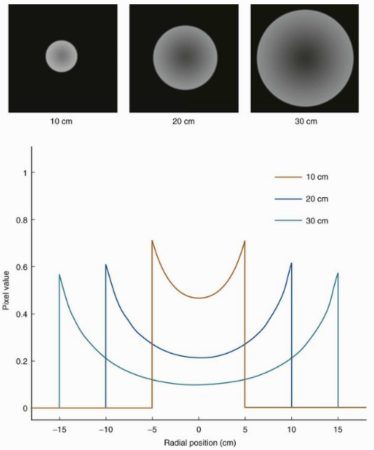
More projections improve image, but, attribution of activity leads to apparent activity outside the object and blurring of the image

More sophisticated reconstruction algorithms (e.g. filtered back projection, see later) have been developed to overcome this defect

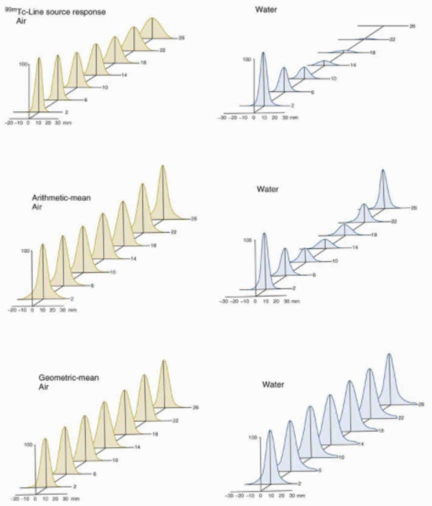
Attenuation depends on γ energy and depth



$E_\gamma^{99m\mathrm{Tc}} = 140$ keV; $E_\gamma^{131\mathrm{I}} = 364$ keV; $E_\gamma^{201\mathrm{Tl}} = 70$ keV



Attenuation

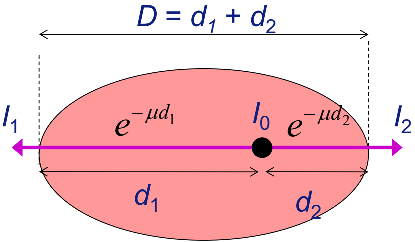


Example:

- High-resolution pin-hole collimator
- Resolution for line source diameter 2.5 mm:
  - As a function of distance source→detector
  - In air (left) and in water (right)
- Corrections applied:
  - Top: no correction
  - Middle: arithmetic mean: $I_A = \frac{1}{2}(I_1 + I_2)$
  - Bottom: geometric mean: $I_G = (I_1 \times I_2)^{\frac{1}{2}}$

Arithmetic mean gives most uniform response

Geometric mean



Intensity measured in two conjugate PMTs, numbered 1 and 2:

$$\begin{aligned} I_1 &= I_{01}\exp(-\mu d_1) \\ I_2 &= I_{02}\exp(-\mu d_2) \end{aligned}$$

Geometric mean; I_G:

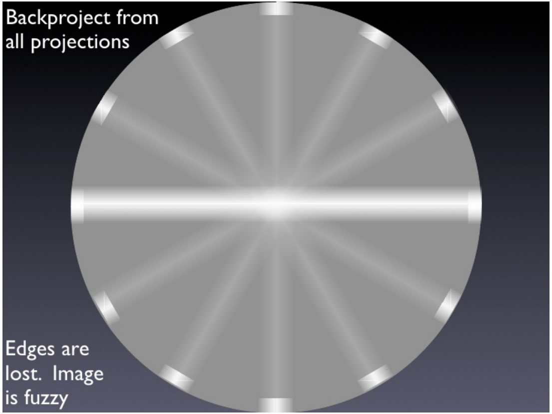
$$\begin{aligned} I_G^2 &= I_{01} \times I_{02} \\ &= I_{01} \times I_{02} \exp\left(-\mu(d_1 + d_2)\right) \end{aligned}$$

If $I_0 = I_{01} + I_{02}$:

$$I_G = I_0 \exp\left(-\mu\frac{D}{2}\right)$$

I.e. I_G depends on total depth D rather than d_1 or d_2.
The result is exact only for homogeneous media and point sources. Corrections can be derived to accommodate these effects.

Attenuation correction



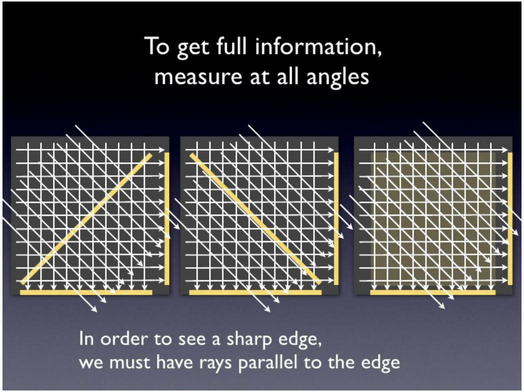
Define, attenuation correction factor, ACF:

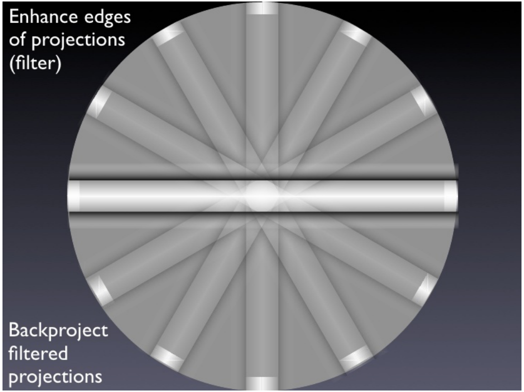
$$\mathrm{ACF} = \exp\left(\mu \frac{D}{2}\right)$$

The corrected intensity I_{corr} is then calculated by evaluating:

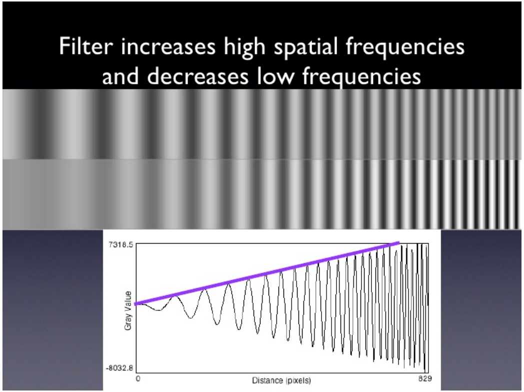
$$I_{\mathrm{corr}} = \mathrm{ACF} \times I_G$$

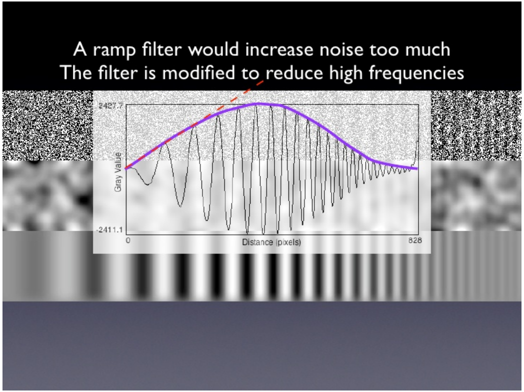
Combination of projections



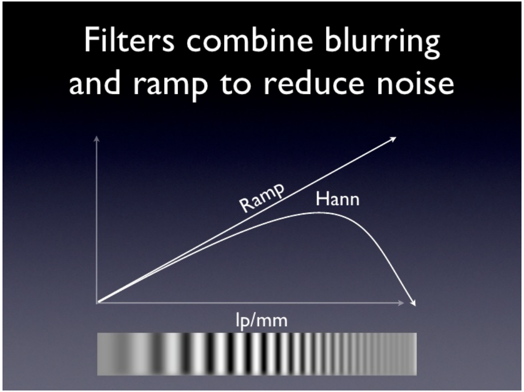


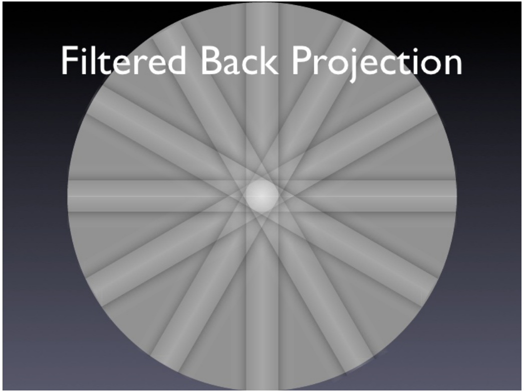
Filtered back projection





Filtered back projection





Attenuation correction strategies

1. Exploit ACF in "Chang's multiplicative method"
2. Generate a transmission map using "attenuation scans"
3. Use mean patient shape
    - Disadvantage "there is no mean (or average) patient"
4. Exploit CT image:
    - X-ray image processed to give transmission map that can be used to calculate ACF as a function of position

Will consider 1 and 2 below

Chang's multiplicative method

Steps:

1. Reconstruct image without any attenuation correction
2. Use reconstructed image to identify contour of patient
3. Assume uniform linear attenuation coefficient, μ, and calculate ACF pixel by pixel
4. ...

Calculation of ACF pixel by pixel

For pixel i at position x_i, y_i, a distance d_i from the surface in the direction of the camera, the pixel's attenuation factor, η_i, is given by:

$$\eta_i = \exp(-\mu d_i)$$

For a pixel at x, y, can now sum attenuation over all pixels between x, y and the surface to obtain the total attenuation factor for the path:

$$\eta = \sum_1^N \exp(-\mu d_i)$$

As before, N is the number of projections. The attenuation correction coefficient, now a function of x and y is given by:

$$\mathrm{ACF}(x, y) = \frac{1}{\sum_1^N \exp(-\mu d_i)}$$

Chang's multiplicative method

Steps:

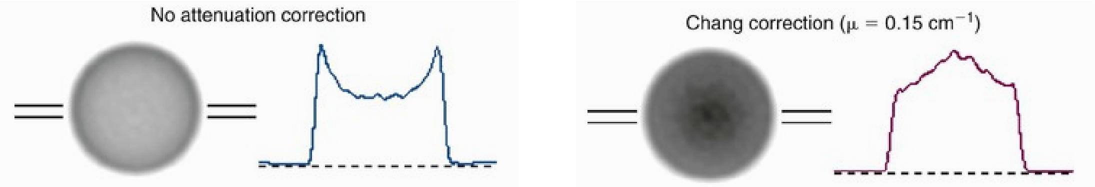
1. Reconstruct image without any attenuation correction
2. Use reconstructed image to identify contour of patient
3. Assume uniform linear attenuation coefficient, μ, calculate ACF(x, y)
4. Apply ACF pixel by pixel:

$$f(x, y) = f'(x, y) \times \mathrm{ACF}(x, y)$$

where $f'(x, y)$ is the uncorrected response reconstructed in the pixel at x, y, and $f(x, y)$ is the corrected response.

Example

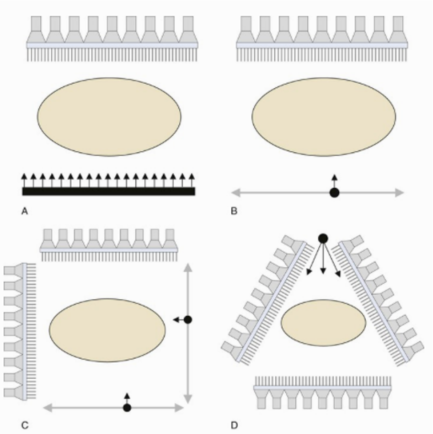
20 cm diameter cylinder with uniform concentration of ^{99m}Tc.



Apparent "over correction" attributed to scattered events.

In this example Chang's method has been applied, followed by a further correction by "forward projecting". The corrected image is projected to the gamma camera. The predicted response of the camera is then compared to the measured response and a further correction is made based on the difference of the forward projection and the measurement.

Transmission scans



A: Flood source
B: Single source
C: 2 orthogonal sources
D: Stationary line source

Reference scan: $I_{\rm ref}$; transmission scan: $I_{\rm trans}$
For a particular projection element:

$$I_{\rm trans} = I_{\rm ref} \exp(-\mu d)$$

Taking the logarithm of the ratio:

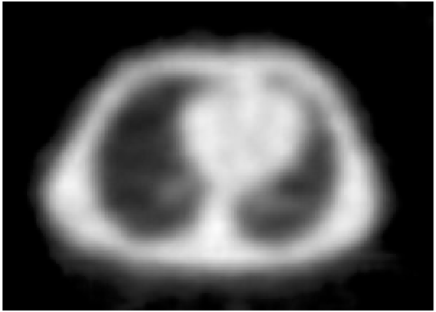
$$\ln\left(\frac{I_{\rm ref}}{I_{\rm trans}}\right) = \mu d$$

Back-projection technique yields

$$\mu d = \sum_i \mu_i d_i$$

where the $i^{\rm th}$ pixel is of size d_i and is characterised by μ_i

Transmission scan: example



Transmission map of thorax using moving line source.

Radionuclides for transmission scans:

- ^{99m}Tc ($E_\gamma = 140\,$keV)
- ^{153}Gd ($E_\gamma = 97\,$keV and $103\,$keV)
- ^{123}Te ($E_\gamma = 159\,$keV)

Long half-life convenient as then source does not need to be replaced frequently

Scatter correction

Primarily due to Compton scattering

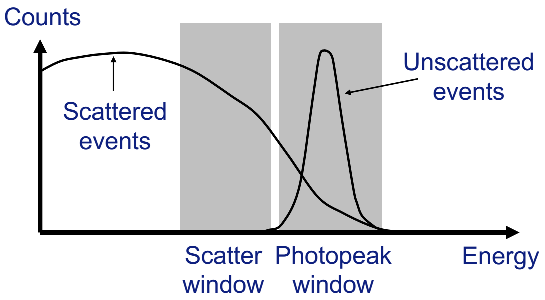
Effect is smaller in magnitude than attenuation

Ratio of scattered to non-scattered photons may be as high as 40%, even when using a narrow energy window

Scatter reduces image contrast as events are put in the "wrong place" and leads to an overestimation of radioactivity in a pixel

Loss of contrast may obscure clinically relevant details

Scatter correction



Weighting factor must be determined experimentally, it depends on:

- Choice of energy detection window (photopeak window)
- Size of object being scanned
- Energy resolution of gamma camera

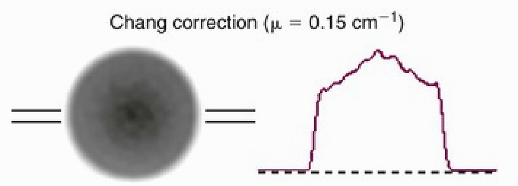
Estimate contribution of scattering events in "photopeak window" by calculating a weighting factor, w_f

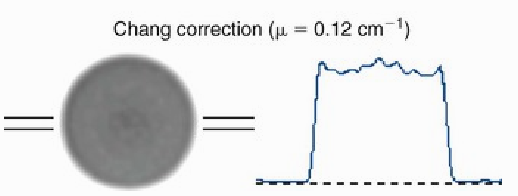
Number of events subtracted from photopeak is w_f times number of events in scatter window

Scatter-correction method limited by differences in spatial distribution of scatter and photopeak

Example of impact of scatter correction

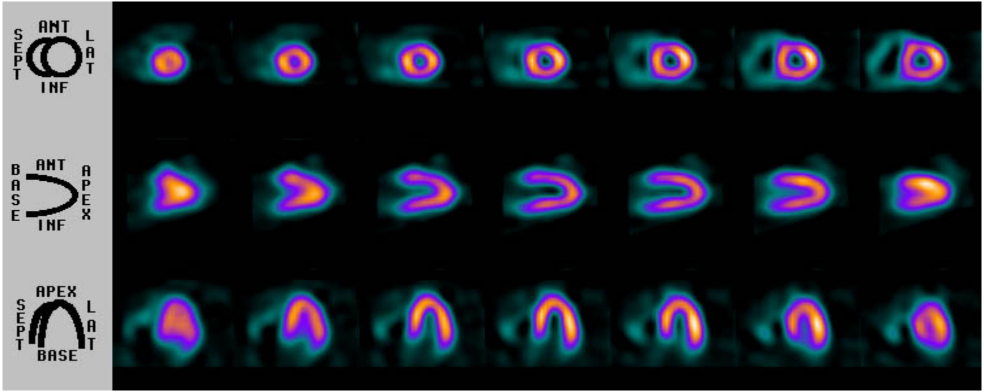
20 cm diameter cylinder with uniform concentration of ^{99m}Tc.





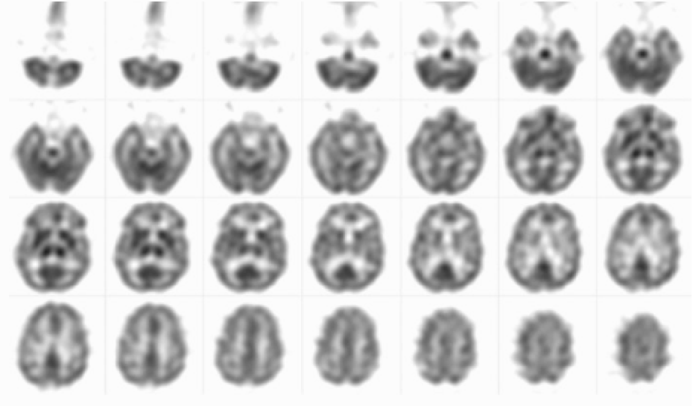
"Over correction" noted above removed by scatter correction

SPECT images of cardiac perfusion



Cardiac perfusion scan using ^{99m}Tc-sestamibi. Images are shown in three slices, as indicated on the LHS. The time sequence (left to right) is in steps of 20 s.

SPECT images of brain perfusion



Brain perfusion scan using ^{99m}Tc-HMPAO. Images were acquired with an exposure of 40 s per view.