Brief Introduction to CT and PET Imaging

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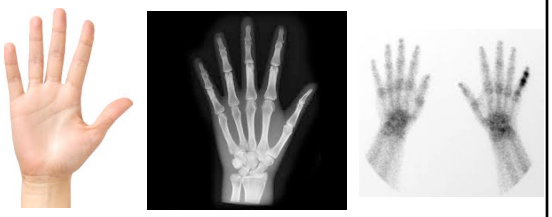
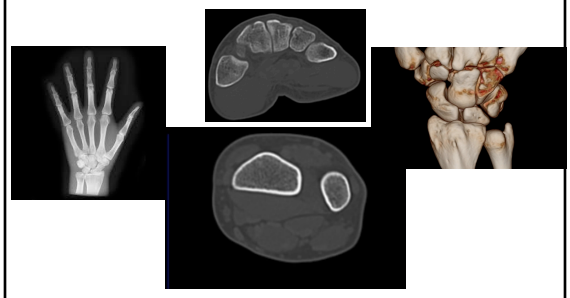
Types of Images



What is this an image of?
What physical property is being imaged?

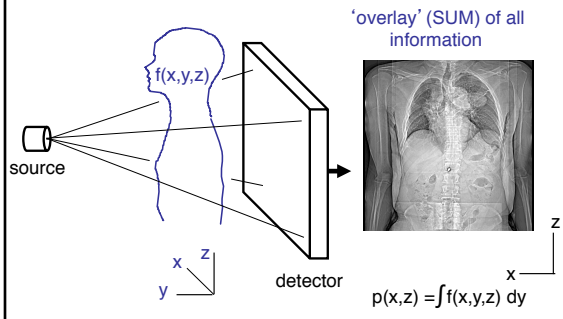
2

X-Ray Projections to CT



3

Projection Imaging



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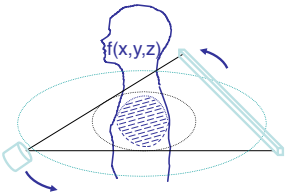
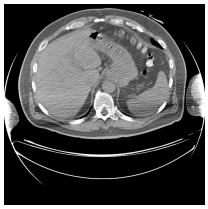
Tomographic Imaging

'Tomo' + 'graphy' = Greek: 'slice' + 'picture'



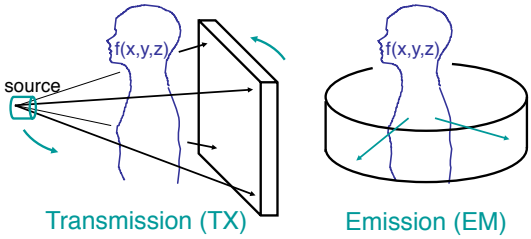
orbiting source + detector data for all angles



true cross-sectional image

5

Types of imaging systems



Transmission (TX) Emission (EM)

… but same mathematics of tomography

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The Nobel Prize in Physiology or Medicine 1979
Allan M. Cormack, Godfrey N. Hounsfield

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The Nobel Prize in Physiology or Medicine 1979

Allan M. Cormack
Prize share: 1/2

Godfrey N. Hounsfield
Prize share: 1/2

- First CT System developed at EMI Central Research Laboratories, 1967-1971
- First CT Scan of patient in 1971

The Nobel Prize in Physiology or Medicine 1979 was awarded jointly to Allan M. Cormack and Godfrey N. Hounsfield *"for the development of computer assisted tomography"*

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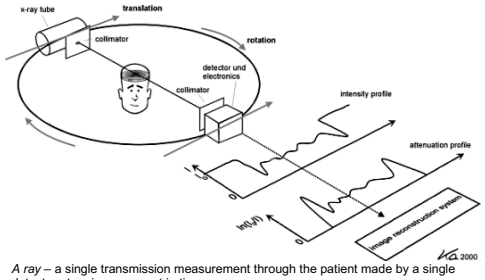
Formation of a CT image – Measurement Data



- *A ray* – a single transmission measurement through the patient made by a single detector at a given moment in time
- A *projection* or *view* - a series of rays that pass through the patient at the same orientation or view angle
- By stacking a series of 2D X-ray CT images we can get a volumetric image or data set, which is then displayed by looking at principal sections through the image volume

Physics of Transmission Imaging

The Electromagnetic Spectrum

longer wavelength ⟷ higher energy

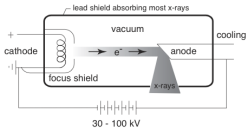
radiofrequency	micro wave	IR	UV	X-ray & gamma-rays	cosmic -ray

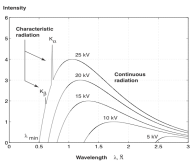
AM FM TV ↑ Optical

Transmission through 10cm of tissue (water)

X-ray Production

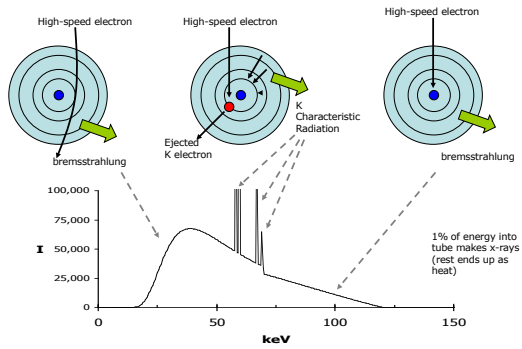
- In evacuated tube, heated cathode current releases electrons
- Electrons are accelerated to anode by voltage
- In field of anode atoms, electrons release their energy as *bremsstrahlung*
- Yields a continuous X-ray spectrum





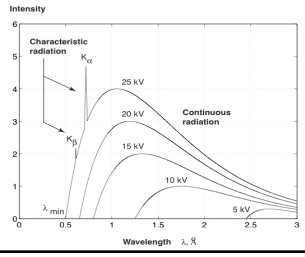
•*Characteristic radiation* from knock-outs in inner electron shell and subsequent filling of the low-energy hole. Energy emitted is characteristic of the energy levels in the anode material.

X-ray Production
accelerated electron interacts with nuclei in target (often tungsten with x-ray tubes)



X-ray Source Parameters

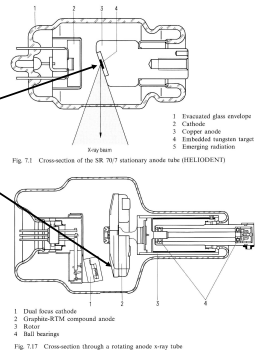
- Amount of x-ray photons = Cathode current · time [mAs]
- Energy of the emitted photons controlled by voltage between anode and cathode [kV] - but a spectrum of energies produced. Often peak photon energy quoted [kVp]



X-ray Source Tubes

- Stationary anode tube (low power version suitable for radiographs)

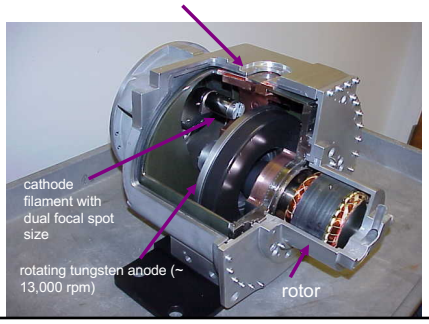
focal spot



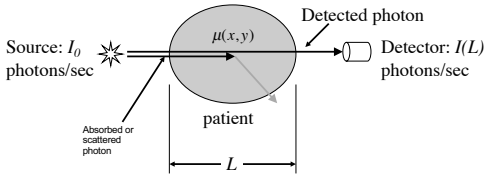
- Rotating anode tube (dissipates heat for higher continuous beam currents needed for CT)

Modern X-Ray Tube (GE Performix)

metal vacuum casing with thin metal foil for X-ray aperture



Physics of transmission imaging



Narrow-beam approximation: $\frac{dI}{dx} = -\mu I \rightarrow I(x) = I_0 e^{-\int_0^L \mu(x,y)dL}$

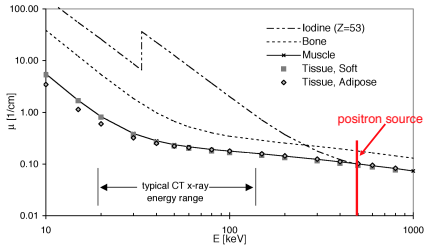
Linear attenuation coefficient (1/cm): $\mu(x,y) = \mu(x,y)_{PE} + \mu(x,y)_{CS} + \ldots$

photo-electric absorption, Compton scatter

most important at ~30-1000 keV

Attenuation of photons is a function of

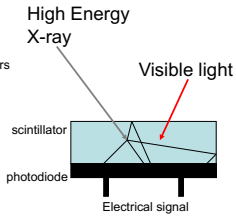
1. energy of the photons
2. Medium (density & atomic number)



CT Detectors

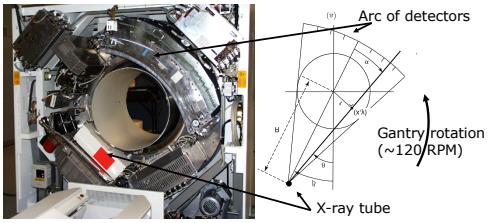
Most modern CT systems use either:

- Xenon Detectors
  - old technology still found in a few single slice scanners
  - Highly directional xenon filled ionization chambers
- Solid-state scintillator detectors
  - Better x-ray absorption (denser and higher Z)
  - Require slight gap between adjacent detector elements reducing geometric efficiency
  - similar to digital radiography systems
  - Specifics of detector materials and fabrication proprietary information (CsI, GE uses "Lumex" or "Garnet", Siemens uses "GOS")



X-ray CT Scanner Geometries

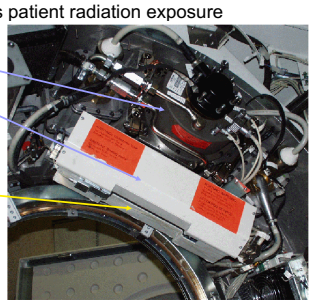
- For efficiency, clinical X-ray CT scanners always use a 'fan-beam' arrangement



- So far we have described how to get a single 2D image, but how do we image a volume?
  - 1. Step and scan (sequential CT, cine CT)
  - 2. Helical (or Spiral CT)

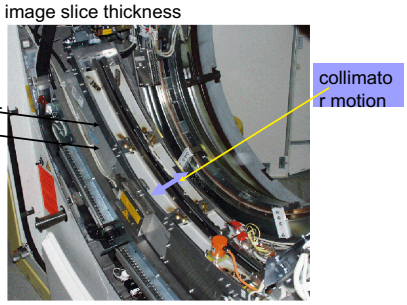
Pre-Patient Collimation

- Controls patient radiation exposure

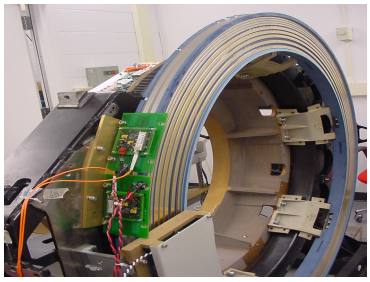


Post-Patient Collimation

- Adjusts image slice thickness

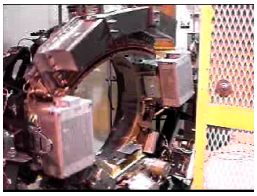


Gantry Slip Rings



- Allows for continuous rotation

CT Movie: 0.3 sec Rotation Time

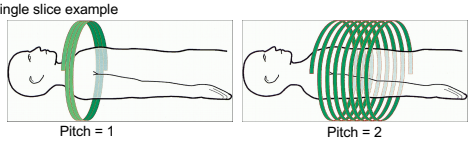


64-slice CT, weight ~ 1 ton, speed 0.33 sec (180 rpm)

Pitch

$$\text{pitch} = \frac{\text{table travel per rotation}}{\text{(number detectors) x (detector width)}} = \frac{\text{table travel per rotation}}{\text{acquisition beam width}}$$

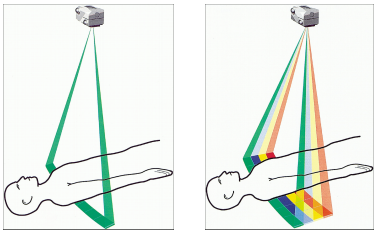
slingle slice example



Pitch = 1 Pitch = 2

Single versus Multislice Detectors

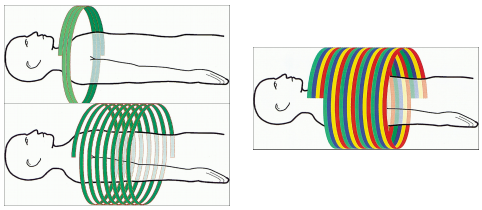
- Can image multiple planes at once



1 detector row 4 detector rows

Helical Multi-Detector CT (MDCT)

- Fastest possible acquisition mode -- same region of body scanned in fewer rotations, even less motion effects
- Single row scanners have to either scan longer, or have bigger gaps in coverage, or accept less patient coverage
- The real advantage is reduction in scan time



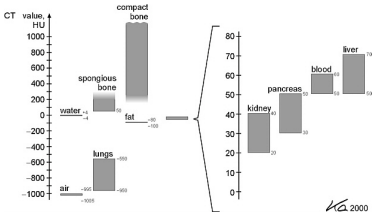
1 detector row: pitch 1 and 2 4 detector rows: pitch 1

X-ray CT Scanning

- To provide a uniform scale, do a scale conversion where all scanners set air = -1000 and water to 0. Units are called Hounsfield units [HU] after Godfrey N. Hounsfield.

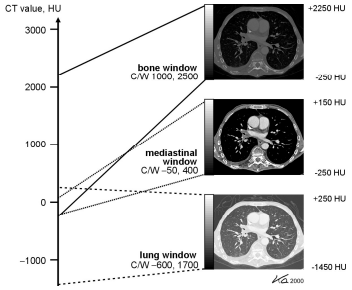
$$\text{CT number} = 1000\,\frac{\mu_{pixel} - \mu_{water}}{\mu_{water}}\,[\text{HU}]$$

- Some typical values



X-ray CT Scanning

- The Hounsfield scale has no upper limit, but scanners will often limit display to a max of 3071 = 4096 (2^{12}) - 1024



Intro to PET

Some History:

- One of the first effective PET tomographs attributed to Mike Phelps and Ed Hoffman of Washington University for blood flow and metabolism in animals 1974
- First Human scanner 1975 (only ~4 years after first human CT)
- In December 2000, The Health Care Financing Administration recommended Medicare reimbursement of PET for diagnosis, staging and restaging of the following:
  - Lung Cancer, Colorectal Cancer, Melanoma, Lymphoma, Head and Neck Cancer (excluding central nervous system and thyroid), Esophageal Cancer, Refractory Seizures, Myocardial Viability, and Breast Cancer.

- PET has 4 major stages:
  - I. Radionuclide Production
  - II. Radiochemistry
  - III. Imaging
  - IV. Data Analysis

Basic Stages of PET

I. Radionuclide Production
- Make radio-isotope

II. Radiochemistry
- Make radiopharmaceutical - Label a tracer

III. Imaging
1. Administer radiotracer
2. Positron decay - annihilation
3. Anti-parallel photons travel through patient (some interact)
4. Photons enter detectors (most interact)
5. Detected photons paired into coincident events
6. Store events in sinogram format (data)

IV. Data Analysis
1. Correct data for errors
2. Reconstruction into images and interpret

I. Radionuclide Production

- Stable nuclei bombarded with high energy particles

PET RADIONUCLIDES PRODUCED WITH A CYCLOTRON

Nuclide	Half-Life	Nuclear Reaction
O-15	2 min	$^{14}N(d,n)^{15}O$; $^{15}N(p,n)^{15}O$
N-13	10 min	$^{12}C(d,n)^{13}N$
C-11	20 min	$^{10}B(d,n)^{11}C$
F-18	110 min	$^{18}O(p,n)^{18}F$

p is proton, n is neutron, d is deuterium (heavy hydrogen with proton and neutron)

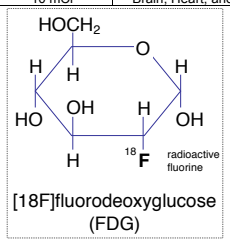
In production of F-18, O-18 is the target material bombarded with a proton causing a neutron to be emitted in the production of F-18

Requires on-site cyclotron or quick access to isotopes

II. Radiochemistry

- Radioactive isotopes are synthesized with biologically meaningful material to form a radiotracer (radiopharmaceutical) - "label" a substance used in the body

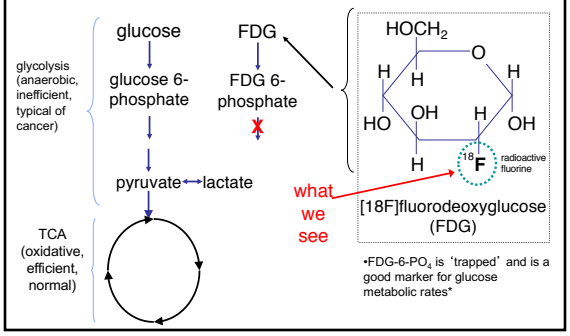
Radiotracer	Injected Dose	Imaging application
[O-15]water	50 mCi	Brain Blood Flow
[N-13]ammonia	25 mCi	Myocardial Perfusion
[C-11]acetate	20 mCi	Myocardial Flow and Metabolism
[F-18]FDG	10 mCi	Brain, Heart, and Tumor Metabolism



III. Imaging

1. Administer Radiotracer - travels to function of interest

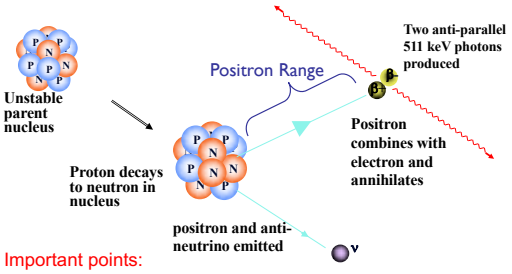
Example: Glucose Metabolism



•FDG-6-PO_4 is 'trapped' and is a good marker for glucose metabolic rates*

III. Imaging

2. Radioisotope Decays



Important points:

- Positron travels finite distance before annihilation (F18 ~ 1 mm water)
- Photons are not at exactly 180° apart

→Result in position inaccuracies

III. Imaging

3. Photons travel through subject

- Photons interact with matter as they travel:
  - Rayleigh Scattering
  - **Photoelectric Absorption**
  - **Compton Scatter**
  - Pair Production

  Occur in PET
- The probability that the photons will continue on a straight line (will not be attenuated) is stated as:

$$P = e^{-\mu x}$$

μ- linear attenuation coefficient, x - distance along a line

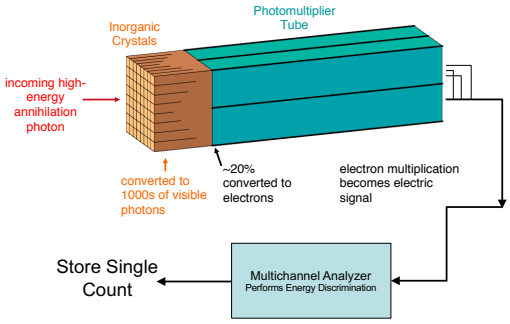
Ex. What is the probability a photon will travel on a straight line through 10 cm of adipose tissue?

$$P = e^{-\mu x} = e^{(-0.087/cm)(10cm)} = 0.41$$

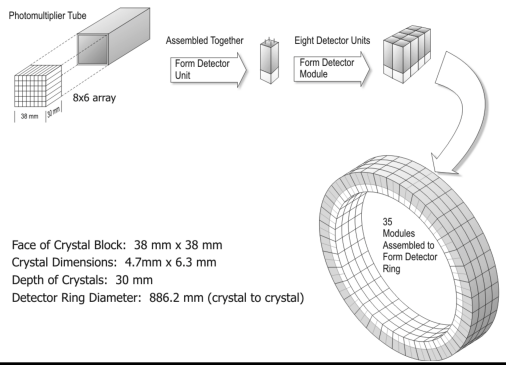
For 511keV photons
Linear Attenuation Coefficients (1/cm):
Body (muscle) = 0.0932
Adipose (fat) = 0.0868
Lung = 0.0267
Wet Spine Bone = 0.0997
Wet Rib Bone = 0.1246

III. Imaging

4. Photons Enter Detectors

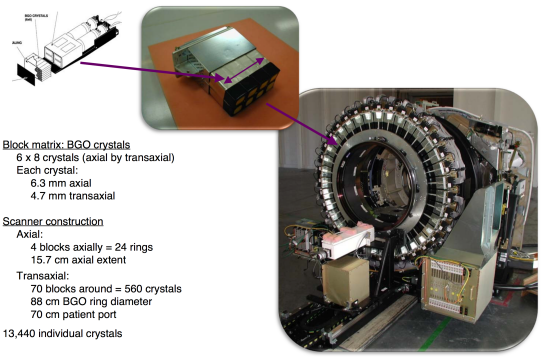


Typical PET Scanner Detector Ring



Face of Crystal Block: 38 mm x 38 mm
Crystal Dimensions: 4.7mm x 6.3 mm
Depth of Crystals: 30 mm
Detector Ring Diameter: 886.2 mm (crystal to crystal)

Inside GE Discovery STE PET/CT



Block matrix: BGO crystals
- 6 x 8 crystals (axial by transaxial)
- Each crystal:
  - 6.3 mm axial
  - 4.7 mm transaxial

Scanner construction
- Axial:
  - 4 blocks axially = 24 rings
  - 15.7 cm axial extent
- Transaxial:
  - 70 blocks around = 560 crystals
  - 88 cm BGO ring diameter
  - 70 cm patient port

13,440 individual crystals

III. Imaging 4. Photons Enter Detectors

a. Photons enter inorganic scintillator - cause scintillation
(One high energy photon makes many low energy photons)
b. Photomultipliers increase light to measurable signal
c. Detection system "counts" photon

Ex. What is the probability a photon will travel through 10 cm of BGO?

Common Detector Materials

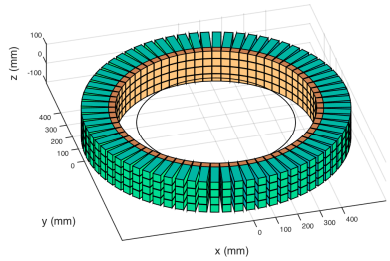
For 511keV photons
Linear Attenuation Coefficients (1/cm):
BGO = 0.955
LSO = 0.833
GSO = 0.674

$$P = e^{-\mu x} = e^{(-0.995/cm)(10cm)} = 7.1 \times 10^{-5}$$

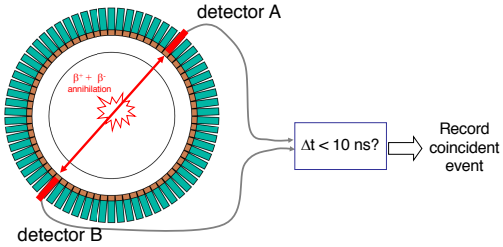
Inorganic Scintillators
(physical characteristics)

	NaI(Tl)	BGO	LSO(Ce)	GSO(Ce)	relevant detector property
Density (gm/cm^3)	3.67	7.13	7.4	6.71	sensitivity
Effective Atomic Number	51	75	66	59	sensitivity
Attenuation Coefficient (@ 511 keV, cm^{-1})	0.34	0.955	0.833	0.674	sensitivity
Light Output (photons/Mev)	40K	~8K	~30K	~20K	energy & spatial resol.
Decay Time	230 ns	300 ns	12 ns 40 ns	60 ns	counting speed
Wavelength	410 nm	480 nm	420 nm	430 nm	photo-sensor matching manufacturing / cost
Index of Refraction	1.85	2.15	1.82	1.85	photo-sensor matching manufacturing / cost
Hygroscopy	yes	no	no	no	photo-sensor matching manufacturing / cost
Rugged	no	yes	yes	no	photo-sensor matching manufacturing / cost

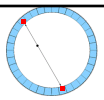
III. Imaging 4. Photons Enter Detectors



III. Imaging 5. Photons paired into coincident events

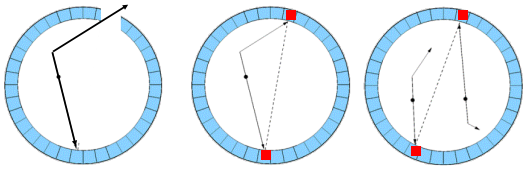


III. Imaging - Some Challenges



True Event

Quantitative errors in measurement



Lost (Attenuated) Event

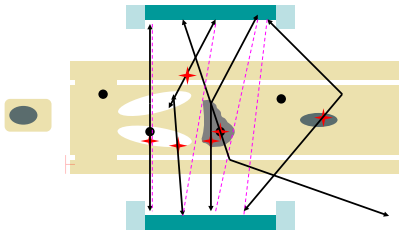
Scattered Coincident Event

Accidental Event (Random)

- Annihilation
- Photon Path
- Detected LOR

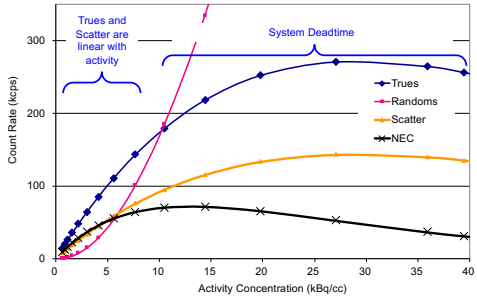
III. Imaging - Some Challenges

Examples of Attenuated, Scatter and Accidental Events



Coronal View

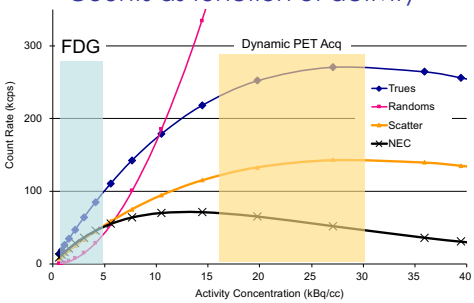
Counts as function of activity



$$NEC = \frac{T^2}{T + R + S}$$

Noise-equivalent count rate provides measure of amount of "good counts" in data

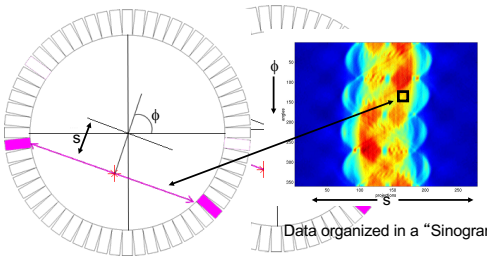
Counts as function of activity



- Randoms limit the maximum usable count rate.
- Lower NEC → Noiser Images → Need to Smooth More → Worse Quantification (both globally and regionally)

III. Imaging

5. Photons paired into coincident events
6. Data organized into sinograms



Data organized in a "Sinogram"

IV. Data Analysis

1. Correct data for errors

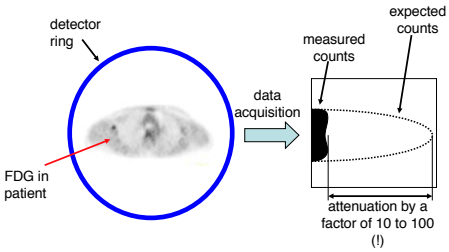
Order of corrections (common application in PET):

Start with Raw Data:

Prompt Events = Trues + Randoms + Scatter

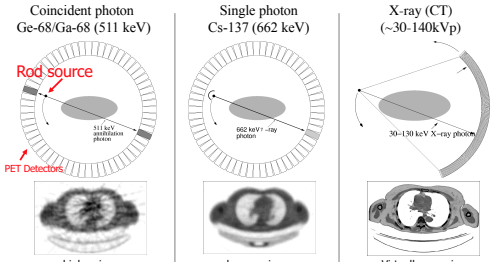
1. Randoms correction (Yr = Prompt-Randoms)
2. Detection efficiency normalization (Yn = Yr * Norm)
3. Deadtime (Yd = Yn * Dead)
4. Scatter (Ys = Yd - Scat)
5. Attenuation (Ya = Ys * ACF) attenuation correction factors
6. **Image Reconstruction**

For PET, the most significant error in data is attenuation



- Attenuation is mainly due to Compton scatter
- It is by far the most important effect for both noise (due to reduced counts) and qualitative image appearance

PET Attenuation Correction-Transmission Scan Methods



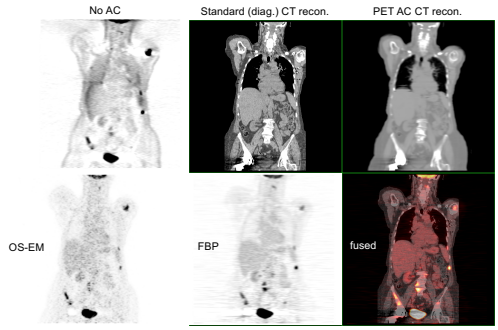
high noise
~12-45 min whole body scan
no bias
post injection contamination

lower noise
~4-20 min scan time
some bias
post injection contamination

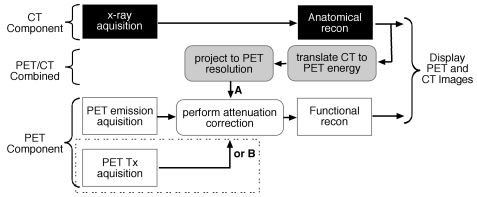
Virtually no noise
~20 sec scan time
potential for significant bias
no post injection contamination

PET Attenuation Correction (CTAC)

PET AC is now performed with a CT image



Flowchart of typical PET/CT operation

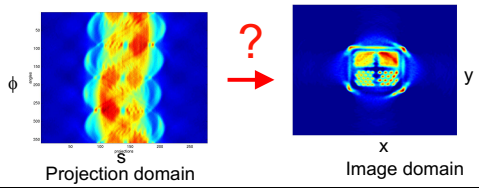


Attenuation correction factors can be obtained from either

(A) CT-based or

(B) conventional PET transmission source (only on original/older systems)

IV. Data Analysis 2. Sinograms reconstructed into images



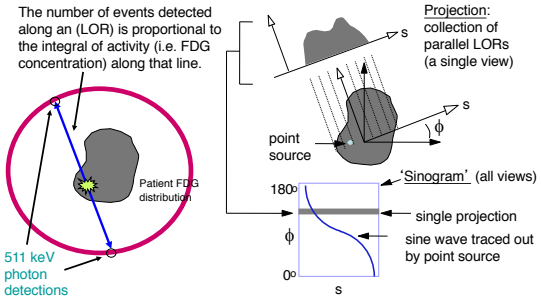
Reconstruction Problem :

All modalities that collect line-integral data (sinograms) can employ the same methods for reconstruction:

1. Analytical Methods - most popular: FBP- Filtered Backprojection
2. Iterative Methods - most popular: OSEM - ordered subset expectation maximization

From photon detection to data in form of **Sinograms**

The number of events detected along an (LOR) is proportional to the integral of activity (i.e. FDG concentration) along that line.



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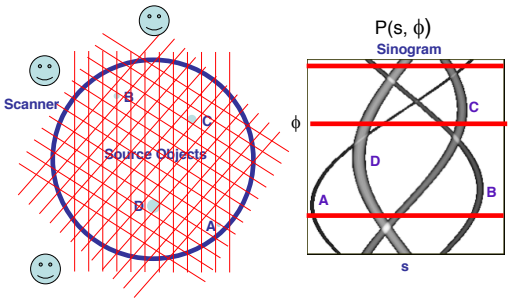
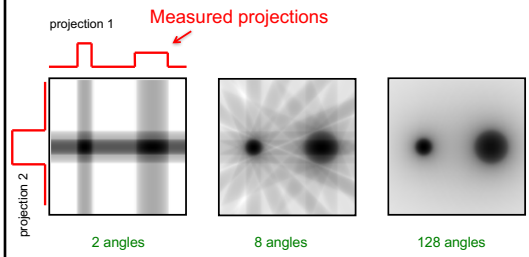
Sinogram Example



- The sinogram is P(s, ϕ) organized as a 2D histogram - Radon Transform of the object

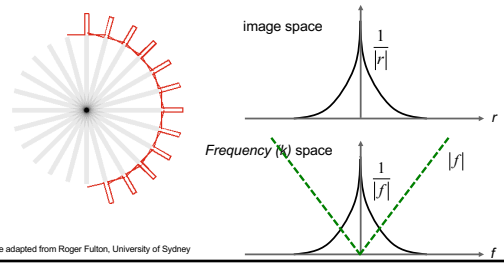
Simple Back Projection



Slide adapted from Roger Fulton, University of Sydney

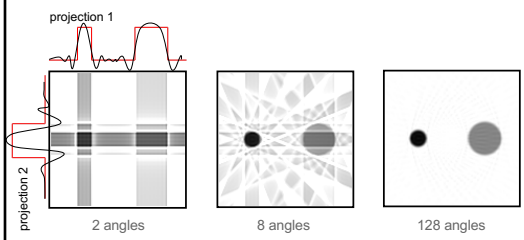
Simple Back Projection

- leads to radial (1/r) blurring in the spatial domain
- this is equivalent to a 1/f function in the frequency domain
- what type of function would we need to multiply 1/f by to result in a perfect point source reconstruction?



Slide adapted from Roger Fulton, University of Sydney

Filtered Back Projection

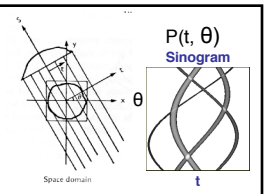


Slide adapted from Roger Fulton, University of Sydney

Why Ramp Filter?

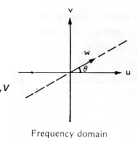
Projections:

$$P_\theta(t) = \int_{(\theta,t)\,line} f(x,y)\,ds$$



Inverse Fourier Transform Image:

$$f(x,y) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} F(u,v)e^{j2\pi(ux+vy)}\,du\,dv$$



We didn't sample *F(u,v)* in cartesian coordianate *u,v*
We sampled in polar coordinates *w,theta:*

$$u = w\cos\theta$$
$$v = w\sin\theta$$
$$du\,dv = w\,dw\,d\theta$$

Origin of Ramp Component

$$f(x,y) = \int_0^{2\pi}\int_0^{\infty} F(w,\theta)e^{j2\pi w(x\cos\theta + y\sin\theta)}\,w\,dw\,d\theta$$

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Why Ramp Filter?

$$f(x,y) = \int_0^{2\pi}\int_0^{\infty} F(w,\theta)e^{j2\pi w(x\cos\theta + y\sin\theta)}\,w\,dw\,d\theta$$

Use Symmetry Properties: $F(w,\theta+180) = F(-w,\theta)$

Substitution: $t = x\cos\theta + y\sin\theta$

FBP Algorithm: $f(x,y) = \int_0^{\pi}\left[\int_{-\infty}^{\infty} F(w,\theta)|w|e^{j2\pi wt}\,dw\right]d\theta$

1. Take Fourier Transform of a projection at θ
2. Weight with "ramp" w
3. Take Inverse Fourier Transform at θ
4. Add up contributions from all θ

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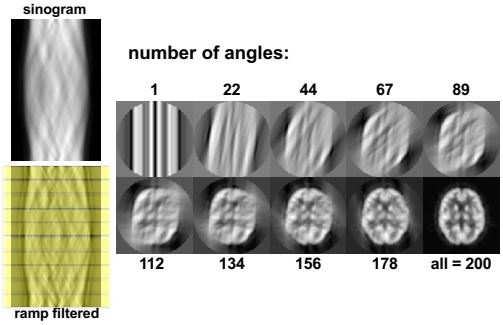
Filtered Back Projection



Slide from Roger Fulton, University of Sydney

Filtered Back Projection

CT Example

Filtered Back Projection

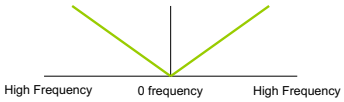
BP | FBP without noise | FBP with noise

Works well when there is no noise, but real projections contain noise...

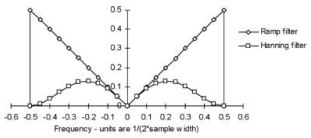
Slide adapted from Roger Fulton, University of Sydney

Analytical Method: Filtered back-projection

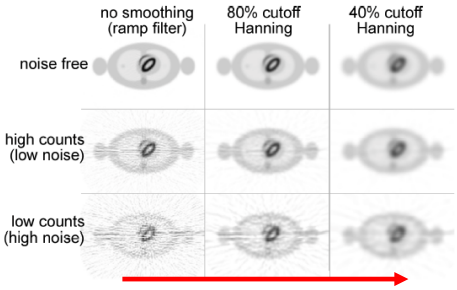
Ramp filter accentuates high frequency - Not good for noise



Either clip ramp filter or often use filters to roll-off ramp to reduce noise: ex: Hanning filter



FBP Examples



Reduce Variance & Increase Bias

(Reduce noise & Increase quantitative error)

The Reconstruction Problem: An Inverse Problem

$$y = \mathbf{P}x + n$$

Unknown image

Observed PET data

system matrix

Error in observations

- x is N x 1 image (PET typically N ~ 128 x 128)
- y is M x 1 data (PET typically M ~ 280 x 336)
- P is M x N system matrix (provides probability entry j from x will be placed in entry i of y)
- If we could just invert P, we could solve this. But P is 16,000 x 95,000 entries…Need to solve iteratively.

Iterative Reconstruction: Basic Components

1. Description of the form of the image (voxels, blobs…)
2. System model relating unknown image to each detector measurement : relates image to data (Can include detector response, corrections for attenuation, efficiencies, etc…)
3. Statistical Model describing how each measurement behaves around its mean (Poisson, Gaussian,…)
4. Objective Function defining the "best" image estimate (Log-likelihood, WLS, MAP…)
5. Method for maximizing the objective function (OSEM)

Primary unique component to each vendor

Main Point: Not all "OSEM" algorithms the same

Not all Vendors Recon algorithms are the same…

EM for NM Reconstruction in a Nutshell

Expectation Maximization is a general method for solving all kinds of statistical estimation problems, in tomography results in easy maximization step

Back project

$$\text{EM}\left\{ \text{Image}_{NEW} = \text{Image}_{OLD} \times \text{BP}\left[\frac{\text{Measured\_Data}}{\text{FP}(\text{Image}_{OLD})}\right] \right.$$

Forward project

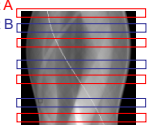
Tomography EM: Method for maximizing the Poisson Likelihood Function (ML-EM)

STEPS:
1. Forward Project (FP) Image_{OLD}
2. Take ratio with measured data
3. Back Project (BP) ratio
4. Multiply by Image_{OLD} to get Image_{NEW}
5. Repeat

Common Methods

1. **EM** (ML-EM - maximum likelihood method)
2. **OSEM** (Ordered Subset Expectation Maximization)
   - Variant of EM (still Maximum Likelihood) - Pro: Fast | Con: Does not converge
   - Uses subsets of the data to compute each estimate…Subset A then B then …then repeat

Subset A
Subset B



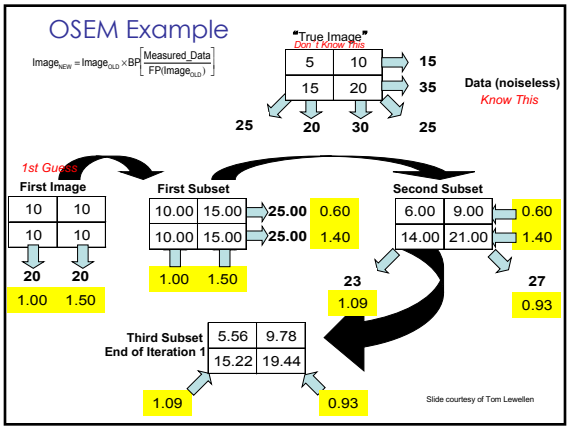
$$\text{Image}_{NEW} = \text{Image}_{OLD} \times \text{BP}\left[\frac{\text{Measured\_Data}_A}{\text{FP}_A(\text{Image}_{OLD})}\right]$$

Then…

$$\text{Image}_{NEW} = \text{Image}_{OLD} \times \text{BP}\left[\frac{\text{Measured\_Data}_B}{\text{FP}_B(\text{Image}_{OLD})}\right]$$

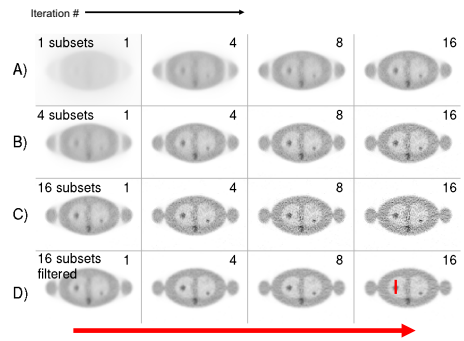
OSEM Example

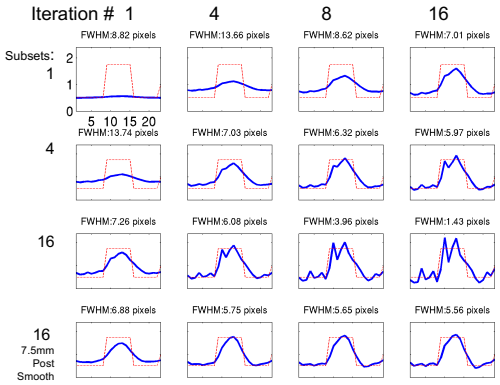
$$\text{Image}_{NEW} = \text{Image}_{OLD} \times \text{BP}\left[\frac{\text{Measured\_Data}}{\text{FP}(\text{Image}_{OLD})}\right]$$



OSEM "Convergence"

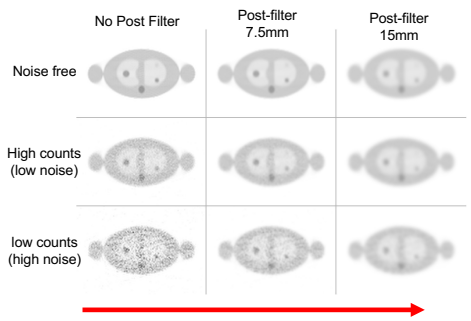
(Implicit Smoothing-Stop algorithm before convergence)





OSEM Examples

(Explicit Smoothing-Post Smooth) 12 subsets, 7 iterations



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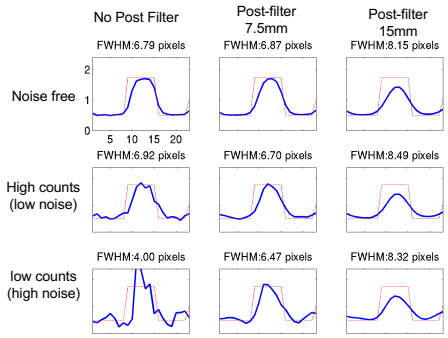
OSEM Examples

(Explicit Smoothing) 12 subsets, 7 iterations



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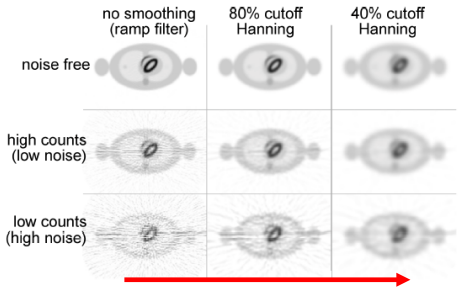
FBP Examples



Reduce Variance & Increase Bias

(Reduce noise & Increase quantitative error)

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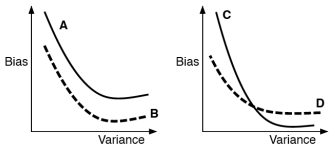
Each method can tradeoff Bias and Variance

FBP - can vary cutoff on filter which modifies the ramp filter

OSEM - Can vary number of iterations (stop early) or vary amount of post-recon smoothing



Hypothetical Tradeoff Curves-Which is better:
method A or B?
method C or D?

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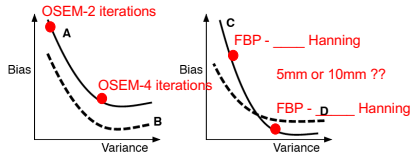
Each method can tradeoff Bias and Variance

FBP - can vary cutoff on filter which modifies the ramp filter

OSEM - Can vary number of iterations (stop early) or vary amount of post-recon smoothing



Hypothetical Tradeoff Curves

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