Lecture 3: X-ray Computed Tomography

- Based on chapter 5 of Suetens, but not in order.
- Homework for next week:
  - Read chapter 8 of Suetens up to and including section 8.4
  - Find 2 medical images of abnormal anatomy or physiology (pathology) formed using Nuclear Medicine (planar gamma camera images, not PET or SPECT). Place these images in a document. Write 1-2 brief sentences describing each image. Write 1-2 brief sentences describing differences between the images. Write 1-2 sentence what the image values represent physically.
- **Discussion of class project**
  - Form groups of 2-3 by next week. I will assign groups if you want
  - Investigate a specific problem involving medical imaging and present your conclusions in a written report and 15 minute oral presentation (on Dec 6th). The investigation should have the following three components: 1) it should target a specific organ, disease, and/or other condition, 2) it should specify one or more of the imaging modalities discussed in class, and 3) it should define the objective behind the use of medical imaging.

Class Project

Example of each of the three components are:

Organ / Disease	Modality	Objective
Brain tumor	x-ray	Detection / diagnosis
Lung cancer, breast cancer, etc.	CT	Progression
Coronary artery disease	Nuclear medicine	Registration
Stroke	Ultrasound	Image guided surgery
Joint injuries	MRI	Segmentation

Your report should address these questions:

1. What is the problem being investigated and why is it important?
2. Why is the chosen modality or modalities the best choice to address the problem? This should be an argument based on the technical benefits (e.g. resolution, SNR, speed, etc.) of your choice as compared to other options.
3. How is the problem currently addressed using this modality? What image processing is required? Cite appropriate references from the literature.
4. What could be done to better address this problem in the future? What about the peripheral or support equipment? That is, if you were asked to improve the methodology, what avenues would you pursue first?

Class Project

Example Projects:

1. Brain tumor / MRI / Progression – The goal is to measure changes in tumor size (progression) over time. This can be used to assess the response of the tumor to treatment. MRI provides good soft tissue definition necessary to identify tumor boundaries

2. Liver disease / CT and ultrasound / Registration – The goal is to align the CT and ultrasound images (registration) so they can be displayed in a combined image. This can be useful in minimally invasive surgery where the ultrasound is used in real time to guide the surgeon and the CT provides high definition images of the anatomy

Deadlines:

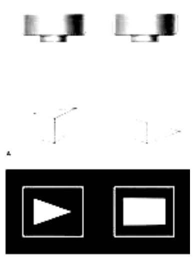
Nov 1:	Outline due	30% of mark for project
Nov 29:	Final report due	50% of mark for project
Dec 6:	Class presentation	20% of mark for project

Projection Imaging Versus Tomography

The need for more than one projection:

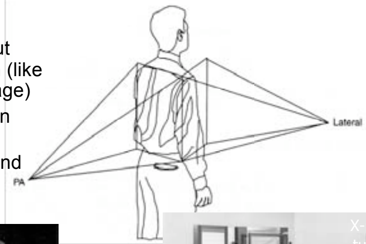
- Depending on the view angle, the same object looks different
- A corollary is that two different objects can look the same in one, or a few views (examples?)

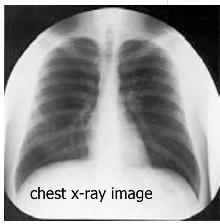
- So, how many views do we need to uniquely identify an object?
- And, what can we do with all the information?



Basic Principles - Conventional Radiograph

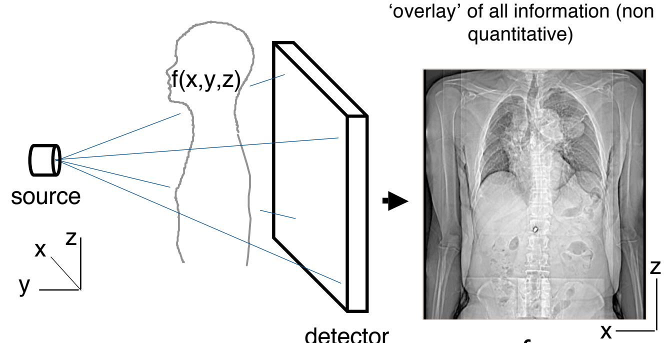
- Forms a 'collapsed' projection view without any depth information (like a gamma camera image)
- Example of "Projection Imaging"
- Here two views (PA and RL lateral) are shown





Projection Imaging

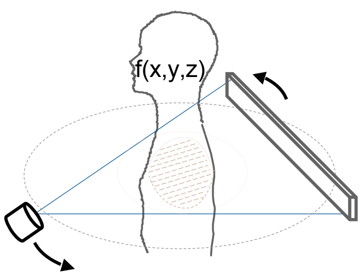
'overlay' of all information (non quantitative)



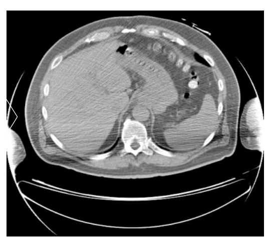
$p(x,z) = \int f(x,y,z)\, dy$

Tomographic Imaging

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

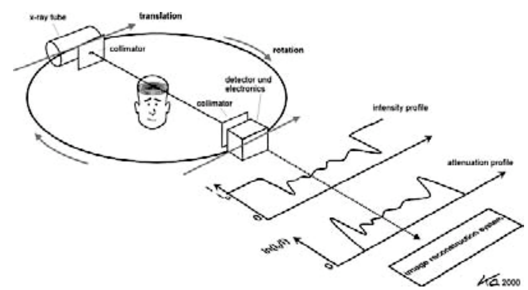


orbiting source + detector data for all angles



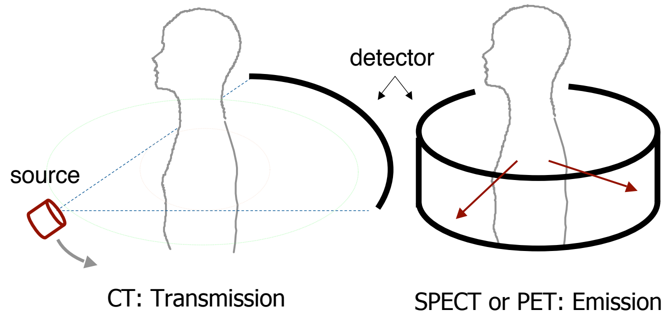
true cross-sectional image

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

Two Types of Tomography: Transmission and Emission



nuclear Magnetic resonance imaging (MRI or MR) and ultrasound (US) are somewhere in between in that they use emission stimulated by an external source

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Major Medical Imaging Modalities

Modality	Resolution	TX or EM	Mode
X-ray	0.2 – 1.0 mm	TX	Planar
Nuclear Medicine	10 – 20 mm	EM	Planar
X-ray CT	1 mm	TX	Tomographic
Ultrasound	3 mm	TX/EM (sound)	Tomographic
MRI	1 mm	TX/EM (RF)	Tomographic
PET/SPECT	5 - 10 mm	EM	Tomographic

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X-ray Imaging

- X-ray imaging has been used for medical imaging for over a century, and plane-film x-ray scanning is the workhorse of radiology departments. With radiographs, however, it is difficult to see low-contrast objects.
- Computed Tomography (CT) was developed in the early 1970s at EMI by Godfrey Hounsfield to generate cross-sectional (tomographic) images of a patient. At that time EMI had large profits as the Beatles were recording under their 'Parlorphone' label.
- Terms such as “computerized transverse axial tomography”, “computed-assisted tomography” or “computerized axial tomography” (CAT) have been used. The term “computed tomography” (CT) was standardized by the Radiological Society of North America

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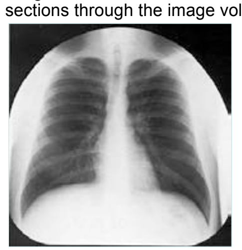
Comparing Projection to Tomographic Images

- Hounsfield's insight was that by imaging all the way around a patient we should have enough information to form a cross-sectional image
- Radiographs typically have higher resolution but much lower contrast and no depth information
- 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



Chest radiograph

Coronal section of a 3D CT image volume

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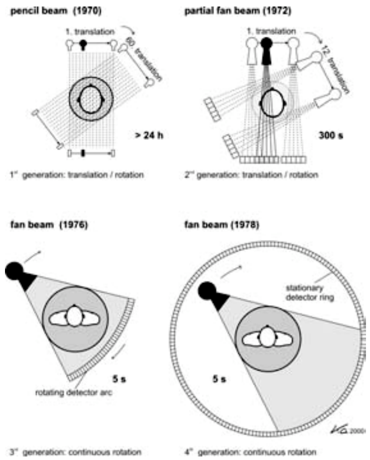
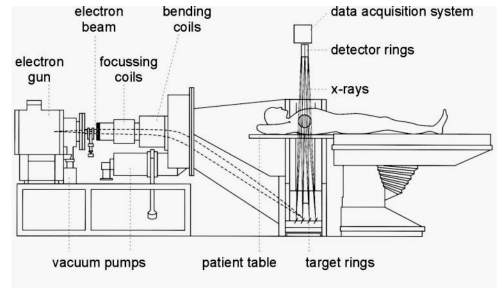
Historical Development of CT



Historical Development of CT

- So-called 'Fifth' Generation CT: Stationary/Stationary (1990's)
- AKA Electron beam scanner
- Primarily for cardiologists, 50 msec scan times
- Uses tungsten target and high-energy electron beam
- Now largely obsolete!

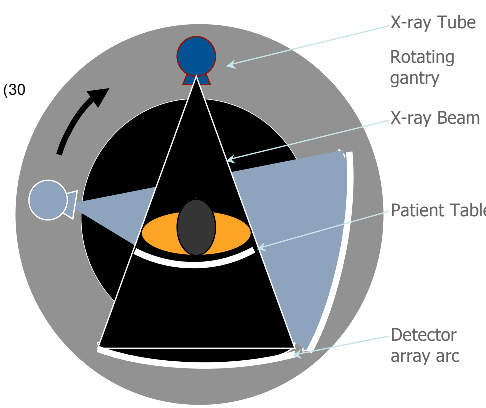


CT Developments

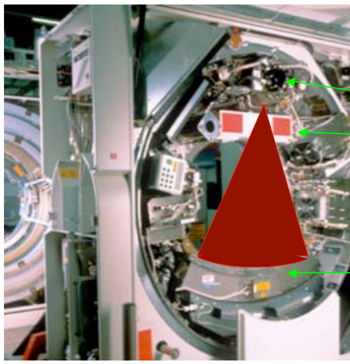
1972 Invention of CT
1978 Head scan takes 30 min
1986 Slip-ring technology, 1 second scan
1989 Helical CT
1998 Multi-detector CT, 1/2 second scan
2000 57 million CT examinations done in 7645 facilities
2001 Commercial PET/CT
2002 4, 8, and 16 slice CT
2003 32 slice CT
2003 Head scan takes 3 seconds
2004 64 slice CT, 0.3 second scan
2006 Dual tube CT scanners

How it works: CT Scan Concept

Third Generation CT
- Rotating fan beam
- 0.3 to 2 seconds to acquire an image (30 rpm to 200 rpm!)
- Workhorse for CT scanners



CT Tube and Detectors



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CT Scanner in Operation

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

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X-ray CT Scanning

Stationary anode tube (low power version suitable for radiographs)

focal spots

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

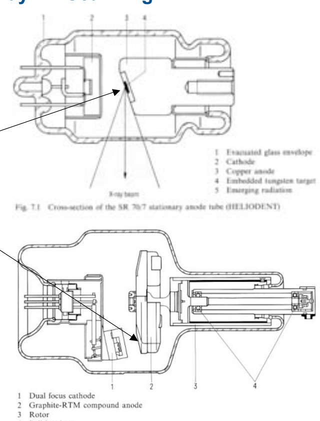

Fig. 7.1 Cross-section of the SR 70/7 stationary anode tube (HELIODENT)

Fig. 7.17 Cross-section through a rotating anode x-ray tube

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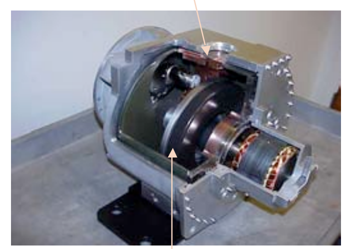
Modern X-Ray Tube

Electron Collector: reduce off-focal radiation
- Lower patient dose



High Peak-Power Target & Bearings
- High peak-mA for fast rotation

Rotation speed (s)	typical mAs	mA needed	
0.5	200	400	
0.4	200	500	
0.4	240	600	Large Patient
0.35	200	571	
0.35	240	686	Large Patient

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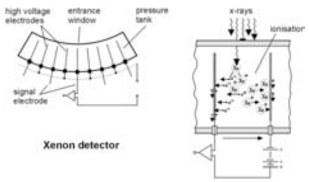
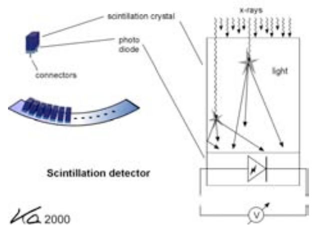
Detectors Used in CT Scanners

- Older generation used Xenon gas detectors, i.e. fast ionization detectors
- More recently solid state scintillators, such as CsI are used, which allow for construction of 'multirow' detector systems

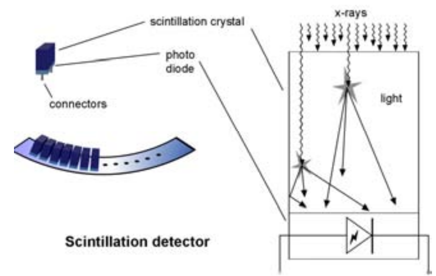




Ka 2000

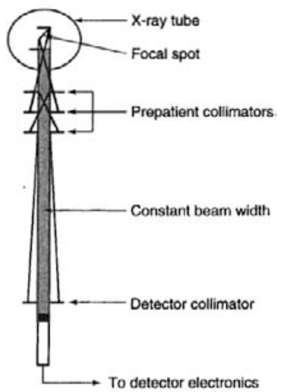
X-ray CT Physics

- The detectors are basically similar to methods used in nuclear imaging systems: scintillation followed by light collection
- The scintillator (e.g. CsI) converts the high-energy photon to a light pulse, which is detected by photo diodes



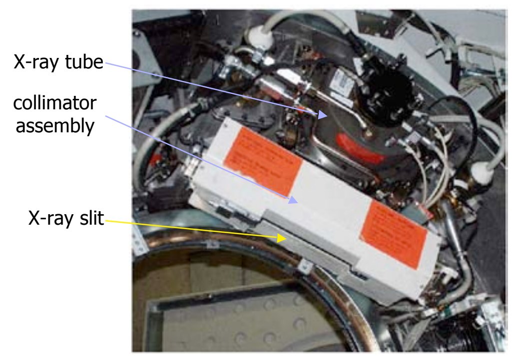
Collimation

- Collimators are used to protect the patient by restricting the x-ray beam to the anatomy of interest
- Prepatient collimation is influenced by the focal spot size because of penumbra
- The larger the focal spot size, the greater the penumbra and more complicated collimator design
- Detector or postpatient collimation shapes the x-ray beam and removes scattered radiation
- Collimation also helps define slice thickness, 0.5 mm to 10 mm depending on scanner



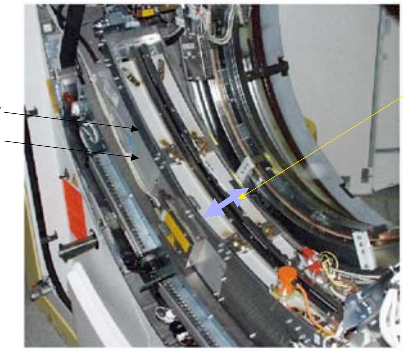
Pre-Patient Collimation

- Controls patient radiation exposure



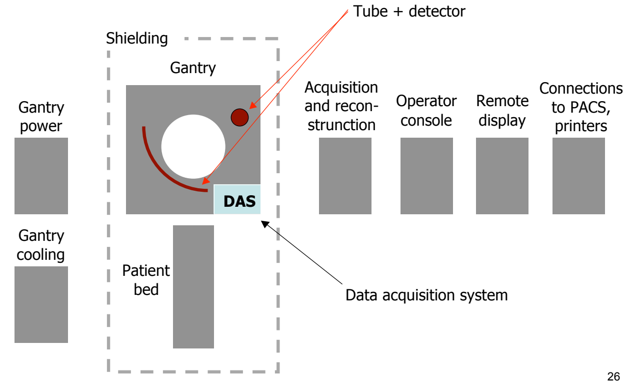
Post-Patient Collimation

- Adjusts image slice thickness



Major Component of a CT Scanner System

- Typical configuration



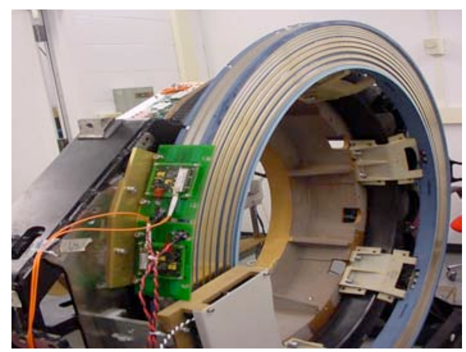
Patient Couch

Couches are typically capable of supporting 400-450 pounds with an table positioning accuracy/reproducibility of ±0.25 mm

Horizontal movement range is typically 170 to 200 cm

Maximum horizontal movement speeds of 100 to 150 mm/s

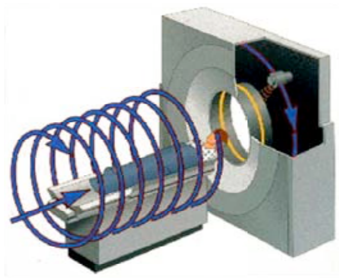
Gantry Slip Rings



Allows for continuous rotation

Helical CT Scanning

- Patient is transported continuously through gantry while data are acquired continuously during several 360-deg rotations
- The ability to rapidly cover a large volume in a single-breath hold eliminates respiratory misregistration and reduces the volume of intravenous contrast required



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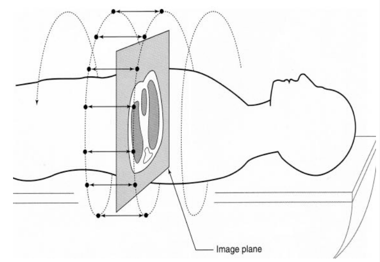
Helical Scanning and Image Interpolation

- The helical data set is interpolated into a series of planar image data sets
- Production of additional overlapping images with no additional dose to the patient
- Images can be reconstructed at any level and in any increment but must have a thickness equal to the collimation used



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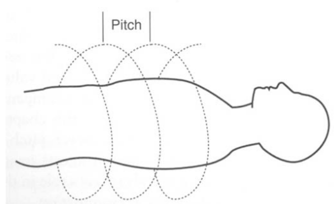
Pitch



$$\text{Collimator Pitch} = \frac{\text{distance that the table travels per } 360^{\circ} \text{rotation}}{\text{slice thickness}}$$

- A pitch of 1.0 implies axial scanning
  - ✓ best image quality in helical CT scanning
- A pitch of less than 1.0 involves overscanning
  - ✓ some slight improvement in image quality, but higher radiation dose to the patient
- A pitch greater than 1.0 is not sampling enough to avoid artifacts
  - ✓ Faster scan time often more than compensates for undersampling artifacts)
  - ✓ Also reduction in patient radiation dose

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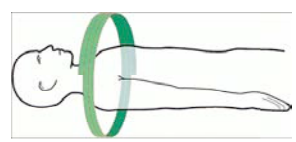
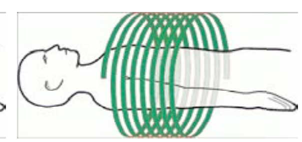
Increasing Pitch to Reduce Scan Duration

- Faster acquisition mode -- same region of body scanned in fewer rotations, thus less motion effects



Pitch = 1



Pitch = 2

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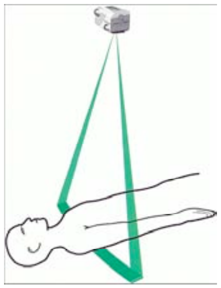
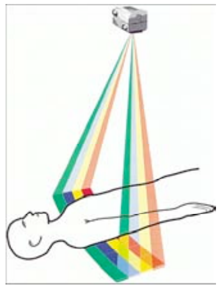
Single versus Multislice Detectors

Can image multiple planes at once



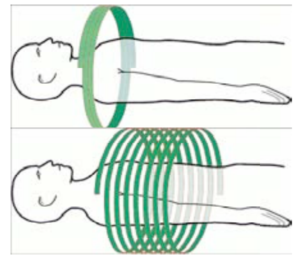
1 detector row



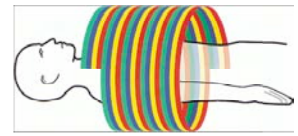
4 detector rows

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

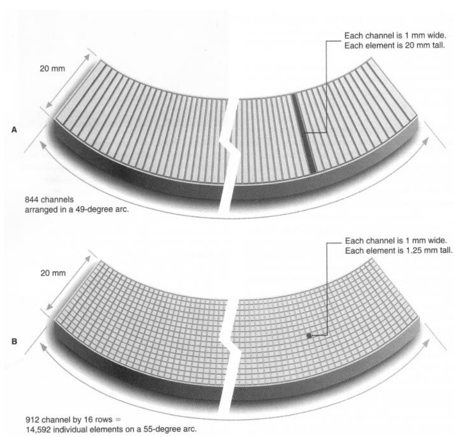


1 detector row: pitch 1 and 2



4 detector rows: pitch 1

CT Detectors: Single Slice vs. Multislice CT

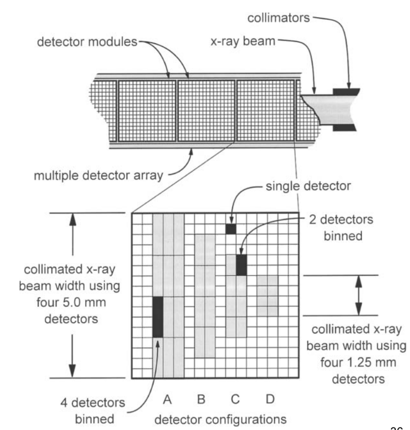


Single-slice CT detectors are based on a single-row detector array

Multislice CT uses multiple-row detector array, with each row consisting of individual elements

Multi-row Detector Arrays

- Four channel system
- Although there are 16 detectors in axial direction (in this example) only four groups can be read out at a time
- This is called a '4-slice' CT scanner
- Detector collimators (i.e. post-patient) is adjusted to match beam width



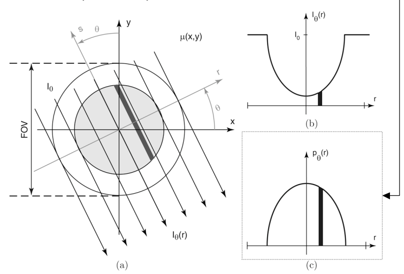
X-ray Computed Tomography

- For an ideal narrow beam of monoenergetic photons, the fractional reduction of the beam intensity *I* is given by $-dI\ /\ I = \mu\, dt$
- This can be integrated to give $I(t) = I_0 \exp\left(-\int_0^t \mu\, dt'\right)$
- The solution is Lambert-Beer law $I(t) = I_0 \exp(-\mu t)$

X-ray Computed Tomography

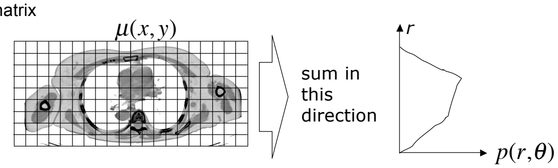
- For spatially varying attenuation coefficients $\mu(x,y)$ (which is what we really want --right?) we can convert it to a simple integral with reference to the initial (unattenuated) beam intensity I_o

$$-\ln\left(\frac{I(r,\theta)}{I_0}\right) = \int_{-\infty}^{\infty} \mu(x,y)\, ds \triangleq p(r,\theta)$$



X-ray Computed Tomography

- The function p(r,θ) is formally called the <u>projection</u> of μ(x,y) along the direction θ. This can is more easily visualized as a row or column sum of a matrix

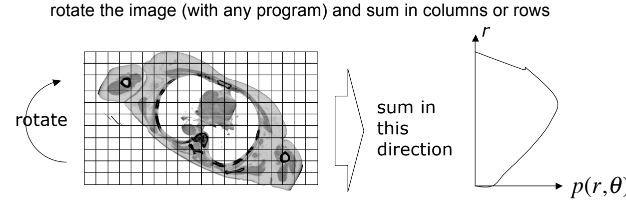


We need to have p(r,θ) for all (r,θ) to determine the original object μ(x,y).

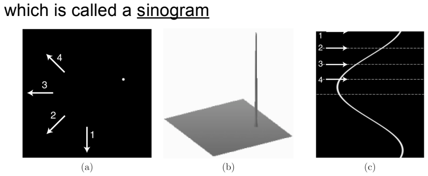
Estimating μ(x,y) from p(r,θ) knowing the relation $p(r,\theta) = \int_{-\infty}^{\infty} \mu(x,y)\, ds$ is a classic <u>inverse problem</u>

X-ray Computed Tomography

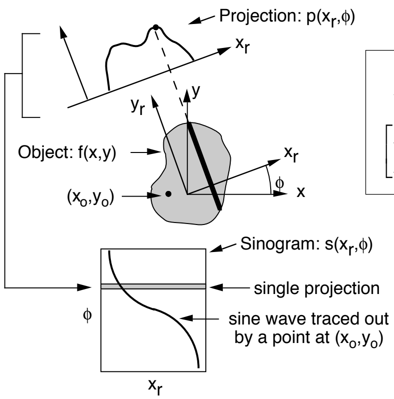
- Actually the easiest way to calculate projections at other angles is to rotate the image (with any program) and sum in columns or rows



- We can group all the data for p(r,θ) in a 2D array to make a sort of image, which is called a <u>sinogram</u>



One-dimensional projections



$$p(x_R,\phi) = \int_{-\infty}^{\infty} dy_R\, f(x,y)$$

$$\begin{bmatrix} x_R \\ x_R \end{bmatrix} = \begin{bmatrix} \cos\phi & -\sin\phi \\ \sin\phi & \cos\phi \end{bmatrix} \begin{bmatrix} x \\ y \end{bmatrix}$$

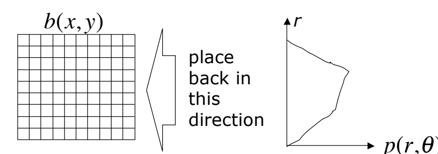
X-ray Computed Tomography

Given the sinogram p(r,θ), how do we get back to μ(x,y)?

The details are beyond the scope of our course, but are described in section 5.3.2

The core concept is <u>backprojection</u>, which is:

placing values back into the image matrix

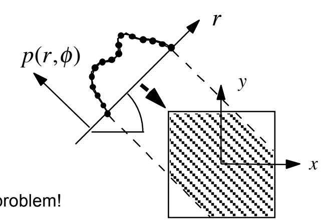


Then repeat for all angles θ.

X-ray Computed Tomography

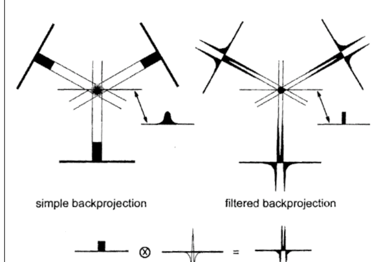
We can write this as

$$b(x,y) = \int_0^{2\pi} d\theta\, p(r,\theta)$$



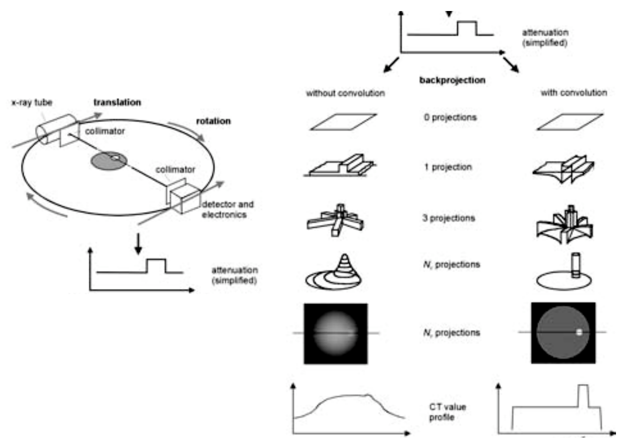
But there is a problem!

Formation of a CT image – Filtered back projection



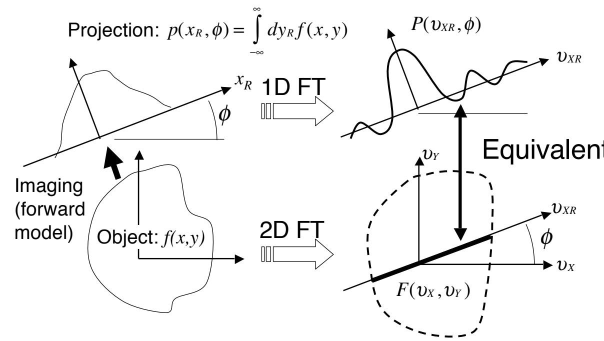
- Simple backprojection produces an image that is blurred
- Raw data must first be filtered using a mathematical filter, or convolution kernel

Formation of a CT image – Filtered back projection



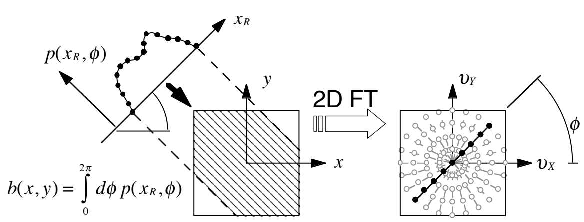
2D Central Section Theorem

- Probably the simplest way to understand the necessary convolution 'filter' and 2D image reconstruction -- compressed into 3 slides
- AKA the projection-slice theorem



2D Backprojection

- The result of backprojecting a single projection is equivalent to 'placing' the the Fourier transformed values into the reconstruction matrix, which represents magnitude and phase of the the spatial frequencies of the image.

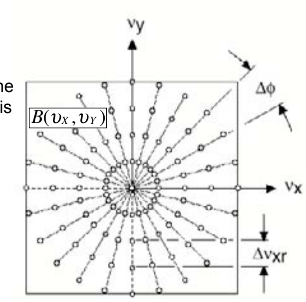


$$B(\upsilon_X,\upsilon_Y) = F_{2D}\{b(x,y)\}$$

$$= \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} dx\, dy \exp(-2\pi i(x\upsilon_X + y\upsilon_Y))\, b(x,y)$$

2D Backprojection and Filtering

- The 2D Fourier transform is built up from the 1D FTs of the projections, in the limiting case as $\Delta\nu_{XR}$ and ϕ ->0, the sampling density in frequency space is proportional to $1/|\nu|$.
- Thus $B(\upsilon_X,\upsilon_Y) = F(\upsilon_X,\upsilon_Y)/|\upsilon|$
- so $F(\upsilon_X,\upsilon_Y) = |\upsilon| B(\upsilon_X,\upsilon_Y)$

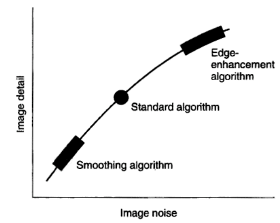


- We normalize the unevenly sampled Fourier transform of the object with the 'ramp' filter $|\nu|$
- This sampling is the same for PET, SPECT, CT (not MR)
- If we do not use this normalization, we get heavily blurred images

Reconstruction Demo 1

Filtered Backprojection

Image Noise vs. Resolution



- Bone filter - fine detail, increased noise
- Soft tissue filters - smoothing, decreases image noise and spatial resolution
- The choice of the best filter to use with the reconstruction algorithm depends on the clinical task

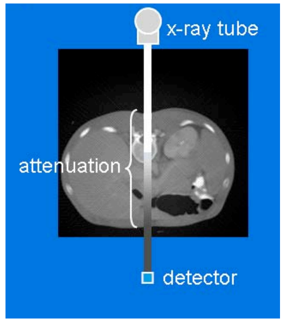
Image Noise vs. Resolution

Reconstruction Demo 2

Noise versus Resolution

What Is Being Measured?

During acquisition, each detector element is related to the average linear attenuation coefficient of the tissue contained in each voxel along the line of response (LOR) from tube to detector



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High-Energy Photons and Interactions with Matter

The names for high-energy photons are determined by the source, not the energy

- X-rays come from bremsstrahlung: electrons are accelerated from cathode to an anode by a voltage (Vp) and bend around orbital electrons in the anode target. The bending is an acceleration that releases energy (mostly heat)
- γ-rays come from nuclear decay processes that release energy
- Annihilation photons come from the mutual annihilation of electrons and positrons. Their mass is converted to energy according to $E = mc^2$. (often called γ-rays by mistake)

Interactions in the energy range of 30-1000 keV are

- Photoelectric absorption: Photon energy absorbed by electron. Dominates at lower energies or for materials with high Z values
- Compton scatter: photon scatters off a "free" electron and changes direction and loses energy. Dominates at higher energies or for materials with low Z values (biological materials)

Charged particles interact in less than a mm, photons take many cm

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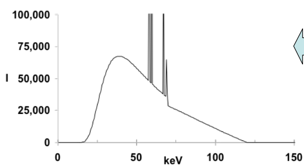
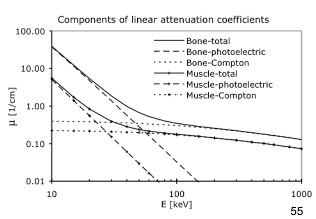
X-ray CT Image Values

- With CT attempt to determine $\mu(x,y)$, but due to the bremsstrahlung spectrum we have a complicated weighting of $\mu(x,y)$ at different energies, which will change with scanner and patient thickness due to differential absorption.



Input x-ray bremsstrahlung spectrum (intensity vs. photon energy) for a commercial x-ray CT tube set to 120 kVp

Energy dependent linear attenuation coefficients ($\mu(x,y)$) for bone and muscle



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CT Numbers or Hounsfield Units

- We can't solve the real inverse problem since we have a mix of densities of materials, each with different Compton and photoelectric attenuation factors at different energies, and a weighted energy spectrum
- The best we can do is to use an *ad hoc* image scaling
- The CT number for each pixel, (x,y) of the image is set to:

$$CT(x,y) = 1000\left[\frac{\mu(x,y)-\mu_{water}}{\mu_{water}}\right]$$

- $\mu(x, y)$ is the attenuation coefficient for the voxel, μ_{water} is the attenuation coefficient of water and CT (x,y) is the CT number (or Hounsfield unit) that comprises the final clinical CT image

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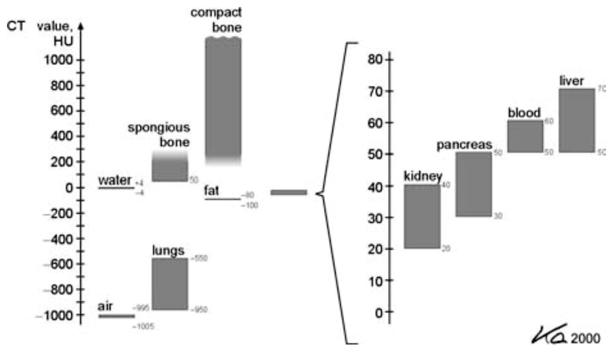
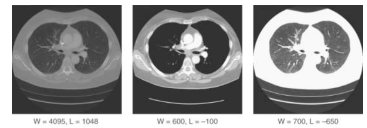
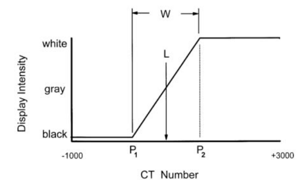
CT Numbers or Hounsfield Units

Typical values



Digital Image Display: Window/Level





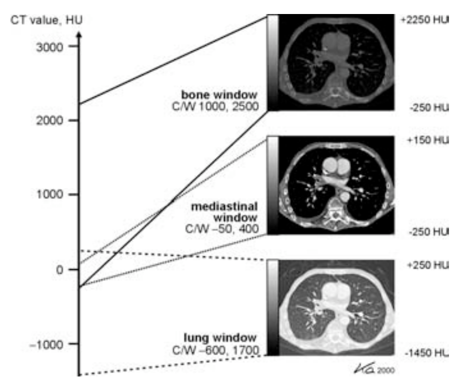
- The window width (W) determines the contrast of the image, with narrower windows resulting in greater contrast
- The level (L) is the CT number at the center of the window

$P_1 = L - ½ W$

$P_2 = L + ½ W$

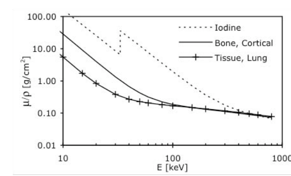
Digital Image Display: Window/Level

- The dynamic range of X-ray images is broader than can be displayed or perceived, thus we use the window/level settings to control what we see



Contrast Agents

- Iodine- and barium-based contrast agents (very high Z) can be used to enhance small blood vessels and to show breakdowns in the vasculature
- Enhances contrast mechanisms in CT
- Typically iodine is injected for blood flow and barium swallowed for GI, air is now used in lower colon



CT scan without contrast

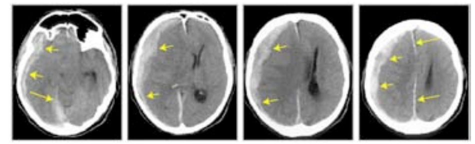
CT scan with contrast

CT Applications

A wide range of abnormalities or diseases in any part of the body
Calcium Scoring
Radiation treatment planning
Cancer
Trauma
Infection/inflamation
Follow-up of conventional chest X-ray findings
Pneumonia
Tuberculosis
Emphysema
Angiography
Stroke
Sinusitis
Bone fracture
Spinal column damage

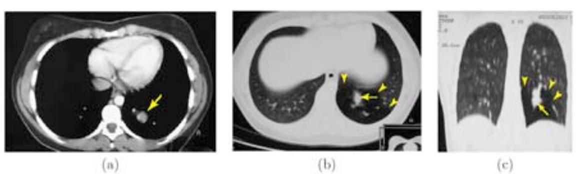
X-ray CT Scanning: Clinical Uses

- The advantage of CT compared to projection radiography is improved contrast and detail for complex structures, so it is used in cases where this detailed information is important enough to outweigh the cost.
- It can also be used in all parts of the body (unlike US or MRI), and is excellent at detecting bone fractures to to the density range for imaging
- With contrast agents can be used for finding regions of abnormal blood flow (e.g. cancer in the liver)



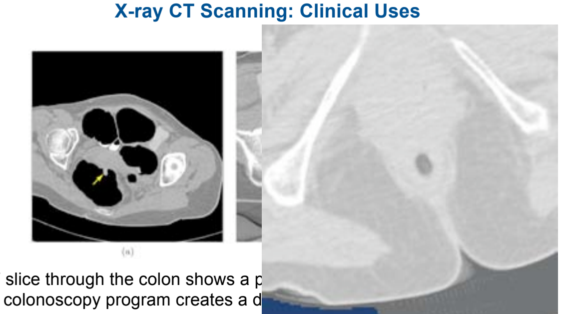
CT slices through the brain show a subdural hemorrhage as a hyperdense region blood more dense than brain tissue) along the inner skull wall

X-ray CT Scanning: Clinical Uses



- CT of the chest. (a) Mediastinal and (b) lung window/level settings, and (c) coronal resliced image
- The images show a congenital malformation of the lung located in the left lower lobe. Notice the two components of the lesion: a dense multilobular opacity (arrow) surrounded by an area of decreased lung attenuation (arrow heads)

X-ray CT Scanning: Clinical Uses



CT slice through the colon shows a p
colonoscopy program creates a d
Air or barium is used as a contrast agent and segmented from the GI tract

'Technique'

- Technique refers to the factors that control image quality and patient radiation dose
- kVp (kV potential) - energy distribution of X-ray photons (recall lower energy photons are absorbed more readily
- mA - number of X-ray photons per second (controlled with tube current)
- s - gantry rotation time in seconds
- mAs - total number of photons (photons per second X seconds)
- pitch
- slice collimation
- filtration - filters placed between tube and patient to adjust energy and/or attenuation (not discussed here)

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Radiation Dose in CT

- CT acquisition requires a high SNR to achieve high contrast resolution and therefore the dose to the slice volume is higher because the techniques used are higher
  - PA Chest x-ray – 120 kVp, 2 mAs
  - Chest CT – 120 kVp, 200 mAs
- Radiation dose is linearly related to mAs
- At same kVp and mAs, number of detected photons increases linearly with slice thickness, SNR improves
- Larger slice thickness at same technique yields better contrast resolution (higher SNR) but spatial resolution in the slice thickness dimension is reduced
- Smaller slice thickness improves spatial resolution in slice thickness dimension and reduces partial volume averaging
- Noise will increase with thinner slices unless mAs is also increased to compensate for loss of x-ray photons from collimation
- In CT, there is a well-established relationship between radiation dose, pixel dimensions Δ, SNR, and slice thickness T:

$$D \propto \frac{SNR^2}{\Delta^3 T}$$

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Signal to Noise Ratio (SNR)

- Signal = mean photons used to produce the image/unit area
- Noise injects a random or stochastic component into an image – many sources
- SNR = signal/noise = increases with increase in the number of photons detected
- Quantum noise is the statistical fluctuation in the number of photons detected
- Quantum noise and structure noise both affect the *conspicuity* of a target

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Radiation Dose
kVp

kVp not only controls the dose but also controls other factors such as image contrast, noise and x-ray beam penetration through patient

Parameter	80 kV	120 kV	140 kV
Image Contrast	***Best***	Intermediate	Poor
Noise	Most	Average	***Least***
Penetration	Least	Average	***Most***
Patient Dose per mAs	***Lowest***	Intermediate	Highest

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Effective Dose Comparison with Chest PA Exam

Procedures	Eff. Dose [mSv]	Equivalent no. of chest x-rays	Approx. period of background radiation
Chest PA	0.02	1	3 days
Pelvis	0.7	35	4 months
Abdomen	1.0	50	6 months
CT Chest	8	400	3.6 years
CT Abdomen or Pelvis	10-20	500	4.5 years

Typical Background Radiation - 3 mSv per year

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Some CT Artifacts

- Physics based
  - beam-hardening
  - partial volume effects
  - photon starvation
  - scatter
  - undersampling
- Scanner based
  - center-of-rotation
  - tube spitting
  - helical interpolation
  - cone-beam reconstruction
- Patient based
  - metallic or dense implants
  - motion
  - truncation

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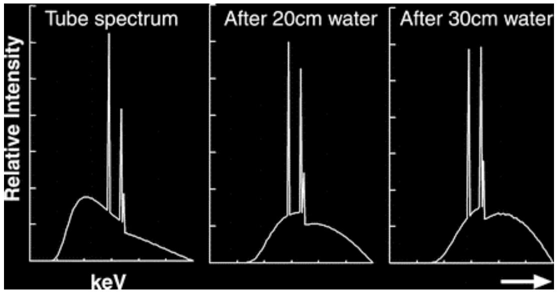
Physics-based CT Artifacts

Beam Hardening

- unique to Xray due to preferential energy absorption of lower energy photons (unlike nuclear imaging)



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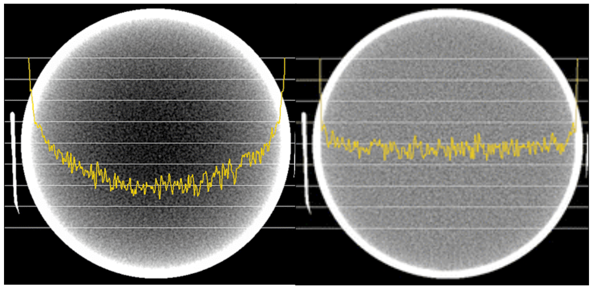
Beam-Hardening

Causes 'dishing' artifact in uniform cylinders
Can be corrected by scanner calibration tables



Uncorrected Corrected

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Photon Starvation

Occurs when insufficient photons reach the detectors
typically across the shoulders
usually corrected by 'adaptive filtration'

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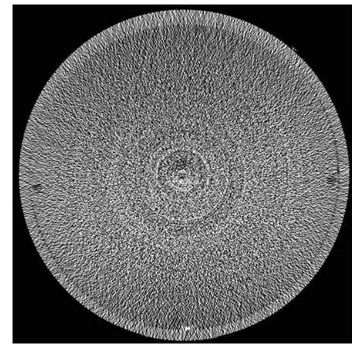
Center-of-Rotation

Similar artifacts occur with deficient detector channels and tube 'spitting'



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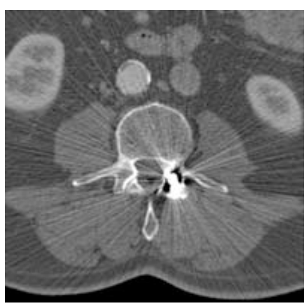
Metallic Objects

Occur because the density of the metal is beyond the normal range that can be handled
Additional artifacts from beam hardening, partial volume, and aliasing are likely to compound the problem



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Patient Motion

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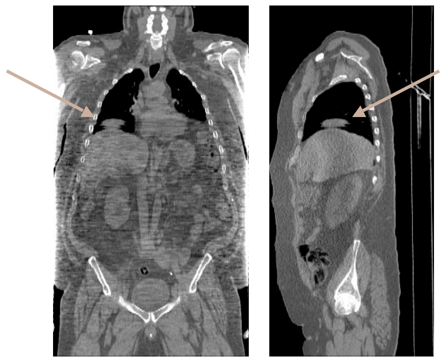
Breathing artifacts in CT

- Respiratory motion during a helical CT scan can lead to artifacts at the dome of the diaphragm (and other areas)



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Truncation

- For accurate topographic image reconstruction, the entire patient must be viewed from every direction
- Unfortunately, the patient port diameter is typically 70 cm in diameter, while the imaging field of view (FOV) is typically 50 cm in diameter

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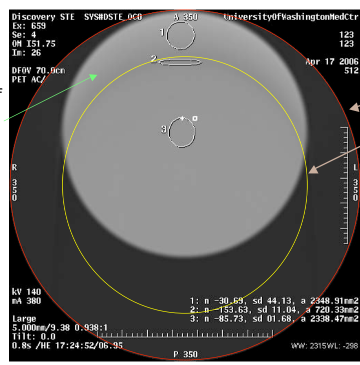
Extended Field of View (EFOV) CT Reconstruction

Extended Field of View (EFOV) CT reconstruction for an offset 50 cm diam plastic phantom



Port (70 cm)

CT FOV (50 cm)

Mean value (m) not constant for regions 1, 2, and 3

Not used for diagnostic CT yet

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Some Challenges in CT

- Patient radiation exposure
- Information overload from modern multi-slice scanners
- The pace of technological progress

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