Questions

• I know the reasons why everything is moving towards digital systems, but based on image quality alone, which is better for these systems, film or digital?

• Not sure how to interpret the left illustration on slide 25. Can you explain?

• Regarding to Voltage determining the X-ray energy Kvp, what is the unit Kvp is equivalent to typical voltage unit?

Email questions to jackie24@uw.edu by Friday April 26
The subject line should be "Phys 428 Lecture 4 Question"
Class Project

• Pick:
  – An imaging modality covered in class
  – A disease or disease and treatment

• Review:
  – what is the biology of the imaging
  – what is the physics of the imaging
  – what are the competing imaging (and non-imaging) methods
  – what is the relative cost effectiveness of your imaging modality for this disease?

• Form groups (or let me know) by Friday April 26

• 1 page outline Friday May 3 (20%)

• Background summary Friday May 10 (15%)
  (what background material you will use & capsule summaries)

• Rough draft Friday May 17 (15%)

• Final version Friday May 31 (30%)

• Presentation / slides Friday June 7 (10%)

• Presentation Tuesday June 11 (10%)
X-ray Computed Tomography
Types of Images: Projection Imaging
Types of Images: Tomography Imaging

tomographic acquisition

reconstruction of multiple images

form image volume

transaxial or axial view

coronal view

sagittal view
Comparing Projection and 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

• Sir Godfrey Hounsfield shared the 1979 Nobel Prize with Allan Cormack (of FBP fame), funded by the EMI and the Beatles

• Radiographs typically have higher resolution but much lower contrast and no depth information (i.e. in CT section below we can see lung structure)
CT Scanner Geometry

source to isocenter distance

source to detector distance

isocenter

datactor arrays

x-ray tube

max FOV
CT Scanner Geometry

gantry rotation
CT Scanner Components

- Data acquisition in CT involves making transmission measurements through the object at angles around the object.
- A typical scanner acquires 1,000 projections with a fan-beam angle of 30 to 60 degrees incident upon 500 to 1000 detectors and does this in <1 second.
CT X-ray Tube

- In a vacuum assembly
- A resistive filament is used to 'boil off' electrons in the cathode with a carefully controlled current (10 to 500 mA)
- Free electrons are accelerated by the high voltage towards the anode
X-ray tubes

• Voltage determines maximum and x-ray energy, so is called the kVp (i.e. kilo-voltage potential), typically 90 kVp to 140 kVp for CT
• High-energy electrons smash into the anode
  – More than 99% energy goes into heat, so anode is rotated for cooling (3000+ RPM)
  – Bremmstrahlung then produces polyenergetic x-ray spectrum
Typical X-ray spectra in CT

scaled to peak fluence
Mass attenuation coefficient versus energy
Pre-Patient Collimation

• Controls patient radiation exposure

X-ray tube

Collimator and filtration assembly

X-ray slit
Need for x-ray beam shaping
Addition of 'bow-tie' filters for beam shaping
Use of 'Bow-tie' beam shaping

periphery dose in patient is reduced
with no loss of image quality
Radiation dose considerations

- No bow tie
- Perfect bow tie
- Small bow tie
Pre-Patient Collimation

- Controls patient radiation exposure

X-ray tube

'fan' of X-rays
X-ray Detector Assembly

detectors
collimators
X-ray CT Detectors

- The detectors are similar to those used in digital flat-panel imaging systems: scintillation followed by light collection
- The scintillator converts the high-energy photon to a light pulse, which is detected by photo diodes
X-ray CT Detectors

Typically composed of rare-earth crystals (e.g. Gd$_2$O$_2$S)

Sintered to increase density
X-ray CT Detectors

Detector module sits on a stack of electronic modules
- pre-amp
- ADC
- voltage supply
Gantry Slip Rings

• Allows for continuous rotation
CT Scanner in Operation

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

- The attenuation depends on material (thus position of material) and energy
- With bremsstrahlung radiation, there is a weighted distribution of energies
- We combine previous results to get the imaging equation

$$I(x) = \int_{E=0}^{E_{\text{max}}} E' S'_0(E') e^{-\int_0^x \mu(x',E') \, dx'} \, dE'$$

Beam intensity along a line with $\mu = \mu(x)$

$S'_0(E)$

$\mu(E)$

$\mu(E)$ graph showing attenuation coefficient with different materials:
- Bone
- Muscle
- Fat
- Soft tissue

$S'_0(E)$ graph showing relative x-ray intensity with different energies:
- Characteristic radiation (tungsten) at 57.984 keV, 59.321 keV, 66.950 keV, 67.244 keV, 69.081 keV
- Bremsstrahlung radiation

$X$-ray photon energy, keV
Imaging Equation

• Similar to x-ray projection systems (ignoring geometric effects etc.) for intensity at a detector location $d$

$$I_d = \int_0^{E_{\text{max}}} S_0(E) E e^{-\int_0^d \mu(s,E) \, ds} \, dE$$

• In this case $I_d$ is our measured data, and we want to recover an image of $\mu(x,y)$

• Unfortunately, the integration over energy presents a mathematically intractable inverse problem

• We work around this approximately by assuming an effective energy

$$\overline{E} = \frac{\int_0^{E_{\text{max}}} ES(E) \, dE}{\int_0^{E_{\text{max}}} S(E) \, dE}$$
Approximate Imaging Equation

- Using an effective energy, we can write the imaging equation as
  \[ I_d = I_0 e^{-\int_0^d \mu(s,E) ds} \]

- A further simplification comes from defining
  \[ g_d \triangleq -\ln \left( \frac{I_d}{I_0} \right) \]

- Giving an x-ray transform
  \[ g_d = -\int_0^d \mu(s,E) ds \]

(we can solve this imaging equation)

- We need to measure the reference intensity \( I_0 \), typically done with a detector at the edge of the fan
- Although we can use FBP, the effective energy will be object dependent, so the reconstructed \( \mu(x,y) \) will only be approximate
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](image)

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

![Components of linear attenuation coefficients](image)
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 scaled to give us a fixed value for water (0) and air (-1000) according to:

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

• \(\mu(x, y)\) is the reconstructed attenuation coefficient for the voxel, \(\mu_{\text{water}}\) is the attenuation coefficient of water and CT(x,y) is the CT number (using Hounsfield units) of the voxel values in the CT image
CT Numbers

• Typical values in Hounsfield Units
CT scan showing 'apparent' density
Helical CT Scanning

- The 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.
**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**

- A pitch of 1.0 is roughly equivalent to axial (i.e. one slice at a time) 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, relative to detector axial extent, to avoid artifacts
  - Faster scan time, however, often more than compensates for undersampling artifacts (i.e. patient can hold breath so no breathing artifacts).
Image Reconstruction from Helical data

• Samples for the plane-of-reconstruction are estimated using two projections that are $2\pi$ apart

$$p'(\gamma, \beta) = wp(\gamma, \beta) + (1-w)p(\gamma, \beta + 2\pi)$$

where \( w = (q-x)/q \)
Single versus Multi-row Detectors

- Can image multiple planes at once
Single versus Multi-row Detectors

- Can image multiple planes at once
Multi-row Detectors
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
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 showing 'apparent' density

CT scan with iodine-based contrast enhancement
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)
Radiation dose versus kVp

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

<table>
<thead>
<tr>
<th>Parameter</th>
<th>80 kVp</th>
<th>120 kVp</th>
<th>140 kVp</th>
</tr>
</thead>
<tbody>
<tr>
<td>Image Contrast</td>
<td>Best</td>
<td>Intermediate</td>
<td>Poor</td>
</tr>
<tr>
<td>Noise</td>
<td>Most</td>
<td>Average</td>
<td>Least</td>
</tr>
<tr>
<td>Penetration</td>
<td>Least</td>
<td>Average</td>
<td>Most</td>
</tr>
</tbody>
</table>
Effective Dose Comparison with Chest PA Exam

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

Typical Background Radiation - 3 mSv per year
Types of 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
Beam Hardening

- Energy spectrum of an x-ray beam as it passes through water (rescaled)
- Mean energy increases with depth
- More photons get through, so measured attenuation is less than we would expect
Beam Hardening

- If there are significant contrast changes, beam-hardening can be difficult to correct
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
Patient Motion

- Respiratory motion effects during helical CT scans lead to well known artifacts at the dome of the diaphragm.
Truncation

- Standard CT field of view is 50 cm, but many patients exceed this
- Not often a problem for CT, but can be a problem when a truncated CT is used for PET attenuation correction