Chapter 3: X-ray Radiography and Computer Tomography

X-ray Physics
A Brief history
X-ray terminology
X-ray generation
  • Bremsstrahlung and characteristic X-rays
  • Basic structure of X-ray tube
  • Practical considerations
Attenuation of X-rays photons in media
  • Interaction of X-rays
  • Measures of X-ray attenuation
X-ray Terminology

- Film Radiography
- Fluoroscopy
- Digital Radiography
- Angiography
- Computed Tomography
Wilhelm Röntgen, Wurtzburg

Nov. 1895 – Announces X-ray discovery
Jan. 13, 1896 – Images needle in patient’s hand
– X-ray used presurgically
1901 – Receives first Nobel Prize in Physics
– Given for discovery and use of X-rays.
Roentgen’s experimental apparatus (Crookes tube) that led to the discovery of the new radiation on 8 Nov. 1895 – he demonstrated that the radiation was not due to charged particles, but due to an as yet unknown source, hence “x” radiation or “x-rays”

Known as “the radiograph of Bera Roentgen’s hand” taken 22 Dec. 1895
X-ray Imaging Examples

3D image of the blood vessels viewed by means of stereoscopic glasses.
X-ray Imaging Examples

Double mandibular fracture with strong displacement to the left.

Solitary humeral bone cyst known as "fallen leaf sign"
X-ray Imaging Examples

Cerebral angiogram obtained by injecting a iodine containing fluid into the arteries. The contrast dye subsequently fills the cerebral arteries, capillaries and veins.

Cerebral angiogram showing an aneurysm or saccular dilation of a cerebral artery.
K-edge Subtraction X-ray Imaging

Figure 3. Image of intracerebral arteries acquired in radiography mode with the germanium detector in antero-posterior projection, using synchrotron K-edge digital subtraction angiography: (A) early filling phase, (B) late filling phase.

Synchrotron-based intravenous cerebral angiography in a small animal model, PMB, 2004
K-edge Subtraction X-ray Imaging

Figure 4. Images acquired below (A) and above (B) the K-edge of iodine, both without easily discernible contrast in the cerebral arteries. The subtracted image (C), however, shows very good contrast in the cerebral arteries.

Synchrotron-based intravenous cerebral angiography in a small animal model, PMB, 2004
Questions

Questions concerned in this lecture:

• Why X-ray is the most used ionizing radiation for diagnostic imaging?
• What are we measuring with X-ray radiography?
• What are the physical factors that contribute to the contrast in an X-ray image?
• What is the fundamental limitation of X-ray for diagnostic purpose?

To answer these questions, we need to know the basics of X-ray physics.

• How X-rays are generated?
• What are the characteristics of these particles?
• How are X-rays attenuated in objects and in X-ray detector?
A Typical X-ray Radiography System
X-ray Sources
X-ray Generation – Bremsstrahlung

- e⁻ released from the cathode are accelerated towards the anode with a gain in KE as the e⁻ drops through the applied potential difference (kilovoltage potential - kVp)

Unfortunately,
- About 99% of the KE converted to heat via collision-like interactions
- About 1% of the KE converted into x-rays via strong Coulomb (electrostatic) interactions → Bremsstrahlung
X-ray Generation – Bremmstrahlung

- Target nucleus positive charge \((Z \cdot p^+)\) attracts incident \(e^-\)
- Deceleration of an incident \(e^-\) occurs in the proximity of the target atom nucleus
- \(E\) lost by \(e^-\) is gained by the EM photon (x-ray) generated
  - The impact parameter distance, the closest approach to the nucleus by the \(e^-\) determines the amount of \(E\) loss
  - The Coulomb force of attraction varies strongly with distance \((\propto 1/r^2)\); ↓ distance → ↑ deceleration and \(E\) loss → ↑ photon \(E\)
  - Direct impact on the nucleus determines the maximum x-ray \(E\) \((E_{max})\)
Interestingly, this process creates a relatively uniform spectrum.

\[ \text{Intensity} = n \hbar \nu \]

Photon energy spectrum
X-ray Generation – Bremsstrahlung

• The unfiltered Bremsstrahlung spectrum (intensity) contains a large number of very low E photons and ↓ approx. linearly as photon E ↑.

• The peak voltage (kVp) applied across the electrodes of the x-ray tube determines the highest x-ray E (E_{max})

• The lowest E of the unfiltered x-ray spectrum is not easily determined, due to severe attenuation of these photons by the material and thickness of the x-ray tube envelope

• X-ray production efficiency is influenced by the target Z and acceleration potential (kVp)

\[-\left(\frac{dE}{dx}\right)_r = \frac{NEZ(Z+1)e^4}{137m_e^2v^4}\left(4\ln\frac{2E}{m_ec^2} - \frac{4}{3}\right)\]

Specific energy loss by Bremsstrahlung
The Unfiltered Bremsstrahlung Spectrum
Thick Target X-ray Formation

We can model target as a series of thin targets. Electrons successively loses energy as they moves deeper into the target.

Each layer produces a flat energy spectrum with decreasing peak energy level.
X-ray Generation – Characteristic X-rays

Electron binding energy

Hydrogen  $Z = 1$

Tungsten  $Z = 74$
X-ray Generation – Characteristic X-rays

- $e^-$ of the target atom have a binding energy (BE) that depends on atomic $Z$ (rem: $BE_K \propto Z^2$) and the shell ($BE_K > BE_L > BE_M > \ldots$)

- When $e^-(KE)$ incident on the target exceeds the target atom $e^-(BE)$, it’s energetically possible for a collisional interaction to eject the bound electron and ionize the atom.

- What would happen then?
X-ray Generation – Characteristic X-rays

TABLE 5-1. ELECTRON BINDING ENERGIES (keV) OF COMMON X-RAY TUBE TARGET MATERIALS

<table>
<thead>
<tr>
<th>Electron Shell</th>
<th>Tungsten</th>
<th>Molybdenum</th>
<th>Rhodium</th>
</tr>
</thead>
<tbody>
<tr>
<td>K</td>
<td>69.5</td>
<td>20.0</td>
<td>23.2</td>
</tr>
<tr>
<td>L</td>
<td>12.1/11.5/10.2</td>
<td>2.8/2.6/2.5</td>
<td>3.4/3.1/3.0</td>
</tr>
<tr>
<td>M</td>
<td>2.8–1.9</td>
<td>0.5–0.4</td>
<td>0.6–0.2</td>
</tr>
</tbody>
</table>

- The target materials used in x-ray tubes for diagnostic medical imaging include: W (Z=74), Mo (Z=42) and Rh (Z=45): $BE \propto Z^2$.

- As the $E$ of the incident $e^-$ increases above the threshold $E$ for characteristic x-ray production, the % of char. x-rays $\uparrow$ (5% @ 80 kVp vs. 10% @ 100 kVp for W).

- A variety of $E$ transitions occur from adjacent ($\alpha$) and non-adjacent ($\beta$) $e^-$ shells $\rightarrow$ discrete $E$ lines superimposed on the continuous bremsstrahlung spectrum.
X-ray Generation – Characteristic X-rays

• Within each shell (other than K) there are discrete E orbitals \( \ell = 0, 1, \ldots, n-1 \) → characteristic x-ray fine E splitting
• Characteristic x-rays other than those generated through K-shell transitions are unimportant

<table>
<thead>
<tr>
<th>Shell Transition</th>
<th>Tungsten</th>
<th>Molybdenum</th>
<th>Rhodium</th>
</tr>
</thead>
<tbody>
<tr>
<td>( K_{\alpha 1} )</td>
<td>59.32</td>
<td>17.48</td>
<td>20.22</td>
</tr>
<tr>
<td>( K_{\alpha 2} )</td>
<td>57.98</td>
<td>17.37</td>
<td>20.07</td>
</tr>
<tr>
<td>( K_{\beta 1} )</td>
<td>67.24</td>
<td>19.61</td>
<td>22.72</td>
</tr>
</tbody>
</table>

*Note: Only prominent transitions are listed.
X-ray Generation – Characteristic X-rays

Superimposed multiple flat spectrum with decreasing cutoff energy

Low energy X-rays suffer attenuation inside the anode

Further attenuation by the source package.

External filtering to reduce low E photons $\rightarrow$ lower dose

Beam hardening

Figure 5.5
Relative intensity of x-ray photons. (Adapted from Webster, 1998. This material is used by permission of John Wiley & Sons, Inc.)
Practical Considerations on X-ray Source
X-ray Generation – X-ray Tube

Figure 5.3
An x-ray tube.

Figure 5.4
Schematic diagram of an x-ray tube.

Motor, Why?

Rotating target

Electron beam? How are electrons generated?

X-ray Generation – Anode Design

The size of the focus spot varies with angle.

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.
X-ray Generation – Filament

FIGURE 5-7. The x-ray tube cathode structure consists of the filament and the focusing (or cathode) cup. Current from the filament circuit heats the filament, which releases electrons by thermionic emission.

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.
X-ray Generation – Cathode

- Shapes $e^-$ distribution when at same V as filament (unbiased)
- Isolation from filament and application of a negative bias V constrains $e^-$ distribution further (biased)
- Focusing cup slot width determines the focal spot width
- Filament length determines focal spot length
- Small and large focal spot filaments (usu. 0.6 and 1.2 mm)

What limits the focal spot size? The power output of the tube!
X-ray Generation – Cathode

Geometrical Unsharpness

The **Focal Spot MTF** may be measured using a pinhole to determine the PSF or a slit to determine the LSF, and calculating the normalised modulus of the Fourier Transform of the spread function.

(Left) A pinhole image a 2.0 mm focal spot showing a typical distribution of the X ray intensity
(Right) the corresponding 2D MTF
Power Rating Versus Focal Spot Size

<table>
<thead>
<tr>
<th>Nominal X-ray Tube Focal Spot Size (mm)</th>
<th>Typical Power Rating (kW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.2–1.5</td>
<td>80–125</td>
</tr>
<tr>
<td>0.8–1.0</td>
<td>50–80</td>
</tr>
<tr>
<td>0.5–0.8</td>
<td>40–60</td>
</tr>
<tr>
<td>0.3–0.5</td>
<td>10–30</td>
</tr>
<tr>
<td>0.1–0.3</td>
<td>1–10</td>
</tr>
<tr>
<td>&lt;0.1 (micro-focus tube)</td>
<td>&lt;1</td>
</tr>
</tbody>
</table>

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.
Factors Affecting X-ray Generation

1) Target material - affects quality and quantity of radiation emitted
2) Tube voltage – kVp - determines maximum energy of photons emitted in the bremsstrahlung spectrum, thus affecting quality of the beam and the overall exposure.

Rule of thumb 1: Exposure $\propto (\text{kVp})^5$
Rule of thumb 2: Exposure $\propto \text{mAs}$

3) Tube current (mA)
4) Exposure time (s) – duration of x-ray production. Often the current and exposure time are expressed together as a product, in mAs.
5) Beam filtration – modifies the x-ray energy spectrum and the overall number of photons in the beam.
6) Generator waveform – affects the spectrum and quantity of photons emitted. Single phase system provides lower average energy and number of photons than does a three phase system.
7) Focal spot size – affects the number of photons being produced.
X-ray Generation – Tube Output

• Filament current (A) → filament temperature (T) → thermionic emission rate
• When kVp is low, an e⁻ cloud (space charge cloud) forms around filament
• Space charge cloud shields the electric field for tube voltages of \( \leq 40 \text{kVp} \) → only some e⁻ are accelerated towards the anode: space charge limited
• \( \geq 40 \text{kVp} \) the space charge cloud effect overcome by kVp applied and tube current (mA) limited only by the emission of e⁻ from the filament: emission-limited operation
• Tube current about 5-10 times less than the filament current in the emission-limited range

Increasing filament current does not increase tube current.
X-ray Generation – Heel Effect

- Reduction of x-ray beam intensity towards the anode side of the x-ray field although x-rays generated isotropically.
- Self-filtration by the anode and the anode bevel causes greater intensity on the cathode side of the x-ray field.
- Can use to advantage, e.g., chest exposure.
- Orient chest to anode side and abdomen to cathode side.
- Less pronounced as SID ↑
X-ray Generation – Filtration

Filtration for geometrical compensation

Figure 5.7
Various compensation filters. (Adapted from Carlton and Adler, 2001, and Wolbarst, 1993.)
X-ray Generation – Collimation

- Collimators adjust size and shape of x-ray beam
- Parallel-opposed lead shutters
- Light field mimics x-ray field
- Reduces dose to patient; ALARA: as low as reasonably achievable
- Reduced scatter radiation to image receptor: better image contrast