Chapter 16
Diagnostic Radiology

Radiation Dosimetry I

Text: H.E Johns and J.R. Cunningham, The physics of radiology, 4th ed.

http://www.utoledo.edu/med/depts/radther

Introduction

- In diagnostic radiology we are interested in the beam of x-rays transmitted through the patient
- Difference in beam attenuation results in a shadow picture registered by the detector
- The objective is to obtain the best picture quality with the minimal dose to the patient

Table 1

Adult Effective Doses for Various Diagnostic Radiology Procedures

Examination	Average Effective Dose (mSv)	Values Reported in Literature (mSv)
Skull	0.1	0.03–0.22
Cervical spine	0.2	0.07–0.3
Thoracic spine	1.0	0.6–1.4
Lumbar spine	1.5	0.5–1.8
Posteroanterior and lateral study of chest	0.1	0.05–0.24
Posteroanterior study of chest	0.02	0.007–0.050
Mammography	0.4	0.10–0.60
Abdomen	0.7	0.04–1.1
Pelvis	0.6	0.2–1.2
Hip	0.7	0.18–2.71
Shoulder	0.01	. . .
Knee	0.005	. . .
Other extremities	0.001	0.0002–0.1
Dual x-ray absorptiometry (without CT)	0.001	0.001–0.035
Dual x-ray absorptiometry (with CT)	0.04	0.003–0.06
Intravenous urography	3	0.7–3.7
Upper gastrointestinal series	6*	1.5–12
Small-bowel series	5	3.0–7.8
Barium enema	8*	2.0–18.0
Endoscopic retrograde cholangiopancreatography	4.0	. . .

* Includes fluoroscopy.

Mettler et al., Special Report: Effective doses in radiology and Nuclear Medicine, Radiology **248**, 254-263 (2008).

Table 2

Adult Effective Doses for Various CT Procedures

Examination	Average Effective Dose (mSv)	Values Reported in Literature (mSv)
Head	2	0.9–4.0
Neck	3	. . .
Chest	7	4.0–18.0
Chest for pulmonary embolism	15	13–40
Abdomen	8	3.5–25
Pelvis	6	3.3–10
Three-phase liver study	15	. . .
Spine	6	1.5–10
Coronary angiography	16	5.0–32
Calcium scoring	3	1.0–12
Virtual colonoscopy	10	4.0–13.2

Background radiation: ~3 mSv/y

Table 3

Adult Effective Doses for Various Interventional Radiology Procedures

Examination	Average Effective Dose (mSv)*	Values Reported in Literature (mSv)
Head and/or neck angiography	5	0.8–19.6
Coronary angiography (diagnostic)	7	2.0–15.8
Coronary percutaneous transluminal angioplasty, stent placement, or radiofrequency ablation	15	6.9–57
Thoracic angiography of pulmonary artery or aorta	5	4.1–9.0
Abdominal angiography or aortography	12	4.0–48.0
Transjugular intrahepatic portosystemic shunt placement	70	20–180
Pelvic vein embolization	60	44–78

* Values can vary markedly on the basis of the skill of the operator and the difficulty of the procedure.

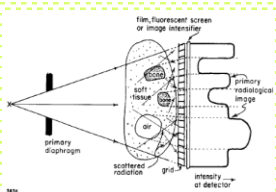
Table 4

Adult Effective Dose for Various Dental Radiology Procedures

Examination	Average Effective Dose (mSv)	Values Reported in Literature (mSv)
Intraoral radiography	0.005	0.0002–0.010
Panoramic radiography	0.01	0.007–0.090
Dental CT	0.2	. . .

Mettler et al., Special Report: Effective doses in radiology and Nuclear Medicine, Radiology **248**, 254-263 (2008).

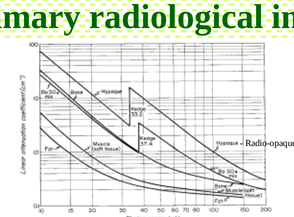
Primary radiological image



- The kV energy source size needs to be small to provide good resolution
- Only the primary beam carries the information about the object being imaged

- A detector registers the primary radiological image and converts it into a visible image
- The scatter component is considerably higher than the primary – have to use the grid to reduce the scatter

Primary radiological image



I: Z=53
Ba: Z=56

Hypaque - Radio-opaque agent containing iodine

Figure 16-2. Linear attenuation coefficient in cm^{-1} for fat, soft tissue, bone, 75% Hypaque, and $BaSO_4$ "mix." The coefficients are the total less coherent scattering and were calculated using data in the Appendix. The compositions for fat, soft tissue, and bone are given in Table 5-3. 75% Hypaque contains, per cm^3, 0.25 g $C_{18}H_{26}I_3N_3O_9$, 0.50 g $C_{11}H_8I_3N_2O_4$, and 0.65 g water, giving a density of 1.40 g/cm^3. $BaSO_4$ "mix" consists of 450 g $BaSO_4$ in 2500 ml water to give a suspension of density 1.18 g/cm^3.

- The difference in attenuation coefficients between tissue, fat, and bone is large enough to produce an image
- To resolve different soft tissues often need contrast media

Images with contrast media

- In nuclear medicine injected radioactive material is imaged through detection of decay products
- In radiology contrast media having significantly larger attenuation coefficients is used for soft tissue visualization
- Liquid compounds containing iodine (Z=53, k-edge=33.2keV) or barium (Z=56, k-edge 37.4keV)
  - 1 mm iodine-filled artery reduces the photon fluence through 13 cm soft tissue by > 60% for 90 kVp beam, easily visible in the image

Diagnostic radiology modalities

- Screen-film radiography
- Fluoroscopy
- Digital radiography
- Computed tomography
- MRI
- Ultrasound

(MRI, Ultrasound) } Do not utilize x-ray source

Radiographic film

- Only a small fraction of x-rays (~2%) is absorbed within a film
- Film is sandwiched between two fluorescent screens packed into a light-tight cassette
  - Both front and back surfaces of the film contain photosensitive emulsion
  - Image is created with optical or UV photons emitted from both screens

Radiographic film

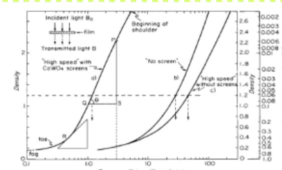


Figure 16-12. Typical characteristic curves of x ray films. (a) "high speed" film with calcium tungstate screen; (b) "no-screen" film; (c) "high speed" film without screen. The right hand scales show the density corresponding to the fraction of light transmitted.

Film characteristics:

Density

$$D = \log_{10}\left(\frac{B_0}{B}\right)$$

Gamma (resolution)

$$\gamma = \frac{D_2 - D_1}{\log_{10}(X_2 / X_1)}$$

- After being developed and fixed the film is viewed in front of a lightbox; transmitted brightness B is used to find density
- Characteristic (H&D after Hurter and Driffield) curves: density vs. exposure
- Film speed (sensitivity): the exposure required for D=1.0 above the background (base + fog)

Image intensifier

- The main purpose is to increase the brightness of an image
- Two processes are used:
  - (1) minification, in which a given number of light photons emanates from a smaller area
  - (2) flux gain, where electrons accelerated by high voltages produce more light as they strike a fluorescent screen

Image intensifier

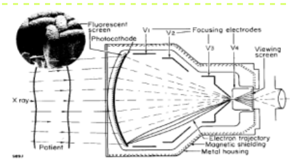


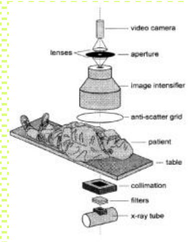
Figure 16-9. Schematic diagram of image intensifier tube. The insert shows a photograph of the CsI layer. This layer is in the form of many crystal light "pipes," which prevent the light from spreading and direct it towards the photocathode. (Diagram

- Convex fluorescent input screen, gain ~ 2000 to 3000
- Optical photons fall onto a photo-cathode, generating electrons (0.15 – 0.2 efficiency ratio)
- Electrons are accelerated and focused onto the output screen (similar to the input, but with smaller phosphor granules)

Fluoroscopy

- If transmitted x-rays are converted into optical photons - images can be viewed in real time
- Old days – used fluoroscopic screens, producing very dim images
- Image intensifier makes the image very bright, and much easier to view and analyze
- The brightness gain of image intensifiers varies from 1000 to over 6000

Fluoroscopy

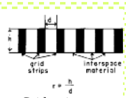


Fluoroscopic imaging chain

C-arm

Grids

- The scattered radiation spoils the radiograph
- Scatter can be removed by a grid placed between the film (detector) and the patient
- The ability of the grid to discriminate against scatter is measured by the grid ratio = h/d
- Use of grids increases the required exposure



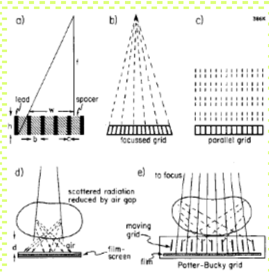
Grid ratio

TABLE 16-3

Grid Factors for Primary (P) and for Primary plus Scattered Radiation (P + S) for X ray Beams Passing through 20 cm Water

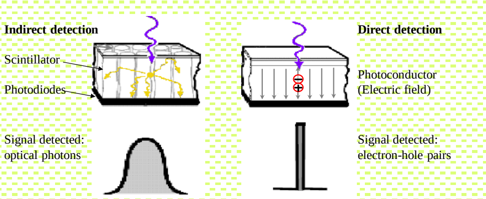
Grid ratio		60 kV	80 kV	100 kV	120 kV
8:1	P	1.9	1.8	1.8	1.7
	P + S	4.3	4.0	3.7	3.4
12:1	P	2.1	2.1	2.0	2.0
	P + S	5.3	5.0	4.8	4.4

Grids



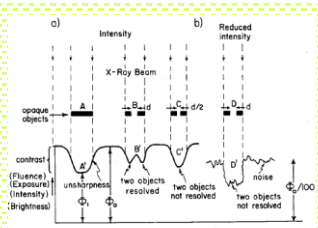
- Parallel grid may produce image cut-off
- Focused grid requires alignment with x-ray tube
- Moving grid allows complete removal of its own image (traveling period should be coordinated with the exposure time and pulses)
- 10 to 20 cm thick air gap reduces the scatter

X-ray detector configurations



- Typical configuration for high-energy x-rays is a two-stage indirect detection
- Direct detection is more desirable: simpler, less signal spreading

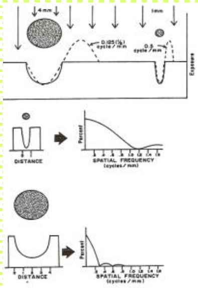
Description of image quality



Parameters:

- Sharpness
- Resolution
- Contrast $(\Phi_1-\Phi_0)/\Phi_0$
- Noise

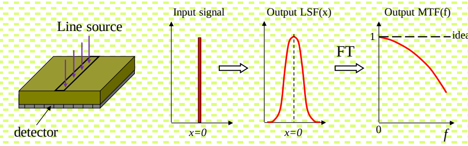
Object spectrum



- Frequency $f = 1/(2x)$
- Units of cycles/mm or lp/mm
- Small features correspond to high frequencies
- All parameters characterizing performance of a detector are frequency-dependent
- Sample spacing and aperture size limit the frequency range

LSF and MTF

- MTF(f)=FT{LSF(x)}
- Sharp line source is registered as a distribution of signal with coordinate
- After FT is performed - obtain information on how all frequencies are degraded by the system



Imaging system characterization: MTF

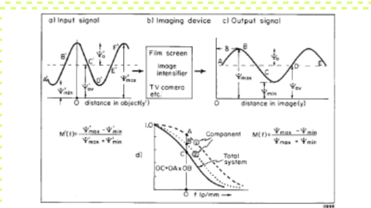


Figure 16-19. Schematic diagram to illustrate how various components in an imaging system alter the input pattern A'B'C'D'E' to yield an output pattern ABCDE. Panel d shows how the MTF of the total system can be calculated from the MTF values for two components of the system.

The output signal modulation is always less than the input value: imaging always results in loss of information

Imaging system characterization: MTF

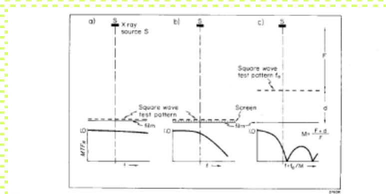
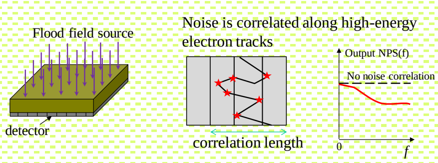


Figure 16-20. Diagrams to illustrate how imaging devices alter the MTF. (a) Square wave test pattern in contact with film. (b) Square wave test pattern in contact with screen, which is in contact with film. (c) Square wave test pattern some distance from the film. The square wave test pattern could equally well be a slit of x ray fluence.

- Loss of modulation at higher frequencies (blurring)
- Any feature of detection system (slit, magnification, etc.) will be translated into characteristic frequency in MTF plot

Noise power spectrum (NPS)

- The noise transfer properties: level and correlation between neighboring pixels
- Synthesized slit technique: absorbed energy distributions in separate slits are averaged and analyzed using FT



Detective quantum efficiency

- Imaging system is characterized by how well it transmits the signal
- Parameter - Detective Quantum Efficiency $DQE = SNR_{in}/SNR_{out}$
- DQE(0) – corresponds to absorption efficiency
- DQE(f) – characterizes the ability of system to image objects of different sizes

Detective quantum efficiency

- Putting it together:
  - MTF(f)=FT{LSF(x)}
  - NPS(f)=FT{Noise(x)}
  - *D* – average energy deposited
  - q_0 – number of incident photons per unit area

$$DQE(f) = \frac{D^2 \cdot [MTF(f)]^2}{q_0 \cdot NPS(f)}$$

- The objective is always to maximize DQE

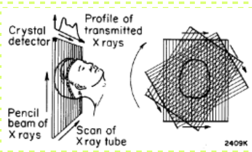
Computed tomography

- In a 2-D radiograph transmitted intensity

$$I = I_0 e^{-\sum_{i=1}^{n} \mu_i x_i}$$

- Values of μ_i and x_i are not known
- If we take many images in the same plane, at different angles, it is possible to find μ_i and x_i and reconstruct a 3-D image

Computed tomography



- An x-ray tube emitting a pencil-like beam is coupled to a radiation detector
- The two are moved together sot hat the head is scanned by a series of a parallel x-rays as the translation takes place
- The fraction of radiation transmitted is stored for each ray

Computed tomography

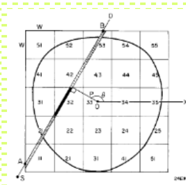
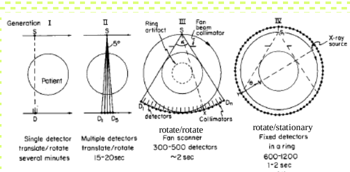


Figure 16-39. Schematic representation of pixels with a cross section of the head superimposed. A typical ray from the source S to the detector D is shown. It can be represented by the parameter p and the angle θ. The path length through each of the pixels is illustrated.

- Ray SD can be described by two parameters: p and θ
- Image is split into pixels
- Path length through each pixel contributes to the final ratio of I_0/I, with its own μ_i and x_i
- A set of equations can be solved to find all μ_i and x_i and reconstruct the original image

Computed tomography



With development of each generation of CT scanners
- Scan time was drastically decreased
- The resolution was improved, artifacts eliminated
- Further generations: helical movement, detector arrays

CT image reconstruction

- Even though a set of linear equations can in principle be solved to find μ_i and x_i for each pixel, it is not practical
- Instead image reconstruction algorithms are used
  - Simple backprojection
  - Filtered backprojection (convolution)
  - Fourier transform
  - Series expansion

CT image reconstruction

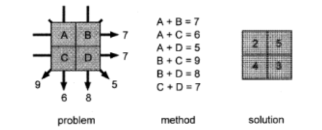


FIGURE 13-27. The mathematical problem posed by computed tomographic (CT) reconstruction is to calculate image data (the pixel values—A, B, C, and D) from the projection values (*arrows*). For the simple image of four pixels shown here, algebra can be used to solve for the pixel values. With the six equations shown, using substitution of equations, the solution can be determined as illustrated. For the larger images of clinical CT, algebraic solutions become unfeasible, and filtered backprojection methods are used.

Bushberg, et al., The essential physics of medical imaging, 2nd edition

- A modern CT image contains ~200,000 pixels; each of 800,000 projections represents an individual equation

CT image reconstruction

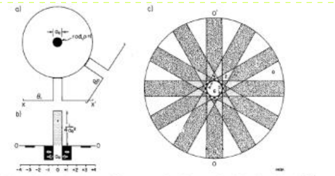


Figure 16-42. (a) Transmission profiles at θ_i and θ_n for a rod of density 1.0 and diameter a_0 in air. (b) Ramachandran filter function. (c) Back projection of a for θ = 0, 30, 60, 90, 120, and 150 degrees giving a starlike pattern.

- Filtered backprojection (convolution) results in more accurate image
- There is a number of different convolution functions used

Spatial resolution

TABLE 1-1. THE LIMITING SPATIAL RESOLUTIONS OF VARIOUS MEDICAL IMAGING MODALITIES: THE RESOLUTION LEVELS ACHIEVED IN *TYPICAL* CLINICAL USAGE OF THE MODALITY

Modality	Δ (mm)	Comments
Screen film radiography	0.08	Limited by focal spot and detector resolution
Digital radiography	0.17	Limited by size of detector elements
Fluoroscopy	0.125	Limited by detector and focal spot
Screen film mammography	0.03	Highest resolution modality in radiology
Digital mammography	0.05–0.10	Limited by size of detector elements
Computed tomography	0.4	About ½-mm pixels
Nuclear medicine planar imaging	7	Spatial resolution degrades substantially with distance from detector
Single photon emission computed tomography	7	Spatial resolution worst toward the center of cross-sectional image slice
Positron emission tomography	5	Better spatial resolution than with the other nuclear imaging modalities
Magnetic resonance imaging	1.0	Resolution can improve at higher magnetic fields
Ultrasound imaging (5 MHz)	0.3	Limited by wavelength of sound

Bushberg et al., The essential physics of medical imaging, 2nd edition.