Lecture 5 –X ray Instrumentation

This lecture will cover:

- – Instrumentation of planar radiography (*CH2.6-2.7*)
  - Anti-Scatter grid
  - X ray Detector
- – Characteristics of X-ray images (*CH2.8*)
  - SNR
  - Resolution
  - CNR

(*Supplementary reading: The Essential Physics of Medical Imaging CH7.1-7.10, 7.12*)

A radiography system

- Source: X-ray tube
- Filters (滤过)
  - Aluminum and copper
  - Absorbing low energy
  - Beam hardening (射束硬化效应)
- Collimator (准直器)
  - Sheets of lead between source and patient
  - Limited the patient area to field-of-view (FOV, 视野)
  - reduce radiation dose and Compton scattered X-ray
- Anti-Scatter grid (防散射滤线栅)
- Detector (探测器)

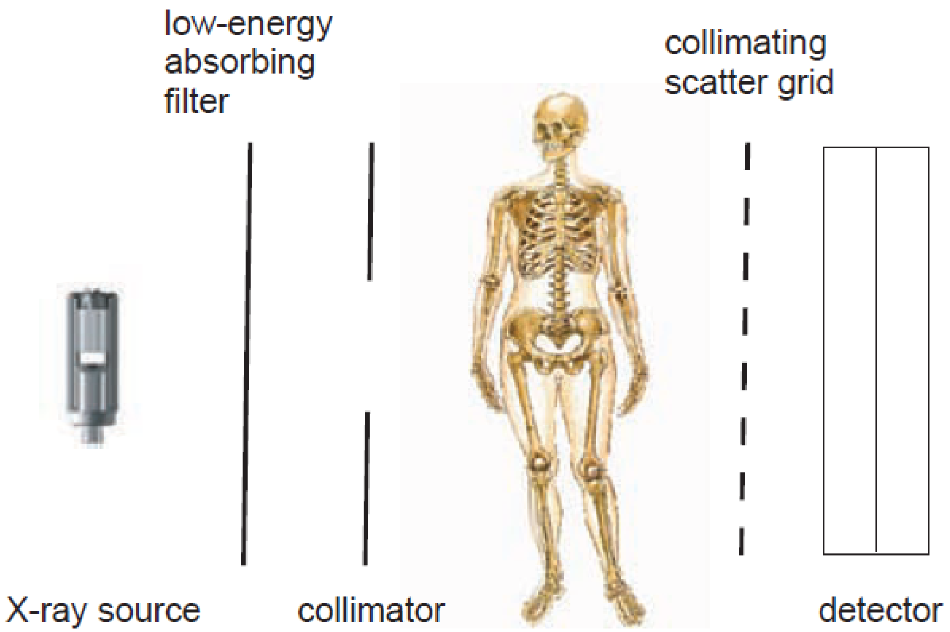


Figure. Schematic representation of the radiographic imaging chain.

Anti-Scatter grid

- Compton scattered X-rays contribute to a background signal which reduces the image.

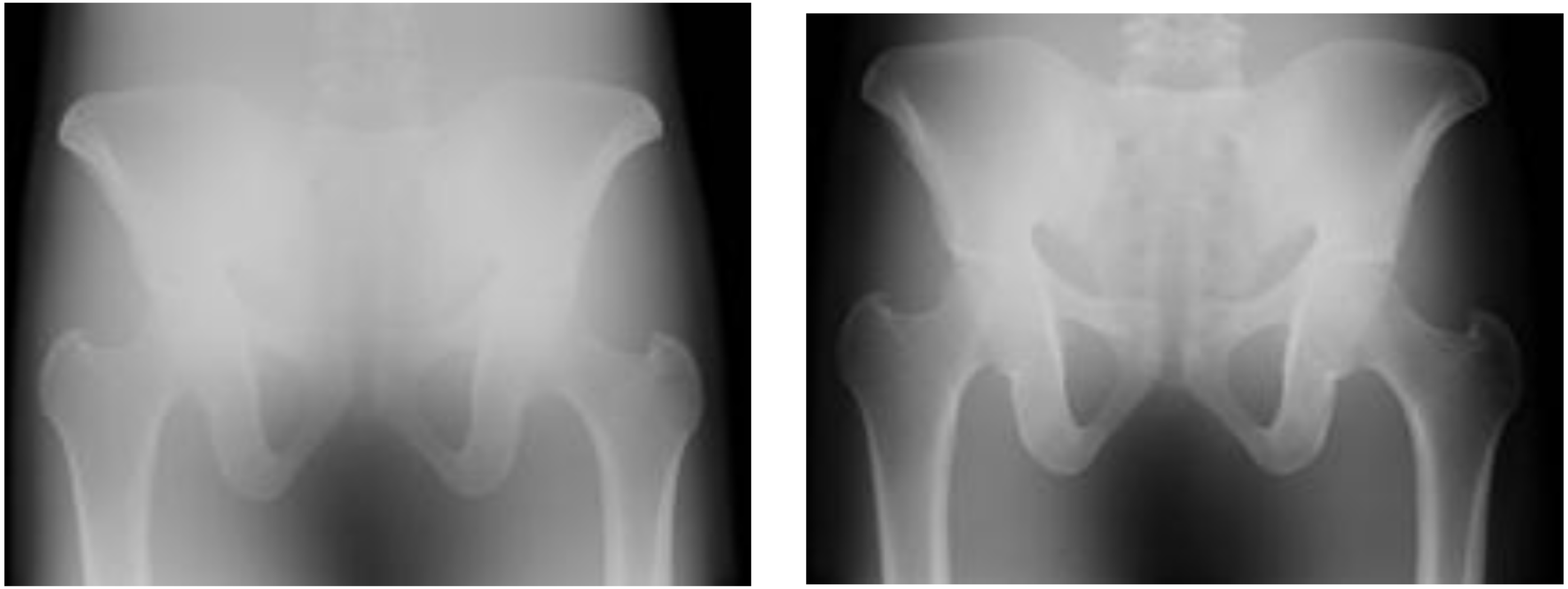


Figure. Images showing the effects of an anti-scatter grid on the CNR of a planar X-ray image. The images are produced from a pelvic phantom, which simulates the absorption properties of the human pelvis. (a) No anti-scatter grid: there is a large background signal from Compton-scattered X-rays which reduces the CNR of the image. (b) With an anti-scatter grid in place the overall signal intensity of the image is reduced, but the CNR is improved significantly.

Anti-Scatter grid

- Between patient and detector
- Parallel strips of lead foil
- grid ratio $= \frac{h}{d}$ grid frequency $= \frac{1}{d+t}$

  where

  h : length of lead strip

  d : separation of lead strip

  t : thickness of lead strip
- Bucky Factor (BF): the dose delivered to the patient increase by BF with grid (~4-10)

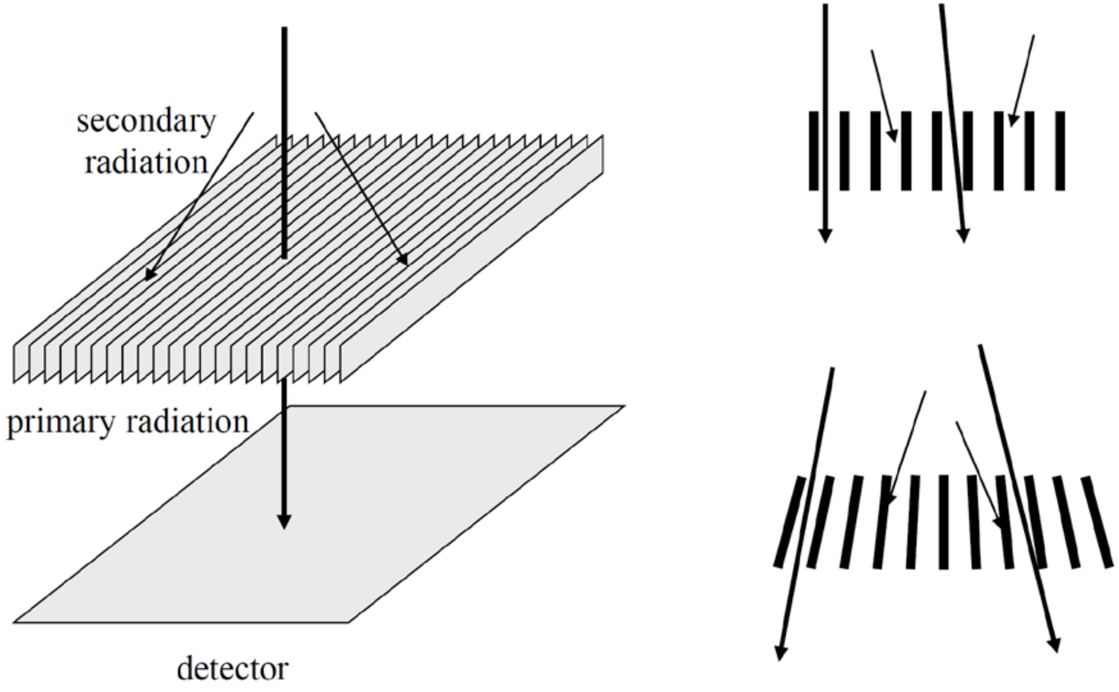


Figure. (left) Basic design of an anti-scatter grid, with thin lead septa aligned in either a parallel (top right) or slightly diverging (bottom right) geometry. The thick arrows show primary radiation which passes through the anti-scatter grid, and the thin arrows correspond to secondary Compton-scattered radiation which is stopped by the grid.

Detector

- ➢ Traditional X-ray film
- ➢ Digital detector
  - Advantages
    - ✓ Higher quality image
    - ✓ Store and transfer easily
      - Picture Archiving and Communication System (PACS)
      - Digital Imaging and Communications in Medicine (DICOM)
  - Types
    - ✓ Computed Radiography (CR, 计算机X线摄影)
    - ✓ Digital Radiography (DR, 数字X线摄影)

Computed Radiography

- ➢ CR plate / Imaging plate (成像板)
  - Standard: 35*43cm
  - Mammography: 18*24cm
- ➢ CR reader
- ➢ Image display and storage

Figure. Illustration of a CR system.

Computed Radiography

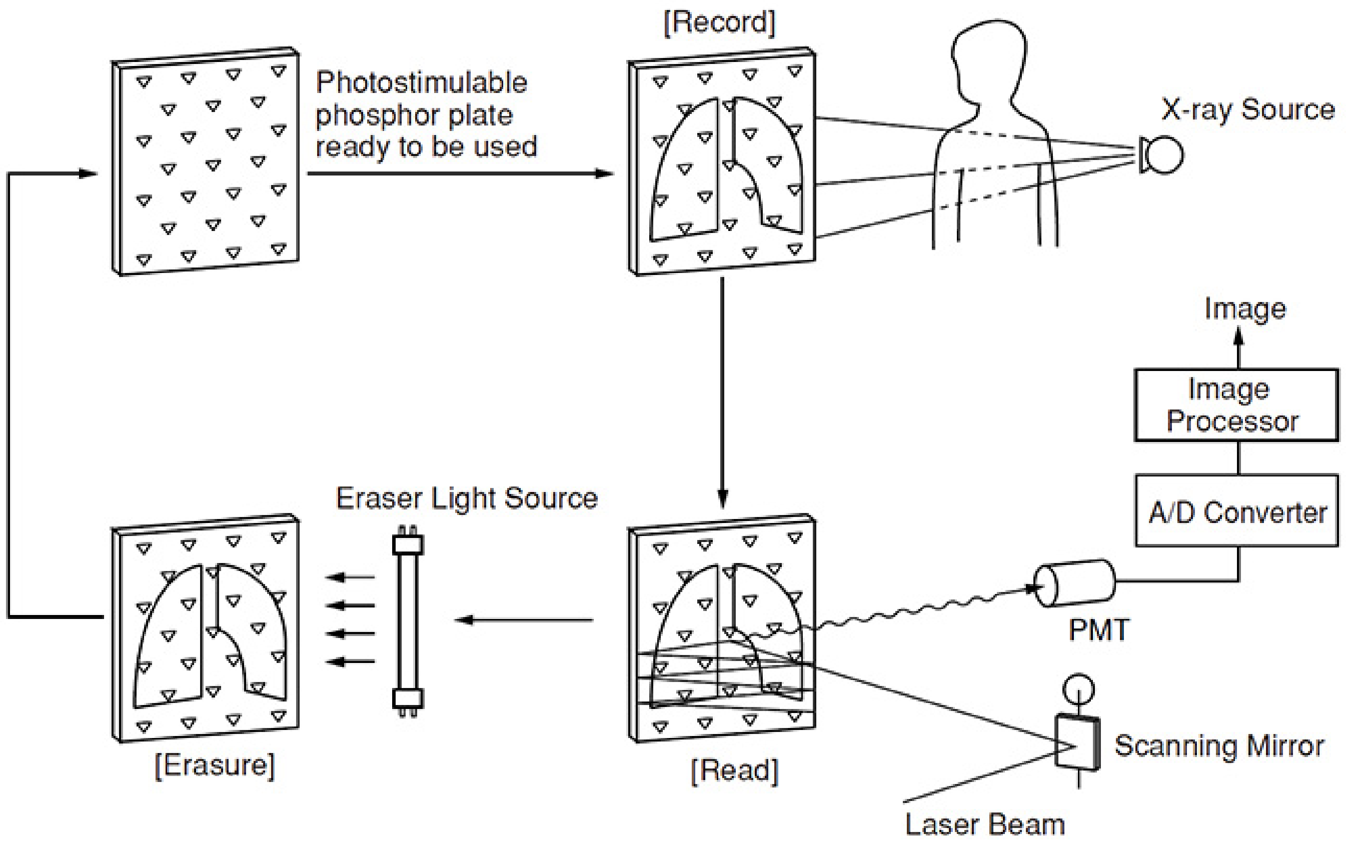


Figure. Image recording and reading using a CR plate.

Imaging Plate

- **Protective layer** – protection of the phosphor layer
- **Phosphor or active layer** (荧光层) – image acquisition
- **Reflective layer** – improve the detection efficiency
- **Support Layer** – provide strength for the imaging plate
- **Backing Layer** – protect back of the cassette

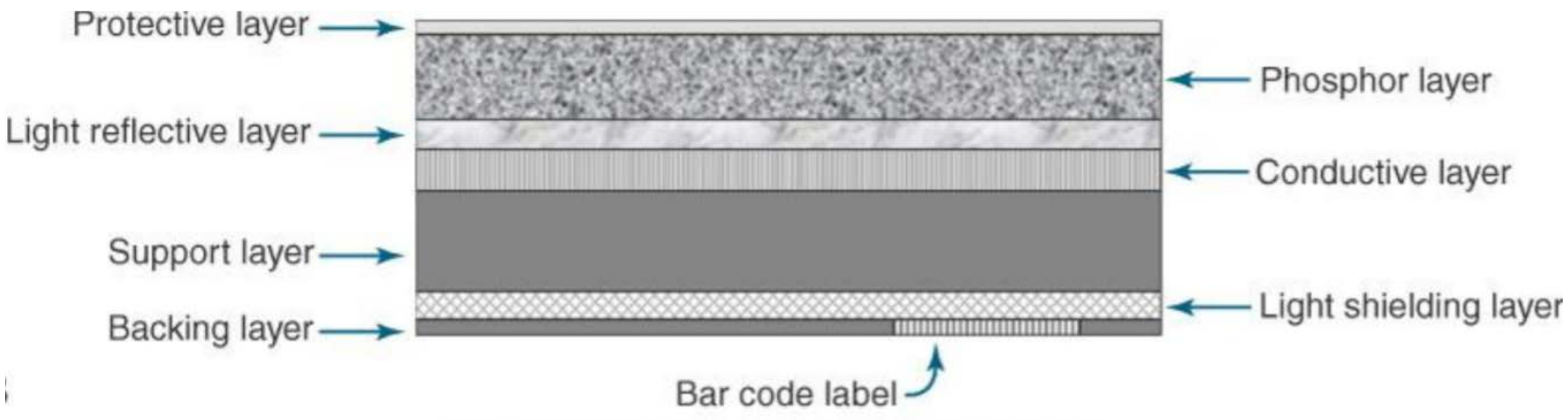
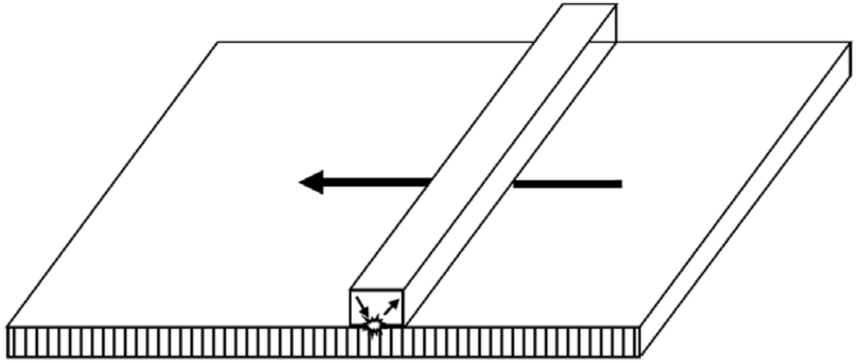


Figure. Structure of image plate.

Photo stimulated luminescence

- **Photo stimulated luminescence (PSL, 光激励发光)**
  1. By X-ray : release electron and trapped as a “latent” image (潜影)
  2. By Laser :
     ① trapped electron return to ground states from the excited states and transfer energy to EU^{2+}
     ② emit light from the excited states of EU^{2+} detected by photodiodes and lens
     ③ convert light intensity to voltages
  3. Light intensity is dependent on energy of incident X-ray.
  4. PSL Material: $BaFX:EU^{2+}$, $CsBr:EU^{2+}$



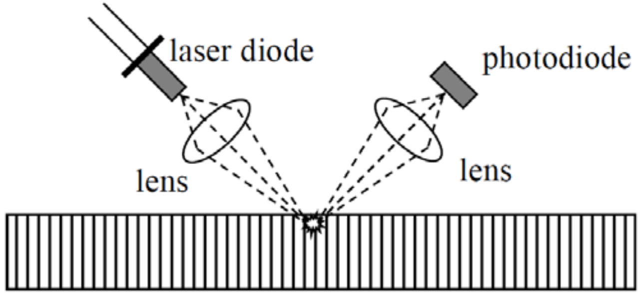
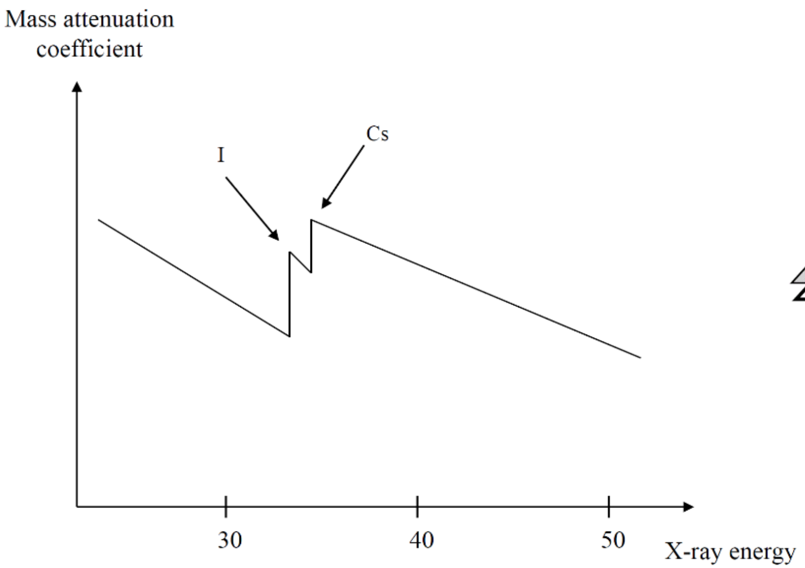
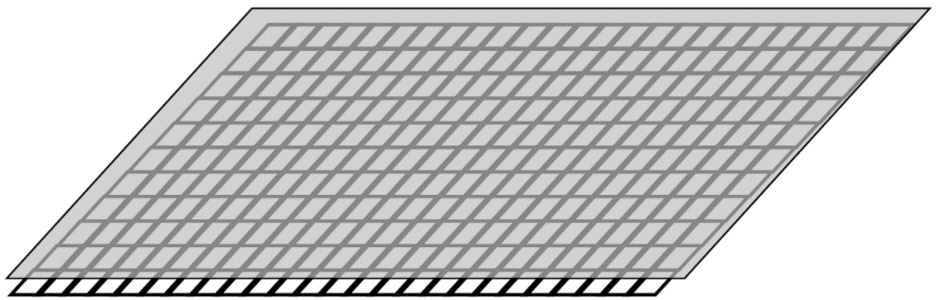


Figure. A computed radiography reader. The reader consists of a large array of laser diodes and photodiodes, and this array is rapidly moved from right-to-left across the plate to produce the entire image.

Digital Radiography

- **Indirect DR detector :** commonly used DR detector
  - X-ray → light → voltage
  - CsI:Tl scintillator : needle crystal
  - Flat-panel detector (FPD，平板探测器) + thin-film transistor (TFT, 薄膜晶体管)
  - 43*43cm FDP with a TFT array of 3001*3001 elements, and pixel interval is 143μm
- **Direct DR detector**
  - X-ray → voltage
  - Amorphous selenium: not efficient on X-ray absorption





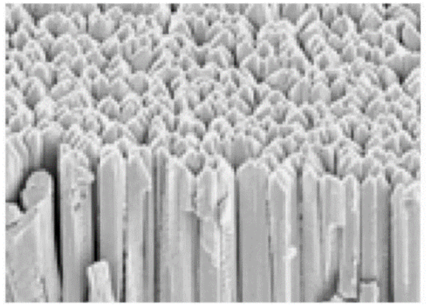


Figure. (left) The attenuation coefficient of CsI:Tl as a function of the incident X-ray energy. Each of the elements has a K-edge which increases absorption significantly. (middle) A thin CsI:Tl layer (shaded) placed on top of a SiH active matrix array. (right) An electron micrograph showing the needle-like structure of the crystals of CsI:Tl.

Lecture 5 –X ray Instrumentation

This lecture will cover:

- Instrumentation of planar radiography
  - Anti-Scatter grid
  - Detectors
- **Characteristics of X-ray images**
  - **SNR**
  - **Resolution**
  - **CNR**

Signal-to-noise ratio

N X-ray strike the detector per unit area

(Poisson distribution)

$$P(N) = \frac{\mu^N e^{-\mu}}{N!}$$

Where variation $\sigma = \sqrt{\mu}$

$$\text{SNR} \propto \sqrt{N}$$

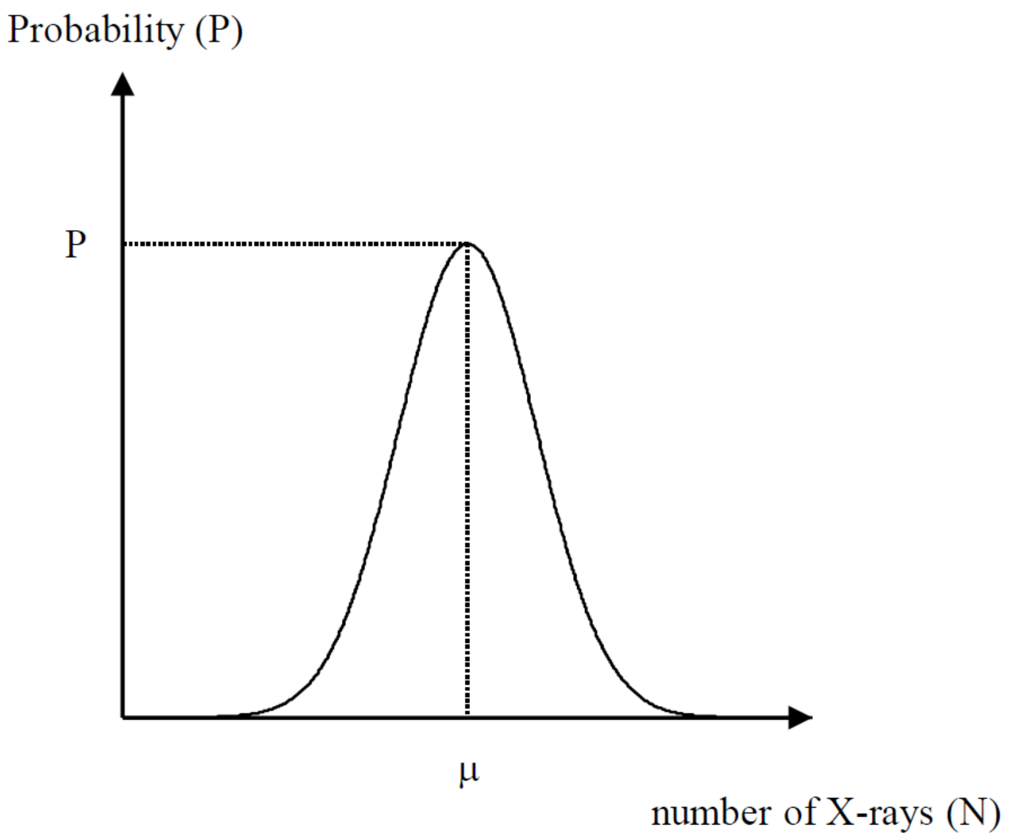


Figure. The probability of a certain number of X-rays striking a unit area of the detector. The statistical uncertainty in this number is represented by the standard deviation (r) of the Poisson distribution.

Signal-to-noise ratio

- $\text{SNR} \propto \sqrt{N}$

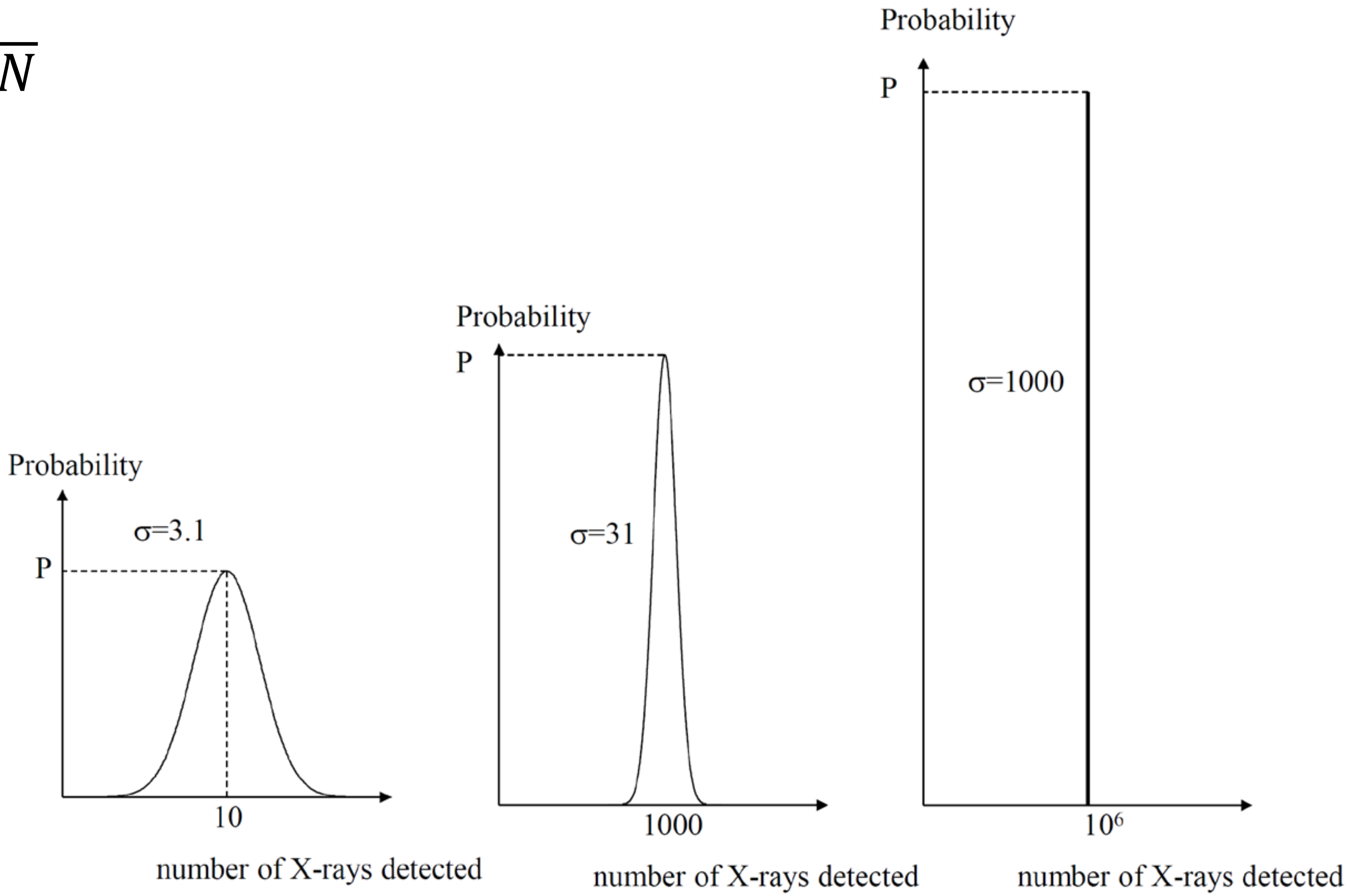


Figure. Plots of a Poisson distribution with increasing number of X-rays produced. As the number increases, the distribution becomes narrower, with the mean number of events and the total number of events converging to the same value.

Factors affecting SNR

- **The tube current and exposure time:** square root of the product of these two quantities
- **The tube kVp:** increase SNR with higher voltage, non-linear effect
- **The patient size and part of the body being images:** the bigger, the lower SNR
- **The geometry of the anti-scatter grid:** large grid ratio, lower SNR.
- **The efficiency of the detector:** detector quantum efficiency (DQE)

$$\mathrm{DQE} = \left[\frac{\mathrm{SNR_{out}}}{\mathrm{SNR_{in}}}\right]^2$$

Signal-to-noise ratio

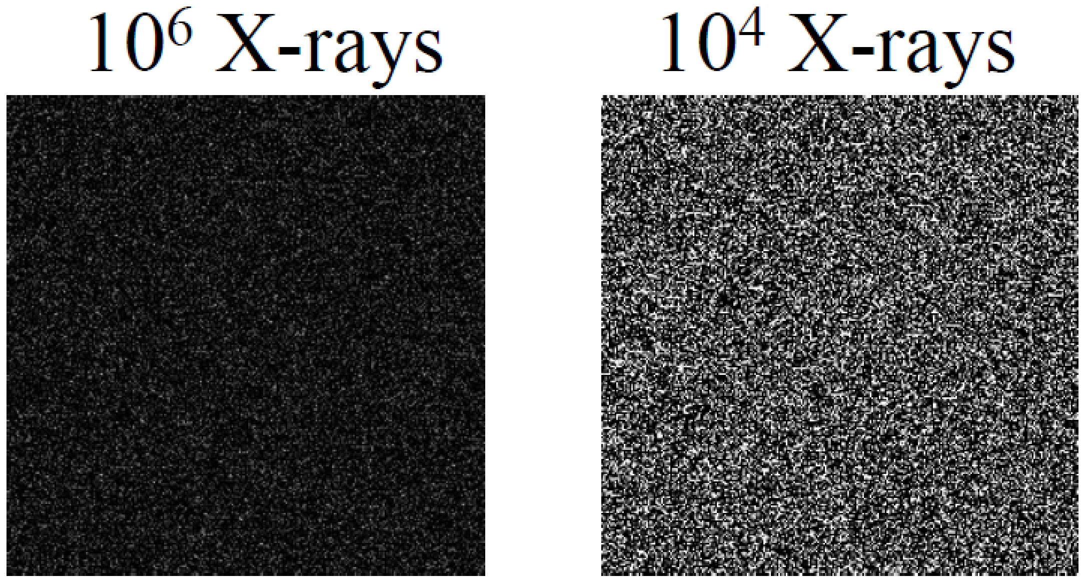


Figure. The effect of radiation dose (total number of incident X-rays) on the SNR of a planar image with no patient in place. The image on the left has 100 times the number of X-rays per pixel, and so has 10 times higher SNR.

Spatial resolution

➢ A combination of the contributions from each part of imaging process:

- Size of focal spot:

$$f = F \sin \theta$$

- Distance between X-ray tube and the patients (S_0) and distance between X-ray tube and detector (S_1)

$$P = \frac{f(S_1 - S_0)}{S_0}$$

- The properties of X-ray detector: number of pixels in the images

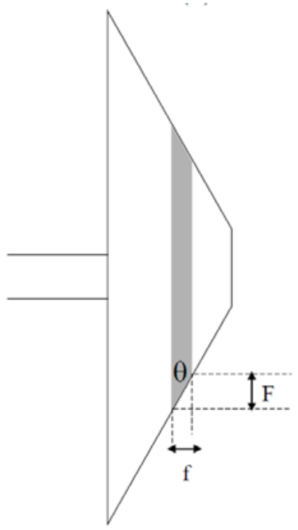


Figure. The effect of the bevel angle h on the effective focal spot size (f) as a function of the width of the electron beam (F).

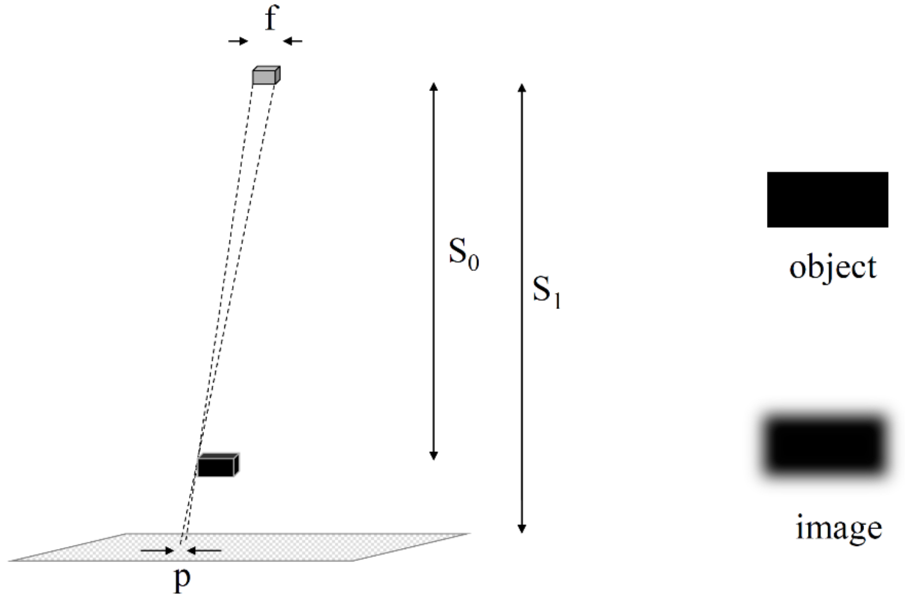


Figure. A finite effective spot size as well as the tube–patient (S0) and tube–detector (S1) distances determine the spatial resolution of the image. The ‘geometric unsharpness’ or penumbra (P) causes features and edges in the image to become blurred, as shown on the right..

Contrast-to-noise ratio

Factors affecting CNR

- Compton scattered X-ray
  - X-ray energy spectrum
  - The field of view (FOV, 视野) of X-ray image — linear in 10-30cm, constant above 30cm
  - The thickness of the body being imaged
  - The geometry of the anti-scatter grid
- SNR: $\mathrm{CNR}_{AB} = |\mathrm{SNR}_A - \mathrm{SNR}_B|$
- Spatial resolution