EUROPEAN SOCIETY OF RADIOLOGY

eBook for Undergraduate Education in Radiology

CHAPTER: Conventional X-ray Imaging

Preface

Undergraduate teaching of radiology in Europe is provided according to national schemes and may vary considerably from one academic institution to another. Sometimes, the field of radiology is considered as a "cross-cutting discipline" or taught within the context of other clinical disciplines, e.g., internal medicine or surgery.

This e-book has been created in order to serve medical students and academic teachers throughout Europe to understand and teach radiology as a whole coherent discipline, respectively. Its contents are based on the *Undergraduate Level of the ESR European Training Curriculum for Radiology* and summarize the so-called ***core elements*** that may be considered as the basics that every medical student should be familiar with. Although specific radiologic diagnostic skills for image interpretation cannot be acquired by all students and rather belong to the learning objectives of the *Postgraduate Levels of the ESR Training Curricula*, the present e-book also contains some ***further insights*** related to modern imaging in the form of examples of key pathologies, as seen by the different imaging modalities. These are intended to give the interested undergraduate student an understanding of modern radiology, reflecting its multidisciplinary character as an organ-based specialty.

We would like to extend our special thanks to the authors and members of the ESR Education Committee who have contributed to this eBook, to Carlo Catalano, Andrea Laghi and András Palkó who initiated this project, and to the ESR Office, in particular Bettina Leimberger and Danijel Lepir, for all their support in realising this project.

We hope that this eBook may fulfil its purpose as a useful tool for undergraduate academic radiology teaching.

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Chapter Outline

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eBook for Undergraduate Education in Radiology

Based on the ESR Curriculum for Undergraduate Radiological Education

Chapter: Conventional X-ray Imaging

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Chapter Outline

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Principles of X-Ray Imaging

X-ray imaging is a major diagnostic technique based on the interactions of X-rays in a body to produce images of organs and tissues.

Three main X-ray imaging modalities are used:

- Projection radiography
- Fluoroscopy
- Computed tomography (CT)

As shown in Fig. 1, these three imaging techniques are based on the:

- Production of X-rays in an X-ray tube
- Transmission of an X-ray beam through a patient
- Detection of the transmitted photons on a detector
- Image processing

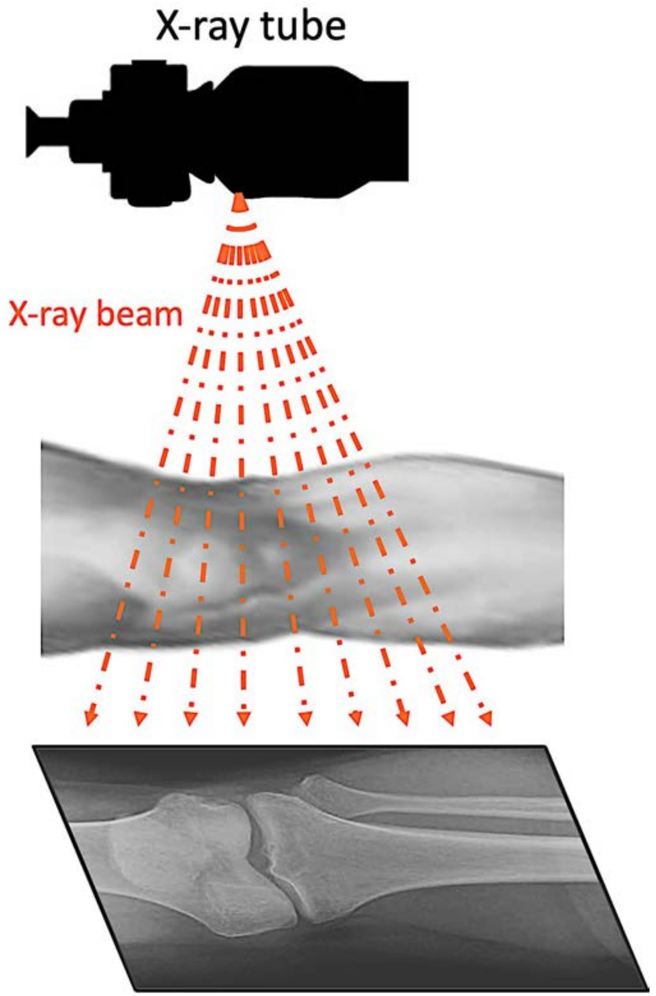

Fig. 1 – Principle of X-ray imaging

Chapter Outline

Principles of X-Ray Imaging

X-ray imaging modalities provide two-dimensional projections or slices of the attenuating properties of the tissues traversed by X-rays.

- Radiography gives a single static projection acquired on an X-ray flash (Fig. 2B).
- Fluoroscopy produces temporal series of projections at an adjustable image rate (0.5 - 30 images per second) and give access to dynamic imaging.
- Computed tomography (CT) acquires single projections at many angles over 360° around the patient to reconstruct slices of the anatomy (Fig. 2C) and volume rendering (Fig. 2D).

This chapter explains the principle of projection X-ray imaging, also called "conventional X-ray imaging".

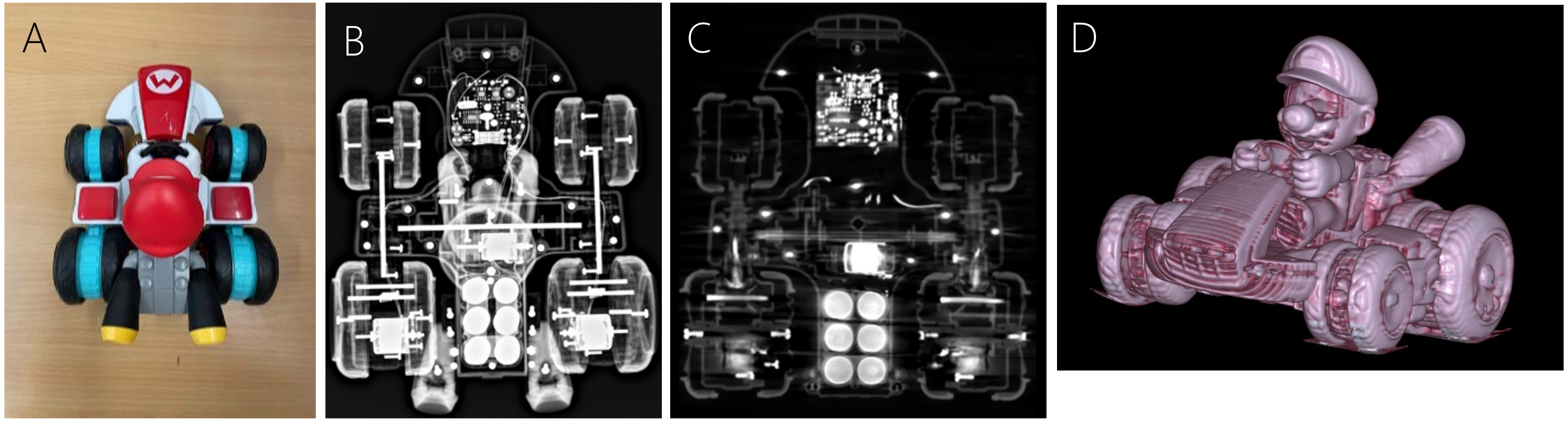

Fig. 2 – Difference between an X-ray projection (B) and a CT slice (C) of an imaged object (A). D illustrates a 3 dimensional reconstruction from the CT slices. Figure courtesy Davide Cabral, Department of Radiology, Geneva University Hospitals.

The X-ray Tube

X-rays are high-frequency electromagnetic waves produced in an *X-ray tube* when highly energetic electrons interact with matter.

Major tube components are (Fig. 3) :

1. Cathode : Negative electrode comprised of an electron emitter and a focusing cup.
2. Anode : Metal target electrode at a positive potential difference relative to the cathode
3. Rotor/stator
4. Glass or metal envelope
5. Tube housing comprising a lead shielding

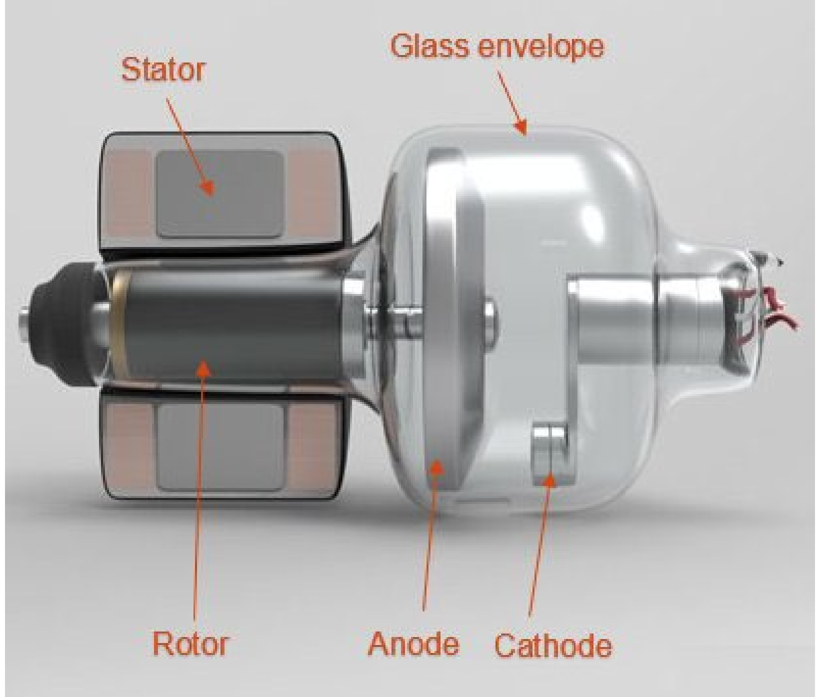

Fig. 3 – X-ray tube

The X-ray Tube

Cathode (Fig. 4) :

- The cathode usually contains tungsten *filaments* electrically connected to the X-ray generator
- Most X-ray tubes are referred to as *dual-focus tubes* because they have two filaments: a large filament and a small filament.
- The small or the large filament can be manually or automatically selected, depending on the voltage (kV) and time-current product (mAs)
- The filament is heated by an electrical resistance
- A static electron cloud is formed around the filament
- When voltage is applied, electrons from the filament are accelerated toward the anode.
- The electrons flux corresponds to the X-ray tube current.

The X-ray Tube

Anode (Fig. 4):

The anode is a metal target electrode maintained at a positive potential difference relative to the cathode.

- Tungsten is the most widely used anode material because of its high melting point (3,000°C) and high atomic number (Z = 74) which provides high X-ray production.
- The anode area impacted by the electrons is the *focal spot*.
- Dental x-ray units, mobile x-ray machines, and mobile fluoroscopy systems use fixed-anodes.
- Rotating anodes allow higher x-ray output by spreading the heat over a larger surface.
- The *actual focal spot size* is the area on the anode struck by electrons, determined by the size of the filament selected in the cathode.
- The *effective focal spot size* is the projection of the actual focal spot size on the image plane, determined by the anode angle.

The X-ray Tube

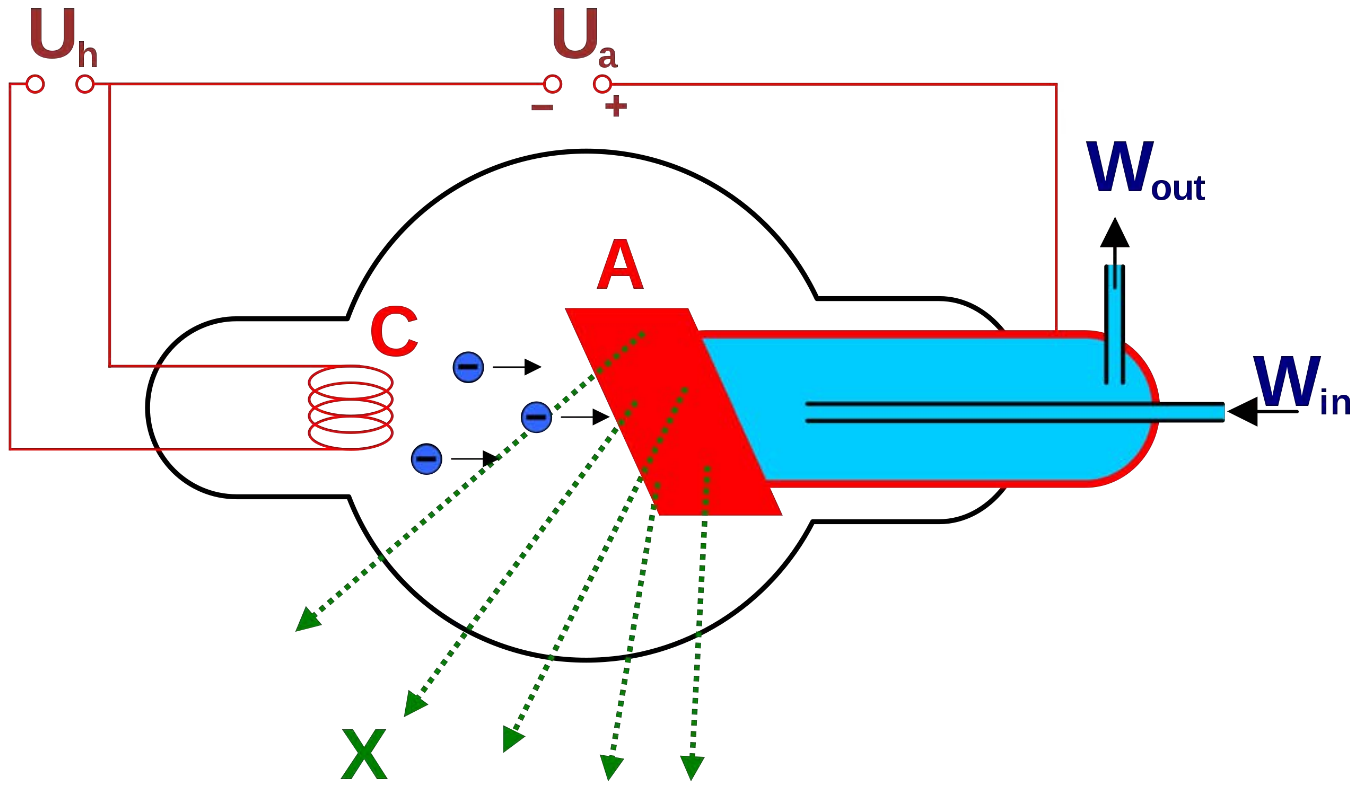

Fig. 4 – Schematic drawing of the original Coolidge side window X ray tube, where electrons are produced by heating a tungsten filament by electric current. C: filament/cathode (-); A: anode (+); W_{in} and W_{out}: water inlet and outlet of the cooling device. U_h: voltage potential for heating the cathode; U_a: voltage potential between anode and cathode. The electrons produced by the cathode are accelerated in the vacuum tube towards the anode. X: X-rays produced by the anode. Image reproduced from: https://commons.wikimedia.org/wiki/File:WaterCooledXrayTube.svg

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X-ray Spectra

- X-rays are produced through two processes: *braking* and *characteristic radiations*.
- *Braking X-rays* are emitted from the anode in a continuous range of energies, the maximum energy being determined by the tube voltage (Fig. 5).
- Electrons can **eject** other electrons from the inner shells of the atoms in the anode. These vacancies are filled when electrons descend from higher energy levels and emit ***characteristic X-rays*** (Fig. 4).
- *Characteristic X-rays* have well-defined energies determined by the difference between the atomic energy levels of the atoms of the anode.
- A **filtration in aluminum** placed at the tube output cuts off low-energy X-rays which would increase the patient dose but would never reach the patient exit nor the detector.

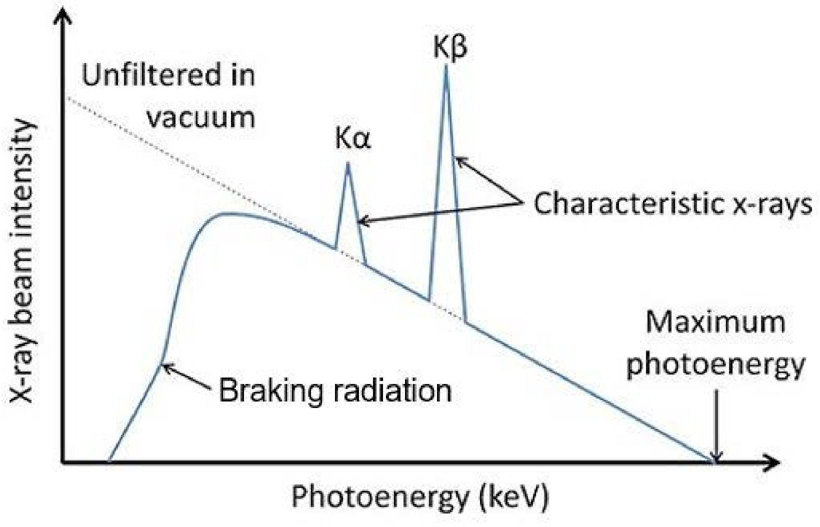

Fig. 5 – Typical X-ray spectrum

Setting Parameters

Focus size (Fig. 6)

- A *small focus* size (small cathode filament) helps reducing geometric blur (penumbra) when magnification is employed.
- A *large focus* size helps reducing motion blurring when high exposure rate (high mA) is needed for shortest exposure times.

Voltage (*kV*)

- X-ray tube high voltage is applied between the cathode and the anode
- The mean energy of the X-ray spectrum and the quantity of X-rays produced increases with the tube voltage.
- The tube high voltage is set between:
  - 40 and 150 kV in standard radiography and fluoroscopy
  - 23 and 40 kV in mammography

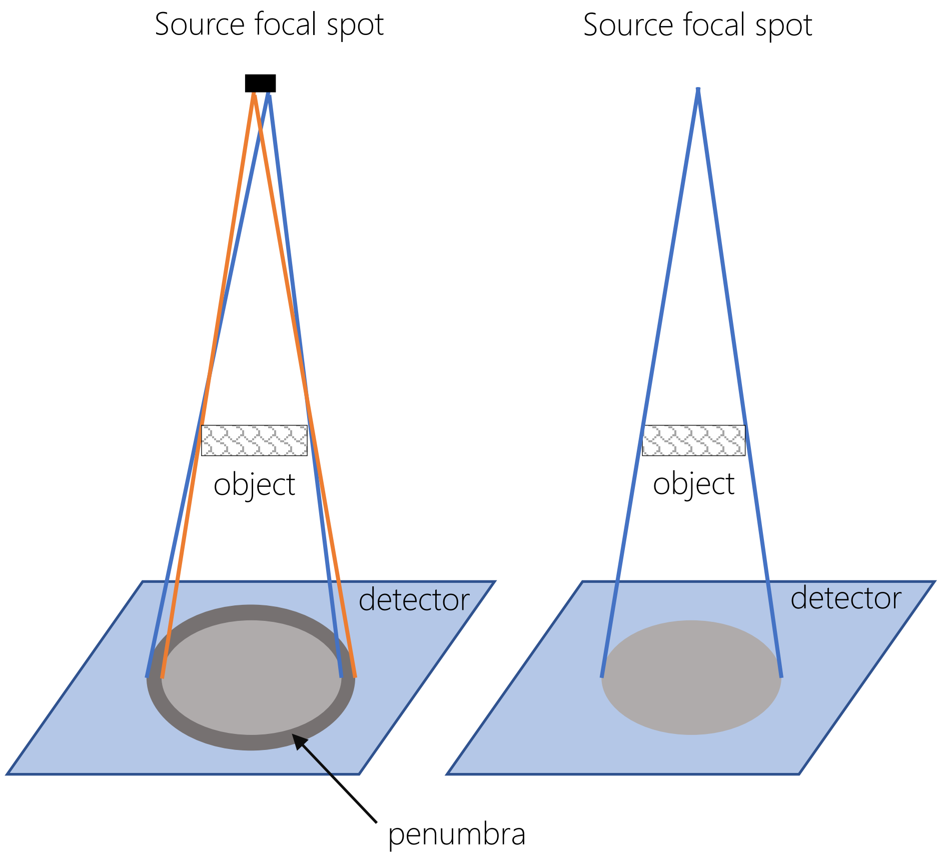

Fig. 6 – Geometric blurring

Chapter Outline

Setting Parameters

Current time product

- The electrical current in the X-ray tube is the charge of electrons per unit time (mA)
- The current time product represents the electrical charge going from the cathode to the anode during the exposure time.

Filtration (Fig. 9)

- X-ray tube filtration absorbs the low-energy X-rays as they only produce an irradiation to the patient and do not reach the detector.
- *Inherent filtration* results from the composition of the tube and housing
- *Additional filtration* consists of aluminum or copper plates of different thicknesses placed between the window and the collimator which can be inserted or removed depending on the imaging protocol.
- *Total filtration* is the sum of all the filtrations

The total filtration is expressed in equivalent millimeters of aluminum and must be at least 2.5 mm aluminum

Interaction of X-rays with Matter

There are 3 outcomes of the passage of X-rays through matter (Fig. 7):

- T: Transmission (no interaction)
- A: Absorption
- S: Scatter

- The absorption of X-rays is caused by the ***photoelectric effect***.
- The photoelectric effect produces the *contrast* in the radiological image.
- It constitutes the basis of X-ray imaging.

- There are two mechanisms for producing scattered radiation:
  - Incoherent scattering : *Compton effect*
  - Coherent scattering : *Rayleigh effect*
- Scattered radiation does not produce contrast in the radiological image. It is an "unwanted effect".

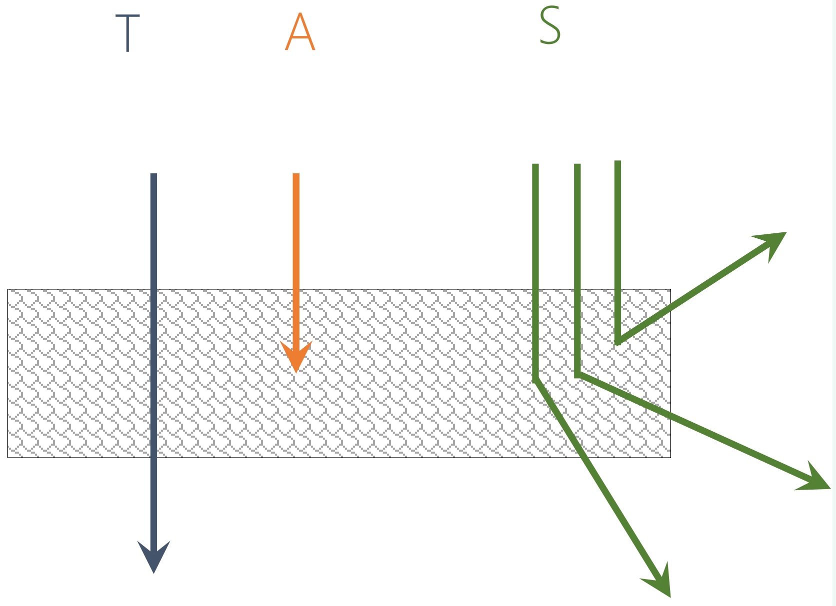

Fig. 7 – Interactions of X-rays with matter

Interaction of X-rays with Matter

In the photoelectric effect (absorption) (Fig. 8)

- An X-ray hits an electron, which is ejected from the atom (photoelectron)
- The X-ray stops and the atom is *ionized*, with an inner-shell electron vacancy.
- The electron vacancy is filled with an electron resulting from a cascade from outer to inner shells.
- The difference in binding energy is released as either *characteristic X-rays* or *Auger electrons*

The probability of a photoelectric effect :

- Decreases with the beam energy, which explains why image contrast decreases with X-ray energy E.
- Increases in materials of high atomic number Z.
- Is approximately proportional to Z^3/E^3.

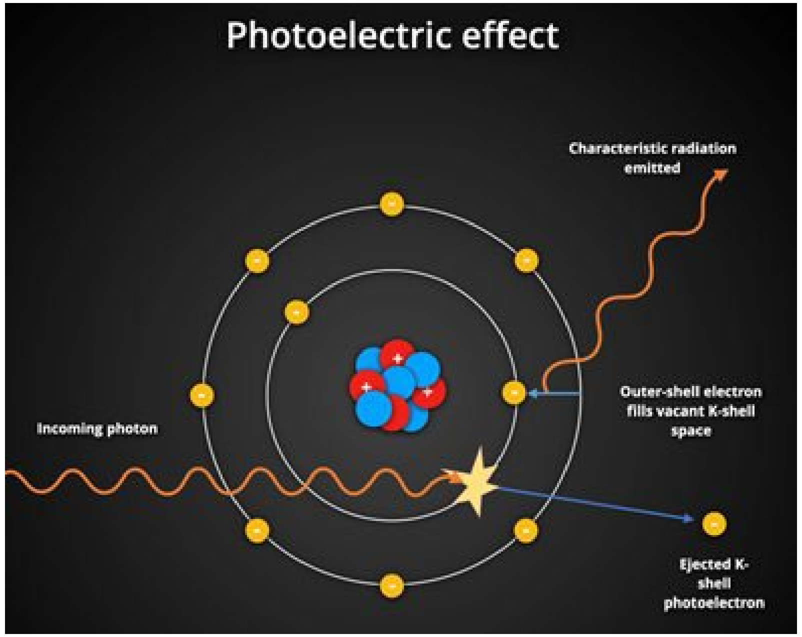

Fig. 8 – Photoelectric interaction

Case courtesy of Frank Gaillard, https://radiopaedia.org/articles/photoelectric-effect

If the photon energies are doubled, the probability of photoelectric interaction is decreased eightfold: $(½)^3 = 1/8$

Chapter Outline

Interaction of X-rays with Matter

In Compton scattering (inelastic scattering)

- An X-ray hits an electron which is ejected from the atom (atom ionized) (Fig. 9).
- A scattered X-ray is emitted at a different angle with respect to the incident photon.
- The scattered X-ray has a reduced energy due to the transfer of energy to the electron.
- The scattered X-ray may undergo subsequent interactions such as Compton or Rayleigh scattering, or photoelectric absorption.
- Compton scattering is the main interaction of X-rays with soft tissues in the diagnostic energy range.

Scattered X-rays degrade image contrast and signal-to-noise ratio.

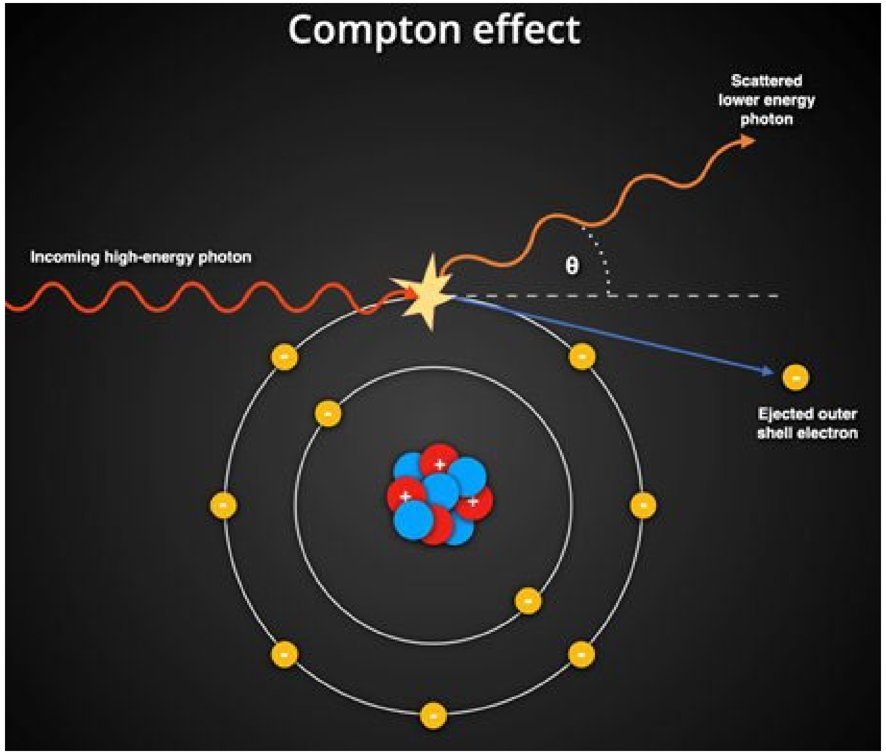

Fig. 9 – Compton interaction

Case courtesy of Frank Gaillard, https://radiopaedia.org/articles/compton-effect

Interaction of X-rays with Matter

In Compton scattering (inelastic scattering)

- At the X-ray energies used in diagnostic imaging (15 - 150 keV), the incident X-ray energy is mainly transmitted to the scattered X-ray.
- The average scattering angle decreases as the X-ray energy increases (Fig. 10).
- The scattering angle of the ejected electron cannot exceed 90°, whereas that of the scattered X-ray can be any value including a 180° backscatter.
- In contrast to the scattered X-ray, the ejected electron is usually reabsorbed near the scattering site.

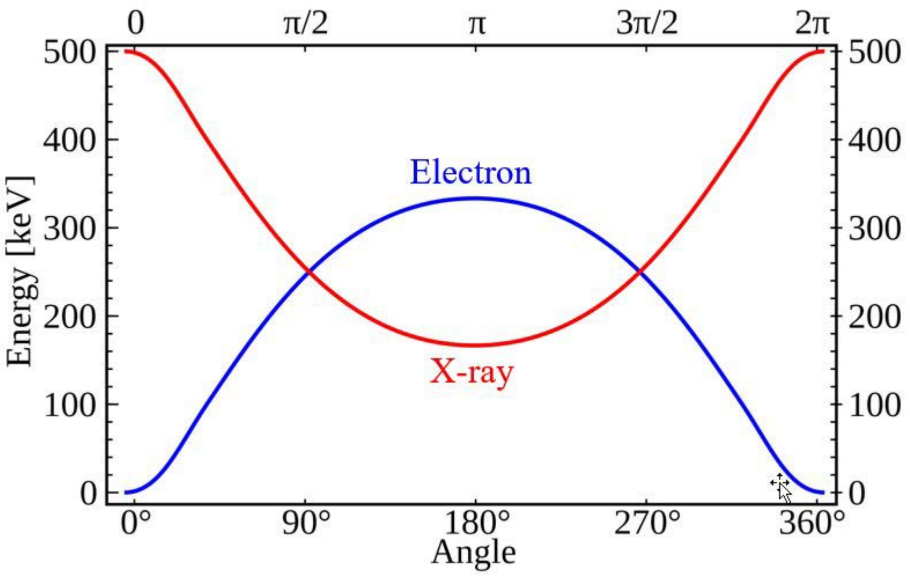

Fig. 10 – Deviation angle of the scattered X-ray and the emitted electron as a function of the incident X-ray energy

Case courtesy: https://en.wikipedia.org/wiki/Compton_scattering

- The probability of Compton scattering is
  - nearly independent of *Z*
  - approximately proportional to the density of the material

Interaction of X-rays with Matter

In Rayleigh scattering (elastic scattering)

- The incident X-ray excites the *total atom*, as opposed to Compton scattering or photoelectric effect .
- Electrons are not ejected, and no ionization occurs.
- This interaction occurs mainly with low energy x-rays, such as those used in mammography (15 - 30 keV).
- The atom's electron cloud in the scattering atom oscillates in phase and immediately radiates this energy, emitting a scattered X-ray of the same energy but in a slightly different direction (Fig. 14).
- The average scattering angle decreases as the X-ray energy increases.

Rayleigh scattering accounts for only 10% of interactions in mammography and 5% in standard radiography.

Chapter Outline

Attenuation of X-rays

- X-rays are attenuated in matter due to absorption and scattering.
- Due to X-ray absorption, the quantity of X-rays decreases exponentially. X-ray absorption depends on matter thickness:

$$N(x) = N_0 \cdot e^{-\mu x}$$

where N and N_0 are the number of X-rays at depth x and at the surface (depth zero) of the traversed matter, and μ is the linear attenuation coefficient, which gives the probability of interaction per unit length of matter in (cm^{-1}) (Fig. 11).

- Most of X-ray attenuation in the diagnostic energy range is due to the photoelectric effect and is proportional to $(Z/E)^3$
- The linear attenuation coefficient depends on each material and:
    - o increases with the atomic number of matter
    - o decreases with the energy of X-rays

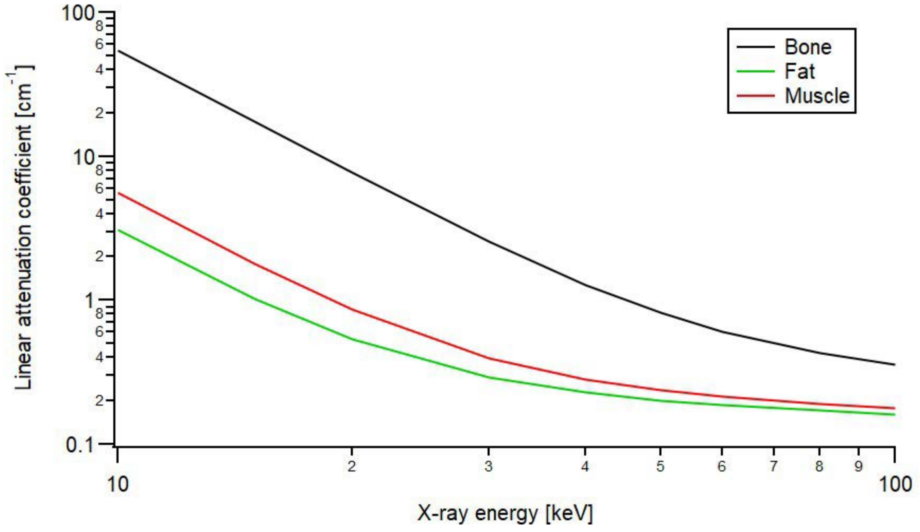

Fig. 11 – Linear attenuation coefficient of bone, fat and muscle for X-rays between 10 and 100 keV

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Half Value Layer (HVL)

- The HVL of an X-ray beam is the thickness of absorbing material that is needed to reduce the beam intensity by half of its initial value (Fig. 12).
- HVL is an indirect measure of the mean beam energy, and is inversely proportional to the linear attenuation coefficient μ :

$$HVL = \frac{ln(2)}{\mu}$$

- Low energy X-rays are stopped faster than high-energy X-rays, causing the mean beam energy of polyenergetic beams to increase in the depth of the traversed material. This effect is called beam hardening.
- Beam hardening causes X-rays to decay in the depth of the traversed matter less rapidly than exponentially.
- The HVL of polyenergetic X−ray beams defines the effective attenuation coefficient :

$$\mu_{eff} = ln(2)/HVL$$

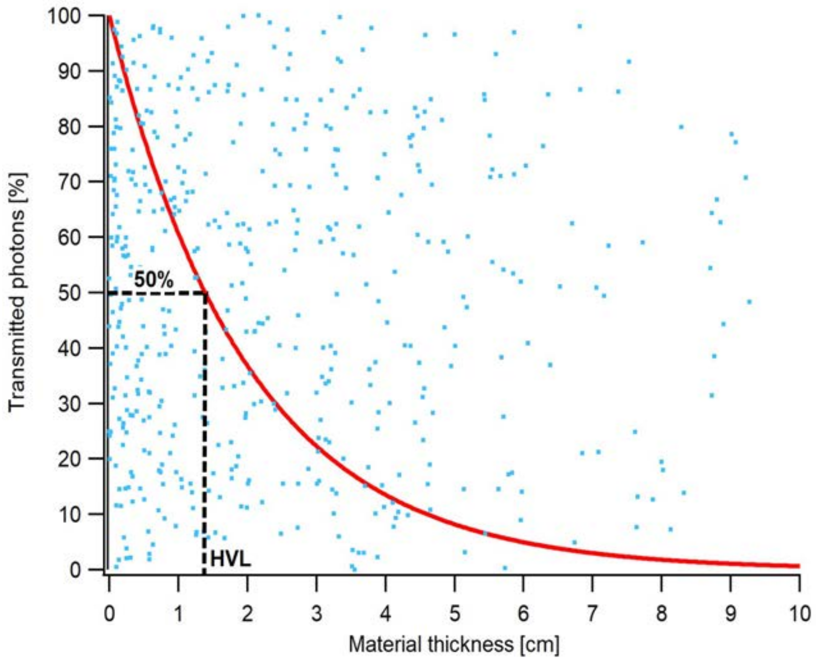

Fig. 12 – Half value layer

Scatter Fraction

- The amount of scatter detected in an image is characterized by the scatter-to-primary ratio (SPR) or the scatter fraction (SF), which is expressed as a percentage.

- The SF increases with the volume of tissue irradiated by the X-ray beam
  - o with the beam size (field of view)
  - o with the thickness of the patient

- For a typical 30 × 30 cm^2 abdominal irradiation in a 25-cm-thick patient, the SF is about 80% (Fig. 13).

The contrast in the image is inversely proportional to the SF, and a scatter rejection technique must be used.

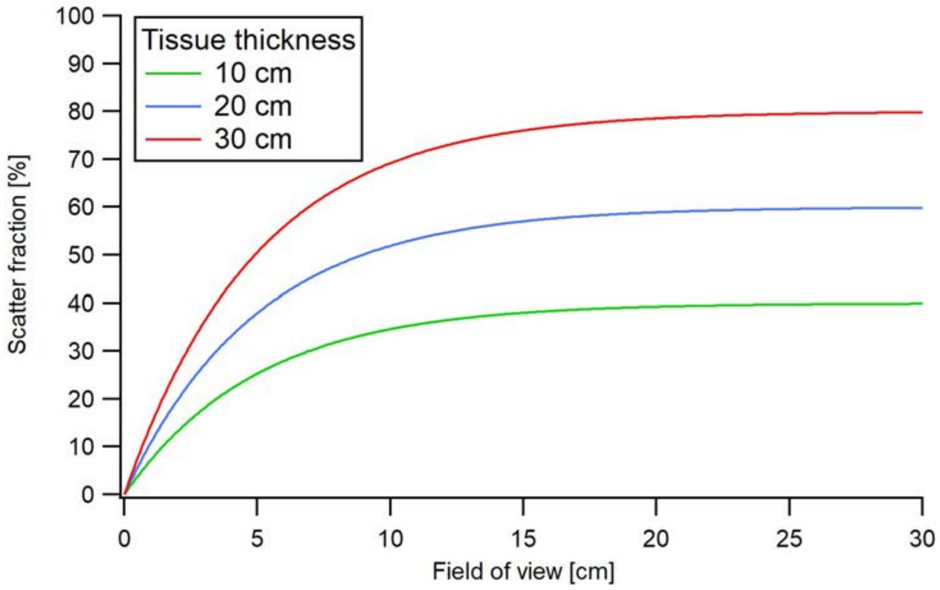

Fig. 13 – Scatter fraction of X-ray beams for different field of view and three patient tissue thicknesses

Chapter Outline

Scattered Radiation Rejection

- Rejection of X-rays scattered in the patient is based on their oblique orientation relative to primary X-rays.
- Rejection of scatter is important for enhancing contrast in projection radiography.

Anti-scatter grid (Fig. 14)

- The anti-scatter grid, placed between the patient and the detector, is the most widely used technology for reducing scatter in radiography, fluoroscopy, and mammography.
- Grids are typically manufactured with lead strips oriented along one dimension separated by a low attenuating interspace material such as carbon fiber or aluminum.
- Parallel grids have lead strips that are focused to infinity
- Focused grids have lead strips oriented towards the focal point of the grid, located at the focal distance of the grid.
- Typical focal distances are 100, 150 and 180 cm.

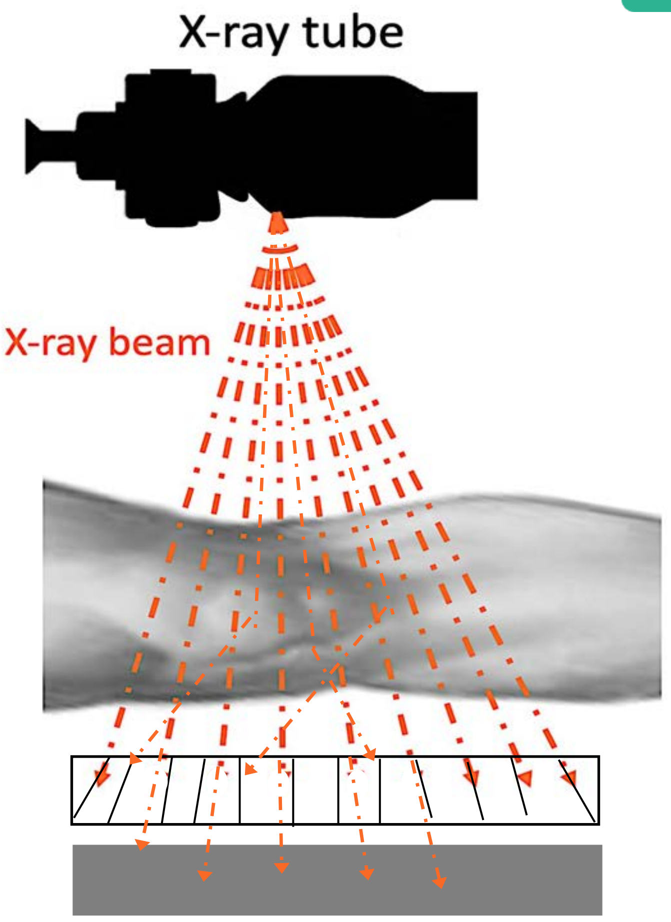

Fig. 14 – Scatter rejection in front of the detector

Scattered Radiation Rejection

Anti-scatter grid

- The grid attenuates some of the primary X-rays that are incident directly on the lead strips.
- The transmission of primary X-rays through the grid is the primary transmission (T_p).
- The grid allows transmission of some scattered X-rays that have a small scattering angle, or scatter in a direction parallel to the lead strips.
- The transmission of scattered X-rays through the grid is the scatter transmission (T_s).
- The primary and scatter transmissions of the grid determine the grid efficiency, quantified by the grid selectivity Σ.

- Grids are characterized by the parameters (Fig. 15) :
    - o Grid ratio r: the ratio between the height of the lead strip to the interspace distance
    - o Grid frequency f: the number of grid lines per cm
    - o Grid focal distance: The distance to the focal point

$$\Sigma = \frac{T_p}{T_s}$$

$$r = \frac{h}{D}$$

$$f = \frac{1}{t + D}$$

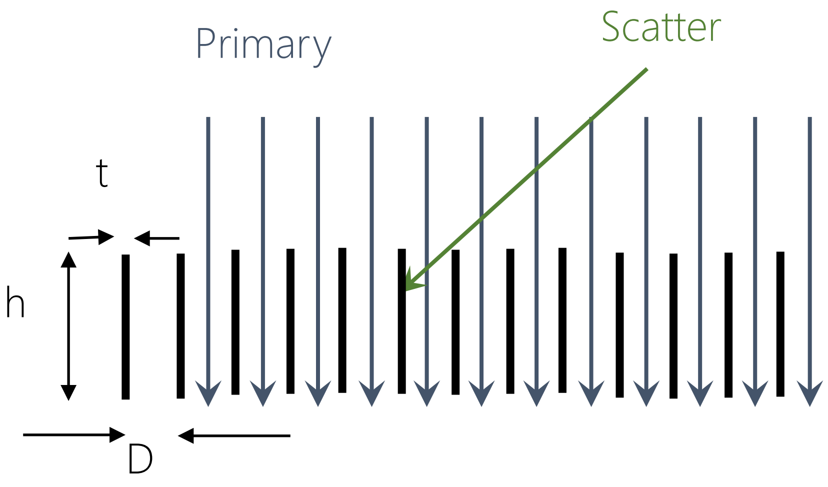

Fig. 15 – Grid characteristics

Chapter Outline

Scattered Radiation Rejection

Anti-scatter grid

- The total transmission (T_t) of the grid depends on T_p, T_s and the scatter fraction (SF).
- The grid factor or bucky factor, the inverse of T_t, is the increase in patient dose when using a grid compared to not using a grid to match the same detector dose.
- A grid with a higher grid ratio has a lower T_s, due to a more limited transmission angle, but also a lower T_p, and a high contrast improvement factor (Fig. 16).
- The focal range is an indicator of the flexibility of grid positioning distance from the focal spot, and is a function of the grid ratio and frequency.

- Grid artifacts arise from bad grid positioning:
  - tilting the grid to the incident X-ray beam
  - bad centering to the beam central axis
  - using a focused grid outside the specified focal range

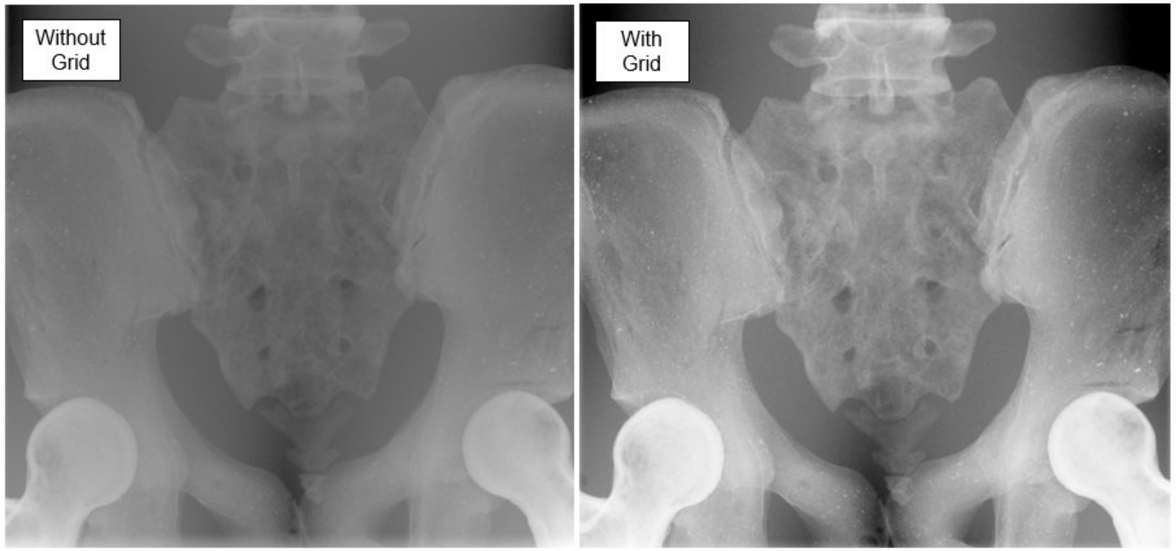

Fig. 16 – Contrast improvement due to the grid

Scattered Radiation Rejection

Air gap (Fig. 17)

- An air gap distance between the patient and the detector lets the X-rays scatter out of the image field of view.
- The scattered x-ray intensity decreases with the air gap distance.

- Practical factors limit the use of the air gap :
  - magnification of the patient anatomy
  - increased geometrical blurring due to the focal spot size
  - may require extending the focus-to-detector distance to decrease magnification, while increasing exposure time and motion blur

- The advantages of the air gap over the grid:
  - a primary transmission at 100%
  - a lower increase in patient dose
  - a higher efficiency (selectivity) for median scatter fractions

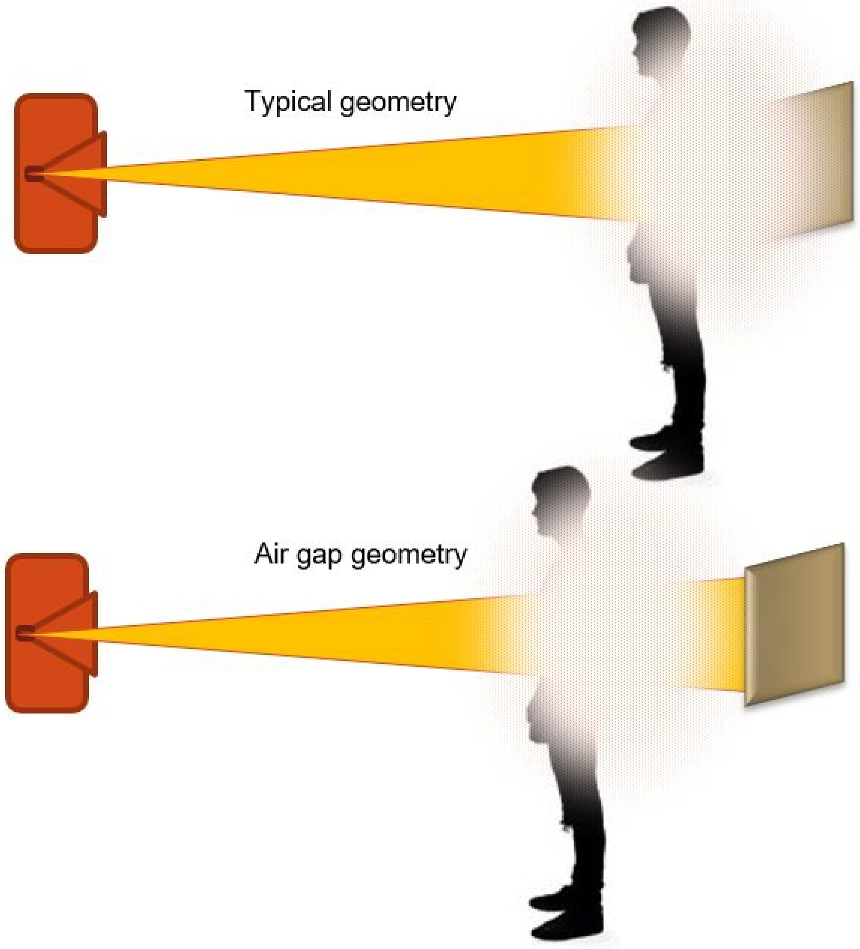

Fig. 17 – Principle of the air gap

Projection

- Conventional radiological image formation results from the projection of the differential transmission of the primary X-ray beam through the different anatomic tissues.
- Two different processes contribute to beam attenuation: absorption and scattering.
- The transmission characteristics of the anatomic parts are determined by their
    - thickness
    - linear attenuation coefficient
- The X-ray transmission T decreases exponentially with the linear attenuation coefficient μ and the thickness x of the irradiated tissue: $$\mathrm{T} = e^{-\mu x}$$
- The output radiation modulation that interacts with a detector is the **latent image**.
- Once detected, on a radiographic image (Fig. 18):
    - A strongly attenuating tissue is bright
    - A low attenuating tissue is dark

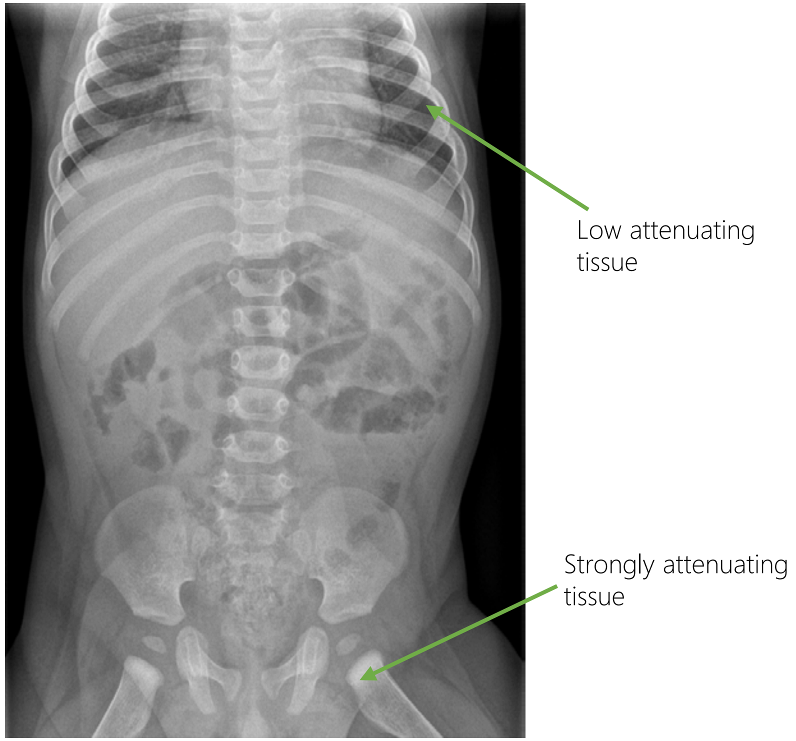

Fig. 18 – A conventional radiograph is a projection of the differential transmission of X-rays in the tissues

Projection

Magnification:

- The focal spot creates a divergent beam in which X-rays travel in straight line.
- The patient-to-detector distance in the divergent projection creates a magnification of the anatomy on the image.
- The magnification M is defined as (Fig. 19):

$$\mathrm{M} = \frac{Image\ size}{Object\ size} = \frac{SID}{SOD}$$

- Low magnification occurs for:
  - A long source-to-detector distance
  - A short object-to-detector distance

The size of the object on projections also strongly depends on the orientation of the object relative to the detector plane!

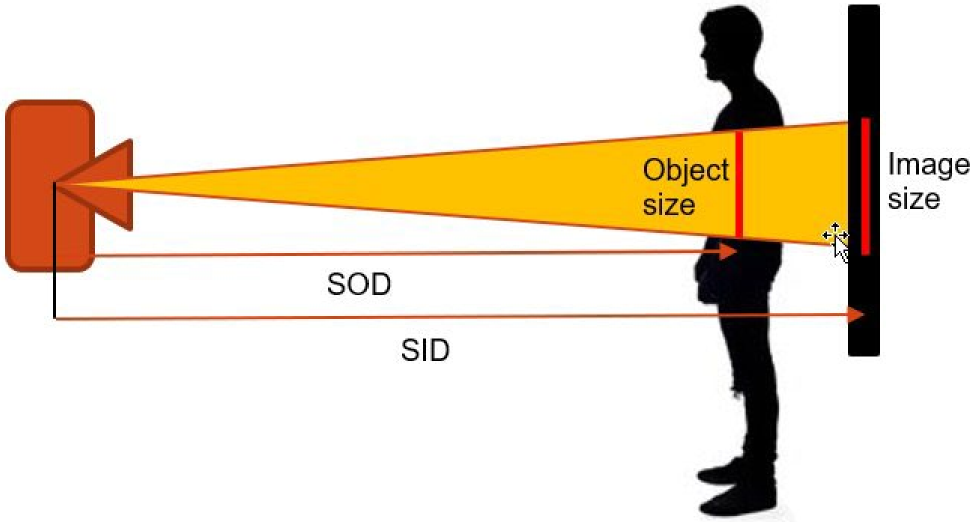

Fig. 19 – Magnification factor in conventional radiography.
SOD = Source Object Distance
SID = Source Image Distance

Projection

Geometrical blur:

- The finite size of the focal spot is the reason why the X-rays "are coming out" from an area and not just from a single point.
- As the X-rays come from the whole area of the focal spot, a **penumbra** appears on the edge of objects.
- The larger the size of the focal spot, the greater the blur will be on the detector.

- The geometrical blur depends on the size of the focal spot, and the system geometry (Fig. 20):

$$B = f \cdot \frac{OID}{SOD} = f \cdot (M - 1)$$

- Low geometrical blur occurs for:
  - o A small focal spot size
  - o A long source-to-detector distance
  - o A short object-to-detector distance

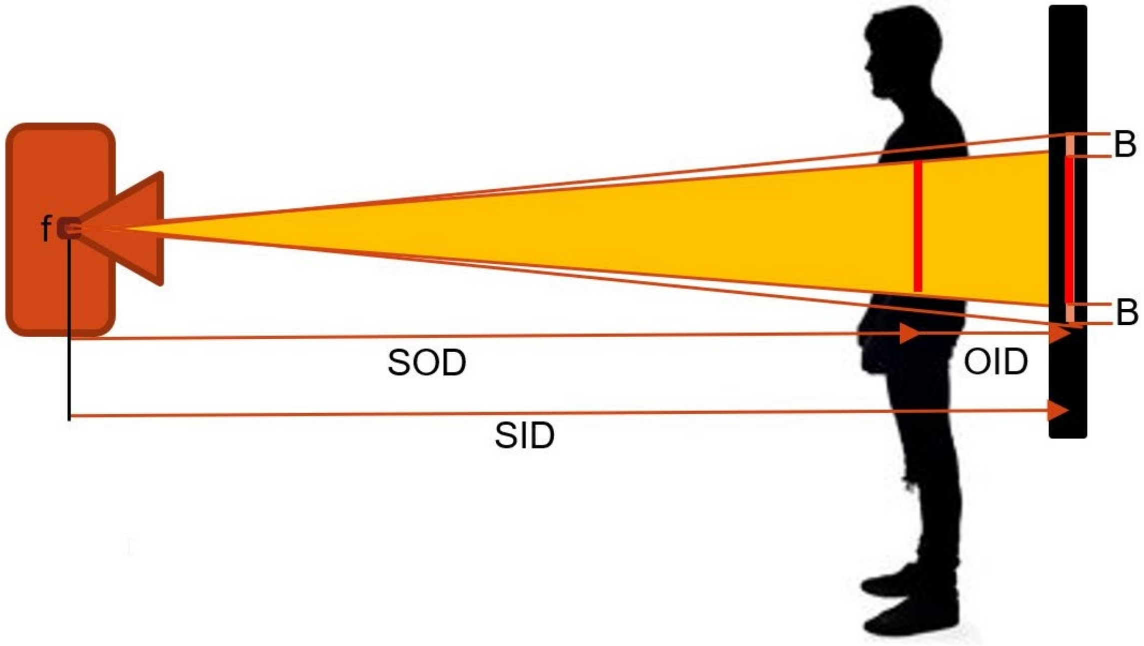

Fig. 20 – Geometrical blur (B) in conventional radiography (penumbra).
SOD = Source Object Distance
SID = Source Image Distance

Digital X-ray Detectors

- Flat panel detectors (FPD) make an electronic conversion of X-rays to digital signal on a flat area
- Two technologies are used: indirect and direct detectors

Indirect FPD

- A scintillator converts X-rays to light
- The scintillator is coupled to a matrix of photodiodes made with amorphous silicon (a-Si)
- Electric charges of the photodiodes are collected by thin-film transistors (TFT) for signal processing of each pixel
- Light in the scintillator is spread, which causes degradation of the spatial resolution (Fig. 21).
- Columnar-shaped caesium iodide (CsI) scintillators have been used to reduce light spread, thus allowing to obtain ultra-high resolution images.

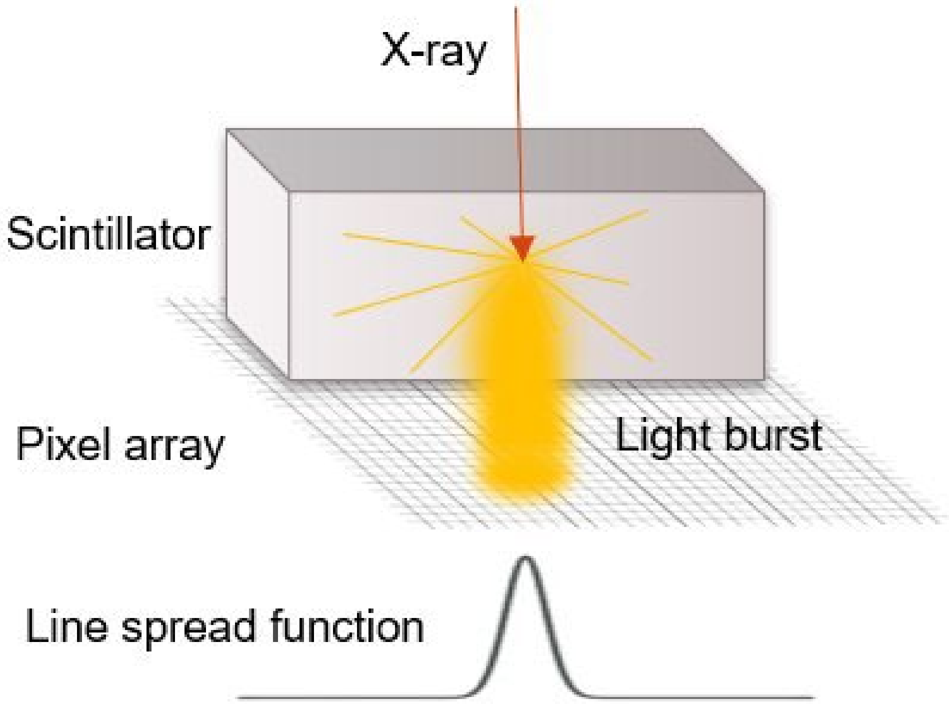

Fig. 21 – Light spread in the scintillator

Digital X-ray Detectors

Direct FPD

- A semiconductor in amorphous selenium (a-Se) directly converts X-rays to electrons
- Electric charges are collected on a capacitor matrix coupled to a TFT for signal processing of each pixel
- Direct conversion gives a high spatial resolution (Fig. 22).
- The detection efficiency of a-Se (Z=34) is low at high energy.
- For this reason, direct conversion detectors are used mainly for mammography.

New X-ray detectors, solid-state C-MOS imagers or GEM (gas detectors), have paved the way for photon-counting imaging, in which each photon is detected individually and its energy is estimated with high efficiency and without image noise.

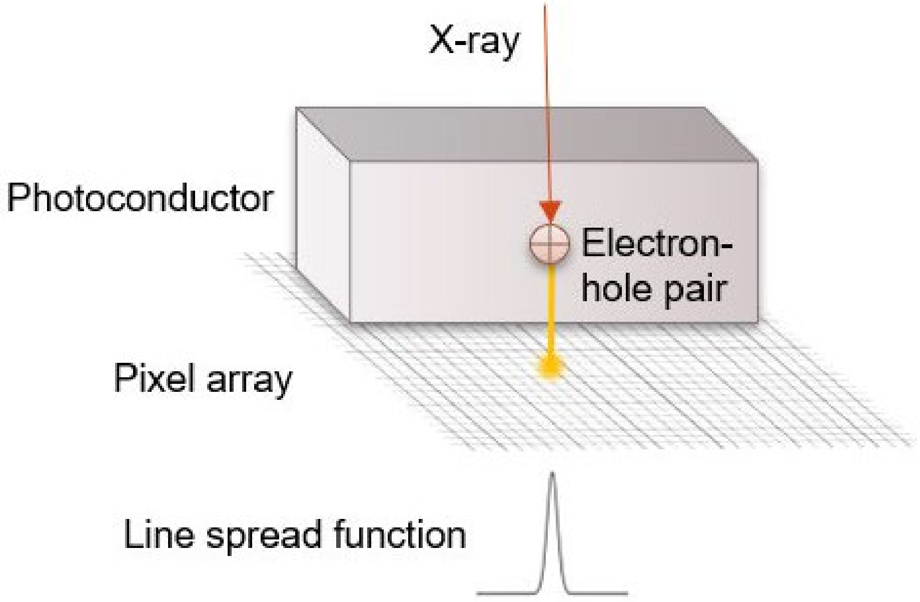

Fig. 22 – High spatial resolution of direct conversion detectors

Chapter Outline

Automatic Exposure Control (AEC)

- The automatic exposure control (AEC) device controls the radiation dose reaching the detector by regulating the length of exposure.
- The AEC device terminates the exposure when the detector's target dose is reached.
- Most of AEC systems consist of three or five radiation-measuring sensors, two lateral and one central, as shown in Fig. 23.
- The radiographer selects the configuration of the three AEC sensors, determining which of the three individually or in combination measures the detector dose.
- The detector's target dose of AEC devices can be adjusted using the control panel buttons numbered −2, −1, 0, +1, +2, …
- The anatomic area of interest must cover the selected detectors, in order to avoid over or under exposition.

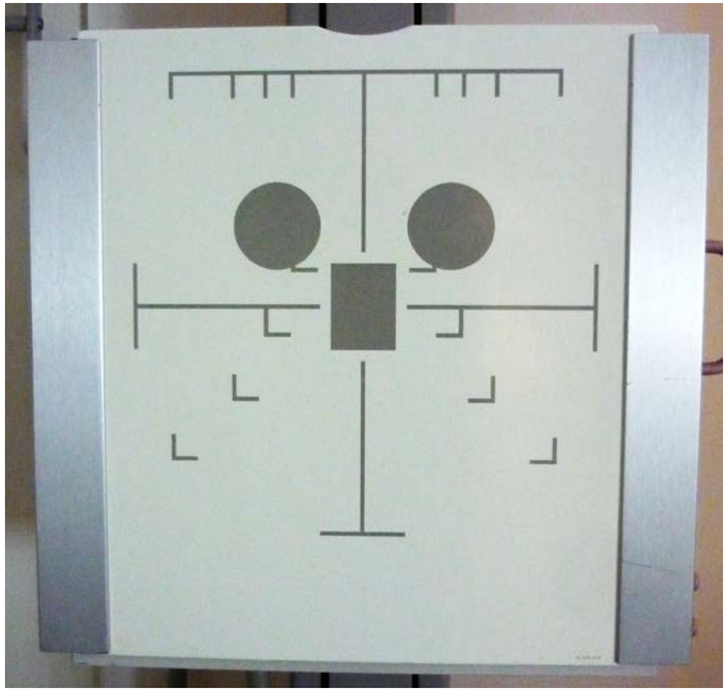

Fig. 23 – Three cells of the automatic exposure control system. Source: Automatic exposure control - Wikipedia

Exposure Index (EI)

- The exposure index gives the user feedback about the detector dose of a radiography.
- The exposure index is calculated from signals in the acquired image itself, and describes the detector dose.

EI is not an equivalent for patient entrance exposure !

- For some systems, the exposure index can scale differently to dose quantities.
- The definition of exposure index was standardized in 2008*:
  - ➢ EI = 100 x detector dose in μGy
- Different combinations of patient body constitutions and exposure can result in the same detected signal and EI.
- Variation in EI can occur due to varying imaging content, even if the same exposure setting was used and patient entrance exposure was the same.

*Medical electrical equipment—exposure index of digital X-ray imaging systems—Part 1: definitions and requirements for general radiography. International Electrotechnical Commission (IEC), International Standard IEC 62494-1:2008-08, Geneva, Switzerland (2008)

Chapter Outline

Air Kerma of the X-ray Beam

- X-rays deposit energy in matter, and hence dose, through interactions with electrons, making ionizations.
- The energy of X-rays transferred to electrons is transformed into kinetic energy.
- These energetic electrons interact with other electrons in matter, depositing most of their energy in a very small volume.
- The air kerma (Kinetic Energy Released in Matter) is the energy transferred from non charged particles to charged particles, divided by the mass of air in the measurement volume.
- The air kerma unit is J/kg, called Gray (Gy).
- The air kerma (K_a) of an X-ray beam (Fig. 24) depends on the current time product (Q), the voltage squared (U^2), the inversed square distance to the source (d^{-2}) and the X-ray tube constant C as follows :

$$K_a = C \cdot \left(\frac{U}{100}\right)^2 \cdot \frac{Q}{d^2}$$

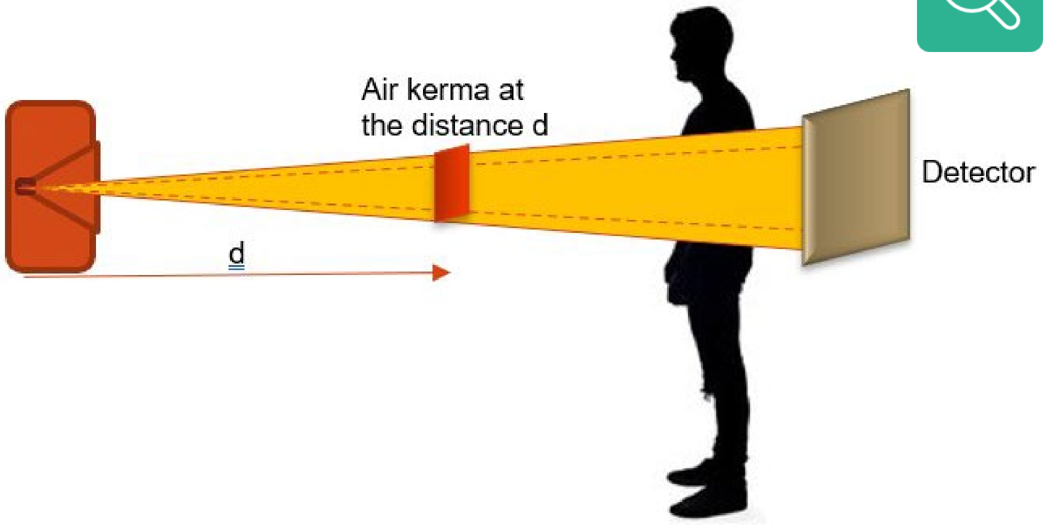

Fig. 24 – Air kerma

The air kerma is defined at a point in the air and does not take into account scattering or beam size. The main use of air kerma is to estimate the peak skin dose in interventional radiology (see next slide). It is of little interest in conventional radiology , where skin doses are below the threshold for deterministic effects.

Chapter Outline

Entrance Skin Dose (ESD)

- The entrance skin dose (ESD) is the dose to a thin layer of skin as the X-rays reach surface of the patient (Fig. 25).
- Backscattered radiations contribute to the ESD in addition to the primary radiation coming from the x-ray tube.
- ESD will therefore be larger by 15-30% than the air kerma at the patient surface.
- The ESD is especially important in prolonged or high dose-rate radiological examinations, such as in interventional procedures using fluoroscopy, because it is related to skin injury.
- The contribution of backscattered radiation to ESD is modelled by the backscatter factor (BSF).rays

$$\mathrm{ESD} = BSF \cdot C \cdot \left(\frac{U}{100}\right)^2 \cdot \frac{Q}{FPD^2}$$

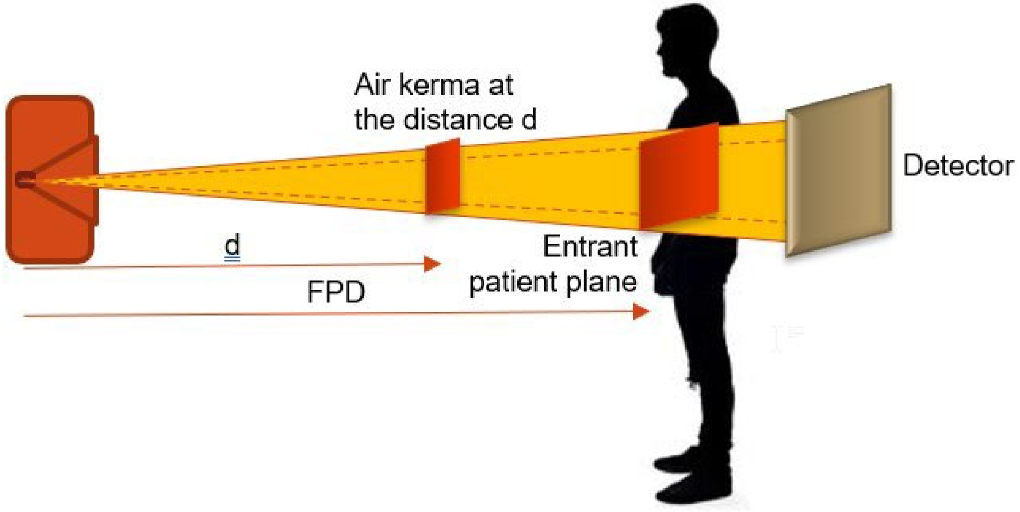

Fig. 25 – Entrance skin dose

The BSF and the ESD depend on beam size and patient thickness.

The ESD is used in plain radiography to establish diagnostic reference levels (DRLs).

DRLs constitute a benchmark for the optimization of radiation protection in medical imaging using X-rays. The Nuclear Safety Agency regularly updates and publishes ESD recommendations.

Dose Area Product (DAP)

- The **dose area product (DAP)**, expressed in mGy×cm^2, is the product of the air kerma (K_a) and the exposed area (A) over a uniformly exposed area.
- The DAP provides a good estimation of the **total radiation dose** delivered to a patient during a radiological procedure.
- The DAP is **independent of the distance to focus**, which facilitates the comparison of measured DAP values in dose survey studies (Fig. 26).
- The DAP is the most commonly used quantification parameter for **monitoring the radiation dose** delivered to patients.
- Radiographic and fluoroscopic systems are equipped with **DAP meters** which measure the DAP at the tube output for each radiological procedure.

The DAP meters at the tube output don't take into account the contribution of scattered radiation to patient dose.

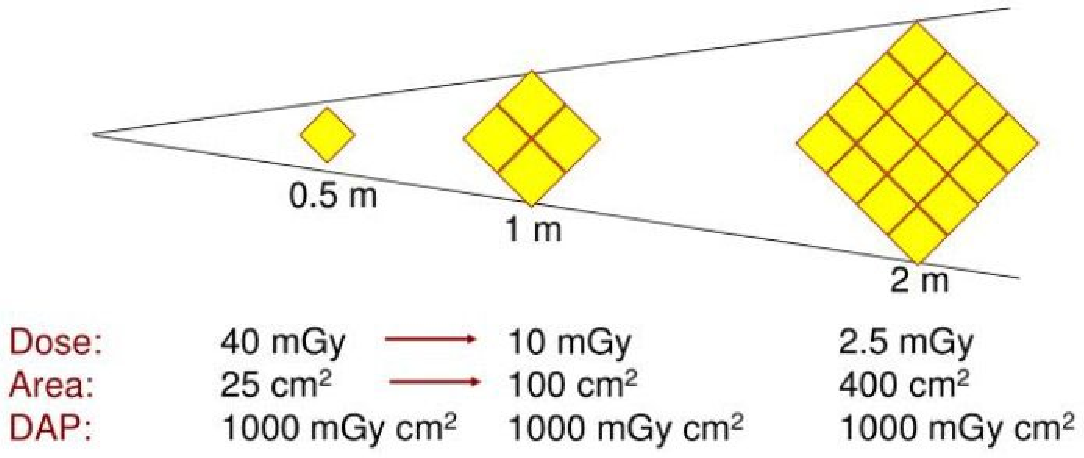

Fig. 26 – Invariance of DAP to the distance from the focus

Diagnostic Reference Levels (DRL)

- The **dose reference level (DRL)** of a radiological examination (e.g., a chest X ray) is the third quartile of the dose distribution reported in a sample of patients (Fig. 27).
- The dose for an X-ray examination may vary depending on the **body mass index (BMI)** of the patient, the **type of detector**, the type of X-ray system and **its settings**.
- **National DRLs** have been established for standard radiology, computed tomography, and for image-guided and interventional procedures.

DRLs give an indication of the expected radiation dose received by an average-sized patient undergoing a given X-ray based imaging procedure.

DRLs are a tool to optimize medical imaging procedures using ionizing radiation.

DRL are <u>not</u> dose limits.

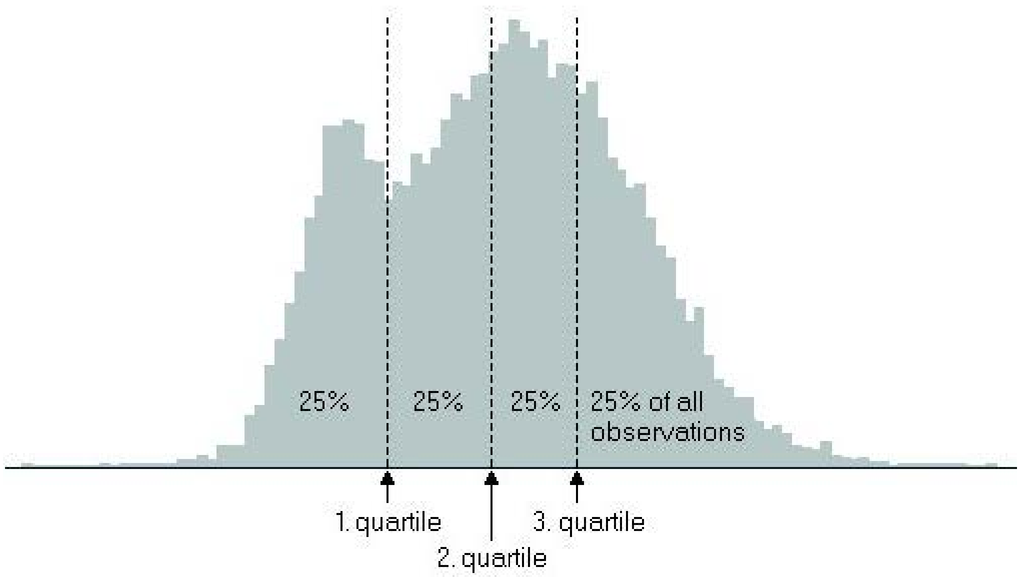

Fig. 27 – DRL is the third quartile of the dose distribution for a radiological examination obtained in a sample of patients.

Contrast

- **The contrast of a radiological image** quantifies the difference of signal among tissues of different densities (Fig. 28).
- The radiographic contrast of a structure is the product of two factors:
  - Difference in X-ray attenuation between tissues
  - Difference in thickness between tissues
- The radiographic contrast decreases with
  - The mean energy of the X-ray beam (tube voltage, additional filtration)
  - The scatter fraction of the X-ray beam on the detector
- The contrast of a digital X-ray can be changed using image processing that changes the histogram of the image.

> Display contrast can be changed by display window setting

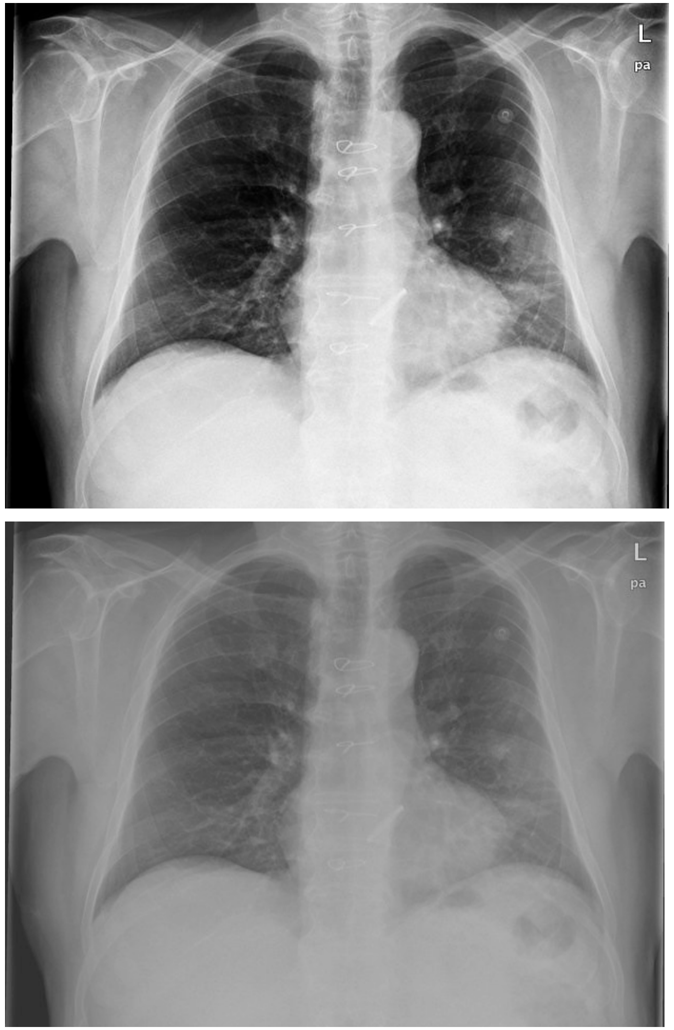

Fig. 28 – Chest X-ray. The same image. Top: high contrast. Bottom: Low contrast

Chapter Outline

Signal-to-Noise Ratio (SNR)

- The signal-to-noise ratio (SNR) of an image is the ratio between the signal and the noise.
- The signal is the mean pixel value, which is related to the number of X-rays converted into a signal by the detector.
- The noise is the level of random variations of pixel values around the mean pixel value, quantified by the standard deviation of pixel values in an homogeneous area.
- The SNR reflects the apparent image noisiness (Fig. 29).
- The SNR of an X-ray increases when increasing
  - The detector dose
  - The pixel size
  - The detective efficiency of the detector
- The SNR can be increased using image processing that decreases the frequency bandwidth of the image.
- The SNR of an X-ray image is a compromise between
  - Patient dose (mAs and anatomical thickness)
  - Spatial resolution (pixel size and image processing)

Fig. 29 – Chest X-ray. Same image. Top: high SNR. Bottom: Low SNR

Chapter Outline

Spatial Resolution

- The spatial resolution of an image **quantifies the sharpness of the signal in the image** (Fig. 30).
- The spatial resolution of an X-ray decreases when increasing:
  - The pixel size
  - The magnification
  - The focal spot size
  - The irradiation time (motion blur)
- The spatial resolution can be increased using image processing that increases the frequency bandwidth of the image.
- The spatial resolution of an X-ray is a compromise with
  - SNR (pixel size and image processing)

A greater source-to-detector distance decreases the geometric blur but increases the exposure time and motion blur.

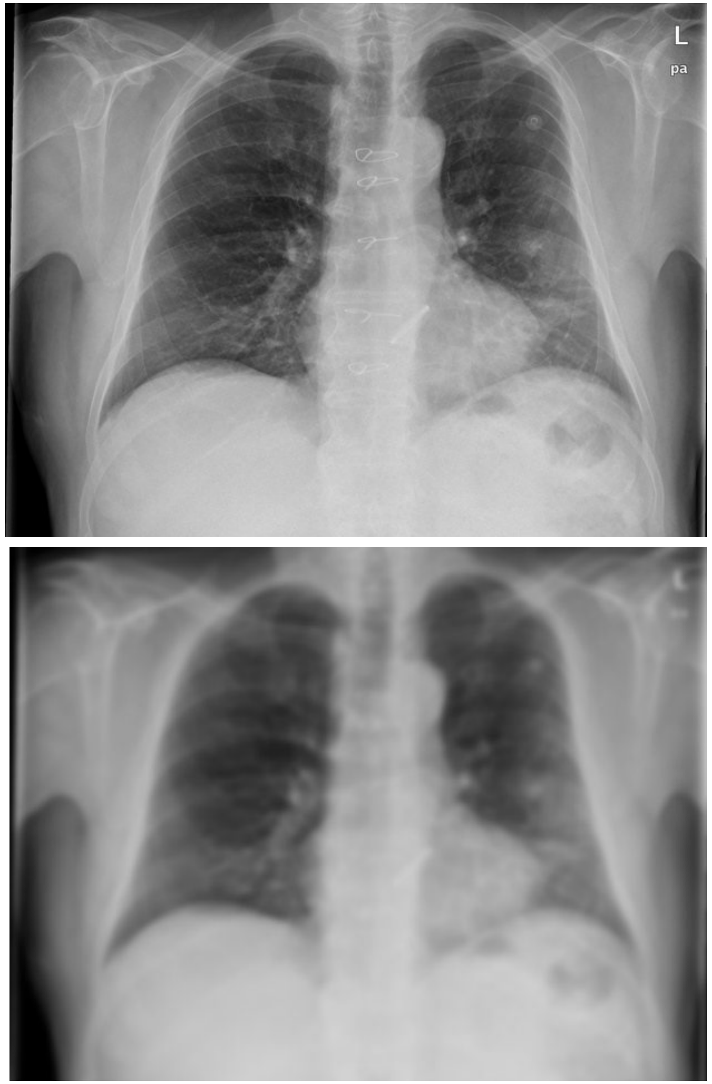

Fig. 30 – Chest X-ray. Top: high resolution. Bottom: Low resolution

Chapter Outline

Take-Home Messages (1):

- X-rays are high-frequency electromagnetic waves produced in an X-ray tube.
- Conventional X-ray imaging modalities provide projections of the attenuating properties of tissues traversed by X-rays.
- The voltage and the current time product set roughly the energy and the quantity of X-rays.
- There are 3 outcomes of the passage of X-rays through matter: transmission, absorption and scatter.
- Rejection of X-rays scattered in the patient is done by an anti-scatter grid or an air gap between the patient and the detector.
- Radiological image formation results from the projection of the differential transmission of the X-rays transmitted through the different anatomic tissues.
- Magnification and blur occur in X-ray imaging because the X-ray beam is divergent and spreads out from a focal spot of finite size.
- Radiological digital detectors convert X-rays to electric signals.

Take-Home Messages (2):

- The automatic exposure control (AEC) device controls the radiation dose reaching the detector by regulating the length of exposure.
- The exposure index gives the user feedback about the detector dose of a radiography.

- The air kerma is the dose to a point in the air in the X-ray beam and does not take into account scattering or beam size.
- The entrance skin dose is the dose to a thin layer of skin at the entrant plane of the patient.
- The dose area product is the product of the air kerma and the exposed area and gives an estimation of the radiation dose delivered to a patient during a radiological examination.
- Three parameters define image quality : contrast, signal-to-noise ratio and spatial resolution.

Chapter Outline

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Chapter Outline

Test Your Knowledge

1 – What is the effective spot size ?

- The area on the anode struck by electrons
- The projection of the focal spot size on the image plane
- The size of the electrons beam on the anode
- The size of the filament selected in the cathode

Chapter Outline

Chapter Outline

Test Your Knowledge

1 – What is the effective spot size ?

- The area on the anode struck by electrons
- ✓ The projection of the focal spot size on the image plane
- The size of the electrons beam on the anode
- The size of the filament selected in the cathode

Test Your Knowledge

2 – What is the role of the additional filtration of a X-ray tube ?

- Cut off low-energy X-rays
- Improve the radiological contrast
- Shape a spatially homogenous X-ray beam
- Stop the X-rays scattered in the X-ray tube window

Test Your Knowledge

2 – What is the role of the additional filtration of a X-ray tube ?

- ✓ Cut off low-energy X-rays
- Improve the radiological contrast
- Shape a spatially homogenous X-ray beam
- Stop the X-rays scattered in the X-ray tube window

Test Your Knowledge

3 – What are characteristic X-rays ?

- X-rays whose energy is characterized by the additional aluminum filter
- X-rays whose energy is characterized by the anode material
- X-rays whose energy is increased by multiple interactions in the anode
- X-rays whose energy is increased by the additional aluminum filter

Chapter Outline

Test Your Knowledge

3 – What are characteristic X-rays ?

- X-rays whose energy is characterized by the additional aluminum filter
- ✓ X-rays whose energy is characterized by the anode material
- X-rays whose energy is increased by multiple interactions in the anode
- X-rays whose energy is increased by the additional aluminum filter

Test Your Knowledge

4 – Which is true about the probability of photoelectric effect ?

- It increases with the electronic density of the material.
- It increases with the material density.
- It increases with the tube voltage.
- It increases with the X-ray energy.

Chapter Outline

Test Your Knowledge

4 – Which is true about the probability of photoelectric effect ?

✓ It increases with the electronic density of the material.

- It increases with the material density.
- It increases with the tube voltage.
- It increases with the X-ray energy.

Test Your Knowledge

5 – What could be done to reduce the geometrical blur of an X-ray ?

- Choose a higher AEC setting.
- Decrease the exposure time.
- Increase the tube-to-detector distance.
- Use an anti-scatter grid.

Chapter Outline

Test Your Knowledge

5 – What could be done to reduce the geometrical blur of an X-ray ?

- Choose a higher AEC setting.
- Decrease the exposure time.
- ✓ Increase the tube-to-detector distance.
- Use an anti-scatter grid.

Test Your Knowledge

6 – A DAP is 1000 $mGy.cm^2$ at 1 m from the X-ray tube. What will be at 2 m ?

- 250 $mGy.cm^2$
- 500 $mGy.cm^2$
- 1000 $mGy.cm^2$
- 4000 $mGy.cm^2$

Chapter Outline

Test Your Knowledge

6 – A DAP is 1000 mGy.cm^2 at 1 m from the X-ray tube. What will be at 2 m ?

- 250 mGy.cm^2
- 500 mGy.cm^2
- ✓ 1000 mGy.cm^2
- 4000 mGy.cm^2

Test Your Knowledge

7 – How does the AEC control the dose ?

- By focusing the energy of electrons in the X-ray tube
- By modifying the intensity of X-rays during the irradiation time
- By regulating the length of exposure
- By setting a target tube current (mA)

Test Your Knowledge

7 – How does the AEC control the dose ?

- By focusing the energy of electrons in the X-ray tube
- By modifying the intensity of X-rays during the irradiation time
- ✓ By regulating the length of exposure
- By setting a target tube current (mA)

Test Your Knowledge

8 – How could we increase the SNR of an X-ray ?

- By choosing the large focus instead of the small focus
- By decreasing the tube voltage
- By increasing the AEC setting to +1
- By reducing the pixel size

Chapter Outline

Test Your Knowledge

8 – How could we increase the SNR of an X-ray ?

- By choosing the large focus instead of the small focus
- By decreasing the tube voltage
- ✓ By increasing the AEC setting to +1
- By reducing the pixel size

Test Your Knowledge

9 – Why is ESD larger than air kerma at the patient surface ?

- Because ESD includes backscatter dose
- Because ESD is not expressed in the same unit
- Because ESD is not measured at the same distance from the X-ray source
- Because ESD takes into account the radiosensitivity of the skin

Test Your Knowledge

9 – Why is ESD larger than air kerma at the patient surface ?

- ✓ Because ESD includes backscatter dose
- Because ESD is not expressed in the same unit
- Because ESD is not measured at the same distance from the X-ray source
- Because ESD takes into account the radiosensitivity of the skin

Test Your Knowledge

10 – What is the detector dose which corresponds to an EI = 250 ?

- 2.5 μGy
- 25 μGy
- 250 μGy
- 250 mGy

Chapter Outline

Test Your Knowledge

10 – What is the detector dose which corresponds to an EI = 250 ?

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- 25 µGy
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Chapter Outline

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