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Medical X-ray Film Processing



2nd Edition

K Thayalan

Foreword
C Amarnath

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2

Production of X-rays

DISCOVERY OF X-RAYS

X-rays were discovered by W.C. Roentgen, the German physicist in 1895, when he was investigating the conduction of electricity through gases at low pressure in glass tubes. He noticed that the positive electrodes in the tubes gave off invisible rays that caused fluorescent screens (barium platinocyanide screen kept near the tube) to glow and that fogged photographic plates. The rays were penetrating; they passed through black paper and even thicker objects. They were not deflected in electric and magnetic field. Therefore, Roentgen concluded that they were not charged particles. As their nature was not known he called them X-rays, later they were known to be electromagnetic radiation of very short wavelength. Roentgen received the first *Nobel Prize* in physics in 1901, for his discovery.

Properties of X-rays

- X-rays are electromagnetic radiations of shorter wavelength (0.1–10 nm).
- They travel in straight line with a velocity equal to light.
- X-rays are not influenced by electric and magnetic fields.
- X-rays penetrate through substances that are opaque to visible light.
- X-rays produce fluorescence in materials like *calcium tungstate* and *zinc-cadmium sulfide*.
- X-rays affect the photographic film and form latent image.
- X-rays produce ionization and excitation in the substances through which they pass.
- X-rays produce chemical changes in substances through which they pass.
- X-rays produce biological effects in living organisms. The cells can be either damaged or killed due to X-ray exposure.

Production of X-rays

X-rays are produced when fast moving electrons are stopped by means of a target material. The moving electron possesses kinetic energy. When the electron is suddenly stopped, its kinetic energy is converted into heat and X-rays. This conversion is taking place in the target material. Therefore, the interaction of electron with the target is the basis for X-ray production.

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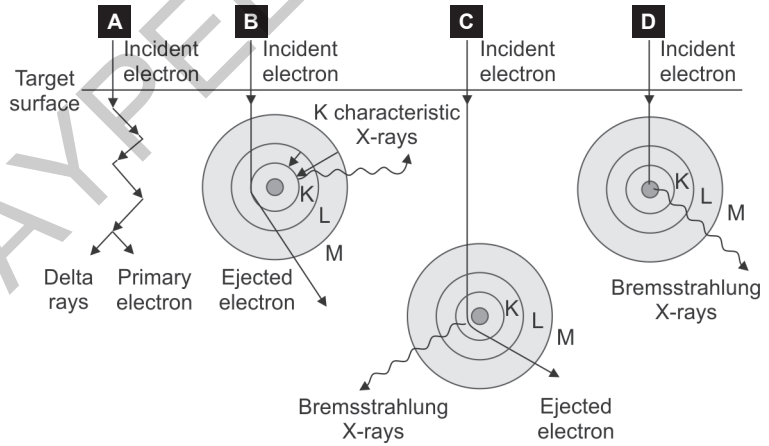
Electron Interaction with the Target

When the electron arrives at the target, it interacts in four ways as follows (Figs. 2.1A to D):

- Ionization of target atoms
- Characteristic X-rays
- Interaction with nuclear field
- Interaction with nucleus.

Ionization of target atoms: Fast moving electrons enter the surface layer of the target and undergo collisions. In this process, the incident electron transfers sufficient energy and removes an electron from the atom. This involves small energy transfer, resulting in ionization of target atoms. The incident electron may undergo number of such collisions and each time its direction gets altered. A 100 keV electron may encounter 1,000 of such interactions, before coming to rest and most of its energy appears as heat in the target. The displaced electron, known as a secondary electron, may have sufficient energy and produce further ionization of target atoms. They are few in number and produce their own track, known as *delta rays*.

Characteristic X-rays: This is an interaction between the incident electron and the electron in the K-shell. In this process, the incident electron directly hit the K-shell, transfers sufficient energy and removes the K-shell electron. The vacancy in the K-shell is filled by an electron moving inward from the outer shell. During this transition, the difference in binding energies of the two shells is given out as X-ray photon. This photon is known as the characteristic X-ray. The ejected electron may produce further interaction in other target atoms.



Figs. 2.1A to D: Interaction of electron with target atoms: (A) Ionization of target atoms; (B) Characteristic X-rays; (C) Interaction with nuclear field; and (D) Interaction with nucleus.

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Interaction with nuclear field: The incident electron occasionally reaches nearer to the nucleus of an atom in the target. Since the electron is a negative particle, it is attracted by the positive nucleus. It is made to orbit partially around the nucleus, decelerates and goes out with reduced energy. The loss of energy appears in the form of X-ray photons, known as *bremsstrahlung*. In German, bremsstrahlung means breaking radiation. The maximum energy of the incident electron is determined by the applied voltage of the X-ray tube. An electron can lose all of its kinetic energy in the form of bremsstrahlung radiation. The energy of the X-ray photon depends on the degree to which the electron is decelerated by the nuclear attraction. The photon energy can take any value from zero to a maximum. This process is unlikely at low energies, but dominant at high energies.

Interaction with nucleus: The electron may hit the nucleus directly and is stopped completely in a single collision. The entire electron energy appears as bremsstrahlung radiation. This type of interaction is very rare, but capable of giving high energy X-rays.

In general, the interaction of B, C, and D are very rare in the diagnostic range of energies, leading to lesser amount of X-ray production (<1%). The ionizational collision (A) dominates (>99%) the interaction process and produces heat. Thus, the X-ray tube is inefficient in the conversion of electron energy into X-rays.

X-ray Spectra

X-ray photons produced by an X-ray tube are heterogeneous in energy. A graph drawn between the photon energy and X-ray intensity is called *X-ray spectrum*. There are two types of X-ray spectrum, namely:

- Bremsstrahlung or continuous spectrum
- Characteristic or line spectrum.

A bremsstrahlung spectrum consists of X-ray photons of all energies up to maximum in a continuous fashion, which is also known as white radiation, because of its similarity to white light. Bremsstrahlung X-ray production increases with increase of applied voltage and atomic number of the target material. Most of the X-ray imaging in hospitals are bremsstrahlung X-rays, except in mammography. A characteristic spectrum consists of X-ray photons of low energy. The position of the characteristic radiation depends upon the atomic number of the target. Only K-shell characteristic X-rays are very important in radiology. To eject a K-shell electron, the incident electron must have energy greater than the binding energy of the K-shell. Characteristic X-rays occur only at discrete energy levels, capable of producing *Auger electrons*. It finds application mammography imaging.

The intensity of the X-rays can be plotted against photon energy in a graph (Fig. 2.2). The area under the curve is proportional to the total

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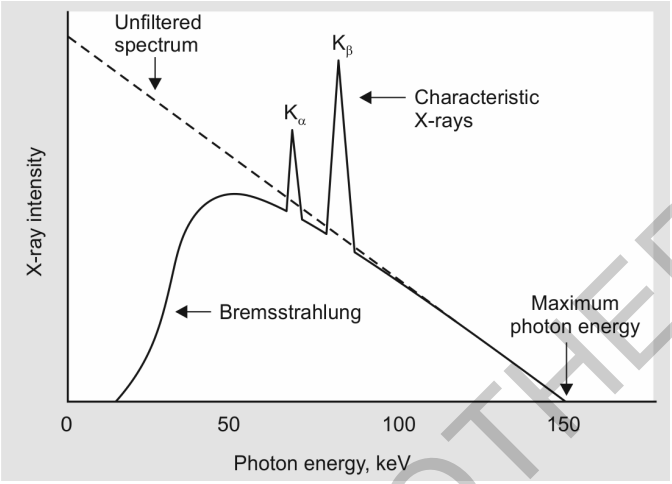

Fig. 2.2: X-ray spectrum consisting bremsstrahlung and characteristic X-rays.

number of photons emitted. The highest X-ray energy is determined by the peak voltage (kV_p) applied in the X-ray tube. The characteristic spectrum is superimposed on the continuous spectrum. An unfiltered beam spectrum (theoretical) will be a straight line and mathematically given by **Kramer's** equation:

$$I_E = KZ\,(E_m - E)$$

where, I_E is the intensity of photons with energy E, Z is the atomic number of the target, E_M is the maximum photon energy and K is a constant. The unfiltered X-ray spectrum looks like a ramp.

In practice, the X-ray beam is a filtered beam, due to inherent and added filtration. The filtration hardens the beam, by absorbing the low energy X-rays up to 10 keV, which is evident by the bremsstrahlung spectrum in the graph. The number of photons increase initially, with photon energy and later decrease linearly up to maximal photon energy. The X-ray spectrum is influenced by the following factors:

- Applied voltage
- Target material
- Tube current
- Exposure time
- Bremsstrahlung process
- Filtration.

To specify the quality of X-rays, a rule of thumb is used, which states that the effective energy is about 1/3 to 1/2 of maximum X-ray energy. K-shell characteristic X-ray contribution is about 10% at 100 kV photon energy. There is no presence of characteristic X-ray at <70 kV.

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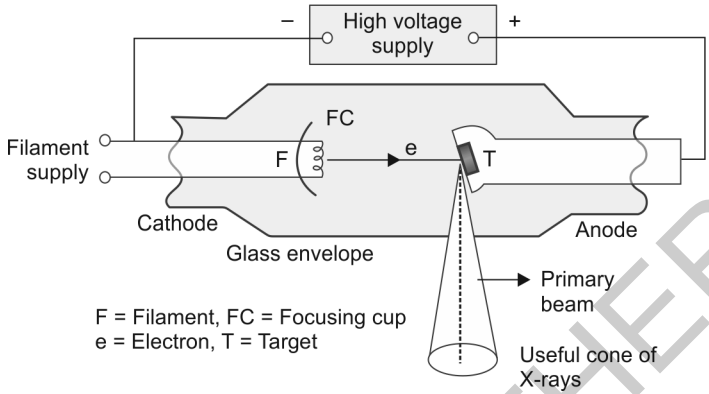

Fig. 2.3: Modern X-ray tube (stationary anode).

MODERN X-RAY TUBE

In earlier periods gas tubes were used to produce X-rays. But these tubes suffered with disadvantages. Hence, **Coolidge** proposed a prototype X-ray tube based on the thermionic emission principle (1913). On the basis of Coolidge tube, several X-ray tubes have been designed. The stationary anode X-ray tube is one of the modern X-ray tube in which the anode is stationary.

The stationary anode X-ray tube consists a cathode and an anode, which are kept in an evacuated glass bulb made up of Pyrex glass (Fig. 2.3). This is kept in a housing that provides insulation, cooling (oil) and shielding to leakage radiation. The primary X-ray emerges through the window. The cathode consists of a tungsten filament in the form of a coil placed in a shallow focusing cup. The filament is made of tungsten wire, about 0.2 mm in diameter, that is coiled to form a vertical spiral about 0.2 cm in diameter and 1 cm in length. The filament is heated by passing an electric current (10 V,4A) through it from a low voltage supply. Sometimes, one can have two filaments in a single tube, to have two focal spots.

The anode is made of copper block in which a small tungsten plate is embedded. The tungsten is a square or rectangular plate of 2 mm or 3 mm thick and the dimension is greater than 1 cm. The tungsten plate serves as a target. The target is positioned at online focus principle, in order to increase the ratio of the actual focal area to the effective focal area. The area of the target in which the electron bombards and produce X-rays is called *focal spot*. It is viewed by the patient direction. There may be small and large focal area. Small focal spot produces sharp images with good spatial resolution. Large focal spot is recommended for high heat loading and short exposures. Typical focal spot size range is about 0.1 to 1.2 mm.

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The anode angle is usually 7–20°. A high voltage supply is applied between the cathode and anode to accelerate the electrons. A vacuum of the order of 10^{-5} mm Hg is maintained in the tube.

When the filament is heated to white light, it emits electrons by thermionic emission principle. The focusing cup (made up of nickel) produces an electric field that focuses the electron to the focal area. The focusing cup also protects the adjacent parts of the tube wall from damage by electron bombardment. If the anode is made positive with respect to the filament, these electrons will be attracted to the anode. This will constitute an electron current around the circuit in the anticlockwise direction. The tube current is measured by the milliammeter (mA). Usually, there is an electron cloud (space charge) between the anode and cathode. The applied voltage should overcome the electron cloud, then only current flows in the tube. The minimum applied potential to create the tube current is 40 kV. Tube current increases with filament heating/current. The tube current ranges from 1–1200 mA.

Since the space between anode and cathode is a high vacuum, the electrons do not collide with gas molecules in crossing the gap and so acquire very high velocities. The electrons which are accelerated by the applied voltage possess high kinetic energy. When they suddenly stopped in the target, X-rays are emitted in all directions. About one-half of these are absorbed in the target itself. The remaining portion emerges as a useful primary X-ray beam. During the production of X-rays, large amount of heat is produced in the target. The tube is also provided with suitable cooling system to remove the heat very quickly.

Stationary anode tubes have a small target area that limits the heat dissipation and this limits the X-ray output. Dental X-ray units, portable X-ray units, and portable fluoroscopy systems use stationary anode X-ray tubes. Alternatively, rotating anode X-ray tubes are used in mammography, cardiac catheterization laboratory and CT scans. These tubes employ rotor and induction motor to rotate the anode, thereby increasing the effective target area and heat capacity.

Quality and Intensity of X-rays

Quality

The term quality describes the penetrating power of the radiation. If a radiation consists of photons of single energy (monoenergetic), then the quality can be described either by photon energy or wavelength. But X-ray beam consists of many photon energies (heterogeneous), and hence, its quality cannot be described by the photon energy. Therefore, the X-ray beam quality is usually specified by the following:

- Half-value layer (HVL)
- Applied voltage (kV)
- Filtration
- Effective photon energy.

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Half-value Layer

The half value layer (HVL) or half-value thickness (HVT) of a radiation beam is the required thickness of a material which reduces the beam intensity to one half. The HVL is always stated together with the value of the applied voltage and the filtration. *Aluminum* and *copper* are the materials commonly used to specify HVL.

Intensity

The intensity is a measure of quantity of radiation, that is the number of X-rays in the beam. The intensity of a radiation beam is the energy flowing in unit time through a unit area. It is equal to the number of photons in the beam multiplied by the energy of each photon. The intensity is commonly measured in roentgens per minute ($R\ min^{-1}$) or air kerma (mGy). The X-ray intensity is proportional to the tube current (Fig. 2.4A) and exposure time. Doubling the tube current or exposure time will double the X-ray beam intensity. The product of tube current (mA) and exposure time (s), is called mAs. The mAs affects the beam intensity, not the photon energy.

Increase of kV also increases X-ray beam intensity in a quadratic manner (Fig. 2.4B). It also increases the photon energy. Any variation in the applied voltage will change the shape of the X-ray spectrum. The maximum photon energy is determined by the applied voltage.

The term exposure is often used in radiology, which is proportional to the energy fluence of the X-ray beam. It refers both quality and quantity of the beam. The term quantity refers the number of X-ray photons in the beam.

Filter

A filter is a metallic sheet introduced in the path of X-rays, in order to reduce the patient dose. Diagnostic X-rays consist of both low energy and high energy. When X-rays pass through a patient, only high energy X-rays penetrate through the patient and form the radiological image. However, the low energy X-rays are absorbed in the first few cm of tissue, thereby increasing the radiation dose. The introduction of filters absorb these low energy X-rays and reduce the patient dose. This process of removing the low energy X-rays, by introducing metallic sheets is called *filtration* (Fig. 2.5). Filtration has two components namely:

- Inherent filtration
- Added filtration.

Filtration resulting from the absorption of X-rays by the X-ray tube and its housing is called inherent filtration. It usually varies between 0.5 to 1.0 mm of Al equivalent. Added filtration results from absorbers placed externally in the path of the X-ray beam. The sum of the inherent and added filtration gives the total filtration:

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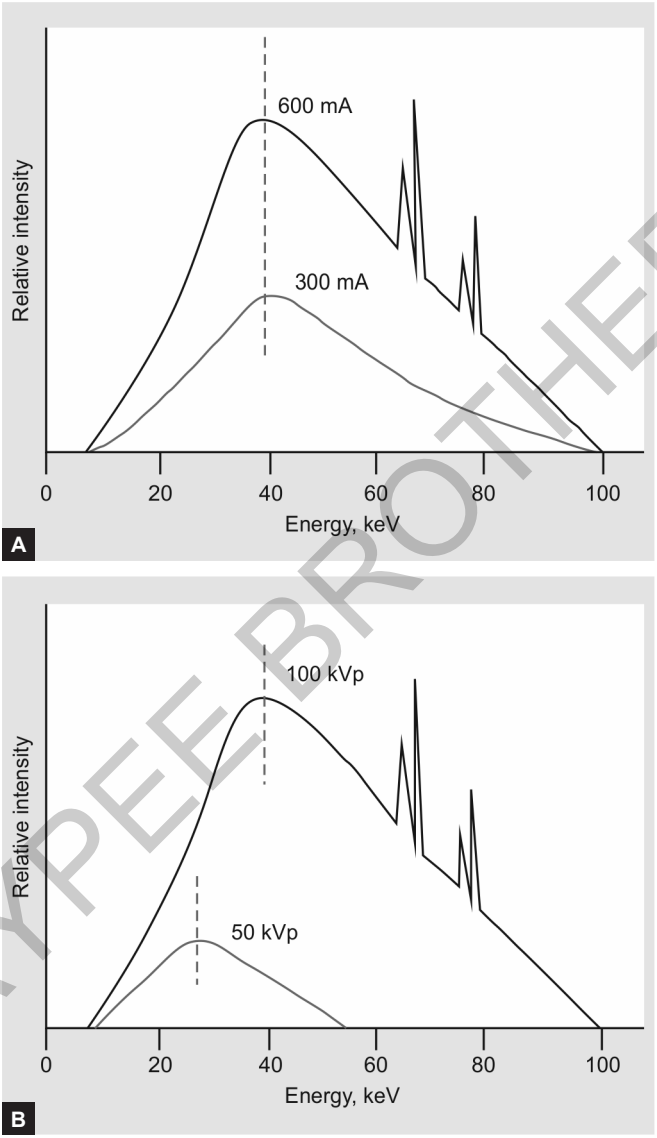

Figs. 2.4A and B: Effect of: (A) tube current (mA); (B) applied voltage (kV).

Total filtration = Inherent filtration + Added filtration.

Al and Cu are the materials usually used in diagnostic radiology. The thickness of added filter varies from 1.0 to 1.5 mm of Al. Aluminum

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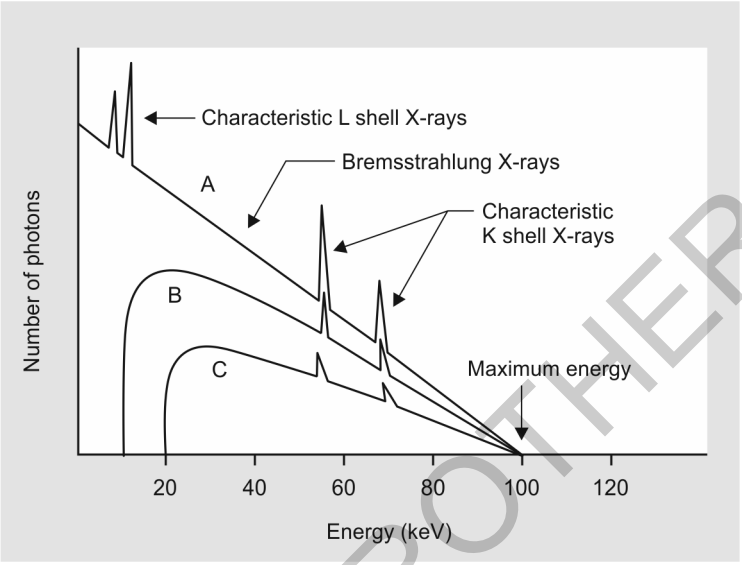

Fig. 2.5: Effect of filter: (A) Unfiltered spectrum; (B) inherent filter; (C) added filter.

(Z = 13) is an excellent filter material for low energy X-rays. Copper (Z = 29) is a better filter for high energy radiation. Copper is always used in combination with aluminum as a compound filter. A compound filter consists of two or more layers of different metals. The layers are arranged in such a way that the high Z layer always faces the X-ray tube.

The recommended total filtration for diagnostic X-ray unit of >70 kVp is 2.5 mm Al. X-ray units with proper filter, reduces the patient dose significantly, up to 80%. Filters are simple and inexpensive. Though filters reduce the intensity of X-ray beam significantly, it does not affect the maximum energy of the X-ray beam spectrum. Heavy metal filters (Gd, Ho) are also used in general radiography. These filters make use of the K-edge, and offer increased absorption of X-rays, while imaging with contrast agents. They enhance contrast for iodine and barium, reduce patient dose and increase tube loading.

In mammography Mo target with Rh filter is commonly used, whereas, Mo cannot be used as filter in mammography X-ray tubes with Rh targets.

Beam Restrictor/Collimator

An X-ray beam restrictor is a device that is attached to the X-ray tube housing, to regulate the size and shape of an X-ray beam. They can be classified into three categories namely:

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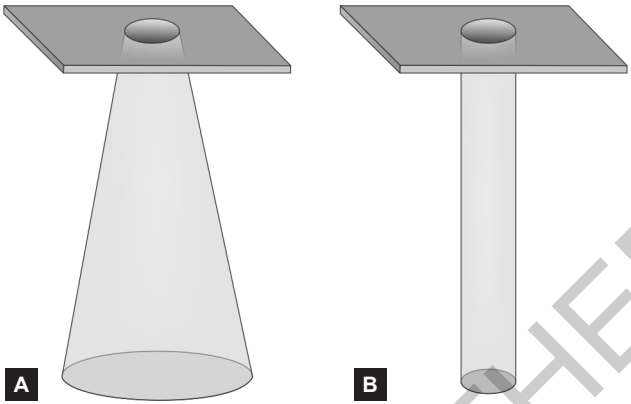

Figs. 2.6A and B: Beam restrictors: (A) Cone; (B) Cylinder.

- Aperture diaphragm
- Cone and cylinder
- Collimator.

Aperture diaphragm consists of a sheet of lead with a hole in the center. The size of the hole determines the size and shape of the X-ray beam. It is simple and the aperture can be altered to any size and shape. The disadvantage of an aperture diaphragm is that it produces large penumbra. The penumbra can be reduced by keeping the aperture diaphragm far away from the X-ray target. Aperture diaphragm is used in dental radiography with rectangular collimation. In addition, it is used in trauma and chest radiography.

The use of cone and cylinder will reduce the penumbra considerably (Figs. 2.6A and B). Both have extended metal structures that restrict the useful circular beam to the required size. The position and size of the distal end determine the field size. If the X-ray source, cone and film are not aligned properly, then one side of the film may not be exposed, which is called as *cone cutting*. Cone is a ideal beam restrictor, but the flare of the cone is greater than the flare of the X-ray beam. These systems provide only limited number of field sizes.

The collimator is the best X-ray beam restrictor. It defines the size and shape of the X-ray field that emerges from the X-ray tube. The collimator assembly is attached to the tube housing at the tube port. A collimator consists of two sets of shutters, which can be moved independently. Each shutter consists of four or more lead plates, which can absorb X-rays completely, to provide a well defined X-ray field. When the shutters are closed, they meet at the center of the X-ray field.

The collimator also has a light and mirror arrangement, to illuminate the X-ray field. The light bulb is positioned laterally and the mirror is

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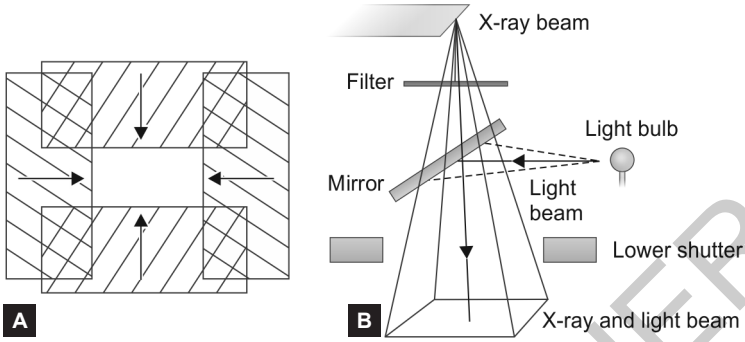

Figs. 2.7A and B: Collimator: (A) Collimator shutters; (B) Light and mirror arrangement for the alignment of X-ray and light field.

mounted in the path of the X-ray beam at an angle 45° (Figs. 2.7A and B). The target and the light bulb should be kept at equal distance from the center of the mirror. The collimator provides variety of rectangular X-ray fields and the light beam shows the center of the X-ray field. The light field and radiation field should match exactly with each other. The variation must be within 4% of target to film distance (TFD). The alignment of light beam and X-ray beam should be checked periodically. A well collimated beam, cover lesser area of the patient, giving less patient dose. It also generates less scatter radiation, which improves the image quality.

Collimator that automatically limits the X-ray field size to the useful area of the detector is also available. These are called *positive beam limitation (PBL)* collimators. A sensor in the cassette holder, adjust the collimator opening, equal to the cassette dimensions. Thus, PBL collimator limits the irradiated volume and reduces the patient dose.

Scattered Radiation and Grid

When the beam of X-rays pass through the patient, the beam is absorbed and scattered. The absorbed primary beam gives a useful shadow, while the scattered radiation will tend to spoil the shadow. Scattered radiation contributes a constant background fog to the film image. This will increase the noise in the image. The ratio between the amounts of scattered radiation energy to the amount of primary radiation energy at a point is called scatter to primary ratio (SPR). The SPR increases with thicker patient and larger field size. For example, in abdomen radiography, only 20% of the photons contributes to the image formation and the other 80% goes as scattered radiation. Hence, scattered radiation must be removed, in order to increase the image contrast.

The scattered radiation can be removed by a grid, placed in-between the film and the patient. The grid consists of a series of parallel lead or tantalum strips of thickness **c** (50 μm) and of height **h** separated by spacers

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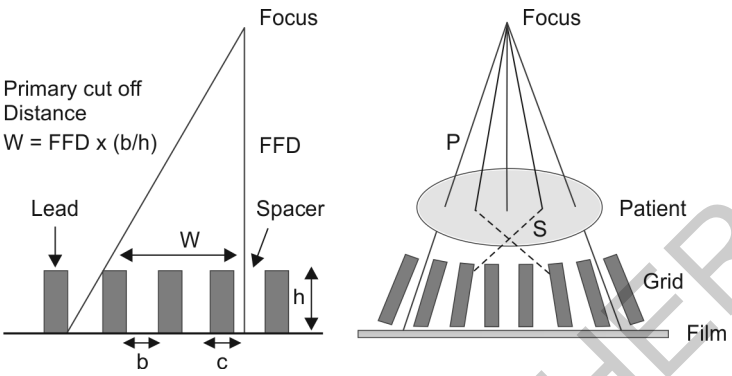

Fig. 2.8: Grid design and principle.

of low attenuating material of width **b** (350 μm) as shown in Figure 2.8. Aluminum or plastic fibers are used as low attenuating spacers. The grid is positioned between the patient and the detector, so that its long axis is pointed towards the X-ray beam. The primary X-rays coming out of the patient, passes through the inter space, since it is parallel in direction. The scattered X-rays, which are in non-parallel direction, strike the grid bars and being absorbed. The ratio of the primary transmission to the scatter transmission of a grid is called the *selectivity*.

Grid Ratio

The ability of the grid to discriminate against scattered radiation is measured by the grid ratio, which is defined as the ratio of the height to the width of the spacer between the lead strips.

$$\text{Grid ratio} = h/b$$

As the grid ratio increases, the grid removes more scatter radiations. The strip line density is 1/(b+C) lines per unit length. Typical grid ratio ranges from 4:1 to 16:1 and strip line densities from 25 to 60 lines per cm. The performance of a grid can be understood by *contrast improvement factor*. It is the ratio between image contrast with grid and image contrast without grid at 100 kVp. Higher grid ratio provides higher contrast improvement factor. However, it increases patient dose, as it employs higher exposure techniques. *Bucky factor* (**Gustave Bucky, 1913**) is another parameter which relates to the patient dose. It is the ratio between the patient dose with grid and patient dose without grid. Bucky factor increases with increase of kVp and grid ratio.

Types of Grid

Grids may be classified as:

- Parallel grid
- Crossed grid

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Medical X-ray Film Processing

Salient Features

- *Medical X-ray Film Processing* consists of 10 chapters. Chapter 1 discusses about the fundamental concepts such as units and measurements, to understand the later concepts better
- Chapter 2 talks about production of X-rays, to refresh the knowledge of medical X-rays
- In Chapter 3, medical X-ray film is discussed, which describes the film structure, design, characteristics, handling and storage
- Chapter 4 talks about the darkroom, where the film is loaded and unloaded, cassettes and intensifying screens, which are used along with X-ray film
- In Chapter 5, the various film processing methods, including developing, rinsing, fixing, washing and drying are given
- Chapter 6 describes about the screen-film radiography image quality, artifacts and quality assurance
- Digital radiography, including computed radiography, direct and indirect radiography systems are explained in Chapter 7
- The digital radiography image quality, viewing and recording are explained in Chapter 8
- Chapter 9 talks about digital radiography image artifacts and quality assurance
- The final Chapter 10 talks about radiological health and safety. It gives the various aspects of safety including biological effect, annual dose limits, methods of radiation control, regulatory issues and personnel monitoring.

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He has 34 publications to his credit, presented 74 scientific papers, delivered 160 invited lectures, in national and international conferences and an author of six books. He was an examiner for UG and PG courses and currently guiding students for PhD. He was the Task Group Member of the Atomic Energy Regulatory Board (AERB), Mumbai, Maharashtra, India for the preparation of safety codes and guides.

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