

Production of X-rays and Interactions of X-rays with Matter

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Electrons traveling from the filament (cathode) to the target (anode) convert a small percentage (1%) of their kinetic energy into x-ray photons by the formation of bremsstrahlung and characteristic radiation.

A. BREMSSTRAHLUNG RADIATION

Bremsstrahlung interactions, the primary source of x-ray photons from an x-ray tube, are produced by the sudden stopping, breaking or slowing of high-speed electrons at the target. When the electrons from the filament strike the tungsten target, x-ray photons are created if they either hit a target nucleus directly (rare) or their path takes them close to the nucleus. If a high-speed electron hits the nucleus of a target atom, all its kinetic energy is transformed into a single x-ray photon. (**Total** absorption has occurred). Thus, the energy of the resultant photon (keV) is numerically equal to the energy of the electron. This in turn is equal to the kilovoltage applied across the x-ray tube at the instant of its passage. This happens rarely.

Most high-speed electrons have near or wide misses with the nuclei. In these interactions, a negatively charged high-speed electron is attracted toward the positively charged nucleus and loses some of its velocity. This **deceleration** causes the electron to lose some kinetic energy, which is given off in the form of a photon. The closer the high-speed electron approaches the nuclei, the greater is the electrostatic attraction on the electron, the braking effect, and the greater the energy of the resulting Bremsstrahlung photon.

Bremsstrahlung interactions generate x-ray photons with a **continuous spectrum** of energy. i.e. different energies. The energy of an x-ray beam may be described by identifying the peak operating voltage (in kVp). A dental x-ray machine operating at a peak voltage of 70,000 volts (70 kVp) for example, applies to a fluctuating voltage of as much as 70 kVp across the tube. This tube therefore produces x-ray photons with energies ranging to a maximum of 70,000 keV (70 keV). The reasons for this continuous spectrum are as follow:

1. The continuously varying voltage difference between the target and the filament, which is characteristic of half wave rectification, causes the electrons striking the target to have varying levels of kinetic energy.
2. Most electrons participate in many interactions before all their kinetic energy is expended. As a consequence, an electron carries differing amounts of energy at the time of each interaction with a tungsten atom that results in the generation of an x-ray photon.
3. The bombarding electrons pass at varying distances around tungsten nuclei and are thus deflected to varying extents. As a result, they give up varying amounts of energy in the form of Bremsstrahlung photons.
4. Depth of generation of photons in the target

B. CHARACTERISTIC RADIATION

Characteristic radiation occurs when an electron from the filament displaces an electron from an **inner-shell** of the tungsten target atom, thereby **ionizing** the atom. When this happens, another electron in an outer-shell of the tungsten atom is quickly attracted into the void in the deficient inner-shell. When the displaced electron is replaced by the outer-shell electron, a photon is emitted with an energy equivalent to the difference in the two orbital binding energies. Characteristic radiation from the K-shell occurs only above 70 kVp with a tungsten target and occurs as **discrete increments** compared with Bremsstrahlung radiation. The energies of characteristic photons are a function of the energy levels of various electron orbital levels and hence are characteristic of the target atoms. Characteristic radiation has a higher intensity, is preferred but is only a minor source of radiation from an x-ray tube.

FACTORS CONTROLLING THE X-RAY BEAM

The x-ray beam emitted from an x-ray tube may be modified to suit the needs of the application by altering the beam exposure length (timer), exposure rate (mA), beam energy (kVp and filtration), beam shape (collimation), and target-patient distance (long or short cone)

I. Exposure Time

Portrays the changes in the x-ray spectrum that result when the exposure time is increased while the tube current (mA) and voltage (kVp) remain constant. When the exposure time is doubled, the number of photons generated is doubled, but the **range** intensity of photons energies is **unchanged**. Therefore changing the time simply controls the “quantity” of the exposure, the number of photons generated. The **amount** of radiation that a patient receives is determined by the mAs (mA x time)

II. Tube Current (mA)

Illustrates the changes in the spectrum of photons that result from increasing tube current (mA) while maintaining constant tube voltage (kVp) and exposure time. As the mA setting is increased, more power is applied to the filament, which heats up and releases more electrons that collide with the target to produce radiation. A linear relationship exists between mA and radiation output. The quantity of radiation produced (mAs) is expressed as the product of time and tube current. The quantity of radiation remains constant regardless of variations in mA and time as long as their product remains constant. For instance, a machine operating at 10mA for 1 second (10mAs) produces the same quantity of radiation when operated at 20 mA for 0.5 second (10 mAs)

III. Tube Voltage (kVp)

Increasing the kVp increases the potential difference between the cathode and anode, thus increasing the energy of each electron when it strikes the target. The greater the potential Difference the **faster** the electrons travel from the cathode to the anode. This results in an increased efficiency of conversion of electron energy into x-ray photons, and thus an increase in

- 1) The number of photons generated ~ *Beware in Board Exams*
- 2) Their mean energy.
- 3) Their maximal energy.

The increased number of high-energy photons produced per unit time by use of higher kVp results from the greater efficiency in the production of Bremsstrahlung photons that occurs when increased number of higher-energy electrons interact with the target.

The ability of x-ray photons to penetrate matter depends on their energy. High-energy x-ray photons have a **greater probability of penetrating** matter, whereas relatively low-energy photons have a greater probability of being **absorbed**. Therefore the higher the kVp and mean energy of the x-ray beam, the greater the penetrability of the beam through matter. The radiation that does damage to a patient is the radiation that is absorbed by the patient.

Half Value Layer

A useful way to characterize the penetrating quality of an x-ray beam by its half-value layer (HVL). The HVL is the thickness of an absorber, such as aluminum, required to reduce by one half the **number** of x-ray photons **passing** through it. As the average energy of an x-ray beam increases, so does it HVL. The term quality refers to the mean energy of an x-ray beam.

Half value layer **measures** the intensity of a beam.

IV. Filtration

An x-ray beam consists of a spectrum of x-ray photons of different energies, but only photons with sufficient energy to penetrate through anatomic structures and reach the image receptor (usually film) are useful for diagnostic radiology. Those that are of low-energy (long wavelength) contribute to patient exposure but do not have enough energy to reach the film. The higher the kVp, the less radiation is absorbed by the patient. Consequently, to reduce patient dose, the less-penetrating photons should be removed. This can be accomplished by placing an aluminum filter in the path of the beam. The aluminum **preferentially removes** many of the **lower-energy** (long waves) photons with lesser effect on the higher-energy photons that are able to penetrate to the film.

In determination of the amount of filtration required for a particular x-ray machine, kVp and inherent filtration of the tube and its housing must be considered. Inherent filtration consists of the materials that x-ray photons encounter as they travel from the focal spot of the target to form the usable beam outside the tube enclosure. These materials include the glass wall of the x-ray tube, the insulating oil that surrounds many dental tubes, and the barrier material that prevents the oil from escaping through the x-ray port.

Total filtration = inherent filtration plus external filtration (aluminum disks).

Governmental regulations require the total filtration in the path of a dental x-ray beam to be equal to the equivalent of **1.5 mm of aluminum to 70 kVp**, and **2.5 mm of aluminum** for all higher voltages (i.e. above 70 kVp).

V. Collimation

A collimator is a metallic barrier with an aperture in the middle used to reduce the **size and shape** of the x-ray beam and therefore the **volume** of irradiated tissue within the patient. The round collimator is a thick plate of radiopaque material (usually lead) with a circular opening centered over the port in the x-ray through which the x-ray beam emerges. Typically, round collimators are built into open-ended aiming cylinders. Rectangular

collimators further limit the beam to a size just larger than that of the x-ray film. The size of the beam should be reduced to the size of the film being exposed to reduce further unnecessary patient exposure. Some types of film-holding instruments also provide rectangular collimation of the x-ray beam.

Use of collimation also improves image quality. When an x-ray beam is directed at a patient about 90% of the x-ray photons are absorbed by the tissues and 10% of the photons pass through the patient and reach the film. Many of the absorbed photons generate scattered radiation within the exposed tissues by a process called Compton scattering. These scattered photons travel in all directions. Many for the film and thereby degrade image quality. The detrimental effect of scattered radiation of the images can be minimized by collimating the beam to reduce the number of scattered photons reaching the film.

Inverse Square Law

The intensity of an x-ray beam at a given point (number of photons per cross-sectional area per unit exposure time) depends on the distance of the measuring device from the focal spot. For a given beam the intensity is inversely proportional to the square of the distance from the source. The reason for this decrease in intensity is that the x-ray beam spreads out as it moves from the source. The relationship is as follows:

$$\frac{I_1}{I_2} = \frac{(D_2)^2}{(D_1)^2} \quad \text{or} \quad \frac{1}{I} = D^2$$

Where I is intensity and D is distance. Therefore if a dose of 1 gray (Gy) is measured at a distance of 2 m, a dose of 4 Gy will be found at 1 m, and 0.25 Gy at 4 m.

Therefore changing the distance between the x-ray tube and patient has a marked effect of beam intensity. Such a change requires a corresponding modification of the kVp or mAs if the exposure of the film is to be kept constant.

Thus, be careful when taking a radiograph to place the cone consistently as close to the patient (but never touching) as possible to maintain the same distance while taking or the radiographs.

INTERACTIONS OF X-RAYS WITH MATTER

The intensity of an x-ray beam is reduced by interaction with the matter it encounters. This attenuation results from interactions of individual photons in the beam with atoms in the absorber (patient). The x-ray photons are either absorbed or scattered out of the beam. In scattering, photons are ejected out of the primary beam as a result of interactions with the orbital electrons of absorber atoms. In the case of a dental x-ray beam, three mechanisms exist where these interactions take place. (1) Coherent scattering, (2) Compton scattering, and (3) photoelectric absorption. In addition, about 9% of the primary photons pass through the patient without interaction to produce the image.

1. COHERENT SCATTERING

Coherent Scattering (also known as classical scattering and Thompson Scattering) may occur when a low-energy incident photon passes near an **outer electron** of an atom (which has a low binding energy). The incident photon interacts with the electron in the outer-shell by causing it to vibrate momentarily at the **same frequency** as the **incoming photon**. The incident photon then ceases to exist. The vibration causes the electron to radiate energy in

the form of **another x-ray photon** with the **same frequency** and **energy** as in the incident photon. In effect, the direction of the incident x-ray photon is altered. This interaction accounts for only about 8% of the total number of interactions (per exposure) in a dental examination. Coherent scattering contributes very little to film fog because the total quantity of scattered photons is small and its energy level is too low for much of it to reach the film.

2. Compton scattering occurs when a photon interacts with an **outer orbital** electron, which receives kinetic energy and recoils from the point of impact. The incident photon is then deflected by its interaction and is scattered from the site of the collision. The energy of the scattered photon equals the energy of the incident photon **minus** the kinetic energy gained by the recoil electron plus its bonding energy. As with photoelectric absorption, Compton scattering results in the loss of an electron and **ionization** of the absorbing atom.

Scattered photons travel in all directions. The higher the energy of the incident photon, however, the greater the probability that the angle of scatter of the secondary photon will be small and its direction will be forward. Approximately 30% of the scattered photons formed during a dental x-ray exposure (primarily from Compton scattering) exit the patient's head. This is advantageous to the patient because some of the energy of the incident x-ray beam escapes the tissue, but it is disadvantageous because it causes **nonspecific film darkening** (or fogging of the film). Scattered photons darken the film while carrying no useful information to it because their path is altered.

The probability of Compton scattering is directly proportional to the electron density. The number of electrons in bone is greater than in water, therefore the **probability** of Compton scattering is correspondingly **greater** in bone than in tissue. In a dental x-ray beam, approximately 62% of the photons undergo Compton scattering.

The importance of photoelectric absorption and Compton scattering in diagnostic radiography relates to differences in the way photons are absorbed by various anatomic structures. The number of photoelectric and Compton interactions is greater in hard tissues than in soft tissues. As a consequence, more photons in the beam exit the patient after passing through soft tissue than through hard tissue. This allows a radiograph to provide a clear image of enamel, dentine and bone and also, soft tissue.

3. **PHOTOELECTRIC ABSORPTION**

Photoelectric absorption occurs when an incident photon collides with an inner-shell electron in an atom of the absorbing medium resulting in **total absorption** and the incident photon ceases to exist. The electron is ejected from its shell, resulting in ionization and becomes a recoil electron (photoelectron). The kinetic energy imparted to the recoil electron is equal to the energy of the incident photon minus that used to overcome the binding energy of the electron. In the case of atoms with low atomic numbers (e.g. those in most biologic energy of the incident photon). Most Photoelectric interactions occur in the K shell because the density of the electron cloud is greater in this region and a higher probability of interaction exists. About 30% of photons absorbed from a dental x-ray beam are absorbed by the photoelectric process.

An atom that has participated in photoelectric interaction is ionized. This electron deficiency (usually in the K shell) is instantly filled, usually by an L- or M- shell electron, with the release of characteristic radiation. Whatever the orbit of the replacement electron, the characteristic photons generated are of such low-energy that they are absorbed within the patient and do not fog the film.

The recoil electrons ejected during photoelectric absorptions travel only a short distance in the absorber before they give up their energy. As a consequence, all the energy of incident photons that undergo photoelectric interaction is deposited in the patient. This is beneficial in **producing high-quality radiographs**, because no scattered radiation fogs the film, but potentially **deleterious** for patients because of **increased radiation absorption**.

The frequency of **photoelectric** interaction varies directly with the third power of the atomic number of the absorber. For example, because the effective atomic number of compact bone ($Z = 7.4$), the probability that a photon will be absorbed by a **photoelectric** interaction in bone is approximately 6.5 times greater than in an equal distance of water. This difference is readily seen on dental radiographs. It is this difference in the absorption that makes that production of a radiographic image possible.

SECONDARY ELECTRONS

In both **Photoelectric** absorption and **Compton scattering**, electrons are ejected from their orbits in the absorbing material after interaction with x-ray photons. These secondary electrons give up their energy in the absorber by either of two processes: (1) collisional interaction with other electrons, resulting in ionization or excitation of the affected atom, and (2) radiative interactions, which produce bremsstrahlung radiation resulting in the emission of low-energy x-ray photons. Secondary electrons eventually dissipate all their energy, mostly as heat by collisional interaction, and come to rest.

BEAM ATTENUATION

As a dental x-ray beam travels through matter, individual photons are removed, primarily through **Photoelectric** and Compton interaction. The reduction of beam intensity is predictable because it depends on physical characteristics of the beam and absorber. A monochromatic beam of photons, a beam in which all the photons have the same energy provides a good example. When just the primary (not scattered) photons are considered, a constant fraction of the beam is attenuated as the beam moves through each unit thickness of an absorber. Therefore 1.5 cm of water may reduce a beam intensity by 50%, the next 1.5 cm by another 50% (to 25% of the original intensity), and so on. This HVL described earlier in the chapter is a measure of beam energy describing the amount of an absorber that reduces the beam intensity by half; in the preceding example, the HVL is 1.5 cm. The absorption of the beam depends primarily on the **thickness** and **mass** of the absorber and the **energy of the beam**.

The spectrum of photon energies (as illustrated by the kVp setting) in an x-ray beam is wide. In such a heterogeneous beam the probability of absorption of individual photons depends on their energy. Low-energy photons are much more likely than high-energy photons to be absorbed. As a consequence the superficial layers of an absorber tend to remove the low-energy photons and transmit the higher energy photons. Therefore as an x-ray beam passes through matter, the intensity of the beam decreases but the mean energy of the resultant beam increases. In contrast to the absorption of a monochromatic beam, an x-ray beam is absorbed less and less by each succeeding unit of absorber thickness. For example, the first 1.5 cm of water might absorb about 40% of the photons in an x-ray beam with a mean energy of 50 kVp. The mean energy of the remnant beam might increase 20% as a result of the loss of lower energy photons. The next 1.5 cm of water removes only about 30% of the photons as the average energy of the beam increases another 10%. If the water test object is thick enough, the mean energy of the remnant beam approaches the

peak voltage applied across the tube and absorption becomes similar to that of a monochromatic beam.

In general, as the energy of the beam increases, so does the transmission of the beam through the absorber. When the energy of the incident photon is raised to the binding energy of the k-shell electrons of the absorber, however, the probability of photoelectric absorption increases sharply and the number of transmitted photons is greatly decreased.

This is called k-edge absorption. (The probability that a photon will interact with an orbital electron is greatest when the energy of the photon equals the binding energy of the electron; it decreases as the photon energy increases.) Photons with energy less than the binding energy of K-shell electrons interact photoelectrically only with electrons in the L shell and in shells even farther from the nucleus.

DOSIMETRY

Determining the **quantity** of radiation exposure or dose is termed dosimetry. The term dose is used to describe the **amount** of energy **absorbed per unit mass** at a site of interest. Exposure is a measure of radiation based on its ability to produce ionization in air under standard conditions of temperature and pressure.

UNITS OF MEASUREMENT

In recent years a move has occurred to use a modernized version of the metric system called the SI system (Système International d'Unités). This book uses SI units.

EXPOSURE

Exposure is a measure of radiation quantity, the capacity of radiation to ionize air. The roentgen ® is the traditional unit of radiation exposure measured in air, 1 R is that amount of x-radiation or gamma radiation that produces 2.08×10^9 ion pairs in 1 cc of air (STP). It measures the intensity of radiation to which an object is exposed. No specific SI unit is equivalent to the R, but in terms of other SI units it is equal to coulombs per kilogram (C/kg); $1 \text{ R} = 2.58 \times 10^4 \text{ C/kg}$ equals $3.88 \times 10^3 \text{ R}$. **The roentgen applies only for x-rays and gamma rays.** In recent years the roentgen has been replaced by air kerma, and acronym for kinetic energy released in matter. Kerma measures the KE transferred from photons to electrons and is expressed in units of dose (Gy).

ABSORBED DOSE

Absorbed dose is a measure of the energy absorbed by any type of ionizing radiation per unit mass of any type of matter. The SI unit is the gray (Gy) – 1 Gy equals 1 joule/kg. The traditional unit of absorbed dose is the Rad (radiation absorbed dose), where 1 rad is equivalent to 100 ergs/g of absorber.

One gray equals 100 rads.

EQUIVALENT DOSE

The equivalent dose (H_T) is used to compare the biologic effects of different types of radiation on a tissue or organ.

It is expressed as a sum to allow for the time possibility that the tissue or organ has been exposed to more than one type of radiation. The radiation weighting factor is chosen for the type and energy of the radiation involved. Therefore high – LET radiations (which are more damaging to tissue than low LET radiations) have a correspondingly higher W_R . The unit of equivalent dose is the sievert (Sv). For diagnostic x-ray examinations, 1 Sv equals 1 Gy. The tradition unit of equivalent dose is the rem (roentgen equivalent man).

One Sievert equals 100 rems

EFFECTIVE DOSE.

The effective dose (E) is used to estimate the risk in humans.

The tissue weighting factors include gonads, 0.20; red bone marrow, 0.12; esophagus, 0.05; thyroid, 0.05; thyroid, 0.05; skin, 0.01; and bone surface, 0.01. The unit of effective dose is the sievert (Sv).

RADIOACTIVITY

The measurement of radioactivity (A) describes the decay rate of a sample of radioactive material. The SI unit is the becquerel (Bq). 1 Bq equals 1 disintegration/second. The traditional unit is the curie (Ci), which corresponds to the activity of 1 g of radium.