Requisites for Generating X-rays

X-rays are produced when high-speed electrons are decelerated.

This is the fundamental principle used for the generation of x-rays, whether for diagnostic or therapy purposes. To generate x-rays you need three things:

1. a means to produce electrons; 2. a means to accelerate the electrons to high speed; and 3. a means to stop them abruptly.

The **tube cathode** (filament, eg. tungsten) is heated with a low-voltage current of a few amps.

The filament heats up and the electrons in the wire become loosely held.

A large electrical potential is created between the cathode and the anode by the high-voltage generator.

The electrons that break free of the cathode are strongly attracted to the anode.

The stream of electrons between the cathode and the anode is the tube current.

When the electrons are slowed or stopped by the interaction with the atomic particles of the target, x-rays are produced.

With most x-ray tubes used in diagnostic radiology, electrons are accelerated towards a tungsten anode (target) by applying a large accelerating voltage between the anode and the cathode. After acceleration at impact with the target, the electron will have an amount of energy that is directly proportional to the instantaneous applied voltage. However, very few electrons acquire a kinetic energy numerically equivalent to the kVp applied to the tube.

Requisites for Generating X-rays (cont)

Therefore, even fewer x-rays are emitted with this energy since the **bremsstrahlung process** generally involves the production of a large number of low energy photons rather than the emission of a single photon with energy equal to the incident electron. Thus, the bremsstrahlung spectrum will be continuous with all energies present up to a maximum energy determined by the maximum accelerating voltage applied to the tube.

Producing Electrons

Thermionic emission is the name given to the process whereby a hot metal gives off low energy electrons.

At room temperature the electrons in an atom occupy orbitals around the nucleus and these are usually the lowest energy, most stable orbits closest to the nucleus with the strongest binding energy.

If an atom is heated, the extra energy can result in the electrons moving into higher energy orbitals, further from the nucleus and with lower binding force.

If the temperature of the atom is sufficiently high it is possible for the electron to break free of the nucleus altogether.

The standard way of obtaining electrons is from a metal filament, like that in a light globe, which is heated by passing an electric current through it.

The temperature of the filament governs how many electrons are produced and this is controlled by the magnitude of the electric current.

The x-ray output depends directly on the number of electrons emitted and so controlling the filament current is one way of controlling the x-ray output.

Producing Electrons (cont)

The higher the temperature of the filament, the larger the number of electrons that leave the cathode and travel to the anode, the greater the intensity of the X-ray output.

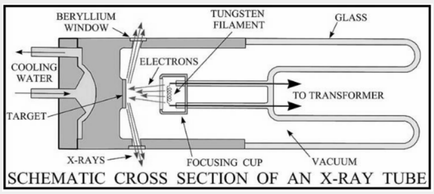
The high-voltage between the cathode and the anode affects the speed at which the electrons travel and strike the anode.

The higher the kilovoltage, the more speed and, therefore, energy the electrons have when they strike the anode.

Electrons striking with more energy results in x-rays with more penetrating power.

The high-voltage potential is measured in kilovolts, an increase in the kilovoltage will also result in an increase in the intensity of the radiation.

Schematic of an x-ray tube



SCHEMATIC CROSS SECTION OF AN X-RAY TUBE

Accelerating Electrons to High Speed

There are two common ways to accelerate electrons to high speed.

The first and simplest is to use a large electric field to supply the force needed - in a process analogous to a mass falling under the influence of a gravitational field.

The second method is to use the electric field associated with electromagnetic radiation. The latter method is analogous to a surfer on a board riding a wave.

Using the first method, it is possible to give electrons energy up to about 500 keV. With the second method the energy attainable can be much higher, with energies of 25 MeV and more achievable.

Decelerating High Speed Electrons: X-ray Targets

An electron that strikes the target with a given kinetic energy will undergo several different interactions with target atoms before it comes to rest and dissipates all of its kinetic energy in the target.

Two classes of electron interactions with a target atom:

1. with orbital electrons of the target atoms
2. with nuclei of the target atoms

Incident electron interaction with orbital electron of a target atom results mainly in collision loss and ionisation of the target atom that may be accompanied by an energetic electron referred to as a delta ray.

The collision loss will be followed by emission of characteristic x-rays and Auger electrons.

Incident electron interaction with the nucleus of a target atom results mainly in elastic scattering events but may also result in radiative loss accompanied with bremsstrahlung production.

As can be seen in Figure 7.3, the peak x-ray intensity occurs at a characteristic angle θmax that depends on the kinetic energy of the incident electrons:

1.In the diagnostic energy range (50 kVp to 120 kVp), the photons are emitted approximately equally in all directions. The θmax is 90°, and so the x-ray tube is constructed with what is called a **reflective target** so that the useful beam is at 90°to the direction of the electrons.

2. However, the direction of the x-ray beam becomes more forward peaked as the energy of the electrons reach the mega-electron voltage range. In the megavoltage radiotherapy range θmax is 0°, and the target is referred to as a **transmission target**, where the generated x-ray photons continue in the same directions as the bombarding electrons.

Decelerating High Speed Electrons: X-ray Targets (cont)

Electrons are light negatively charged particles and interact readily with any atom that they encounter, due to the positive charge of the nucleus.

The larger the charge of the nucleus, in other words the higher the atomic number, the stronger the interaction.

For this reason materials with high atomic number are chosen for the target of the high-energy electrons. As the atomic number of the target material increases, the efficiency of the continuous spectrum x-rays increase.

The discrete spectrum also shifts to the right representing higher energy characteristic radiation.

Unfortunately the production of x-rays is **not very efficient** with most of the energy of the electrons being converted to heat (infrared radiation).

It is therefore an advantage if the target has a high melting point and is a good conductor of heat.

Suitable materials for the target include metals like tungsten, gold, lead, etc. either in the pure form or as alloys. Tungsten is used for general radiography, although some specialty tubes use gold. Molybdenum is used for mammography as it has a lower atomic number so the discrete spectrum is of a lower energy. This is ideal for soft tissue studies such as mammography.

Figure 7.3 Spatial distribution of x-rays

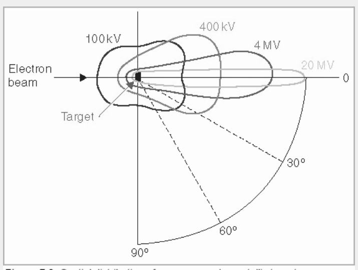


Figure 7.3 Spatial distribution of x-rays around a metallic target.

Rotating Anode

There are two categories of x-ray anodes: stationary and rotating.

As you might guess from the names, one anode stays still (stationary) while the other spins around a fixed point (rotating).

The reason for this difference is primarily related to dispersing heat. A rotating anode promotes cooling between exposures by distributing the intense beam from the cathode over the surface of the anode - the heat is dispersed evenly across the entire surface of the anode.

This enables rotating anode users to perform longer scans and at higher doses. **A rotating anode tube lasts a lot longer than a stationary x-ray tube.**

Design of a Practical X-ray Generator

To accelerate electrons we need a way to produce a high voltage or potential difference. It goes without saying that using 100,000 AA cells (1.5 volt batteries) to get 150 kV is hopelessly impractical, but fortunately we don't have to resort to those sorts of measures. It is possible to transform the normal household AC supply, 240 Volts in Australia, to the levels we need.

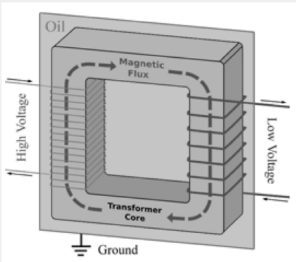
The transformer consists of a primary coil of wire and one or more secondary windings. Alternating current is applied to the primary and this induces a changing magnetic field. The changing magnetic field in turn induces an electric field in the secondary winding/s.

One of the properties of a transformer is that the ratio of the voltage in the primary winding to the voltage in the secondary is equal to the ratio of the turns in the two windings.

Design of a Practical X-ray Generator (cont)

To improve the efficiency of the transformer it is usually wound on a ferro-magnetic core comprised of thin metal sheets laminated together. Transformers are used both to step-up and step-down the input voltage. A step-down transformer is used to supply the low voltage to heat the filament and a step-up transformer to supply the large voltage to accelerate the electrons.

Schematic of a transformer



Properties of an X-ray Beam

Produced by the type of generator discussed here.

The first thing to note is that the beam contains a complete spectrum of x-radiation with energy ranging from very low to the maximum possible, that is the full energy of the electron that produced it.

The energy of an x-ray photon is equal to the amount of energy lost by the electron in its deceleration.

The deceleration of the electron is a result of its interaction, or collision, with the atoms of the target. There is an enormous range of interactions - from a near miss to a head-on collision - and the amount of energy transferred from the kinetic energy of the electron to the resultant photon ranges from almost none to all.

The photons with very low energy are less likely to leave the target.

Properties of an X-ray Beam (cont)

All the photons that are absorbed in the target contribute nothing to the x-ray beam, they merely heat it up. This represents the vast majority of the photons produced. The x-ray production process is only about 1% efficient; the remaining 99% of the energy of the electrons is converted to heat or infrared radiation.

The spectrum of radiation emitted from an x-ray tube extends from x-rays with sufficient energy to escape from the target and tube housing, to those with energy equal to the maximum electron energy.

The majority of the x-rays have an energy between these extremes.

These have a range or spectrum of energies and this is usually displayed as a graph of the number of x-ray photons as a function of their energy. The highest x-ray energy is determined by the peak voltage (kVp) applied between the anode and cathode.

The lower energy x-rays are preferentially absorbed by the x-ray tube and added filters so that no x-rays are seen below about 10 keV. The most probable x-ray energies are typically about one-third to one-half of the maximum energy.

Superimposed on the X-ray spectrum for a tungsten anode are sharply defined radiation lines. These are obtained when the incident electrons remove an electron from a given shell of the atoms within the anode. Once the electron has been removed, a characteristic x-ray is produced when another atomic electron fills the vacancy and emits the energy difference as the characteristic x-ray photon.

The x-rays that emerge through the exit window of the insert and housing ultimately pass through the patient to form the radiographic image.

Properties of an X-ray Beam (cont)

While it is possible to obtain the precise distribution of energies in the emitted spectrum, this is not the most useful information for the radiotherapy clinician. Of more value in radiotherapy is a knowledge of how the dose of radiation is distributed in the patient. For that we need to know how rapidly it is absorbed in soft tissue: its **depth dose curve.**

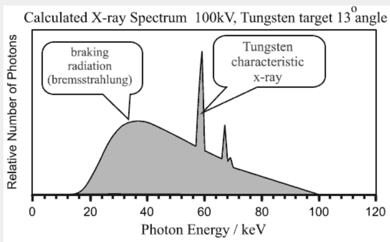
One way of characterising a beam is by its attenuation in various materials:

How much of a particular type of absorber will **reduce its intensity by half**, called the **half-value layer (HVL)**.

The HVL of a beam is a measure that indicates how penetrating a beam is. Common materials used for expressing the HVL are tissue, aluminium, copper and lead.

The standard practice is to quote the HVL in terms of a material that gives convenient numbers of mm, for example 4 mm of Al or 2.5 mm of Cu. There is an equivalence between the different materials and of course to the absorption in soft tissue, which is the most important quantity when it comes to treating a tumour. The HVL is called the quality of the beam.

X-ray spectrum for a tungsten anode



Properties of Kilovoltage Beams

Maximum dose occurs at the surface

A variable rate of dose fall off, depending on beam energy (sharper for lower energy beams)

Very sharp penumbra at the surface

High atomic number inhomogeneity causes markedly increased attenuation

Kilovoltage beams are typically only useful for superficial lesions, as they deposit 100% of their dose on the skin surface; this limits their application for deeper treatments. For energies under 150 keV, treatments are limited to lesions of < 0.5 cm thickness due to rapid dose fall off. Orthovoltage treatments are usually limited to lesions of < 2 cm in thickness.

Summary of Properties of an X-ray Beam

The beam from an x-ray generator comprises a spectrum of energies - it is polychromatic.

The minimum energy is determined by self-absorption in the target and tube.

The maximum energy is determined by the maximum energy of the electrons - the tube kV.

While it is possible to quantify a beam by its spectrum, a more useful way is by its HVL.

One of the quantities that determines the depth dose curve in soft tissue is the quality of the beam (others include the area or field size of the beam, and the treatment distance, to be dealt with later).

Controlling the X-ray Tube Output

Just as there are many different types of cancer and locations of tumours, there are different x-ray beams suitable to treat them.

Skin cancers, where the tumour is located in the superficial layers of the skin, require a beam that deposits dose in the first few millimetres of soft tissue and less of a dose to deeper organs.

Controlling the X-ray Tube Output (cont)

Tumours that lie deep in a patient require a more penetrating beam.

Adjusting the temperature of the filament, or cathode, by changing the current flowing in it alters the rate of electrons ‘boiled off’ the filament.

These electrons form the tube current and the number of electrons per time interval is directly related to the number of x-ray photons over that time interval.

The quantity of x-rays, the **dose-rate**, is directly proportional to the tube current, provided the potential difference across the anode to the cathode is constant. That is, the energy of the electrons as they hit the target is the same.

If this is the case the spectrum of x-ray energies emitted is also the same, as is the quality of the beam in terms of its HVL. If we think of water as an analogy, with temperature being analogous to quality, turning up the tube current is like opening up the tap: we get more water flowing but it's still the same temperature.

If we increase the potential difference between the filament and the target, cathode and anode, but keep the filament at the same temperature, we have the same number of electrons per time interval but they have more energy when they hit the target.

The result is the maximum energy an x-ray photon may have is increased, and, because it is possible for an electron to have more collisions before it loses all its energy, the number of x-rays per time interval also increases. Therefore, both the quality of the beam and the quantity of the beam increase.

Filtration: controlling quality of the x-ray beam

Increasing the accelerating potential of an x-ray tube increases the quality of the beam.

There is a physical limit to how far one can increase the accelerating potential using a transformer before technical difficulties start to arise.

High electric fields can break down the insulators, especially air, and lead to arcing or sparking.

The practical limit for x-ray tubes with length of the order of a metre is around 400-500 kV. The alternative method of increasing the quality of the beam is by filtration.

Filtration relies on the property of preferential absorption of low-energy x-rays compared to high-energy ones. If we put an absorber in the beam the spectrum of the beam is altered, with the low-energy x-rays being absorbed more rapidly than the high energy ones and the effective energy of the beam increases.

At the same time the overall dose-rate decreases, because some of the beam has been absorbed, but its quality, as expressed in its half-value layer, increases.

Many studies have shown that materials like aluminium, tin and copper make excellent filters and when used correctly can optimise the loss in output with the best possible gain in quality.

To control the energy spectrum produced by a kilovoltage machine, filters are placed in the path of the beam. These filters selectively attenuate the desired part of the beam spectrum; this usually **hardens the beam (removes low energy photons).**

Filtration has two major roles in kilovoltage radiotherapy:

Filtration: controlling quality of the x-ray beam (cont)

1. Filtration hardens the beam, attenuating low energy photons and shifting the spectrum towards higher energy photons. The low energy photons are not needed, as they would simply excess dose to the most superficial parts of the skin.

2. Filtration smoothes the beam spectrum, particularly with relation to the characteristic radiation produced in the target. This prevents excessive photons with unwanted energies from contributing to the dose.

Hence, the dose reduction rationale for filtration: **only higher-energy x-rays remain in the beam, these are more penetrating and less likely to be absorbed by tissues …. dose reduction!**

If filtration is absent, very-low-energy x-rays (<20keV) are most likely being totally absorbed by tissues, increasing the patient dose. These x-rays do not contribute to image formation, as they are being almost totally absorbed. Hence, **filtration is a MUST.**

The upper limit in beam quality for a conventional x-ray tube is about 4 mm Cu HVL. This can be achieved with a reasonable dose rate using ~ 300 kV and a combined aluminium, tin and copper filter.

In terms of depth dose in tissue this corresponds to ~65% at a depth of 5 cm for a 10 cm by 10 cm beam, i.e. 65% of the beam incident on the patient's skin reaches a depth of 5 cm. To treat deeper tumours than this requires multiple beams from different angles of incidence to ensure the skin tolerance is not exceeded.

Clearly, this is neither very deep nor very efficient and to treat deep tumours more energetic beams are required.

Superficial and Orthovoltage Radiotherapy

Superficial and Orthovoltage radiotherapy utilise low energy ionising radiation to treat cancer and other conditions that occur either on or close to the skin surface.

Superficial and Orthovoltage Radiotherapy (cont)

Superficial radiotherapy utilises x-ray energies of between 50 and 200 keV, having a treatment range of up to 5mm, and Orthovoltage radiotherapy utilises 200 to 500 keV x-rays penetrating to a useful depth of 4 – 6cm.

The shallow penetrating power of both techniques means that they are often superior to megavoltage external beam radiation for the treatment of superficial lesions.

Orthovoltage and superficial treatment machines are becoming less common, with much of the treatment that was previously delivered with them now being delivered using linear accelerators.

Orthovoltage units use x-rays with energies of 200 - 500 keV. They have a similar design to standard x-ray tubes, but also include:

Extra shielding around the target anode to absorb the higher energy scattered photons.

Increased voltage between the cathode and the anode to increase the energy of generated photons

Jaws may be used to alter the beam shape and size as it emerges from the tube.

Orthovoltage Unit

The radiation from orthovoltage units is referred to as x-rays, generated by bombarding a metallic target (tungsten) with high-energy electrons. This is a relatively low energy radiation source; typically operated at 250 kV.

The maximum dose is deposited at the skin surface and dose falls to 90% at ~2 cm of depth in the tissue. As a result the acute effects to the skin can be severe, but it is difficult to treat deep-seated tumors due to the limitations of the radiation tolerance of the overlying tissues; the skin dose becomes prohibitively large when adequate doses are to be delivered to deep-seated tumors.

Superficial and Orthovoltage Radiotherapy (cont)

Additionally, there is differential absorption of dose in bone versus soft tissue and there is some risk of bone damage or necrosis.

Orthovoltage irradiation is primarily suited for treatment of superficial tumors that do not involve adjacent bone. Applications include primarily skin tumors, and nasal cavity tumors after cytoreductive surgery.

Orthovoltage units are operated at a relatively short source-to-skin distance (usually 50 cm) limiting the size of the treatment field; the field size is defined by the use of different sized/shaped attachments or cones (rectangular, circular, slanted). Orthovoltage units are relatively inexpensive machines, relatively easy to repair and maintain, and less shielding and space is required for operation.

Radiography

Radiographic imaging is not only an important modality for detecting the presence of abnormalities in the body, but in cancer therapy it plays a critical role in identifying the extent of disease and its accurate localisation for radiation therapy.

As well as utilising information obtained from x-ray equipment readily available in the radiology department, radiation therapy technologists operate specially designed equipment that can image the patient in set-ups that simulate the proposed treatment position.

For obvious reasons, this equipment is in fact called a **simulator**.

Additionally, many treatment procedures are checked directly on the treatment machines in processes known as **electronic portal imaging** or **on board imaging**.

Radiographic Grid

Radiography (cont)

When x-rays pass through the patient, they are attenuated by the processes of absorption and scattering.

The contrast produced in a radiograph is the result of photoelectric attenuation.

his is dependent on the atomic number and the density of the tissue. However, Compton scattered radiation is also present and this degrades the quality of the image by randomly irradiating the whole area, increasing the fog level and reducing the contrast.

Scattered radiation can be reduced by the use of a **grid** which is composed of thin equally spaced lead strips in the range of 20 to 40 strips per centimetre.

Focused grids are often used in which the strips of lead are angled from the centre to the outside border to accommodate the divergence of the x-ray beam.

The grid, which is placed between the patient and the film, allows x-rays travelling from the tube focus to pass through unimpeded to the film.

However, the passage of scattered radiation is substantially limited although not completely eliminated.

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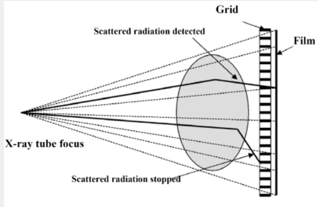
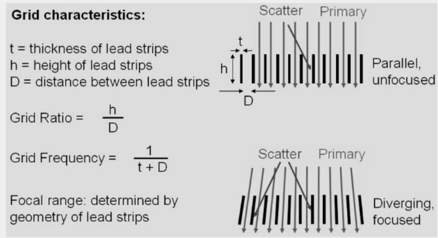
Basic features of parallel radiographic grid



Grid



Radiographic Image

X-rays are extremely penetrating radiations and the degree of penetration in a given medium depends in part on the density of that medium.

In the human body, four distinct types of tissues are present each providing a different degree of attenuation to the x-ray beam:

1. air tissue (lung) which is the least dense or radiolucent
2. bone tissue which is the most dense or radiopaque
3. adipose tissue (fat)
4. liquid tissue

In x-radiography, these differences in attenuation are utilised to provide shadows of varying density on the radiographic image.

A well-exposed radiograph will demonstrate sufficient overall blackening or density, good contrast between the various structures imaged and sharply defined detail with a minimum of distortion.

Magnification Effect

All radiographs will demonstrate magnification since the three-dimensional structure of the human body is being displayed in the two-dimensional format of the image.

Magnification Effect (cont)

Additionally, the magnitude of this magnification varies through the thickness of the patient being a maximum at the beam entrance surface and a minimum at the exit surface.

Since a radiograph displays 3-dimensional structures on a 2-dimensional image it is impossible to tell the depth of any structure from one image. At least one additional radiograph taken orthogonal to the original would be required to gain this information.

Although you may appreciate that the resulting image obtained with conventional radiography is magnified when compared to the original object, it may not be so clear that the magnitude of this magnification varies through the thickness of the patient (depth within the patient), being a maximum at the beam entrance surface and a minimum at the exit surface.

In the more traditional forms of radiation therapy planning, patients were radiographed with lead markers of known dimensions placed on both the anterior and posterior parts of their body (often a circle in one location and a cross in the other for easy identification on the film) so that the magnification at the front and the back could be determined. This then allowed the magnification at any relevant depth (perhaps the site of a tumour) to be determined by interpolation.

Radiographic image quality

This refers to the exactness of representation of anatomic structures on a radiograph.

A radiograph is of high quality if it faithfully reproduces structures and tissues. Radiographic quality depends on a number of complex factors and is not easy to define or measure precisely.

Three important characteristics are: 1. spatial resolution; 2. contrast resolution; 3. noise.

Spatial Resolution

This refers to the ability to image two separate objects that have high subject contrast such as small, calcified lung nodules or breast microcalcifications.

Resolution is measured by the ability to see pairs of lines and is expressed as line pairs per millimetre (lp/mm). Conventional radiography has excellent spatial resolution.

Contrast Resolution

This refers to the ability to distinguish anatomic structures of similar subject contrast such as liver-spleen and grey-white brain matter.

Computed Tomography (CT) scanners and Magnetic Resonance Imaging (MRI) scanners have excellent contrast resolution.

Noise

Radiographic noise is the undesirable fluctuations in the optical density of the image. A reduction in noise results in increased contrast resolution and therefore improved image quality.

Radiotherapy Simulator

A radiotherapy simulator is a specially designed x-ray machine that replicates most of the functions of the **medical linear accelerator**.

The simulator has the manoeuvrability and accuracy of a linear accelerator and can provide radiation beams that are identical in size and position to those used in all treatment plans. Unlike a linear accelerator, the simulator contains a conventional x-ray tube and an image intensifier. These combine to provide real time imaging and high quality radiographs with the patient positioned for the proposed treatment plan. The simulator will always be able to provide improved image quality compared to images obtained with a linear accelerator.

Radiotherapy Simulator (cont)

The image intensifier is a highly evacuated tube that contains a fluorescent screen. This allows us to observe an x-ray image as it changes with time in a technique known as fluoroscopy. The fluorescent screen is contained in a highly evacuated tube known as an image intensifier. It converts the x-ray pattern received from the patient into a light image, which is then viewed by a camera and displayed on a TV monitor.

Consequently, the use of the simulator has greatly improved the processes of tumour localisation, verification and reproducibility of treatment set-up. Radiation treatment planning also makes considerable use of the images obtained using computed tomography (CT) imaging.