CHAPTER 5.

TREATMENT MACHINES FOR EXTERNAL BEAM RADIOTHERAPY

ERVIN B. PODGORSAK
Department of Medical Physics
McGill University Health Centre
Montréal, Québec, Canada

5.1. INTRODUCTION

Since the inception of radiotherapy soon after the discovery of x-rays by Roentgen in 1895, the technology of x-ray production has first been aimed toward ever higher photon and electron beam energies and intensities, and more recently toward computerization and intensity-modulated beam delivery. During the first 50 years of radiotherapy, the technological progress has been relatively slow and mainly based on x-ray tubes, Van de Graaff generators and betatrons.

The invention of the cobalt-60 teletherapy unit by H.E. Johns in Canada in the early 1950s provided a tremendous boost in the quest for higher photon energies, and placed the cobalt unit into the forefront of radiotherapy for a number of years. The concurrently developed medical linear accelerators (linacs), however, soon eclipsed the cobalt unit, moved through five increasingly sophisticated generations, and became the most widely used radiation source in modern radiotherapy. With its compact and efficient design, the linac offers excellent versatility for use in radiotherapy through isocentric mounting and provides either electron or megavoltage x-ray therapy with a wide range of energies.

In addition to linacs, electron and x-ray radiotherapy is also carried out with other types of accelerators, such as betatrons and microtrons. More exotic particles, such as protons, neutrons, heavy ions, and negative π mesons, all produced by special accelerators, are also sometimes used for radiotherapy; however, most of the contemporary radiotherapy is carried out with linacs or teletherapy cobalt units.

5.2. X-RAY BEAMS AND X-RAY UNITS

- Clinical x-ray beams typically range in energy between 10 kVp and 50 MV, and are produced when electrons with kinetic energies between 10 keV and 50 MeV are decelerated in special metallic targets.

- In the target, most of the electron's kinetic energy is transformed into heat and a small fraction of the energy is emitted in the form of x-ray photons which are divided into two groups: *characteristic x-rays* and *bremsstrahlung x-rays*.

5.2.1. Characteristic x-rays

- Characteristic x-rays result from Coulomb interactions between the incident electrons and atomic orbital electrons of the target material (collisional loss).

- In a given Coulomb interaction between the incident electron and an orbital electron, the orbital electron is ejected from its shell and the resulting orbital vacancy is filled by an electron from a higher level shell. The energy difference between the two shells may be emitted from the atom either in the form of a characteristic photon (characteristic x-ray) or is transferred to an orbital electron which is ejected from the atom as an Auger electron.

- The fluorescent yield ω gives the number of fluorescent (characteristic) photons emitted per vacancy in a shell $(0 \le \omega \le 1)$ and ranges from 0 for low Z atoms through 0.5 for copper ($Z = 29$) to 0.96 for high Z atoms for K-shell vacancies that are the most prominent sources of characteristic x-rays.

- The photons emitted through electronic shell transitions have discrete energies that are characteristic of the particular target atom in which the transitions have occurred; hence the term characteristic radiation.

5.2.2. Bremsstrahlung (continuous) x-rays

- Bremsstrahlung x-rays result from Coulomb interactions between the incident electron and the nuclei of the target material.

- During the Coulomb interaction between the incident electron and the nucleus, the incident electron is decelerated and loses part of its kinetic energy in the form of bremsstrahlung photons (radiative loss).

- Photons with energies ranging from 0 to the kinetic energy of the incident electron may be produced, resulting in a continuous bremsstrahlung spectrum.

- The bremsstrahlung spectrum produced in a given x-ray target depends on the kinetic energy of the incident electron as well as on the thickness and atomic number Z of the target.

5.2.3. X-ray targets

- In comparison with the range R of electrons of a given kinetic energy KE in the target material, targets are divided into two main groups: *thin* and *thick*.

- A *thin target* has a thickness much smaller than R, while the thickness of a thick target is on the order of R.

- For thin target radiation, the energy radiated is proportional to the product $(KE) \times Z$, where Z is the target atomic number. The intensity versus photon energy (photon spectrum) is constant from 0 to KE, and 0 for all energies above KE.

- A *thick target* may be considered as consisting of a large number of superimposed thin targets. The intensity $I(h\nu)$ of a thick target spectrum is expressed as:

$$I(h\nu) = CZ(KE - h\nu) \quad , \tag{5.1}$$

where

C is a proportionality constant and
$h\nu$ is the photon energy.

- X-rays are used in diagnostic radiology for diagnosis of disease and in radiation oncology (radiotherapy) for treatment of disease.

- X-rays produced by electrons with kinetic energies between 10 keV and 100 keV are called *superficial x-rays*, with electron kinetic energies between 100 keV and 500 keV *orthovoltage x-rays*, and with electron kinetic energies above 1 MeV *megavoltage x-rays*.

- Superficial and orthovoltage x-rays are produced with x-ray tubes (machines), while megavoltage x-rays are most commonly produced with linacs and sometimes with betatrons and microtrons.

- Typical thin and thick target bremsstrahlung spectra originating from 100 keV electrons striking a thin and thick target, respectively, are shown in Fig. 5.1.

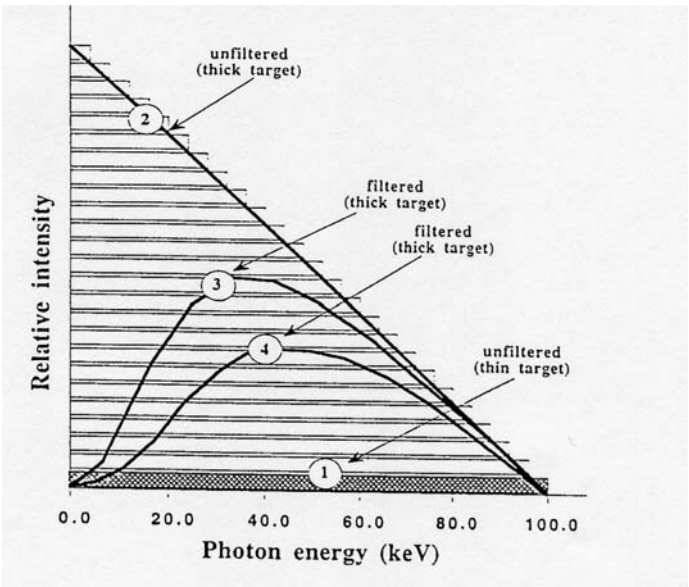


FIG. 5.1. Typical thin target (curve 1) and thick target (curves 2, 3, and 4) spectra for an x-ray tube in which 100 keV electrons strike the target. Curve (1) is for a thin target producing a constant intensity for photon energies from 0 to the kinetic energy of electrons striking the target (100 keV). Curve (2) represents unfiltered spectrum (inside the x-ray tube) for a thick target and represents a superposition of numerous thin target spectra; spectrum of curve (3) is for a beam filtered by x-ray tube window (low energy photons are filtered out); spectrum of curve (4) is for beam filtered by the x-ray tube window and additional filtration.

5.2.4. Clinical x-ray beams

- A typical spectrum of a clinical x-ray beam consists of line spectra that are characteristic of the target material and are superimposed onto the continuous bremsstrahlung spectrum.

- The bremsstrahlung spectrum originates in the x-ray target, while the characteristic line spectra originate in the target and in any attenuators placed into the beam.

- The relative proportion of the number of characteristic photons to bremsstrahlung photons in an x-ray beam spectrum varies with electron beam kinetic energy and atomic number of the target. For example, x-ray beams produced in a tungsten target by 100 keV electrons contain about 20% characteristic photons and 80% bremsstrahlung photons, while in the megavoltage range the contribution of characteristic photons to the total spectrum is negligible.

- In the diagnostic energy range (10 to 150 kV) most photons are produced at 90° from the direction of electron acceleration, while in the megavoltage energy range (1 to 50 MV) most photons are produced in the direction of electron acceleration (forward direction: 0°).

5.2.5. X-ray beam quality specifiers

Various parameters, such as photon spectrum, half-value layer, nominal accelerating potential, beam penetration into tissue-equivalent media, etc., are used as x-ray beam quality indices(ee Sections 9.8.1 and 9.8.2 for details):

- Complete *x-ray spectrum* is very difficult to measure; however, it gives the most rigorous description of beam quality.

- *Half-value layer (HVL)* is practical for beam quality description in the superficial (*HVL* in aluminum) and orthovoltage (*HVL* in copper) x-ray energy range, but not practical in the megavoltage energy range because in this energy range the attenuation coefficient is only a slowly varying function of beam energy .

- The *effective energy* of a heterogeneous x-ray beam is defined as that energy of a monoenergetic photon beam that yields the same *HVL* as does the heterogeneous beam.

- *Nominal accelerating potential (NAP)* is sometimes used for describing the megavoltage beam quality. The *NAP* is determined by measuring the ionisation ratio in a water phantom at depths of 10 and 20 cm for a 10×10 cm^2 field at the nominal source-axis distance of 100 cm.

- Recent dosimetry protocols recommend the use of *tissue-phantom ratios* or *percentage depth doses* at a depth of 10 cm in a water phantom as an indicator of megavoltage beam effective energy (beam quality index).

5.2.6. X-ray machines for radiotherapy

- Superficial and orthovoltage x-rays used in radiation therapy are produced with x-ray machines. The main components of a radiotherapeutic x-ray machine are: an *x-ray tube*; *ceiling* or *floor mount* for the x-ray tube; *target cooling system*; *control console*; and an *x-ray power generator*. A schematic diagram of a typical therapy x-ray tube is shown in Fig. 5.2.

- The electrons producing the x-ray beams in the x-ray tube (Coolidge tube) originate in the heated filament (cathode) and are accelerated in vacuum toward the target (anode) by an essentially constant-potential electrostatic field supplied by the x-ray generator.

- The efficiency for x-ray production in the superficial and orthovoltage energy range is on the order of 1% or less. Most of the electron kinetic energy deposited in the x-ray target (≈ 99%) is transformed into heat and must be dissipated through an efficient target cooling system.

- To maximize the x-ray yield in the superficial and orthovoltage energy range the target material should have a high atomic number Z and a high melting point.

- With x-ray tubes, the patient dose is delivered using a timer and the treatment time must incorporate the shutter correction time (see Section 6.16) that accounts for the time required for the power supply components to attain the steady state operating conditions.

- The x-ray tube current is controlled by hot filament emission of electrons which, in turn, is controlled by the filament temperature (thermionic emission). For a given filament temperature the x-ray tube current increases with the tube (anode) voltage, first rising linearly with voltage in the space charge limited region and saturating at higher voltages when all electrons emitted from the cathode are pulled to the anode.

- Research is currently carried out on cold field emission cathodes produced with carbon nanotubes (CNT). The CNT-based cold cathode x-ray technology may lead to more durable as well as miniature and portable x-ray sources for industrial and medical applications.

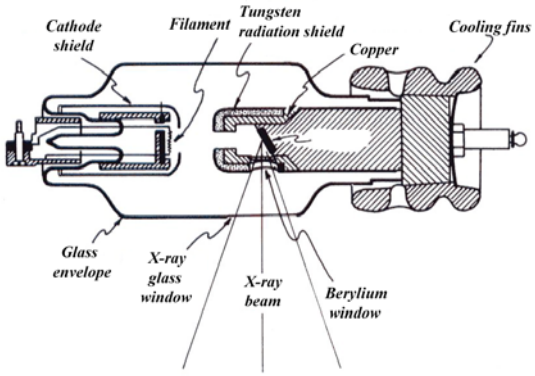


FIG. 5.2. Schematic diagram of a typical therapy x-ray tube (Reprinted from Johns, H.E. and Cunningham, J.R. with permission).

5.3. GAMMA RAY BEAMS AND GAMMA RAY UNITS

5.3.1. Basic properties of gamma rays

- For use in external beam radiotherapy, gamma rays are obtained from specially designed and built sources that contain a suitable, artificially-produced radioactive material.

- The parent source material undergoes a β decay resulting in excited daughter nuclei which attain ground state through emission of gamma rays (gamma decay).

- The important characteristics of radioisotopes in external beam radiotherapy are:

  (i) *High gamma ray energy*;
  (ii) *High specific activity*;
  (iii) Relatively *long half-life*;
  (iv) *Large specific air-kerma rate constant* Γ_{AKR}.

- The *specific activity a* (activity $\mathcal{A}$ per mass *m* of radioactive nuclide) is inversely proportional to half-life $t_{1/2}$, *i.e.*,

$$a = \frac{\mathcal{A}}{m} = \frac{N_{\mathrm{A}} \ln 2}{t_{1/2} A} \quad , \tag{5.2}$$

  where

  N_{A} is Avogadro's number (6.022×10^{23} atoms/g-atom),
  A is the atomic mass number.

- The *air-kerma rate in air* $(\dot{K}_{\mathrm{air}})_{\mathrm{air}}$ is given by the following relationship:

$$(\dot{K}_{\mathrm{air}})_{\mathrm{air}} = \frac{\mathcal{A}\,\Gamma_{\mathrm{AKR}}}{d^2} \quad , \tag{5.3}$$

  where

  $\mathcal{A}$ is the source activity and
  d is the distance between the point of interest and the point source.

- The basic physical properties of the two gamma emitters (cobalt-60 and cesium-137) currently used for external beam teletherapy and a potential source for teletherapy units (europium-152) are listed in the Table 5.I. Of the three radioisotopes cobalt-60 is the most widely used, since it offers the most practical approach to external beam radiotherapy, considering the energy of emitted photons, half-life, specific activity, and means of production.

TABLE 5.I. PHYSICAL PROPERTIES OF RADIONUCLIDES USED IN EXTERNAL BEAM RADIOTHERAPY

Radioactive source	***Cobalt-60*** *Co-60*	***Cesium-137*** *Cs-137*	***Europium-152*** *Eu-152*
Half life (year)	5.3	30	13.4
Specific activity (Ci/g)	1100[(a)] (~ 250[(b)])	80	180[(a)] (~ 150[(b)])
Photon energies (MeV)	1.17 and 1.33	0.662	0.6 - 1.4
Specific gamma rate constant $\Gamma\ \left[\mathrm{R}\cdot\mathrm{m}^2/(\mathrm{Ci}\cdot\mathrm{hr})\right]$	1.31	0.33	1.06
Specific air-kerma rate constant $\Gamma_{\mathrm{AKR}}\left[\mu\mathrm{Gy}\cdot\mathrm{m}^2/(\mathrm{GBq}\cdot\mathrm{hr})\right]$	309	78	250
Half value layer *HVL* (cm Pb)	1.1	0.5	1.1
Means of production	^{59}Co + n in reactor	fission by-product	^{151}Eu + n in reactor

(a) *Theoretical specific activity:* $a = (N_{\mathrm{A}} \ln 2)/(t_{1/2} A)$.

(b) *Practical specific activity is smaller than the theoretical specific activity because the source is not carrier-free, i.e., the source contains stable isotopes in addition to radioactive isotopes (for example, cobalt-59 mixed with cobalt-60).*

5.3.2. Teletherapy machines

- Treatment machines incorporating gamma ray sources for use in external beam radiation therapy are called teletherapy machines.
- Teletherapy machines are most often mounted isocentrically allowing the beam to rotate about the patient at a fixed source-axis distance *(SAD)*.
- The main components of a teletherapy machine are: radioactive source; source housing including beam collimator and source movement mechanism; gantry and stand in isocentric machines or housing support assembly in stand-alone machines; patient support assembly; and machine console.

5.3.3. Teletherapy sources

- The most widely used teletherapy source is cobalt-60.
- The gamma teletherapy source is contained inside a cylindrical stainless-steel capsule and sealed by welding.
- A double-welded seal is used to prevent any leakage of the radioactive material.

- To facilitate interchange of sources from one teletherapy machine to another and from one isotope production facility to another, standard source capsules have been developed.

- Typical diameter of the cylindrical teletherapy source is between 1 and 2 cm, the height of the cylinder is about 2.5 cm. The smaller the source diameter, the smaller is its physical penumbra and the more expensive is the source. Often a diameter of 1.5 cm is chosen as a compromise between the cost and penumbra.

- Typical source activities are on the order of 5,000 to 10,000 Ci (185 to 370 TBq), and provide a typical dose rate at 80 cm from the teletherapy source on the order of 100 to 200 cGy/min. Often the output of a teletherapy machine is stated in Rmm (roentgens per minute at 1 m) as a rough guide for the source strength.

- Teletherapy sources are usually replaced within one half-life after they are installed; however, financial considerations often result in longer source usage.

5.3.4. Teletherapy source housing

- The housing for the teletherapy source is called the *source-head*. It consists of a steel shell with lead for shielding purposes and a mechanism for bringing the source in front of the collimator opening to produce the clinical gamma ray beam.

- Currently two methods are in use for moving the teletherapy source from the BEAM OFF into the BEAM ON position and back: (i) source on a *sliding drawer* and (ii) source on a *rotating cylinder*. Both methods incorporate a safety feature in which the beam is terminated automatically in case of power failure or emergency.

- When the source is in the BEAM OFF position, a light source appears in the BEAM ON position above the collimator opening, allowing an optical visualization of the radiation field, as defined by the machine collimators and any special shielding blocks.

- Some radiation will escape the unit even when the source is in the BEAM OFF position. The head leakage typically amounts to less than 1 mR/hr (0.01 mSv/hr) at 1 m from the source. International regulations require that the average leakage of a teletherapy machine head be less than 2 mR/hr (0.02 mSv/hr) at 1 m from the source.

5.3.5. Dose delivery with teletherapy machines

- The prescribed target dose is delivered with the help of two treatment timers: *primary* and *secondary*. The primary timer actually controls the treatment time, the secondary timer serves as a backup timer in case of the primary timer failure.

- The set treatment time must incorporate the shutter error which accounts for the travel time of the source from the BEAM OFF position toward the BEAM ON position at the start of irradiation and for the reverse travel at the end of irradiation.

5.3.6. Collimator and penumbra

- Collimators of teletherapy machines provide square and rectangular radiation fields typically ranging from 5×5 to 35×35 cm^2 at 80 cm from the source.

- The geometric penumbra, which results from finite source diameter, may be minimized by using small diameter sources and by using penumbra trimmers as close as possible to patient's skin (see Section 6.9 for further discussion of the penumbra).

5.4. PARTICLE ACCELERATORS

Numerous types of accelerators have been built for basic research in nuclear and high-energy physics, and most of them have been modified for at least some limited use in radiotherapy. Irrespective of the accelerator type two basic conditions must be met for particle acceleration:

(1) *Particle to be accelerated must be charged*
(2) *Electric field must be provided in the direction of particle acceleration*.

- The various types of accelerators differ in the way they produce the accelerating electric field and in how the field acts on the particles to be accelerated.

- As far as the accelerating electric field is concerned there are two main classes of accelerators: *electrostatic* and *cyclic*.

  - In *electrostatic accelerators* the particles are accelerated by applying an electrostatic electric field through a voltage difference, constant in time, whose value fixes the value of the final kinetic energy of the particle. Since the electrostatic fields are conservative, the kinetic energy that the particle can gain depends only on the point of departure and point of arrival and, hence, cannot be larger than the potential energy corresponding to the maximum voltage drop existing in the machine. The energy that an electrostatic accelerator can reach is limited by the discharges that occur between the high voltage terminal and the walls of the accelerator chamber when the voltage drop exceeds a certain critical value (typically 1 MV).

  - The electric fields used in *cyclic accelerators* are variable and non-conservative, associated with a variable magnetic field and resulting in some close paths along which the kinetic energy gained by the particle differs from zero. If the particle is made to follow such a closed path many times over, one obtains a process of gradual acceleration that is not limited to the maximum voltage drop existing in the accelerator. Thus, the final kinetic energy of the particle is obtained by submitting the charged particle to the same, relatively small, potential difference a large number of times, each cycle adding a small amount of energy to the kinetic energy of the particle.

- Examples of electrostatic accelerators used in medicine are: *superficial* and *orthovoltage x-ray tubes* and *neutron generators*. The most known example of a cyclic accelerator is the *linear accelerator (linac)*; other examples are *microtrons, betatrons* and *cyclotrons*.

5.4.1. Betatron

The betatron was developed in 1940 by D.W. Kerst as a cyclic electron accelerator for basic physics research; however, its potential for use in radiotherapy was realized soon thereafter.

- The machine consists of a magnet fed by an alternating current of frequency between 50 and 200 Hz. The electrons are made to circulate in a toroidal vacuum chamber (donut) that is placed into the gap between two magnet poles. A schematic diagram of a betatron is given in Fig. 5.3 (a).

- Conceptually, the betatron may be considered an analog of a transformer: the primary current is the alternating current exciting the magnet and the secondary current is the electron current circulating in the vacuum chamber (donut).

- The electrons are accelerated by the electric field induced in the donut by the changing magnetic flux in the magnet; they are kept in a circular orbit by the magnetic field present in the donut.

- In the 1950s betatrons played an important role in megavoltage radiotherapy. However, the development of linacs pushed them into oblivion because of the numerous advantages offered by linacs over betatrons, such as: much higher beam output (up to 10 Gy/min for linacs vs 1 Gy/min for betatrons); larger field size; full isocentric mounting; more compact design; and quieter operation.

5.4.2. Cyclotron

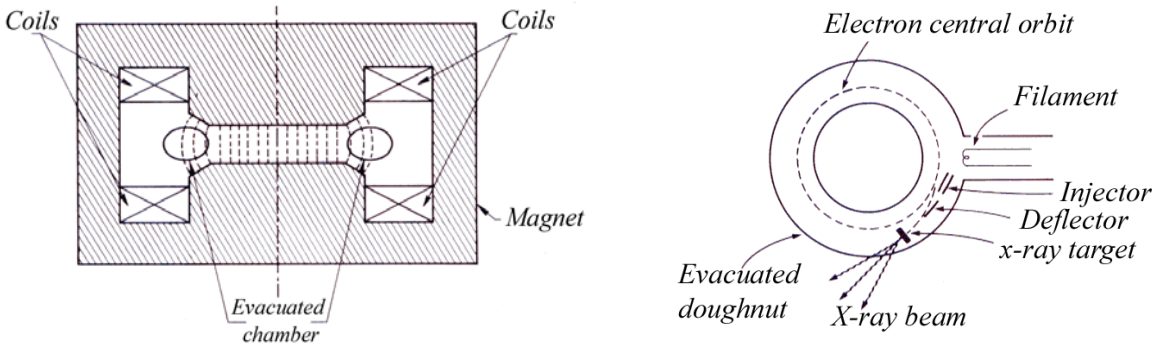
The cyclotron was developed in 1930 by E.O. Lawrence for acceleration of ions to a kinetic energy of a few MeV. Initially, the cyclotron was used for basic nuclear physics research but has later on found important medical uses in production of radioisotopes for nuclear medicine as well as in production of proton and neutron beams for radiotherapy. The recent introduction of the PET/CT machines for use in radiotherapy (see Section 15.10) has dramatically increased the importance of cyclotrons in medicine. The PET/CT machines rely on glucose labeled with positron-emitting fluorine-18 that is produced by proton cyclotrons.

- In a cyclotron the particles are accelerated along a spiral trajectory guided inside two evacuated half-cylindrical electrodes (referred to as dees because of their D-shape form) by a uniform magnetic field (1 tesla) that is produced between the pole pieces of a large magnet. A diagram of the cyclotron is given in Fig. 5.3 (b).

- A radiofrequency voltage with a constant frequency between 10 and 30 MHz is applied between the two electrodes and the charged particle is accelerated while crossing the gap between the two electrodes.

- Inside the electrodes there is no electric field and the particle drifts under the influence of the magnetic field in a semicircular orbit with a constant speed, until it crosses the gap again. If, in the meantime, the electric field has reversed its direction, the particle will again be accelerated across the gap, gain a small amount of energy, and drift in the other electrode along a semicircle of a larger radius than the former one, resulting in a spiral orbit and a gradual increase in kinetic energy after a large number of gap crossings.

(a) ***BETATRON***



(b) ***CYCLOTRON***

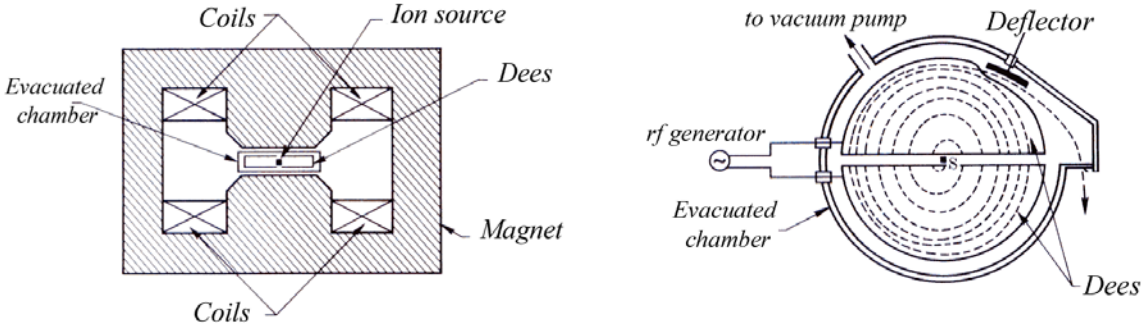


FIG. 5.3. Schematic diagrams for two cyclic accelerators: betatron in part (a) and cyclotron in part (b).

5.4.3. Microtron

The microtron is an electron accelerator, which combines the features of a linac and a cyclotron. The concept of the microtron was developed by V.I. Veksler in 1944 and the machine is used in modern radiotherapy, albeit to a much smaller extent than are linacs.

Two types of microtrons were developed: *circular* and *racetrack*.

- In the circular microtron the electron gains energy from a microwave resonant cavity and describes circular orbits of increasing radius in a uniform magnetic field. To keep the particle in phase with the microwave power, the cavity voltage, frequency, and magnetic field are adjusted in such a way that after each passage through the cavity, the electrons gain an energy increment resulting in an increase in the transit time in the magnetic field equal to an integral number of microwave cycles.

- In the racetrack microtron the magnet is split into two D-shaped pole pieces that are separated to provide greater flexibility in achieving efficient electron injection and higher energy gain per orbit through the use of multi-cavity accelerating structures similar to those used in linacs. The electon orbits consist of two semicircular and two straight sections.

5.5. LINEAR ACCELERATORS

- Medical linear accelerators (linacs) are cyclic accelerators which accelerate electrons to kinetic energies from 4 MeV to 25 MeV using non-conservative microwave *RF* fields in the frequency range from 10^3 MHz (L band) to 10^4 MHz (X band), with the vast majority running at 2856 MHz (S band).

- In a linear accelerator the electrons are accelerated following straight trajectories in special evacuated structures called accelerating waveguides. Electrons follow a linear path through the same, relatively low, potential difference several times; hence, linacs also fall into the class of cyclic accelerators just like the other cyclic machines that provide curved paths for the accelerated particles (*e.g.*, betatron)

- The high power *RF* fields, used for electron acceleration in the accelerating waveguides, are produced through the process of decelerating electrons in retarding potentials in special evacuated devices called *magnetrons* and *klystrons*.

- Various types of linacs are available for clinical use. Some provide x-rays only in the low megavoltage range (4 MV or 6 MV) others provide both x-rays and electrons at various megavoltage energies. A typical modern high energy linac will provide two photon energies (6 MV and 18 MV) and several electron energies (*e.g.*, 6, 9, 12, 16, 22 MeV)

5.5.1. Linac generations

During the past 40 years, medical linacs have gone through five distinct generations, making the contemporary machines extremely sophisticated in comparison with the machines of the 1960s. Each of the five generations introduced the following new features:

- **Low energy photons (4-8 MV)**:
  straight-through beam; fixed flattening filter; external wedges; symmetric jaws; single transmission ionisation chamber; isocentric mounting.

- **Medium energy photons (10-15 MV) and electrons**:
  bent beam; movable target and flattening filter; scattering foils; dual transmission ionisation chamber; electron cones.

- **High energy photons (18-25 MV) and electrons**:
  dual photon energy and multiple electron energies; achromatic bending magnet; dual scattering foils or scanned electron pencil beam; motorized wedge; asymmetric or independent collimator jaws.

- **High energy photons and electrons**:
  computer-controlled operation; dynamic wedge; electronic portal imaging device; multileaf collimator.

- **High energy photons and electrons**:
  photon beam intensity modulation with multileaf collimator; full dynamic conformal dose delivery with intensity modulated beams produced with a multileaf collimator.

5.5.2. Safety of linac installations

The complexity of modern linacs raises concerns as to safety of operation from the point-of-view of patients and operators. IEC, the International Electrotechnical Commission, publishes international standards that express, as nearly as possible, an international consensus of opinion on relevant technical subjects and electron linac is one of the subjects that the IEC addressed in detail. The IEC statement on the safety of linacs (IEC 60601-2-1, p. 13) is as follows: *"The use of electron accelerators for radiotherapy purposes may expose patients to danger if the equipment fails to deliver the required dose to the patient, or if the equipment design does not satisfy standards of electrical and mechanical safety. The equipment may also cause danger to persons in the vicinity if the equipment fails to contain the radiation adequately and/or if there are inadequacies in the design of the treatment room".*

The IEC document addresses three categories of safety issues: *electrical*, *mechanical*, and *radiation*, and establishes specific requirements mainly for the manufacturers of linacs in the design and construction of linacs for use in radiotherapy. It also covers some radiation safety aspects of linac installations in customer's treatment rooms.

5.5.3. Components of modern linacs

- The linacs are usually mounted isocentrically and the operational systems are distributed over five major and distinct sections of the machine:

  (1) gantry;
  (2) gantry stand or support;
  (3) modulator cabinet;
  (4) patient support assembly, i.e., treatment couch;
  (5) control console.

- A schematic diagram of a typical modern S-band medical linac is shown in Fig. 5.4. Also shown are the connections and relationships among the various linac components, listed above. The diagram provides a general layout of linac components; however, there are significant variations from one commercial machine to another, depending on the final electron beam kinetic energy as well as on the particular design used by the manufacturer.

- The length of the accelerating waveguide depends on the final electron kinetic energy, and ranges from ~30 cm at 4 MeV to ~150 cm at 25 MeV.

- The main beam-forming components of a modern medical linac are usually grouped into six classes:

  (1) injection system;
  (2) RF power generation system;
  (3) accelerating waveguide;
  (4) auxiliary system;
  (5) beam transport system; and
  (6) beam collimation and beam monitoring system

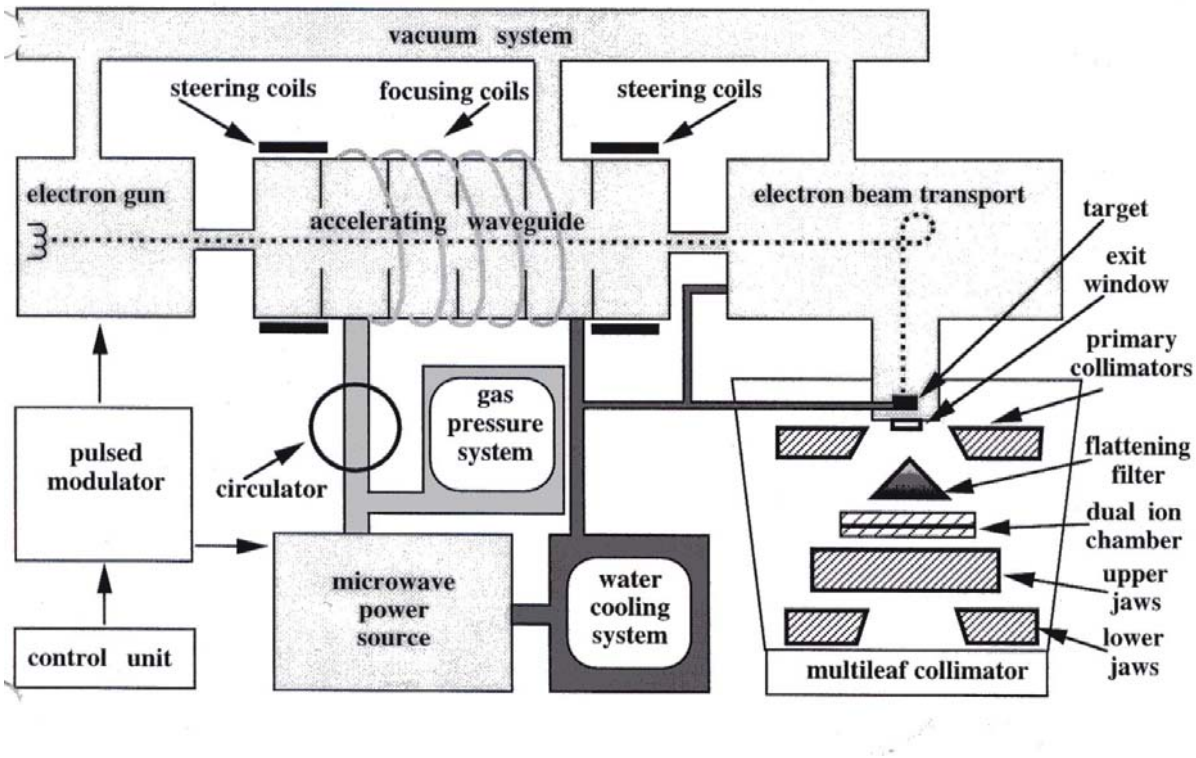


FIG. 5.4. Schematic diagram of a medical linear accelerator (linac).

5.5.4. Configuration of modern linacs

- At megavoltage electron energies the bremsstrahlung photons produced in the x-ray target are mainly forward-peaked and the clinical photon beam is produced in the direction of the electron beam striking the target.

- In the simplest and most practical configuration, the electron gun and the x-ray target form part of the accelerating waveguide and are aligned directly with the linac isocenter, obviating the need for a beam transport system. A straight-through photon beam is produced and the RF power source is also mounted in the gantry.

- The simplest linacs are isocentrically mounted 4 or 6 MV machines with the electron gun and target permanently built into the accelerating waveguide, thereby requiring no beam transport nor offering an electron therapy option.

- Accelerating waveguides for intermediate (8 to 15 MeV) and high (15 to 30 MeV) electron energies are too long for direct isocentric mounting, so they are located either in the gantry, parallel to the gantry axis of rotation, or in the gantry stand. A beam transport system is then used to transport the electron beam from the accelerating waveguide to the x-ray target. The *RF* power source in the two configurations is commonly mounted in the gantry stand. Various design configurations for modern isocentric linear accelerators are shown in Fig. 5.5.

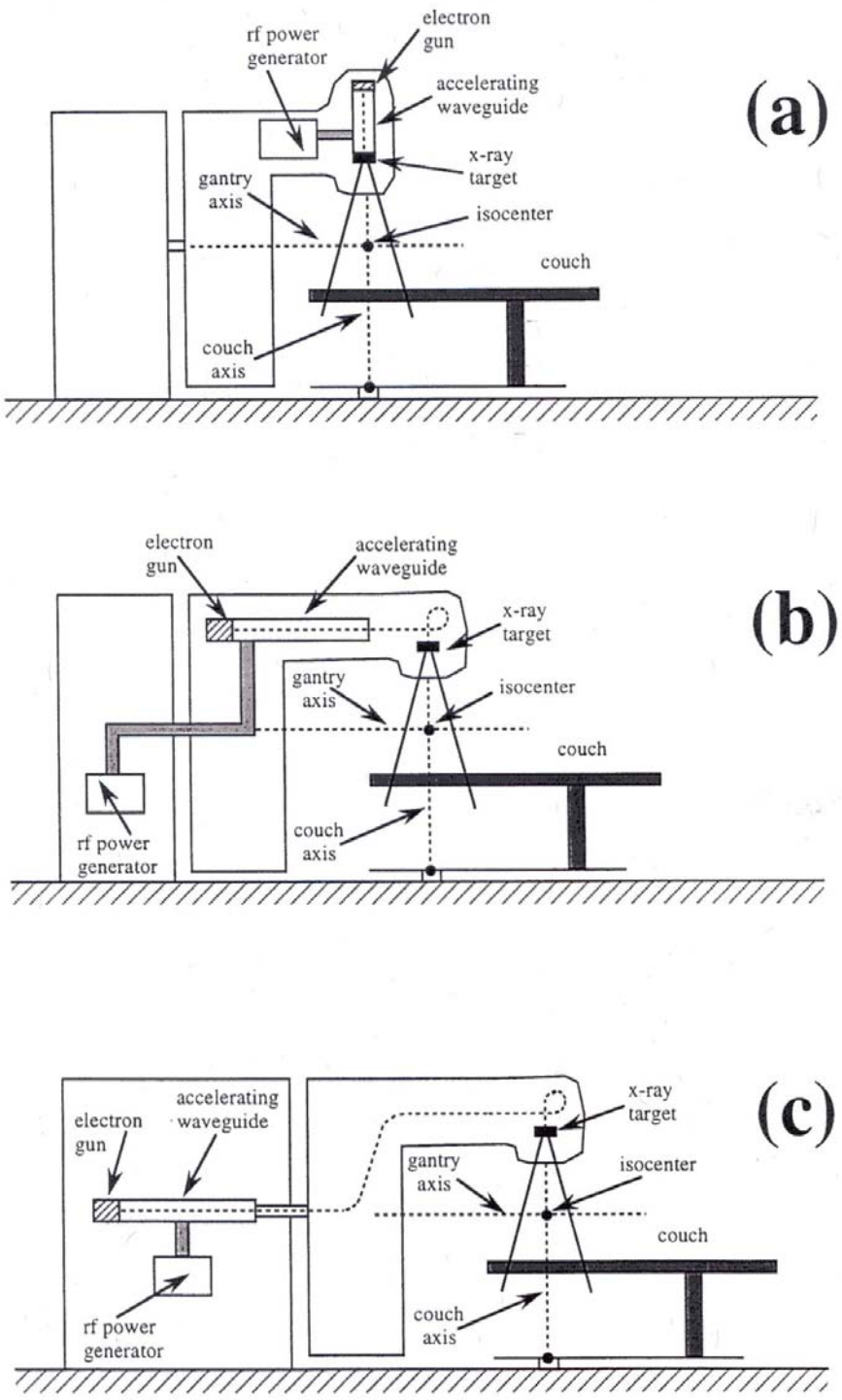


FIG. 5.5. Design configurations for isocentric medical linacs. (a) Straight-through beam design; the electron gun and target are permanently embedded into the accelerating wave-guide; machine produces only x-rays with energies of 4-6 MV; the rf-power generator is mounted in the gantry. (b) Accelerating wave-guide is in the gantry parallel to the isocenter axis; electrons are brought to the movable target through a beam transport system; the rf-power generator is located in the gantry stand; machine can produce megavoltage x-rays as well as electrons. (c) Accelerating wave-guide and rf-power generator are located in the gantry stand; electrons are brought to the movable target through a beam transport system; machine can produce megavoltage x-rays as well as electrons.

5.5.5. Injection system

- The injection system is the source of electrons, essentially a simple electrostatic accelerator called an *electron gun*.

- Two types of electron guns are in use as sources of electrons in medical linacs:

  (1) *Diode* type
  (2) *Triode* type.

  Both electron gun types contain a heated filament cathode and a perforated grounded anode; in addition, the triode electron gun also incorporates a grid.

- Electrons are thermionically emitted from the heated cathode, focused into a pencil beam by a curved focusing electrode, and accelerated toward the perforated anode through which they drift to enter the accelerating waveguide.

- The electrostatic fields which are used to accelerate the electrons in the diode gun are supplied directly from the pulsed modulator in the form of a negative pulse delivered to the cathode of the gun.

- In a triode gun, on the other hand, the cathode is held at a static negative potential (typically -20 kV). The grid of the triode gun is normally held sufficiently negative with respect to the cathode to cut off the current to the anode. The injection of electrons into the accelerating waveguide is then controlled by voltage pulses which are applied to the grid and must be synchronized with the pulses applied to the microwave generator. A photograph of a removable triode gun of a high energy linac is shown in Fig. 5.6 (left photo).

5.5.6. RF power generation system

- The microwave radiation, used in the accelerating waveguide to accelerate electrons to the desired kinetic energy, is produced by the *RF* power generation system which consists of two major components:

  (1) *RF power source*
  (2) *Pulsed modulator*.

- The *RF* power source is either a *magnetron* or a *klystron*. Both are devices using electron acceleration and deceleration in vacuum for production of the high power *RF* fields. Both types use a thermionic emission of electrons from a heated cathode and accelerate the electrons toward an anode in a pulsed electrostatic field; however, their design principles are completely different.

- The high voltage (~100 kV), high current (~100 A), short duration (~1 μs) pulses required by the *RF* power source (magnetron or klystron) and the injection system (electron gun) are produced by the *pulsed modulator*. The circuitry of the pulsed modulator is housed in the modulator cabinet that, depending on the particular linac installation design, is located in the treatment room, special mechanical room next to the treatment room, or in the linac control room.

- A magnetron is a source of high power RF required for electron acceleration, while a klystron is an *RF* power amplifier that amplifies the low power radiofrequency generated by an *RF* oscillator commonly called the *RF* driver.

5.5.7. Accelerating waveguide

- Waveguides are evacuated or gas-filled metallic structures of rectangular or circular cross-sections used in transmission of microwaves. Two types of waveguides are used in linacs: *rf power transmission waveguides* and *accelerating waveguides*. The power transmission waveguides transmit the RF power from the power source to the accelerating waveguide in which the electrons are accelerated.

- The electrons are accelerated in the accelerating waveguide by means of an energy transfer from the high power *RF* fields, which are set up in the accelerating waveguide and are produced by the *RF* power generators.

- The simplest kind of an accelerating waveguide is obtained from a cylindrical uniform waveguide by adding a series of disks (irises) with circular holes at the center, placed at equal distances along the tube. These disks divide the waveguide into a series of cylindrical cavities that form the basic structure of the accelerating waveguide in a linac.

- The cavities serve two purposes:

  (1) To couple and distribute microwave power between adjacent cavities and
  (1) To provide a suitable electric field pattern for acceleration of electrons.

- Two types of accelerating waveguides have been developed for acceleration of electrons:

  (1) *Traveling wave structure*
  (2) *Standing wave structure*.

- In the *traveling wave* structure, the microwaves enter the accelerating waveguide on the gun side and propagate toward the high energy end of the waveguide where they are either absorbed without any reflection or exit the waveguide to be absorbed in a resistive load or to be fed back to the input end of the accelerating waveguide. In this configuration only one in four cavities is at a given moment suitable for electron acceleration, providing electric field in the direction of propagation.

- In the *standing wave* structure, each end of the accelerating waveguide is terminated with a conducting disk to reflect the microwave power resulting in a buildup of standing waves in the waveguide. In this configuration, at all times, every second cavity carries no electric field and thus produces no energy gain for the electrons. Therefore, these cavities serve only as coupling cavities and can be moved out to the side of the waveguide structure, effectively shortening the accelerating waveguide by 50%. A cut-away view of a 6 MV standing wave accelerating wave-guide is shown in Fig. 5.7.

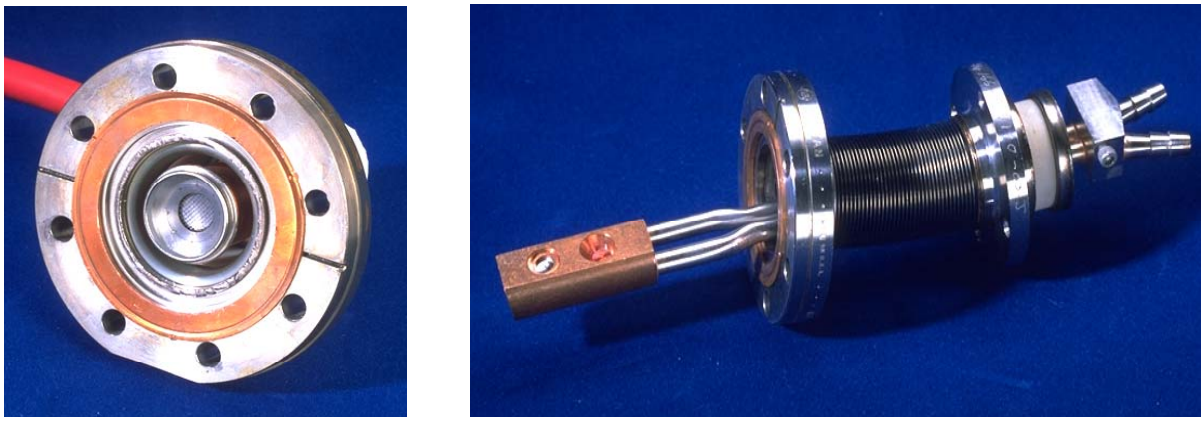


FIG. 5.6. Removable electron triode gun (left photo) and removable x-ray target (right photo for a typical high-energy linear accelerator (Varian Clinac-18), allowing two photon modes and several electron modes. The target is water-cooled and mounted with bellows to allow for movement into the pencil electron beam for x-ray production and movement out of the pencil beam to allow for electron beam production.

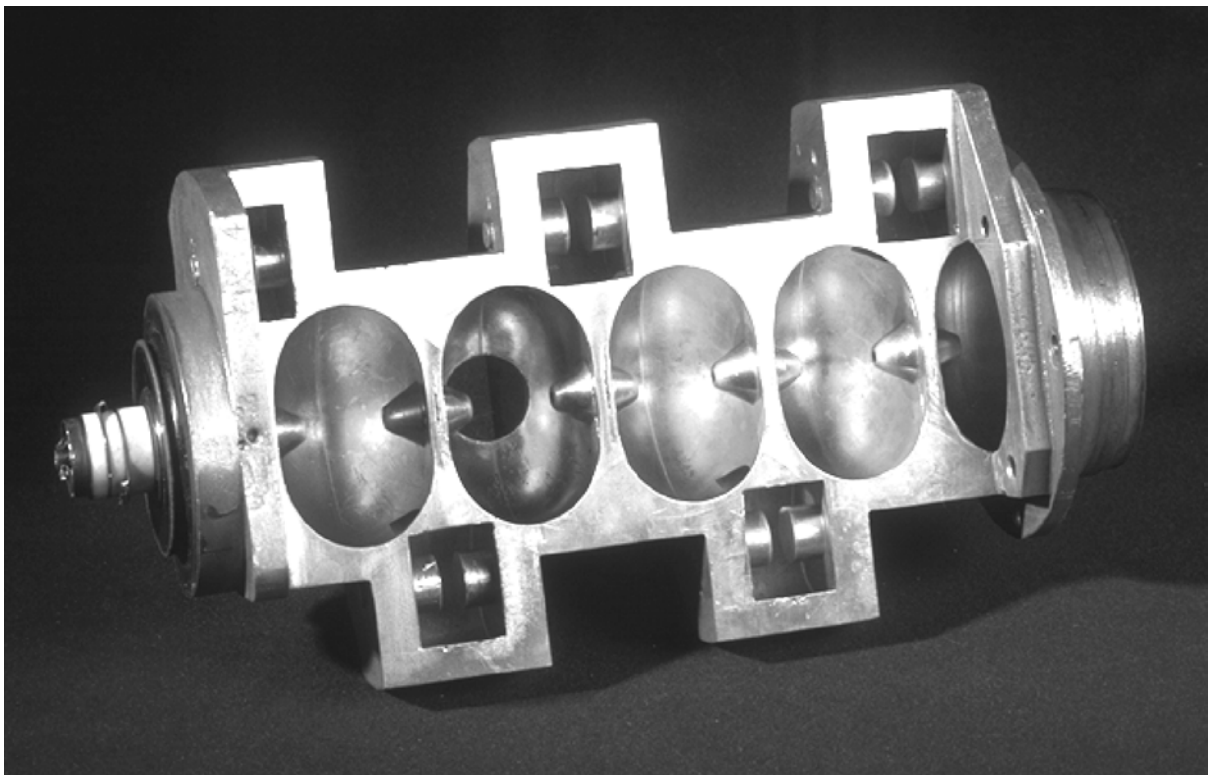


FIG. 5.7. Cut-away view of a standing wave accelerating waveguide for a 6 MV linear accelerator. The cavities as clearly visible: the accelerating cavities are on the central axis, the coupling cavities are off-side. The electron gun is on the left, the target on the right, both permanently embedded.

5.5.8. Microwave power transmission

- The microwave power produced by the *RF* generator is carried to the accelerating waveguide through rectangular uniform S-band waveguides which are either evacuated or, more commonly, pressurized with a dielectric gas (freon or sulfur hexafluoride, SF_6) to twice the atmospheric pressure.

- An important component, which must be inserted into the *RF* power transmission circuit between the *RF* generator and the accelerating waveguide, is a circulator (sometimes referred to as an isolator) which transmits the *RF* power from the *RF* generator to the accelerating waveguide but is impervious to reflected radiation moving in the opposite direction, thereby protecting the RF source from the reflected power.

5.5.9. Auxiliary system

- The linac auxiliary system consists of several services which are not directly involved with electron acceleration, yet they make the acceleration possible and the linac viable for clinical operation.

- The linac auxiliary system comprises four systems:

  (1) Vacuum pumping system producing a vacuum pressure of $\sim 10^{-6}$ tor in the accelerating guide and the *RF* generator;

  (2) Water cooling system used for cooling the accelerating guide, target, circulator, and *RF* generator;

  (3) Optional air pressure system for pneumatic movement of the target and other beam shaping components; and

  (4) Shielding against leakage radiation.

5.5.10. Electron beam transport

- In low energy linacs the target is embedded into the accelerating waveguide and no beam transport between the accelerating waveguide and target is required.

- Bending magnets are used in linacs operating at energies above 6 MeV where the accelerating waveguides are too long for straight-through mounting. Te accelerating waveguide is usually mounted parallel to the gantry rotation axis and the electron beam must be bent to make it strike the x-ray target or be able to exit through the beam exit window. Three systems for electron bending have been developed:

  (1) 90^o bending;
  (2) 270^o bending (achromatic); and
  (3) 112.5^o (slalom) bending.

- In medium (10 MV) and high-energy linacs (above 15 MV) an electron beam transport system is used for transporting the electron beam from the accelerating waveguide to the x-ray target or to the linac exit window for electron beam therapy. The system consists of evacuated drift tubes and bending magnets. In addition, steering coils and focussing coils, used for steering and focussing of the accelerated electron beam, also form components of the beam transport system.

5.5.11. Linac treatment head

- The linac head contains several components, which influence the production, shaping, localizing, and monitoring of the clinical photon and electron beams.

- Electrons, originating in the electron gun, are accelerated in the accelerating waveguide to the desired kinetic energy and then brought, in the form of a pencil beam, through the beam transport system into the linac treatment head, where the clinical photon and electron beams are produced.

- The important components found in a typical head of a fourth or fifth generation linac include:

  (1) Several *retractable x-ray targets*;
  (2) *Flattening filters* and *electron scattering foils* (also called scattering filters)
  (3) Primary and *adjustable secondary collimators*;
  (4) *Dual transmission ionisation chambers*;
  (5) *Field defining light* and *range finder*;
  (6) Optional *retractable wedges*;
  (7) Optional *multileaf collimator* (MLC).

- Clinical photon beams are produced with a target/flattening filter combination.

- Clinical electron beams are produced by retracting the target and flattening filter from the electron pencil beam and:

  (1) either scattering the pencil beam with a single or dual *scattering foil*
  (2) or deflecting and scanning the pencil beam magnetically to cover the field size required for electron treatment.

  Special cones (applicators) are used to collimate the electron beams.

- Each clinical photon beam has its own target/flattening filter combination. The flattening filters and scattering foils (if used for electron beams) are mounted on a rotating carousel or sliding drawer for ease of mechanical positioning into the beam, as required.

- The primary collimator defines a maximum circular field which is then further truncated with an adjustable rectangular collimator, consisting of two upper and two lower independent jaws, and producing rectangular and square fields with a maximum dimension of 40×40 cm^2 at the linac isocenter. The IEC recommends that the transmission of the primary x-ray beam through the rectangular collimator does not exceed 2% of the open beam value.

- The dual transmission ionisation chambers are used for monitoring the photon and electron radiation beam output as well as for monitoring the radial and transverse beam flatness (see Section 5.5.14).

- The field defining light and range finder provide convenient visual methods for correctly positioning the patient for treatment using reference marks. The field light illuminates an area that coincides with the radiation treatment field on the patient's skin, while the range finder is used to place the patient at the correct treatment distance, by projecting a centimeter scale whose image on the patient's skin indicates the vertical distance from the linac isocenter.

5.5.12. Production of clinical photon beams in a linac

- Clinical photon beams emanating from a medical linac are produced in an x-ray target and flattened with a flattening filter. A photograph of a high-energy linac movable target is shown in Fig. 5.6 (right photo).

- At electron energies below 15 MeV (photon beams below 15 MV) optimal targets have a high atomic number Z while at electron energies above 15 MeV (photon beam energies above 15 MV) the optimal targets have a low atomic number Z. Optimal flattening filters have a low Z irrespective of beam energy.

5.5.13. Beam collimation

In a typical modern medical linac, the photon beam collimation is achieved with two or three collimator devices:

(1) *Primary collimator*;
(2) *Secondary movable beam-defining collimators*;
(3) *Multileaf collimator* (MLC) (optional).

In addition to the primary and secondary collimators, clinical electron beams also rely on electron beam applicators (cones) for beam collimation.

- The *primary collimator* defines the largest available circular field size and is a conical opening machined into a tungsten shielding block with the sides of the conical opening projecting onto edges of the target on one end of the block and onto the flattening filter on the other end. The thickness of the shielding block is usually designed to attenuate the average primary x-ray beam intensity to less than 0.1% of the initial value (three *TVL*s). According to IEC recommendations, the maximum leakage should not exceed 0.2% of the open beam value.

- The *secondary beam defining collimators* consist of four blocks, two forming the upper and two forming the lower jaws of the collimator. They can provide rectangular or square fields at the linac isocenter with sides on the order of few millimeters up to 40 cm.

- Modern linacs incorporate independent *(asymmetric)* jaws that can provide asymmetric fields, most commonly one-half or three-quarters blocked fields where one or two beam edges, respectively, are coincident with the beam central axis.

- *Multileaf collimators* (MLCs) are a relatively recent addition to linac dose delivery technology. In principle, the idea behind an MLC is simple; however, building a reliable MLC system presents a substantial technological challenge.

- The number of leaves in commercial MLCs is steadily increasing and models with 120 leaves (60 pairs) covering fields up to 40×40 cm^2 and, requiring 120 individually computer-controlled motors and control circuits are currently available.

- MLCs are becoming invaluable in supplying intensity-modulated fields in conformal radiotherapy either in the step-and-shoot mode or in a continuous dynamic mode.

- Miniature versions of MLCs (microMLCs) projecting 1.5 to 6 mm leaf-widths and up to 10×10 cm^2 fields at the linac isocenter are currently available commercially. They may be used in radiosurgery as well as head and neck treatments.

5.5.14. Production of clinical electron beams in a linac

- The majority of higher energy linacs, in addition to providing single or dual photon energies, also provide electron beams with several nominal electron beam energies in the range from 6 to 30 MeV.

- To activate an electron beam mode both the target and flattening filter of the x-ray beam mode are removed from the electron beam.

- The electron beam currents producing clinical electron beams are two to three orders of magnitude lower than electron currents producing the clinical photon beams in the linac x-ray target.

- The electron pencil beam exits the evacuated beam transport system through a thin window usually made of beryllium, which with its low atomic number Z minimizes the pencil beam scattering and bremsstrahlung production.

- Two techniques are available for producing clinical electron beams from electron pencil beams:

  (1) *Pencil beam scattering*;
  (2) *Pencil beam scanning*.

- The scattering of the electron pencil beam over a relatively large area used in radiotherapy (up to 25×25 cm^2) is achieved by placing thin foils of high Z material (copper or lead) into the pencil beam at the level of the flattening filter in the x-ray mode.

- Electron pencil beam scanning is an alternative, albeit infrequently used, technique for producing clinical electron beams. The technique is usually implemented with two computer-controlled magnets, which deflect the pencil beam in two orthogonal planes, thereby scanning the pencil beam across the clinical treatment field.

5.5.15. Dose monitoring system

The IEC document 60601-2-1 specifies in detail the standards for radiation monitors installed in clinical electron linacs. It deals with standards for the type of radiation detectors, display of MUs, for termination of radiation and for monitoring of beam flatness and dose rate.

- Most common dose monitors in linacs are transmission ionisation chambers permanently imbedded in the linac clinical photon and electron beams to monitor the beam output continuously during patient treatment.

- Most linacs use sealed ionisation chambers to make their response independent of ambient temperature and pressure.

- The customary position of the dose monitor chambers is between the flattening filter or scattering foil and the photon beam secondary collimator.

- For patient safety, the linac dosimetry system usually consists of two separately-sealed ionisation chambers with completely independent biasing power supplies and readout electrometers. If the primary chamber fails during patient treatment, the secondary chamber will terminate the irradiation, usually after an additional dose of only a few percent above the prescribed dose has been delivered.

- In the event of a simultaneous failure of both the primary and secondary ionisation chamber, the linac timer will shut the machine down with a minimal overdose to the patient.

- The main requirements for the ionisation chamber monitors are as follows:

  (1) Chambers must have a minimal effect on clinical photon and electron radiation beams;

  (2) Chamber response should be independent of ambient temperature and pressure (most linacs use sealed ionisation chambers to satisfy this condition); and

  (3) Chambers should be operated under saturation conditions.

- The primary ionisation chamber measures monitor units (MU). Typically, the sensitivity of the chamber electrometer circuitry is adjusted in such a way that 1 MU corresponds to a dose of 1 cGy delivered in a water phantom at the depth of dose maximum on the central beam axis when irradiated with a 10×10 cm^2 field at an *SSD* of 100 cm.

- Once the operator-preset number of MUs has been reached, the primary ionisation chamber circuitry shuts the linac down and terminates the dose delivery to the patient. Before a new irradiation can be initiated, it is necessary to reset the MU displays to zero. Furthermore, irradiation is not possible until a new selection of MUs has been made.

- In addition to monitoring the primary dose in MUs, the dose monitoring system also monitors other operating parameters such the beam energy, beam flatness, and symmetry. Measurement of all these additional parameters requires that the ionisation chamber electrodes of the primary and secondary chambers be divided into several sectors, with the resulting signals used in automatic feedback circuits to steer the electron beam through the accelerating waveguide, beam transport system, and onto the target or scattering foil thereby ensuring beam flatness and symmetry. The particular design of the ionisation chamber electrodes and sectors varies from one manufacturer to another.

- Linacs must be equipped with a monitoring system that continuously displays the machine isocenter dose rate and terminates the beam when the measured dose rate exceeds twice the maximum specified by the technical machine description.

- When the linac is capable of producing more than one beam energy or more than one beam mode (x-rays or electrons), after termination of radiation, further irradiation is prevented until the selection of energy and beam mode has been made afresh and entered into the control console.

- Similarly, for linacs capable of stationary as well as moving beam radiotherapy, after termination of radiation, further irradiation is prevented until stationary radiotherapy or moving beam radiotherapy has been selected afresh and entered into the control console.

5.6. RADIOTHERAPY WITH PROTONS, NEUTRONS AND HEAVY IONS

External beam radiotherapy is carried out mainly with machines that produce either x-rays or electrons. In a few specialized centers around the world, external beam radiotherapy is also carried out with heavier particles such as:

- Neutrons produced by neutron generators and cyclotrons,
- Protons produced by cyclotrons and synchrotrons, and
- Heavy ions (helium, carbon, nitrogen, argon, neon) produced by synchro-cyclotrons and synchrotrons.

These particles offer some distinct advantages over the standard x-ray and electron modalities, such as:

- Considerably lower oxygen enhancement ratio (*OER*) for neutrons (see Section 14.10)
- Improved dose-volume histograms (DVHs) for protons and heavy ions (see Section 7.6).

However, equipment for production of protons, neutrons and heavy ions is considerably more expensive than standard radiotherapy equipment, both in capital costs as well as in maintenance and servicing costs, thus precluding a wide-spread use in standard radiotherapy departments. The decreasing costs of proton cyclotrons are likely to result in a wider use of proton beam therapy in the future.

5.7. SHIELDING CONSIDERATIONS

External beam radiotherapy is carried out mainly with three types of equipment that produces either x-rays or electrons:

(1) *X-ray machines* (*superficial and orthovoltage)*;
(2) *Teletherapy (cobalt-60) machines*;
(3) Linear accelerators (linacs).

All radiotherapy equipment must be housed in specially shielded treatment rooms to protect the personnel and general public in areas adjacent to the treatment rooms. The treatment rooms must comply not only with structural building codes but also with national and international regulations that deal with shielding requirements to render an installation safe from the radiation protection point-of-view. During the planning stage for a radiotherapy machine installation, a qualified medical physicist determines the required thickness of primary and secondary barriers and provides the information to the architect and structural engineer for incorporation into the architectural drawing for the treatment room.

- *Superficial and orthovoltage x-ray therapy rooms* are shielded either with ordinary concrete (2.35 g/cm^3) or lead. In this energy range, the photoelectric effect is the predominant mode of photon interaction with matter, making the use of lead very efficient for shielding purposes.

- *Megavoltage treatment rooms* (often referred to as bunkers or vaults because of the large barrier thickness required for shielding) are most commonly shielded with ordinary concrete in the interest of minimizing the construction costs. The Compton effect is the predominant mode of photon interaction with shielding material in this energy range. To conserve space other higher density materials may be used with the required wall thickness inversely proportional to the density of the shielding material. Thus, the use of high density concrete (5 g/cm^3) will cut the required thickness of ordinary concrete barrier to approximately one half, however, it will also increase the construction material cost by a factor of 30.

- Shielding issues related to linear accelerator bunkers are discussed in more detail in Section 16.17.

5.8. COBALT-60 TELETHERAPY UNIT VERSUS LINEAR ACCELERATOR

Since the inception of radiotherapy soon after the discovery of x-rays by Roentgen in 1895, the technology of radiation production has first been aimed toward ever higher photon energies and intensities and more recently toward computerization and intensity modulated beam delivery. During the first 50 years of radiotherapy, the technological progress has been relatively slow and mainly based on x-ray tubes, Van de Graaff generators and betatrons.

The first truly practical megavoltage therapy machine was the cobalt-60 teletherapy machine developed in Canada in the 1950s. The invention of cobalt-60 teletherapy provided a tremendous boost in the quest for higher photon energies and placed the cobalt-60 unit into the forefront of radiotherapy for a number of years, mainly because it incorporated a radioactive source that is characterized with features extremely useful for radiotherapy.

The important features of cobalt-60 teletherapy machines can be summarized as follows:

(1) Relatively *high energy gamma ray emission*;
(2) Relatively *long half-life*;
(3) Relatively *high specific activity*;
(4) Relatively *simple means of production*.

Figure 5.8 shows a cobalt-60 teletherapy machine (left side) and a stamp issued by Canada Post commemorating Canada's role in the development of the cobalt-60 machine (right side).

Linear accelerators (linacs) were developed concurrently by two groups: W.W. Hansen's group at Stanford University in the U.S.A. and D.D. Fry's group at Telecommunications Research Establishment in the U.K. Both groups were interested in linacs for research purposes and profited heavily from the microwave radar technology developed during World War II and using 3000 MHz as the design frequency.

The potential for the use of linacs in radiation therapy has become apparent in the 1950s and the first clinical linac was installed in 1950s at the Hammersmith Hospital in London, U.K. During subsequent years, the linac eclipsed the cobalt unit and became the most widely used radiation source in modern radiotherapy with several thousand units in clinical practice around the world today. In contrast to a cobalt-60 unit that provides essentially only one gamma energy of 1.25 MeV, a linac can provide either megavoltage electron or x-ray therapy with a wide range of energies. Figure 5.9 shows a modern dual energy linear accelerator.

In comparison to cobalt-60 machines, linacs have become very complex in design:

(1) In part because of the multimodality capabilities that have evolved and are available on most modern machines,

(2) In part because of an increased use of computer logic and microprocessors in the control systems of these machines, and

(3) In part because of added features, such as high dose rate modes, multileaf collimation, electron arc therapy, and the dynamic motion while the beam is ON of the collimators (dynamic wedge), multileaf collimator leaves (intensity-modulated radiotherapy), gantry and couch.

Despite the clear technological and practical advantages of linacs over cobalt-60 machines, the latter still occupy an important place in radiotherapy armamentarium, mainly because of considerably lower capital, installation and maintenance costs of cobalt-60 machines compared to linacs. In the developing world, the cobalt-60 machines, because of their relatively lower costs, simplicity of design, and ease of operation, are likely to play an important role in cancer therapy for the foreseeable future.

Many modern features of linacs, such as multileaf collimators, dynamic wedges and dynamic operation, could also be installed on modern cobalt-60 machines to allow, at a lower cost, a similar sophistication in treatment as linacs do. It is unfortunate that manufacturers of cobalt-60 units are very slow in reacting to new technological developments in radiotherapy, conceding pre-eminence to linac manufacturers even in jurisdictions that would find it much easier and more practical to run cobalt-60 machines than linacs.

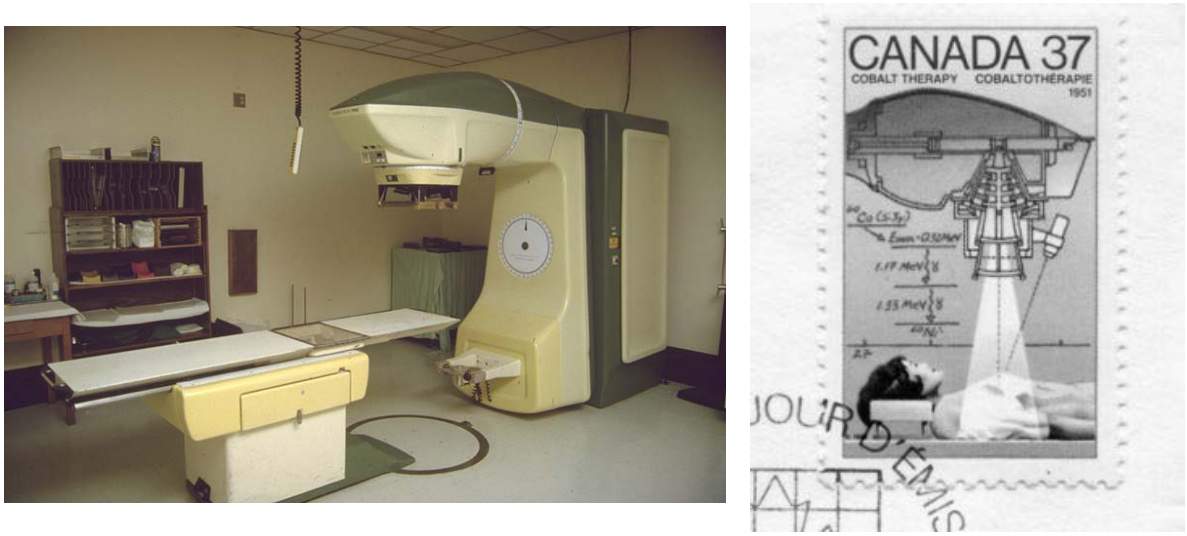


FIG. 5.8. Cobalt-60 teletherapy unit manufactured by Theratronics (now MDS-Nordion), Ottawa, Canada: left – actual photo, right – schematic diagram depicted on a postage stamp issued by Canada Post in 1988 in honour of Dr. Harold E. Johns who invented the cobalt-60 unit in the 1950s.

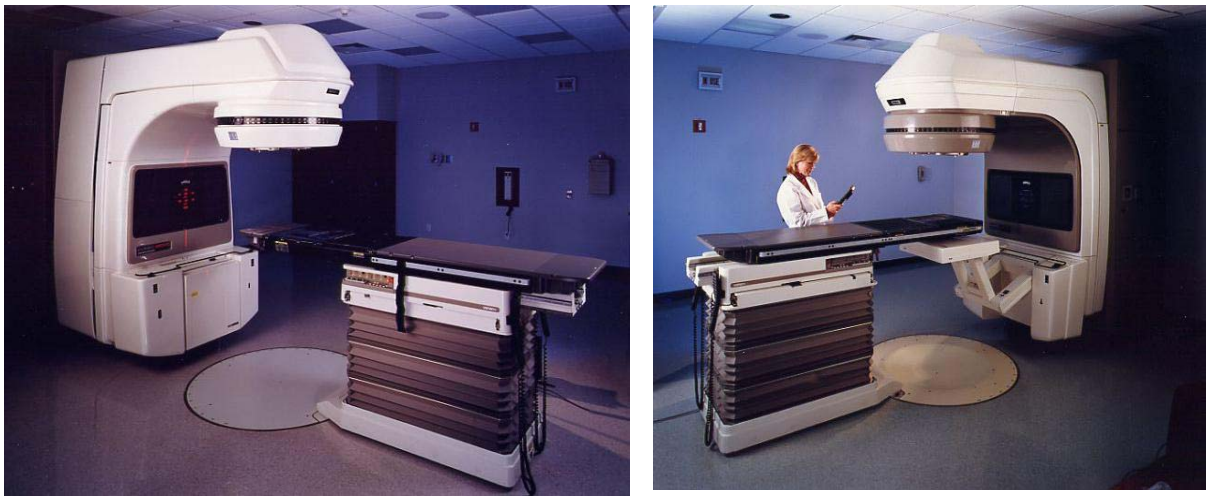


FIG. 5.9. Modern dual photon energy linear accelerator manufactured by Varian, Palo Alto, California; the gantry and the patient support assembly are clearly shown. Left photo: the portal imager is retracted, left photo: the portal imager is activated. Other important manufacturers of linacs are Siemens (Erlangen, Germany) and Elekta (Stockholm, Sweden).

5.9. SIMULATORS AND CT-SIMULATORS

Simulators and CT-simulators are an important component of equipment used in radiation therapy. They cover several crucial steps in the radiotherapeutic process that are not related to the actual dose delivery but are nonetheless very important as they deal with the determination of target location, treatment planning and spatial accuracy in dose delivery. The determination of the target volume that is related to the extent of the disease (See Section 7.2) and its position relative to adjacent critical normal tissues can be achieved with various methods. These range from a simple clinical examination through planar x-ray imaging to the use of complex modern imaging equipment such as CT scanners in conjunction with MR, and PET scanners. Both simulators and CT-simulators incorporate three major systems: (i) mechanical, (ii) x-ray tube, and (iii) imaging equipment.

The major steps in the target localization and field design are:

(1) Acquisition of patient data set.
(2) Localization of target and adjacent structures.
(3) Definition and marking of patient coordinate system.
(4) Design of treatment fields.
(5) Transfer of data to treatment planning system.
(6) Production of image for treatment verification.

The six steps above can be achieved either with a conventional simulator or with a CT-simulator; however, the CT-simulator provides for the more elegant, reliable and practical means to achieve the six steps, in addition to providing reliable external and internal contours and electron density information.

5.9.1. Radiation therapy simulator

The radiation therapy simulator consists of a diagnostic x-ray tube mounted on a rotating gantry, simulating geometries identical to those found on megavoltage therapy machines that are either isocentric teletherapy cobalt-60 units or isocentric linacs. Thus, the simulator enjoys the same degrees of freedom as a megavoltage machine but, rather than providing a megavoltage beam for treatment, it provides a diagnostic quality x-ray beam suitable for imaging, either in the *radiographic mode* (image recorded on radiographic film) or in the *fluoroscopic mode* (image recorded on a TV monitor using an image intensifier).

A modern simulator should mimic all mechanical features and geometric field arrangements of various megavoltage machines, ranging from cobalt-60 machines with SAD = 80 cm to high-energy linacs with SAD = 100 cm.

In megavoltage machines, radiation fields are defined with collimators (upper and lower jaws), while in simulators, the rectangular and square fields are defined with *delineator wires* to enable visualization of the target as well as healthy tissues adjacent to the target.

A modern simulator covers the following processes:

- *Tumour and adjacent normal tissue localization.*
- *Treatment simulation.*
- *Treatment plan verification.*
- *Monitoring of treatment.*

5.9.2. CT-simulator

CT-simulators are CT scanners equipped with special features that make them useful for certain stages in the radiotherapeutic process. The special features typically are:

- *Flat couch top surface* to provide a patient position during simulation that will be identical to position during treatment on a megavoltage machine.

- *Laser patient marking system* to transfer the coordinates of the tumour isocenter, derived from the contouring of the CT data set, to the surface of the patient. Two types of laser marking systems are used: (i) a gantry mounted laser and (ii) a system consisting of a wall mounted moveable sagittal laser and two stationary lateral lasers.

- *Virtual simulator* consisting of software packages that allow the user to: (i) define and calculate a treatment isocenter and then (ii) simulate a treatment using digitally reconstructed radiographs (DRRs).

The CT-simulator essentially obviates the need for conventional simulation by carrying out two distinct functions:

- Physical simulation that covers the first three target localization steps above.
- Virtual simulation that covers the last three steps above.

In CT-simulation the patient data set and target localization are carried out using CT images with fluoroscopy and radiography replaced by DRRs. A laser alignment system is used for marking; and a virtual simulator software package is used for field design and production of verification images. Transfer of all necessary information to the treatment planning system is achieved electronically.

The planar simulation x-ray film provides a beam's eye view (BEV) of the treatment portal but does not provide three- dimensional information about anatomical structures. CT, on the other hand, provides anatomical information and target definition but does not allow a direct correlation with the treatment portals.

A DRR is the digital equivalent of a planar simulation x-ray film (see also Section 7.4.8). It is reconstructed from a CT data set using virtual simulation software available on a CT-simulator or a treatment planning system, and represents a computed radiograph of a virtual patient generated from a CT data set representing the actual patient. Just like conventional radiograph, the DRR accounts for the divergence of the beam.

The basic approach to producing a DRR involves several steps: choice of virtual source position; definition of image plane; ray tracing from virtual source to image plane; determination of CT value for each volume element traversed by the ray-line to generate an effective transmission value at each pixel on the image plane; summation of CT values along the ray-line (line integration); and gray scale mapping.

An extension of the DRR approach is the digitally composited radiograph (DCR) that provides an enhanced visualization of bony landmarks and soft tissue structures. This is achieved by differentially weighting ranges of CT-numbers that correspond to different tissues to be enhanced or suppressed in the resulting DCR images.

5.10. TRAINING REQUIREMENTS

The increased complexity of radiotherapy equipment demands that equipment be used only by highly trained and competent staff in order to minimize the potential for accidents. A recent report by the International Atomic Energy Agency (IAEA) summarized the lessons learned from accidental exposures in radiation therapy and a report by the American Association of Physicists in Medicine specifically addressed medical accelerator safety considerations.

Of vital importance in purchasing, installation and clinical operation of radiotherapy equipment are the following:

(1) *Preparation of equipment specification document*
(2) *Design of treatment room and radiation safety*
(3) *Acceptance testing of equipment*
(4) *Commissioning of equipment*
(5) *Quality assurance program*

Items (1), (3) and (4) are addressed in detail in Chapter 10, item (5) in Chapter 12, and item (2) in Chapter 16.

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