Chapter 5

TREATMENT MACHINES FOR EXTERNAL BEAM RADIOTHERAPY

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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 towards ever higher photon and electron beam energies and intensities, and more recently towards computerization and intensity modulated beam delivery. During the first 50 years of radiotherapy the technological progress was relatively slow and mainly based on X ray tubes, van de Graaff generators and betatrons.

The invention of the ^{60}Co 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 at the forefront of radiotherapy for a number of years. The concurrently developed medical linacs, however, soon eclipsed cobalt units, 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 accelerator, 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 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.

Most of the electron's kinetic energy is transformed in the target 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 (collision loss).

In a given Coulomb interaction between the incident electron and an orbital electron, the orbital electron is ejected from its shell and an electron from a higher level shell fills the resulting orbital vacancy. The energy difference between the two shells may either be emitted from the atom in the form of a characteristic photon (characteristic X ray) or transferred to an orbital electron that 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 ($0 \leq \omega \leq 1$) and ranges from zero for low Z atoms through 0.5 for copper ($Z = 29$) to 0.96 for high Z atoms for K shell vacancies, which 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 zero 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

According to the range R of electrons of a given kinetic energy E_K 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 of the order of R. For thin target radiation, the energy radiated is proportional to the product $E_K Z$, where Z is the atomic number of the target. The intensity versus photon energy (photon spectrum) is constant from zero to the kinetic energy E_K of the incident electron, and zero for all energies above E_K.

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:

$$I(h\nu) = CZ(E_K - h\nu) \tag{5.1}$$

where

C is a proportionality constant;
$h\nu$ 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, those with electron kinetic energies between 100 keV and 500 keV are called orthovoltage X rays, while those with electron kinetic energies above 1 MeV are called 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.

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 that are superimposed on to 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 the 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.

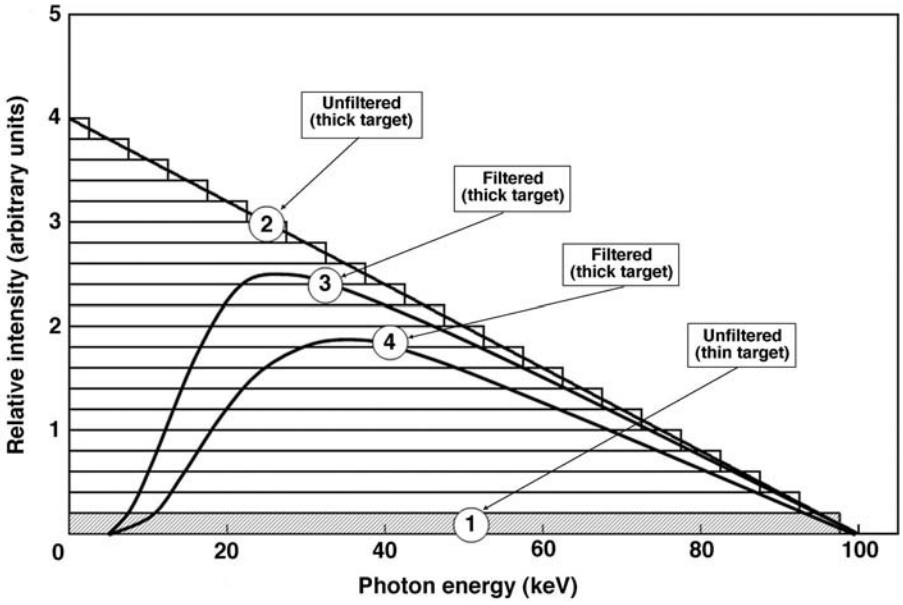


FIG. 5.1. Typical thin target (curve 1) and thick target (curves 2, 3 and 4) spectra for an X ray tube in which 100 keV electrons strike the target. Curve (1) is for a thin target producing a constant intensity for photon energies from zero to the kinetic energy of electrons striking the target (100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

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

5.2.5. X ray beam quality specifiers

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

- A complete X ray spectrum is very difficult to measure; however, it gives the most rigorous description of beam quality.
- The HVL is practical for beam quality description in the superficial (HVL in aluminium) 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.
- The NAP is sometimes used for describing the megavoltage beam quality. The NAP is determined by measuring the ionization ratio in a water phantom at depths of 10 and 20 cm for a 10×10 cm^2 field at the nominal source to axis distance (SAD) of 100 cm.
- Recent dosimetry protocols recommend the use of tissue–phantom ratios or percentage depth doses (PDDs) 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 radiotherapy are produced with X ray machines. The main components of a radiotherapeutic X ray machine are: an X ray tube; a ceiling or floor mount for the X ray tube; a target cooling system; a 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 a

vacuum towards 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 of 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), which 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 a 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.

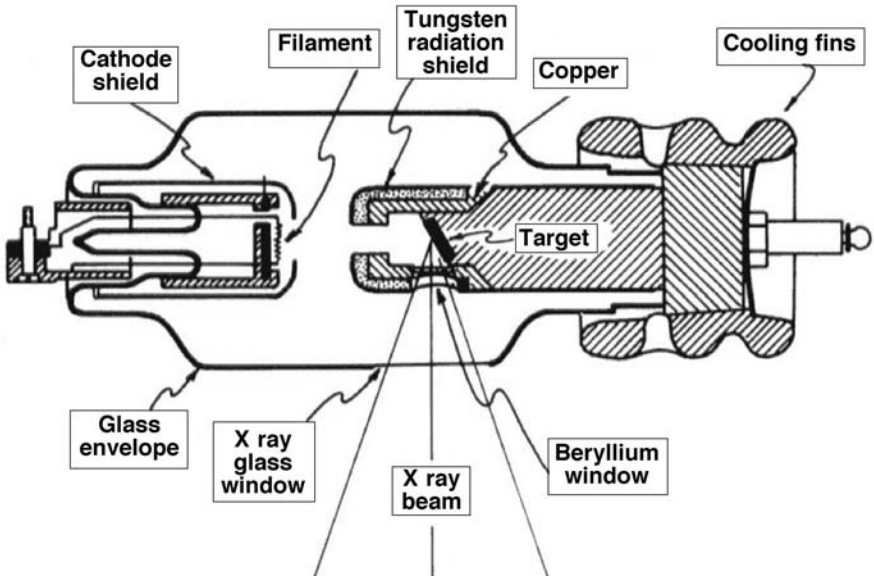


FIG. 5.2. Typical therapy X ray tube (reprinted from Johns, H.E., and Cunningham, J.R., with permission).

- Research is currently being carried out on cold field emission cathodes produced with carbon nanotubes (CNTs). 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.

5.3. GAMMA RAY BEAMS AND GAMMA RAY UNITS

5.3.1. Basic properties of gamma rays

For use in external beam radiotherapy, γ 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 that attain ground state through emission of γ rays (γ decay).
- The important characteristics of radioisotopes in external beam radiotherapy are:
  - High γ ray energy;
  - High specific activity;
  - Relatively long half-life;
  - 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 the half-life $t_{1/2}$:

$$a = \frac{\mathcal{A}}{m} = \frac{N_A \ln 2}{t_{1/2} A} \tag{5.2}$$

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 $(\dot{K}_{air})_{air}$ is given by the following relation:

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

where

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

- The basic physical properties of the two γ emitters (^{60}Co and ^{137}Cs) currently used for external beam teletherapy and a potential source for teletherapy units (^{152}Eu) are listed in Table 5.1. Of the three radioisotopes, ^{60}Co 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.

5.3.2. Teletherapy machines

Treatment machines incorporating γ ray sources for use in external beam radiotherapy are called teletherapy machines. They are most often mounted isocentrically, allowing the beam to rotate about the patient at a fixed SAD. Modern teletherapy machines have SADs of 80 or 100 cm.

The main components of a teletherapy machine are: a radioactive source; a source housing, including beam collimator and source movement mechanism; a gantry and stand in isocentric machines or a housing support assembly in stand-alone machines; a patient support assembly; and a machine console.

5.3.3. Teletherapy sources

The most widely used teletherapy source uses ^{60}Co radionuclides 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.
- The 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 of the order of 5000–10 000 Ci (185–370 TBq) and provide a typical dose rate at 80 cm from the teletherapy source of the order of 100–200 cGy/min. Often the output of a teletherapy machine

TABLE 5.1. PHYSICAL PROPERTIES OF RADIONUCLIDES USED IN EXTERNAL BEAM RADIOTHERAPY

	Co-60	Cs-137	Eu-152
Half-life (a)	5.3	30	13.4
Specific activity (Ci/g)	1100[a] (~250[b])	80	180[a] (~150[b])
Photon energy (MeV)	1.17 and 1.33	0.662	0.6–1.4
Specific γ rate constant Γ [$Rm^2/(Ci{\cdot}h)$]	1.31	0.33	1.06
Specific air kerma rate constant Γ_{AKR} [$\mu Gy{\cdot}m^2/(GBq{\cdot}h)$]	309	78	250
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: $a = (N_A \ln 2)/(t_{1/2}A)$.

[b] The 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 (e.g. ^{59}Co mixed with ^{60}Co)).

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.
- The ^{60}Co radionuclides in a teletherapy source decay with a half-life of 5.26 years into ^{60}Ni with the emission of electrons (β particles) with a maximum energy of 320 keV and two γ rays with energies of 1.17 MeV and 1.33 MeV. The emitted γ rays constitute the therapy beam; the electrons are absorbed in the cobalt source or the source capsule, where they produce relatively low energy and essentially negligible bremsstrahlung X rays and characteristic X rays.

5.3.4. Teletherapy source housing

The housing for the teletherapy source is called the source head, and 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 γ 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) a source on a sliding drawer and (ii) a source on a rotating cylinder. Both methods incorporate a safety feature in which the beam is terminated automatically in the event of a 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 from the unit even when the source is in the beam off position. The head leakage typically amounts to less than 1 mR/h (0.01 mSv/h) at 1 m from the source. International regulations require that the average leakage of a teletherapy machine head be less than 2 mR/h (0.02 mSv/h) 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's failure.

The set treatment time must incorporate the shutter error, which accounts for the travel time of the source from the beam off position towards 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 a finite source diameter, may be minimized by using small diameter sources and by using penumbra trimmers as close as possible to the patient's skin (see Section 6.9 for further discussion of the penumbra).

5.4. PARTICLE ACCELERATORS

Numerous types of accelerator 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:

- The particle to be accelerated must be charged;
- An electric field must be provided in the direction of particle acceleration.

The various types of accelerator 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 accelerator: 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 best known example of a cyclic accelerator is the 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 after.

- 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 (doughnut shaped) vacuum chamber 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 analogue 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 (doughnut).
- The electrons are accelerated by the electric field induced in the doughnut shape by the changing magnetic flux in the magnet; they are kept in a circular orbit by the magnetic field present.
- 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 versus 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 megaelectronvolts. Initially, the cyclotron was used for basic nuclear physics research, but later on found important medical uses in the production of radioisotopes for nuclear medicine as well as in the production of proton and neutron beams for radiotherapy. The recent introduction of positron emission tomography (PET)/computed tomography (CT) machines for use in radiotherapy (see Section 15.10) has dramatically increased the importance of cyclotrons in medicine. PET/CT machines rely on glucose labelled with positron emitting ^{18}F 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 shaped form) by a uniform magnetic field (1 T)

that is produced between the pole pieces of a large magnet. Figure 5.3(b) is a diagram of a cyclotron.

- A radiofrequency (RF) 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.

5.4.3. Microtron

The microtron is an electron accelerator that combines the features of a linac with a cyclotron. The concept of the microtron was developed by

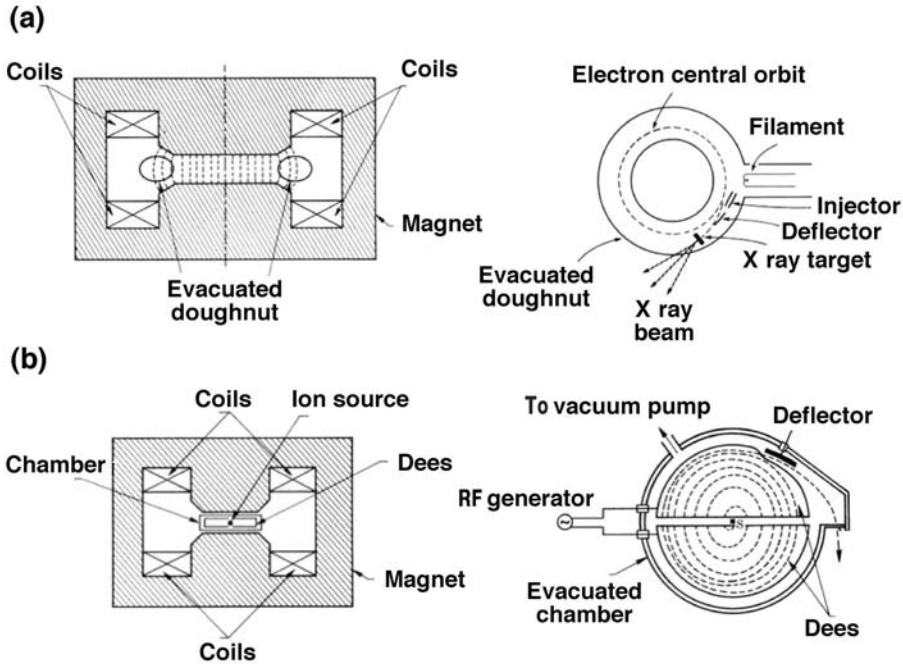


FIG. 5.3. Two cyclic accelerators: (a) a betatron and (b) a cyclotron.

V.I. Veksler in 1944, and the machine is used in modern radiotherapy, albeit to a much smaller extent than linacs.

Two types of microtron have been 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 multicavity accelerating structures similar to those used in linacs. The electron orbits consist of two semicircular and two straight sections.

5.5. LINACS

Medical linacs are cyclic accelerators that accelerate electrons to kinetic energies from 4 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 linac 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. betatrons).

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 linac are available for clinical use. Some provide X rays only in the low megavoltage range (4 or 6 MV), while others provide both X rays and electrons at various megavoltage energies. A typical modern high energy linac will provide two photon energies (6 and 18 MV) and several electron energies (e.g. 6, 9, 12, 16 and 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. 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 ionization chamber; isocentric mounting.
- Medium energy photons (10–15 MV) and electrons: bent beam; movable target and flattening filter; scattering foils; dual transmission ionization 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 (EPID); multileaf collimator (MLC).
- High energy photons and electrons: photon beam intensity modulation with MLC; full dynamic conformal dose delivery with intensity modulated beams produced with an MLC.

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. The International Electrotechnical Commission (IEC) publishes international standards that express, as nearly as possible, an international consensus of opinion on relevant technical subjects; electron linacs are addressed in detail by the IEC. 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 installation in customer's treatment rooms.

5.5.3. Components of modern linacs

Linacs are usually mounted isocentrically and the operational systems are distributed over five major and distinct sections of the machine, the:

- Gantry;
- Gantry stand or support;
- Modulator cabinet;
- Patient support assembly (i.e. treatment table);
- 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 a linac's 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:
  (i) Injection system;
  (ii) RF power generation system;
  (iii) Accelerating waveguide;
  (iv) Auxiliary system;
  (v) Beam transport system;
  (vi) Beam collimation and beam monitoring system.

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 isocentre, obviating the need for a beam transport system. A straight-through photon beam is produced and the RF power source is 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 (8–15 MeV) and high (15–30 MeV) electron energies are too long for direct isocentric mounting and thus 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 linacs are shown in Fig. 5.5.

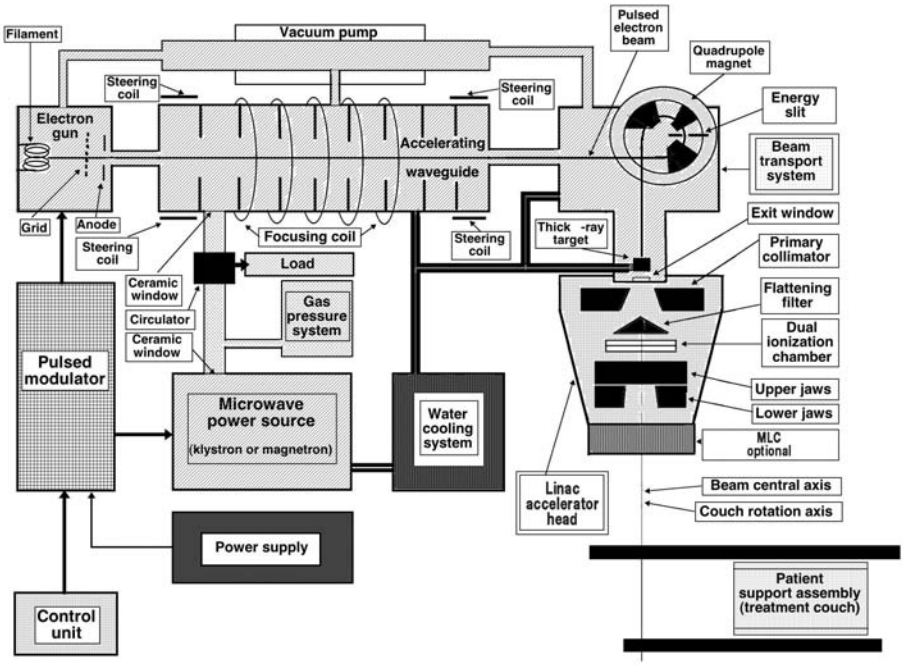
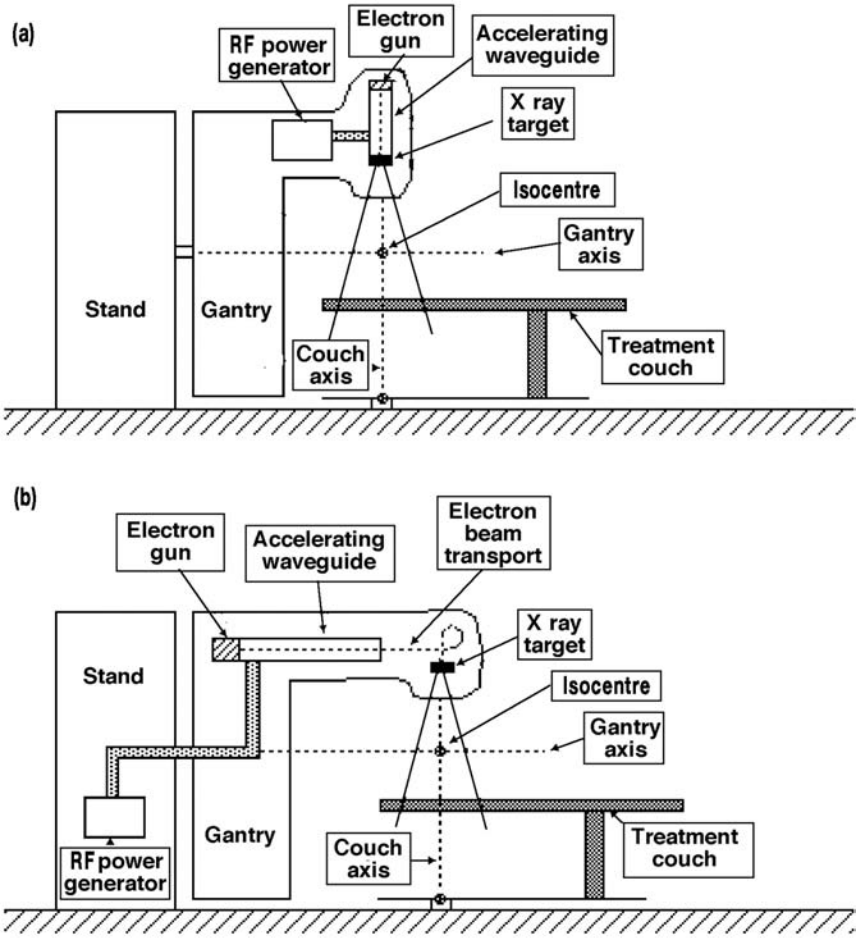


FIG. 5.4. Medical linac.

5.5.5. Injection system

The injection system is the source of electrons; it is essentially a simple electrostatic accelerator called an electron gun.

- Two types of electron gun are in use as sources of electrons in medical linacs:
  - — Diode type;
  - — 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 towards the perforated anode through which they drift to enter the accelerating waveguide.
- The electrostatic fields 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, however, 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 removable triode gun of a high energy linac is shown in Fig. 5.6(a).

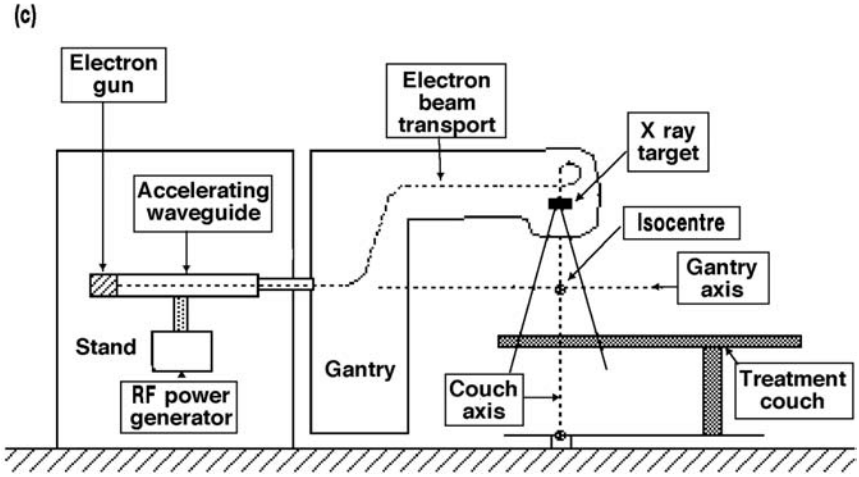
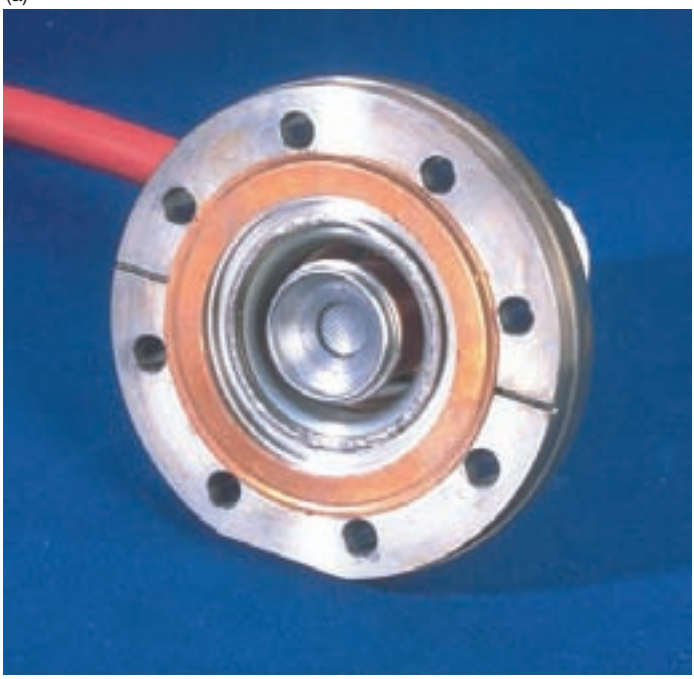


FIG. 5.5. Design configurations for isocentric medical linacs. (a) Straight-through beam design; the electron gun and target are permanently embedded into the accelerating waveguide; the machine produces only X rays with energies of 4–6 MV; the RF power generator is mounted in the gantry. (b) The accelerating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

(a)



(b)

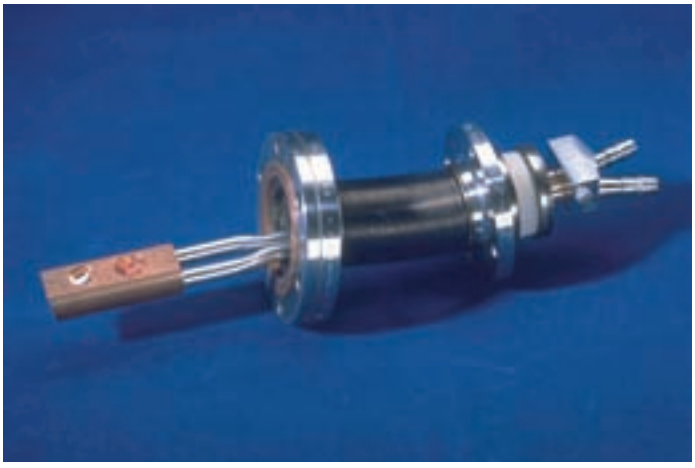


FIG. 5.6. Removable electron triode gun (a) and removable X ray target (b) for a typical high energy linac (Varian Clinac-18), allowing two photon modes and several electron modes. The target is water cooled and mounted with bellows to allow for movement into the pencil electron beam for X ray production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

5.5.6. Radiofrequency 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:

- An RF power source;
- A pulsed modulator.

The RF power source is either a magnetron or a klystron. Both are devices that use electron acceleration and deceleration in a vacuum for the production of high power RF fields. Both types use a thermionic emission of electrons from a heated cathode and accelerate the electrons towards 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 s) pulses required by the RF power source (magnetron or klystron) and the injection system (electron gun) are produced by a pulsed modulator. The circuitry of the pulsed modulator is housed in the modulator cabinet, which, depending on the particular linac installation design, is located in the treatment room, in a 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 RF 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-section used in the transmission of microwaves. Two types of waveguide 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 accelerating waveguide is obtained from a cylindrical uniform waveguide by adding a series of discs (irises) with

circular holes at the centre, placed at equal distances along the tube. These discs divide the waveguide into a series of cylindrical cavities that form the basic structure of the accelerating waveguide in a linac.

The accelerating waveguide is evacuated to allow free propagation of electrons. The cavities of the accelerating waveguide serve two purposes:

— To couple and distribute microwave power between adjacent cavities;
— To provide a suitable electric field pattern for the acceleration of electrons.

Two types of accelerating waveguide have been developed for the acceleration of electrons:

(i) Travelling wave structure;
(ii) Standing wave structure.

In the travelling wave structure the microwaves enter the accelerating waveguide on the gun side and propagate towards the high energy end of the waveguide, where they either are 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 any given moment suitable for electron acceleration, providing an electric field in the direction of propagation.

In the standing wave structure each end of the accelerating waveguide is terminated with a conducting disc 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. These cavities therefore 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 waveguide is shown in Fig. 5.7.

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 that are either evacuated or, more commonly, pressurized with a dielectric gas (Freon or sulphur hexafluoride, SF_6) to twice the atmospheric pressure.

An important component that must be inserted into the RF power transmission circuit between the RF generator and the accelerating waveguide is a

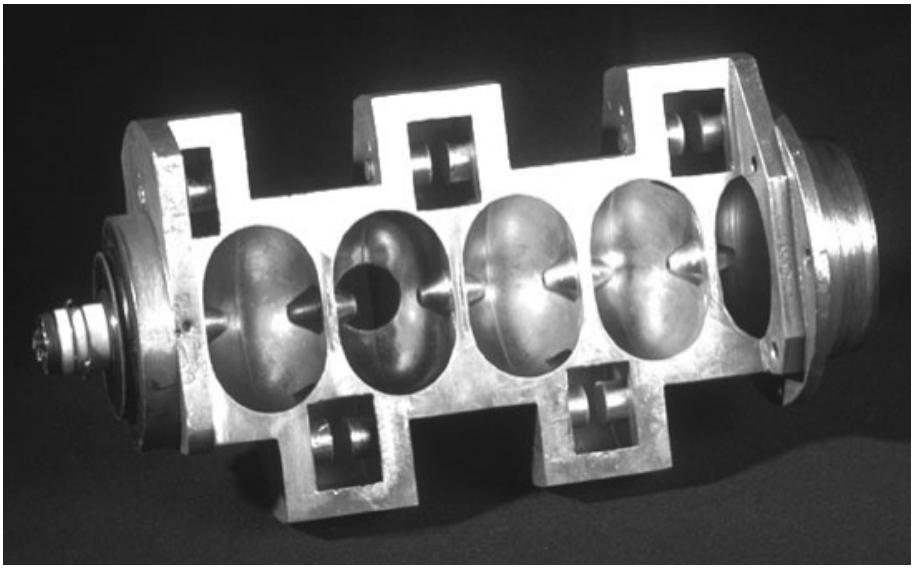


FIG. 5.7. Cutaway view of a standing wave accelerating waveguide for a 6 MV linac. The cavities are clearly visible: the accelerating cavities are on the central axis; the coupling cavities are off-side. The electron gun is on the left, the target on the right, both permanently embedded.

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 that are not directly involved with electron acceleration, yet make the acceleration possible and the linac viable for clinical operation.

The linac auxiliary system comprises four systems:

- A vacuum pumping system producing a vacuum pressure of ~10^{-6} torr in the accelerating guide and the RF generator;
- A water cooling system used for cooling the accelerating guide, target, circulator and RF generator;
- An optional air pressure system for pneumatic movement of the target and other beam shaping components;
- Shielding against leakage radiation.

5.5.10. Electron beam transport

In low energy linacs the target is embedded in 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. The 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:

- 90º bending;
- 270º bending (achromatic);
- 112.5º (slalom) bending.

In medium (10 MV) and high energy (above 15 MV) linacs 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 focusing coils, used for steering and focusing 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 that 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:
  - —Several retractable X ray targets;
  - —Flattening filters and electron scattering foils (also called scattering filters);
  - —Primary and adjustable secondary collimators;
  - —Dual transmission ionization chambers;

— A field defining light and a range finder;
— Optional retractable wedges;
— Optional 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:
  — Either scattering the pencil beam with a single or dual scattering foil; 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 isocentre. The IEC recommends that the transmission of the primary X ray beam through the rectangular collimator should not exceed 2% of the open beam value.
- Dual transmission ionization chambers are used for monitoring the photon and electron radiation beam output as well as the radial and transverse beam flatness (see Section 5.5.14).
- The field defining light and the 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 centimetre scale whose image on the patient's skin indicates the vertical distance from the linac isocentre.

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 high energy linac movable target is shown in Fig. 5.6(b).

At electron energies below 15 MeV (photon beam energies 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:

- A primary collimator;
- Secondary movable beam defining collimators;
- An 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 on to edges of the target on one end of the block and on to 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 tenth-value layers (TVLs)). 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 isocentre, with sides of the order of few millimetres up to 40 cm.
— Modern linacs incorporate independent (asymmetric) jaws that can provide asymmetric fields, most commonly one half or three quarter blocked fields in which one or two beam edges, respectively, are coincident with the beam central axis.
— 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 (micro MLCs) projecting 1.5–6 mm leaf widths and up to 10×10 cm^2 fields at the linac isocentre are currently commercially available. They may be used in radiosurgery as well as for 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 the 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 the 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:
  — Pencil beam scattering. The scattering of the electron pencil beam over the 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.
  — Pencil beam scanning. 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

IEC 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 monitor units (MUs), termination of radiation and monitoring of beam flatness and dose rate.

- Most common dose monitors in linacs are transmission ionization chambers permanently imbedded in the linac clinical photon and electron beams to monitor the beam output continuously during patient treatment.
- Most linacs use sealed ionization 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 ionization 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 per cent above the prescribed dose has been delivered.
- In the event of a simultaneous failure of both the primary and secondary ionization chambers, the linac timer will shut the machine down with a minimal overdose to the patient.
- The main requirements for the ionization chamber monitors are as follows:
  - —Chambers must have a minimal effect on clinical photon and electron radiation beams;
  - —Chamber response should be independent of ambient temperature and pressure (most linacs use sealed ionization chambers to satisfy this condition);
  - —Chambers should be operated under saturation conditions.
- The primary ionization chamber measures MUs. 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 a source to surface distance (SSD) of 100 cm.
- Once the operator preset number of MUs has been reached, the primary ionization 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 as the beam energy, flatness and symmetry. Measurement of all these additional parameters requires that the ionization 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 on to the target or scattering foil, thereby ensuring beam flatness and symmetry. The particular design of the ionization chamber electrodes and sectors varies from one manufacturer to another.

- Linacs must be equipped with a monitoring system that continuously displays the machine isocentre 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 irradiation 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 irradiation 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 centres 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;
- Heavy ions (helium, carbon, nitrogen, argon, neon) produced by synchrocyclotrons 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 and in maintenance and servicing costs, thus precluding a widespread 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:

- X ray machines (superficial and orthovoltage);
- Teletherapy (^{60}Co) machines;
- Linacs.

All radiotherapy equipment must be housed in specially shielded treatment rooms in order to protect personnel and the 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 so as to minimize 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 an ordinary concrete barrier by approximately one half; however, it will also increase the construction material cost by a factor of 30.

Shielding issues related to linac bunkers are discussed in more detail in Section 16.17.

5.8. COBALT-60 TELETHERAPY UNITS VERSUS LINACS

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

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

The important features of ^{60}Co teletherapy machines can be summarized as follows:

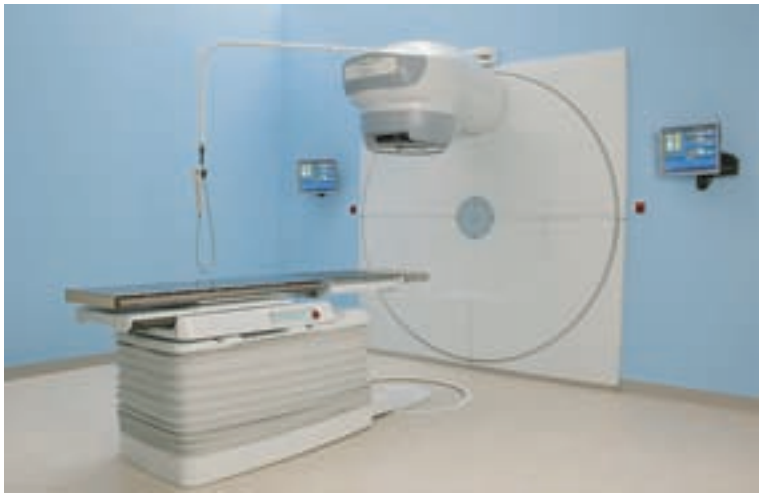
- Relatively high energy γ ray emission;
- Relatively long half-life;
- Relatively high specific activity;
- Relatively simple means of production.

Figure 5.8(a) shows a ^{60}Co teletherapy machine; Fig. 5.8(b) shows a stamp issued by Canada Post commemorating Canada's role in the development of the ^{60}Co machine.

Linacs were developed concurrently by two groups: W.W. Hansen's group at Stanford University in the USA and D.D. Fry's group at the Telecommunications Research Establishment in the UK. Both groups were interested in linacs for research purposes and profited heavily from the microwave radar technology developed during World War II, using 3000 MHz as the design frequency.

The potential for the use of linacs in radiotherapy became apparent in the 1950s, and the first clinical linac was installed in the 1950s at the Hammersmith Hospital in London. 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

(a)



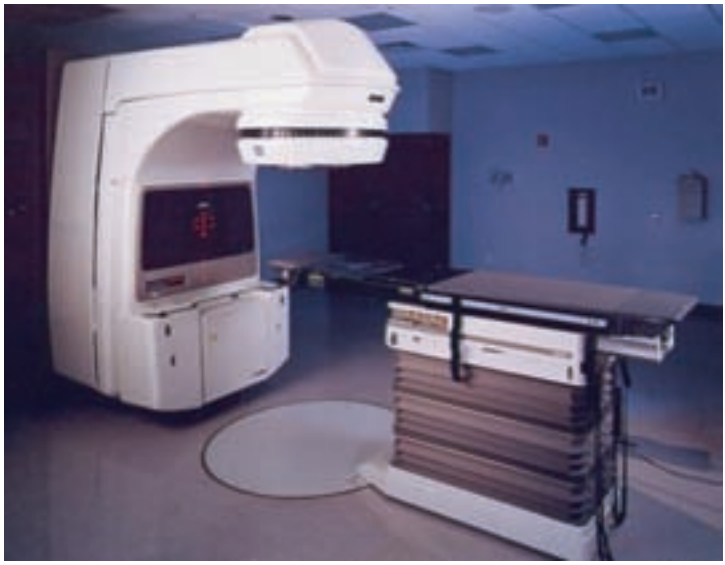
(b)

FIG. 5.8. Cobalt-60 teletherapy machine. (a) Theratron Equinox, a megavoltage external beam therapy system using cobalt technology, manufactured by MDS Nordion, Ottawa, Canada (published with permission from MDS Nordion). (b) Schematic diagram of a cobalt unit depicted on a postage stamp issued by Canada Post in 1988 in honour of H.E. Johns, who invented the ^{60}Co unit in the 1950s (© Canada Post Corporation, 1988; reproduced with permission).

contrast to a ^{60}Co unit, which provides essentially only one γ 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 linac.

In comparison with ^{60}Co machines, linacs have become very complex in design:

(a)



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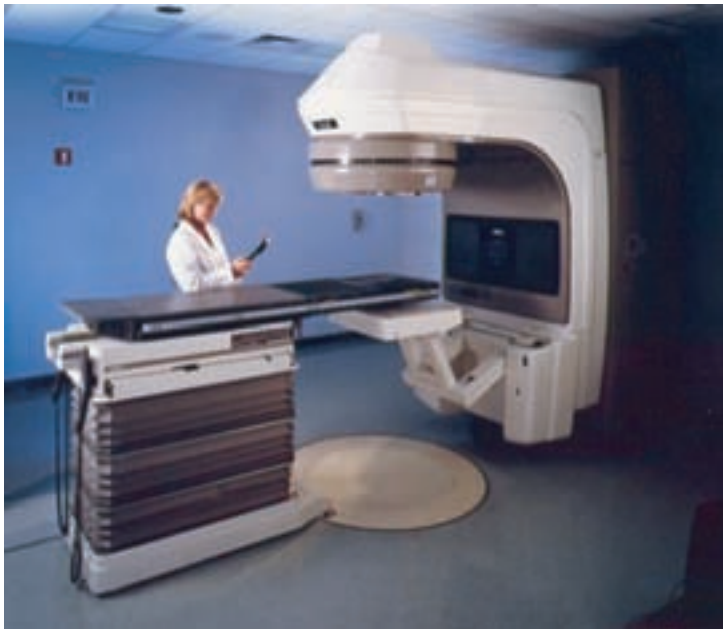

FIG. 5.9. Modern dual photon energy linac manufactured by Varian; the gantry and the patient support assembly are clearly shown. (a) The portal imager is retracted; (b) the portal imager is activated. (Photographs courtesy of Varian Oncology Systems.)

— In part because of the multimodality capabilities that have evolved and are available on most modern machines;
— In part because of an increased use of computer logic and microprocessors in the control systems of these machines;
— In part because of added features, such as high dose rate modes, multileaf collimation, electron arc therapy and the dynamic treatment option, which is characterized by a controlled motion on the collimators (dynamic wedge), MLC leaves (IMRT), gantry or table while the beam is turned on.

Despite the clear technological and practical advantages of linacs over ^{60}Co machines, the latter still occupy an important place in the radiotherapy armamentarium, mainly because of the considerably lower capital, installation and maintenance costs of ^{60}Co machines compared with linacs. In the developing world, ^{60}Co 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 MLCs, dynamic wedges and dynamic operation, could be installed on modern ^{60}Co machines to allow, at a lower cost, a similar sophistication in treatment as linacs. It is unfortunate that manufacturers of ^{60}Co units are very slow in reacting to new technological developments in radiotherapy, conceding pre-eminence to linac manufacturers even in areas where it would be much easier and more practical to run ^{60}Co machines than linacs.

5.9. SIMULATORS AND COMPUTED TOMOGRAPHY SIMULATORS

Simulators and CT simulators are important components of equipment used in radiotherapy. 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 magnetic resonance (MR) and PET scanners. Both simulators and CT simulators incorporate three major systems: the mechanical, X ray tube and imaging equipment.

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

- Acquisition of the patient data set;
- Localization of target and adjacent structures;
- Definition and marking of the patient coordinate system;
- Design of treatment fields;
- Transfer of data to the treatment planning system (TPS);
- Production of an 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. Radiotherapy simulator

A radiotherapy 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 ^{60}Co 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 the mechanical features and geometric field arrangements of various megavoltage machines, ranging from ^{60}Co machines with an SAD of 80 cm to high energy linacs with an SAD of 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 of 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. Computed tomography 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:

- A flat table top surface to provide a patient position during simulation that will be identical to the position during treatment on a megavoltage machine.
- A laser marking system to transfer the coordinates of the tumour isocentre, derived from the contouring of the CT data set, to the surface of the patient. Two types of laser marking systems are used: a gantry mounted laser and a system consisting of a wall mounted moveable sagittal laser and two stationary lateral lasers.
- A virtual simulator consisting of software packages that allow the user to define and calculate a treatment isocentre and then simulate a treatment using digitally reconstructed radiographs (DRRs).

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

— Physical simulation, which covers the first three of the six target localization steps listed above;
— Virtual simulation, which covers the last three of the six target localization steps listed above.

In CT simulation the patient data set is collected and target localization is 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 TPS is achieved electronically.

The planar simulation X ray film provides a beam's eye view (BEV) of the treatment portal but does not provide 3-D 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 TPS and represents a computed radiograph of a virtual patient generated from a CT data set representing the actual patient. Just like a 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 the 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 grey scale mapping.

An extension of the DRR approach is the digitally composited radiograph (DCR), which 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 IAEA summarized the lessons learned from accidental exposures in radiotherapy, and a report by the American Association of Physicists in Medicine (AAPM) specifically addressed medical accelerator safety considerations.

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

(a) Preparation of an equipment specification document;
(b) Design of the treatment room and radiation safety;
(c) Acceptance testing of equipment;
(d) Commissioning of equipment;
(e) A quality assurance programme.

Items (a), (c) and (d) are addressed in detail in Chapter 10, item (e) is addressed in Chapter 12 and item (b) is addressed in Chapter 16.

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