Linear accelerators

Modern linear accelerators (*linacs*) are now found in virtually all radiation therapy departments, having replaced most other therapy units. They are used to treat patients with beams of electrons or Bremsstrahlung x rays following interactions of electrons in a suitable target. Intense electron beam currents are achievable with an accelerator to provide high dose rates for both x-ray and electron treatments. Hence, treatment times are short even at relatively long target-to-patient distances. Many modern linacs provide multiple electron and photon beam energies in the megavoltage energy range. The term *MV* is typically used to describe photon beams (e.g., 6 MV) whereas the term *MeV* is typically used to describe an electron beam (e.g., 6 MeV).

Historical development

The earliest linear accelerators were so-called direct accelerators, in which charged particles were accelerated by an electric field created by placing a high potential difference over an insulated column.

In the simplest form of linear accelerator, such as the simple x-ray tube shown in Figure 4-1, electrons are boiled off the cathode surface (which is heated by a filament) and accelerated toward the anode. The accelerating force is provided by a static electric field produced by maintaining the anode at a positive potential V relative to the cathode. The energy acquired by an electron accelerated in this fashion is determined by the voltage, V; electrons accelerated through a potential difference of 1 V gain an energy of 1 electron volt (eV). This corresponds to 1.6×10^{-19} Joules, although the unit eV (or MeV) is most commonly used.

The velocity of the accelerated electron can be determined from the classic equation for kinetic energy:

$$T = \frac{1}{2} m_e v^2 \tag{4-1}$$

where m_e is the electron's mass and v is its velocity. It is straightforward to calculate the velocity of a 1 eV electron as 1.87×10^7 m/sec, or 6.25% of the speed of light. To accelerate electrons to higher energies requires either that the potential difference be increased (impractical at voltages above a few hundred thousand volts) or that the accelerating force be repeated numerous times. Note that Equation (4-1) cannot be used once the electron becomes *relativistic* (i.e., when its velocity exceeds approximately 20% of the speed of light), because the change in electron mass must be considered.

The first linear accelerator was developed by Wideröe in 1928 to accelerate heavy ions.[18] Wideröe's accelerator consisted of a series of metal cylinders, termed *drift tubes*, with alternate cylinders connected to opposite terminals of an oscillating radio frequency voltage (Figure 4-9).

Because adjacent drift tubes are connected to opposite terminals of the power supply, an electric field develops between the ends of the tubes. Ions are accelerated across the gaps between adjacent drift tubes.

While inside the drift tubes, the ions are shielded from the electric fields produced by the radiofrequency voltage. The length of each tube is sufficient to allow ions to drift undisturbed through the tube each time the radiofrequency voltage changes polarity. In this manner, ions are accelerated from one drift tube to the next only after the correct polarity has been established across the gap between tubes. As the energy, and therefore the velocity, of the particles increases, the drift tubes must increase in length. Although useful for accelerating heavy ions, the Wideröe accelerator was not suitable for accelerating electrons, because the high speed of the electrons would have required inordinately long drift tubes.

Development of the electron linac was made possible by the invention in the late 1930s and early 1940s of the microwave cavity and by the development of klystron and magnetron tubes as sources of microwave power.

Between 1948 and 1955, the first electron linacs were designed and built by groups working independently in England and in the United States. Fry and associates at the Telecommunications Research Group, Great Malvern, England (later to become

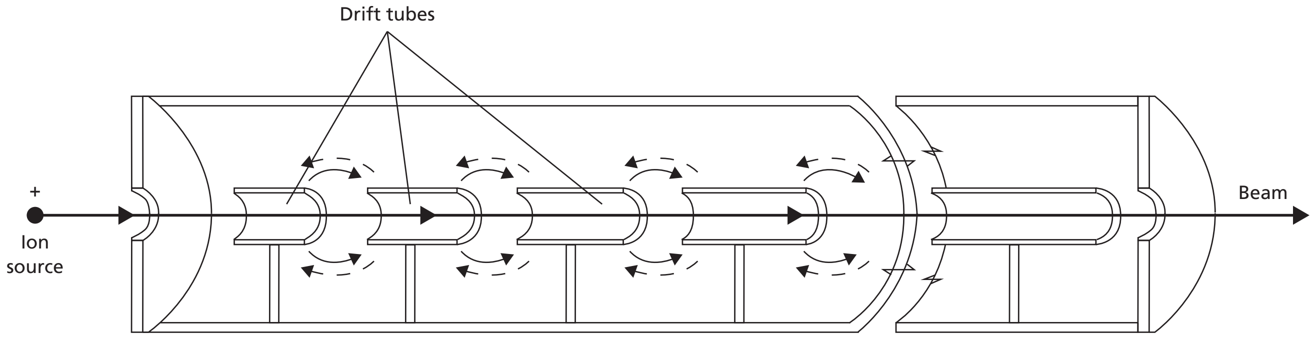

Figure 4-9 An early linear accelerator, in which drift tubes were suspended in an evacuated accelerator waveguide and connected to opposite terminals of a power supply.

part of the Atomic Energy Research Establishment at Harwell) designed a 0.5 MeV linac in 1946 and accelerated electrons later that same year.[19] Independently, Ginzton and associates in Palo Alto, California, USA, developed a 1.7 MeV accelerator, which became operational in early 1947.[20] By late 1947, both groups had achieved energies of 3.5 to 6 MeV. Also during this period, Chodorow, Ginzton, and Hansen constructed multimegawatt klystrons, having many orders of magnitude greater power than those developed for wartime radar applications.[21] These microwave sources made possible the development of traveling-wave linacs with high beam currents and energies of up to 1000 MeV.

In a microwave accelerator, microwaves are guided into a cylindrical metal tube known as an *accelerator waveguide*. Simultaneously, electrons or other charged particles are introduced into one end of the accelerator waveguide. The presence of the electromagnetic field induces an electric current within the walls of the accelerator waveguide. The current generates an electric field, which exerts a force on the particles, accelerating them to high velocities.

The microwave power, P, required to establish the electric field within the accelerator depends on several characteristics of the waveguide:

$$P = \frac{V^2}{ZL}$$

where V is the potential difference developed within the guide through which particles are accelerated, Z is the *shunt impedance* of the accelerator waveguide, and L is the length of the waveguide. The shunt impedance is a measure of the efficiency of the guide. To develop the potential to accelerate particles to an energy of 10 MeV in a structure with a shunt impedance of 100 MΩ/m and a length of 1 m, 1 MW of electromagnetic power is consumed. (*Note*: MΩ represents the quantity "mega ohms," where "ohm" is a unit of impedance.) Additional power is consumed by the accelerated particles themselves, as well as by other components of the linac. Consequently, a source of 2 MW or more of microwave power is required. A 2 MW magnetron is often used in low-energy linacs, whereas klystrons with power ratings of up to 10 MW are used in higher-energy units.

To regulate the phase velocity of the microwave (the velocity at which a peak of the microwave appears to travel), barriers such as metal discs may be placed in the waveguide at regular intervals (Figure 4-10). The term *phase velocity* refers to the velocity of a peak or valley of an electromagnetic wave. A linear accelerator designed by use of this principle allows the crest of the electromagnetic wave to travel at less than the speed of light, permitting the charged particle to keep up. The energy carried by the electromagnetic wave always moves at the speed of light.

The electric field established by the microwaves develops between the discs. In some accelerator structures, a *forward*-directed field develops between one pair of discs, while a *backward*-directed field develops between the adjacent pair of discs. Consequently, a wavelength spans two adjacent cavities (Figure 4-11). Particles in the space between the first pair of discs are accelerated in the forward direction, while particles in the second space (between the second and third discs) are accelerated in the backward direction. Naturally, particles would arrive

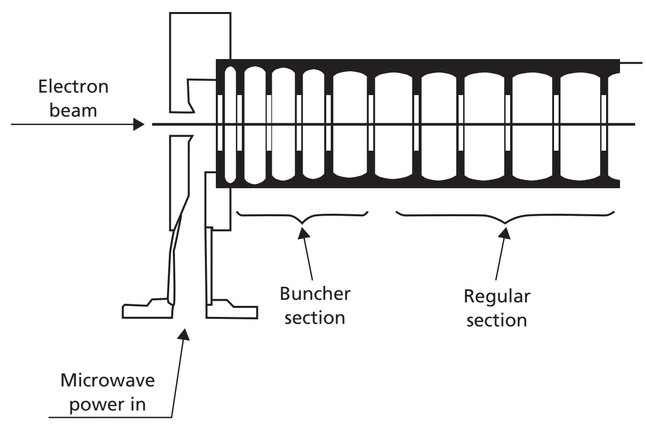

Figure 4-10 A disc-loaded waveguide for electron acceleration. The spacing between discs in the *buncher section* is less than that in the *regular section*. The buncher section accelerates electrons to relativistic energies and forms them into bunches for efficient acceleration by the regular section.

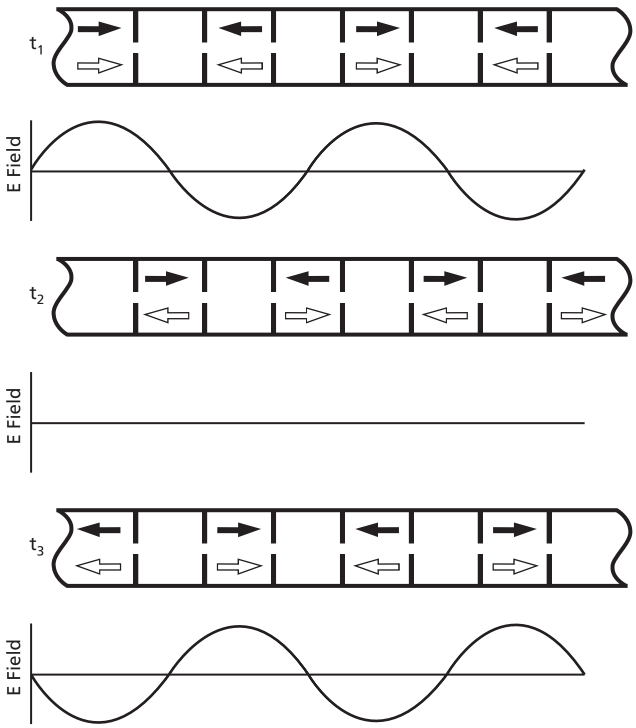

Figure 4-11 A schematic of an accelerator waveguide showing separate forward and backward waves and their superposition. In each diagram, the *solid arrows* indicate the positive and negative peaks of a microwave electric field moving from left to right, whereas the *open arrows* indicate a field reflected from the right moving to the left. At t_1 and t_3, the forward and reflected waves superimpose constructively. At t_2, they interfere destructively, and the electric field intensity is zero everywhere.
Source: Adapted from Karzmark et al. 1993.[23]

at the second space only if they were accelerated in the forward direction while in the first space. The distance between the discs is determined by the velocity of the particles:

$$L_n = \frac{V_n}{2\nu}$$

where L_n is the distance between adjacent discs, V_n, is the velocity of the particles, and ν is the frequency of the microwaves. In medical electron linacs, the electrons quickly approach the speed of light as they are accelerated down the waveguide. Consequently, the spacing between discs quickly reaches a maximum distance. In the more common $\pi/2$ mode, one wavelength spans four cavities, and particles are accelerated only in alternate cavities. The microwave field has zero intensity in the remaining cavities (Figure 4-11). The benefits of this design are described later in this chapter.

Early medical linacs were of the *traveling wave* design. Microwaves were injected into one end of the accelerator and traveled to the other end, where they were extracted. As they traveled, they carried the electrons along with them. The extracted microwaves could be conducted back to the proximal end of the accelerator and re-injected. The disc-loaded design

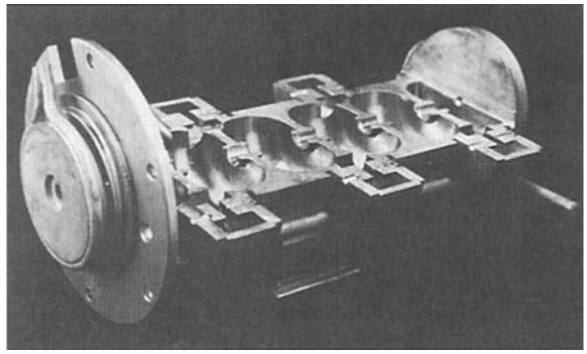

Figure 4-12 An example of a side-coupled standing-wave linear accelerator waveguide.
Source: Karzmark et al. 1993.[23]

limited the shunt impedance to maximum values less than 60 MΩ/m, which restricted the *accelerator gradient*, or attainable electron energy in MeV/m, to fairly low values.

In 1968, Knapp et al. invented the *side-coupled standing wave accelerator*, in which the microwaves were reflected from the ends of the accelerator waveguide.[22] The backward traveling wave interferes with the forward traveling wave, alternately constructively and destructively. The resulting *standing wave* has a magnitude of approximately double that of the traveling wave, and the peak intensity travels along the waveguide at the phase velocity of the traveling wave. In alternate spaces between discs, the magnitude of the wave is always at or near zero (Figure 4-11).

In standing wave structures, the spaces between discs are optimized by changing the shape of the discs. The resulting *cavities* permit the microwaves to resonate, improving the efficiency with which their energy is transferred to the accelerated electrons. An additional improvement is that the microwave electric field intensity is at or near zero in alternate cavities. Although necessary to conduct the microwave energy, these cavities play no role in accelerating particles. By moving them to the side (Figure 4-12), off the waveguide axis, the accelerating cavities can be placed closer together. The overall length of the structure becomes shorter, facilitating the placement of the guide in the treatment unit and improving its efficiency.

Major components of medical electron accelerators

Modern linear accelerators consist of several major subsystems. These components produce the electrical power required to generate microwaves, conduct the microwave power to the accelerator waveguide, and transport the accelerated beam ultimately to the patient.

Modulator and pulse-forming network

Linear accelerators require fairly large amounts of electrical power. The power must be provided in large pulses because

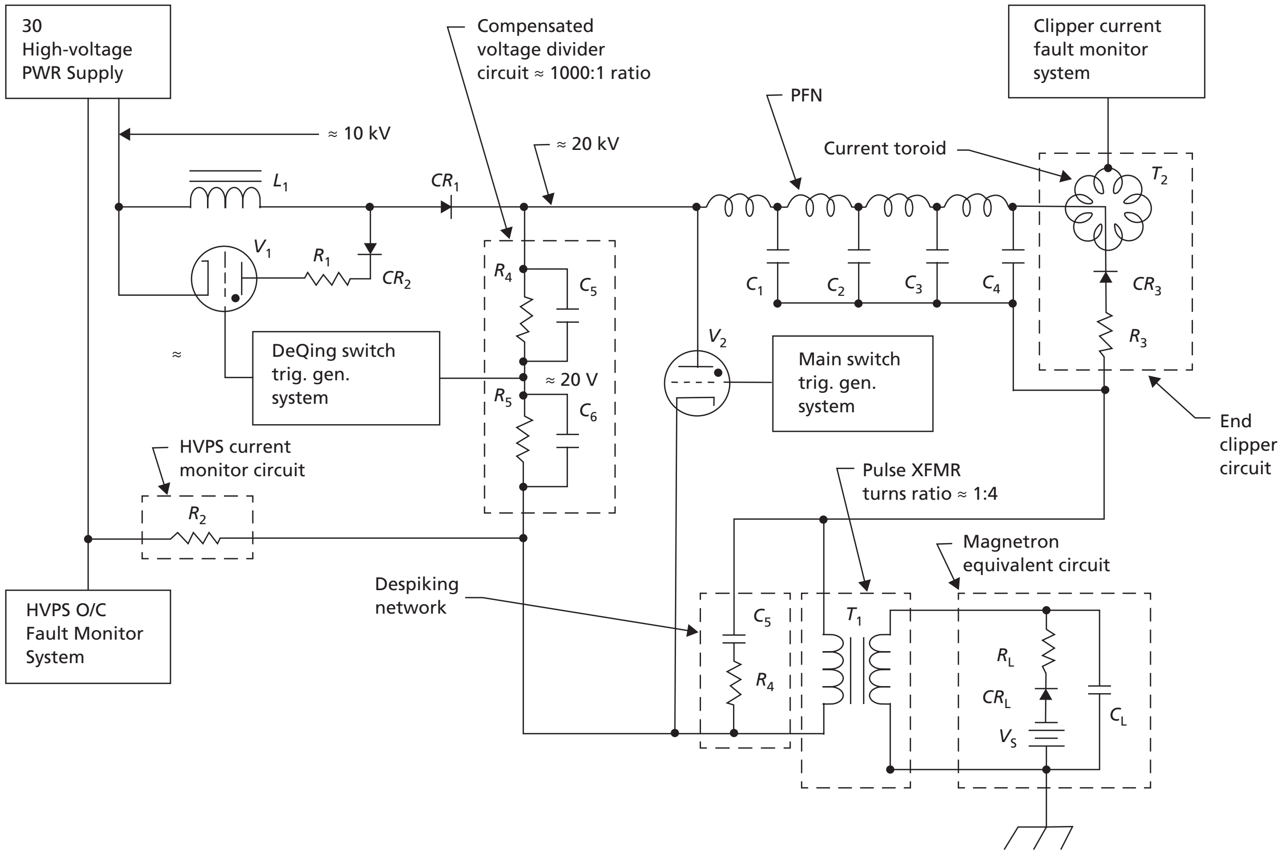

Figure 4-13 The modulator and pulse-forming network of a modern medical electron accelerator.
Source: Karzmark et al. 1993.[23]

linacs accelerate particles in bursts. The *modulator* consists of a power supply that converts the incoming alternating current into direct current, along with a *pulse-forming network* that modulates the current into pulses.

A diagram of a modulator appears in Figure 4-13. The direct current charges a bank of capacitors, which store the charge until a pulse is required. The charging cycle lasts about a millisecond. On receiving a signal from a timing circuit, a switching tube closes, completing a circuit from the capacitor bank, through a transformer, to ground. The capacitors discharge rapidly, but because of the inductor connecting them, they discharge in sequence. The resulting pulse is nearly square.

The switching tube, or *thyratron*, is a gas-filled triode (Figure 4-13). When the grid is charged positively, electrons flow from the cathode to the anode. The gas within the tube ionizes and conducts larger currents than do other switching devices. At the end of the pulse, the grid voltage is removed, preventing further current flow while the pulse-forming network recharges. This cycle is repeated between 50 and 500 times each second.

The output of the pulse transformer is conducted to the microwave-producing tube, either a magnetron or a klystron. The resulting microwave pulse, similar in shape to the electrical pulse, is about 6 μsec long and consists of several megawatts of power.

Magnetrons

The cavity-type magnetron was invented in 1940 by Boot and Randall and made high-definition radar (radio detection and ranging) possible in World War II.[23] Magnetrons are commonly used in low-power linear accelerators. A magnetron of this type consists of a cylindrical diode containing a central cathode, which is heated by internal filaments. The coaxial anode is constructed of solid copper with coupled resonant cavities formed in the wall (Figure 4-14).

An axial magnetic field is supplied by a large permanent magnet. When a DC pulse is applied to the diode, electrons from the cathode are accelerated toward the anode and assume a spiral path because of the magnetic field. The individual electrons follow complex cycloidal paths around the cathode. As the electrons swirl along their spiral pathway, they induce intense local variations in the axial magnetic field. The radiofrequency energy induced in the magnetic field by this process is trapped in the resonant cavities. Oscillation of this trapped energy forms varying electrical fields across the lips of each cavity. These varying electrical fields channel electrons to the more positive regions of the anode, and the spiraling electron cloud appears to sweep around the cathode as it tracks the more positive regions. The sweeping electron cloud induces additional intense variations

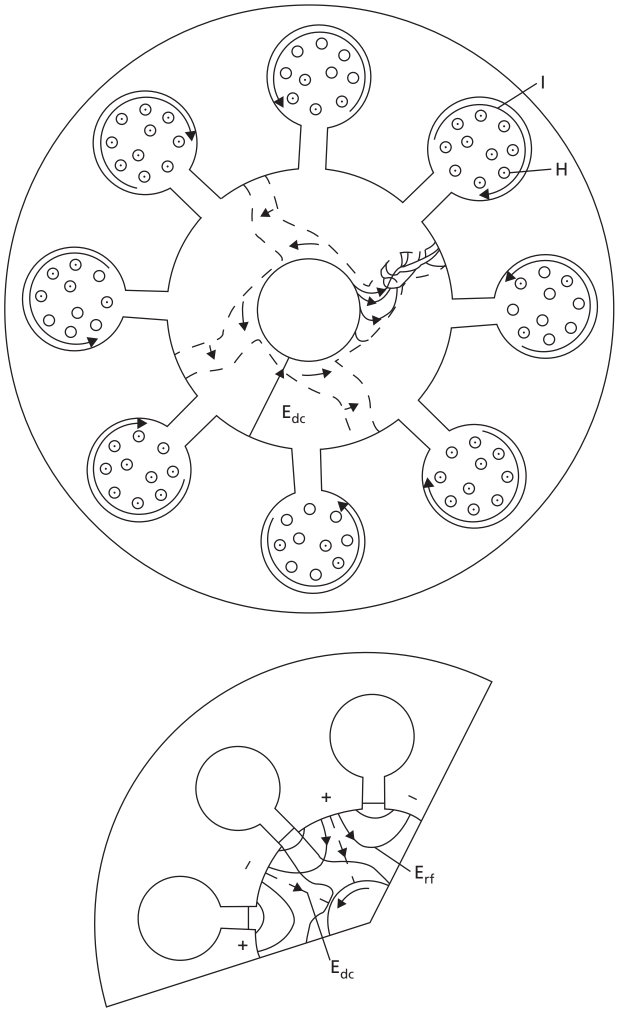

Figure 4-14 A cutaway view of a magnetron microwave power tube. *Source:* Karzmark et al. 1993.[23]

in the magnetic field, which are in resonance and coupled with the cavities of the coaxial anode. A loop antenna inserted in one of the cavities taps the radiofrequency energy in the cavities and transfers it to a waveguide for transmission to the accelerator waveguide. Magnetrons transform DC power to radiofrequency power with efficiencies as high as 60%. Typically, such magnetrons provide 2 MW peak power.

The microwave wavelength must be of an appropriate length so that the accelerator components are reasonably easy to design and manufacture. Most modern medical linacs operate with microwaves of about 3000 MHz, in what is known as the *S-band*. The wavelength in a vacuum may be determined from the frequency ν and speed of light c as:

$$\lambda = \frac{c}{\nu}$$

For microwaves of 3000 MHz, the wavelength in a vacuum is on the order of 10 cm. In a waveguide, the wavelength is reduced somewhat because the phase velocity of the radiation is reduced.

During operation of a linac, the temperature of various components tends to increase. Changes in temperature may adversely affect the operation of the accelerator and must be avoided. In particular, a temperature rise of the accelerator guide of as little as 1°C may cause sufficient expansion to change the resonant frequency by 60 kHz. A change in resonant frequency of as little as 20 kHz can seriously degrade the performance of the accelerator. This sensitivity means that the frequency of the microwaves must be adjusted to compensate. Consequently, the magnetron must be *tunable*, to permit continuous matching of the frequency. Early linacs equipped with fixed-frequency magnetrons required that the operator manually adjusted the flow of water around the accelerator waveguide, to regulate the size of the resonant cavities and keep them matched to the microwave frequency. Modern magnetrons are equipped with motor-driven tuners that are controlled by a circuit that senses the microwave frequency.

Klystrons

The principle of the klystron microwave power tube is illustrated in Figure 4-15. The tube requires a low-power radiofrequency oscillator to supply RF power to the first cavity, termed the *buncher*. The low-power RF source used with a klystron is known as the *RF driver* and typically delivers a power level of a few hundred watts. By application of an accelerating voltage supplied as a DC pulse from the DC power supply (the pulse forming network described earlier), electrons with energies of several keV are injected into the cavity. In the buncher cavity, the velocity of the electrons is modulated by the electric field component of the microwave field. This modulation of velocity causes the electrons to group together into closely spaced electron bunches. As the electron bunches arrive at the second cavity, termed the *catcher cavity*, they are decelerated, and their energy is transformed into a pulse of microwave power. High-power klystrons containing additional cavities have achieved direct current-to-microwave conversion efficiencies of up to 55%, with peak powers as high as 24 MW.

Microwave power-handling equipment

Microwaves from the magnetron or klystron are conducted to the accelerator waveguide by rectangular sections of waveguide (Figure 4-16). The microwaves are confined by the metal walls of the waveguide and propagate through a dielectric gas such as Freon or sulfur hexafluoride. The dimensions of such waveguides are typically 0.6 λ in width by (0.2–0.5) λ in height. The circular accelerator waveguide is normally energized in the TM01 mode, meaning that the magnetic field is transverse to the longitudinal axis of the guide. Consequently, the electric field is axial, and the force exerted by the electric field on the particles accelerates them along the longitudinal axis of the guide.

In single-modality accelerators (those capable of producing only an x-ray beam of selected energy), the waveguide and other microwave power-handling equipment are straightforward.

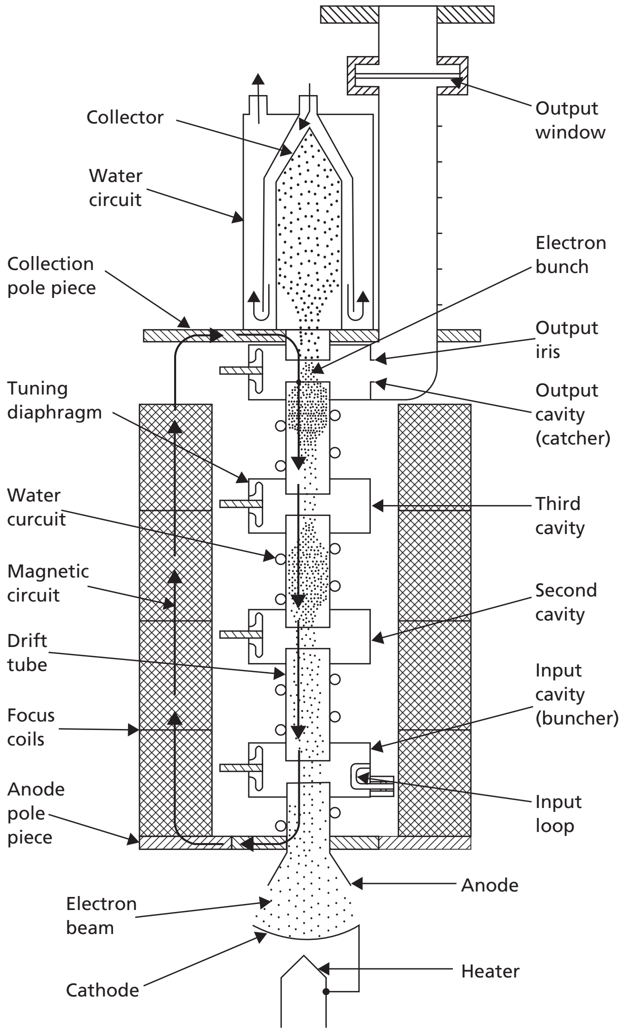

Figure 4-15 Schematic cross-section of a high-power four-cavity klystron power tube.

Multimodality accelerators, capable of producing beams of electrons as well as x-ray beams of one or more energies, require alteration of the microwave power delivered to the accelerator structure, so that particles can be accelerated to different energies. Because magnetrons and klystrons are generally adjustable over only small ranges, other means are required to vary the power. One available device is called a *power splitter*. A variable fraction of the power directed to this device is returned to the accelerator waveguide. The remainder is absorbed in a water-filled *load*. Of the power directed to the accelerator waveguide, a portion may be reflected from the guide. A *circulator* or *directional coupler* directs this reflected power away from the magnetron or klystron, to avoid interfering with its operation.

So-called *dual-energy* or *multi-energy* linacs have the capability of producing x-ray beams of two or more energies. They incorporate accelerator waveguides of sufficient length to accelerate electrons to the energy required for the highest-energy photon beam desired, but must be able also to accelerate electrons to lesser energies. This capability requires radiofrequency power-handling equipment of greater complexity, to ensure that the electron bunches remain focused and that the variation in energy of the electrons remains small. An *energy switch* employed by one manufacturer is shown in Figure 4-17. It provides control over the amount of radiofrequency power passing from the left-hand cavity into the remainder of the guide.

Vacuum pump

The accelerator guide of a linac requires a high vacuum (on the order of 10^{-8} mm Hg) to prevent power loss and electrical arcing caused by interactions of electrons with gas molecules. Although older models used mechanical fore-pumps and oil diffusion vacuum pumps, all newer models use *sputter-ion* (*Vac-Ion*) pumps to maintain good vacuum.

A sputter-ion pump typically consists of multiple cylindrical anodes positioned between two cathodes. The cathodes are sandwiched between the poles of a magnet. The cathodes are composed of a reactive sputtering material such as titanium. Electrons ejected spontaneously from the cathode are attracted toward the anode and assume a spiral path in the magnetic field. As a consequence, they oscillate between the cathodes and collide with gas molecules to produce considerable ionization. The resulting positive ions bombard the cathodes, causing ejection (*sputtering*) of neutral atoms of titanium, which are deposited chiefly on the anodes. By this mechanism, gas molecules are continuously removed from the electron accelerator section.

Bending magnet

Low-energy linacs require short accelerating structures that can be mounted directly in line with the path from the x-ray target to the patient. Higher-energy linacs require longer structures that are often mounted horizontally (or nearly so) within the linac gantry. A *bending magnet* is used to change the direction of the accelerated electron beam from horizontal to vertical. The angle of bend may be 90°, but many accelerators use a 270° *achromatic* magnet. As described below, a magnet with multiple 90° bends provides greater stability of the resulting photon beam.

The electrons accelerated in a linac do not all reach the bending magnet with exactly the same velocity. By equating the force on a particle in an electromagnetic field with the centripetal force, we can show that an electron in a magnetic field follows a curved path with a radius given:

$$r = \frac{mv}{qB}$$

Example 4-7

What is the radius of the path of a 10 MeV electron passing through a magnetic field of strength = 7000 Gauss (0.7 Tesla or 0.7 Weber/m^2)?

$$r = \frac{mv}{qB}$$

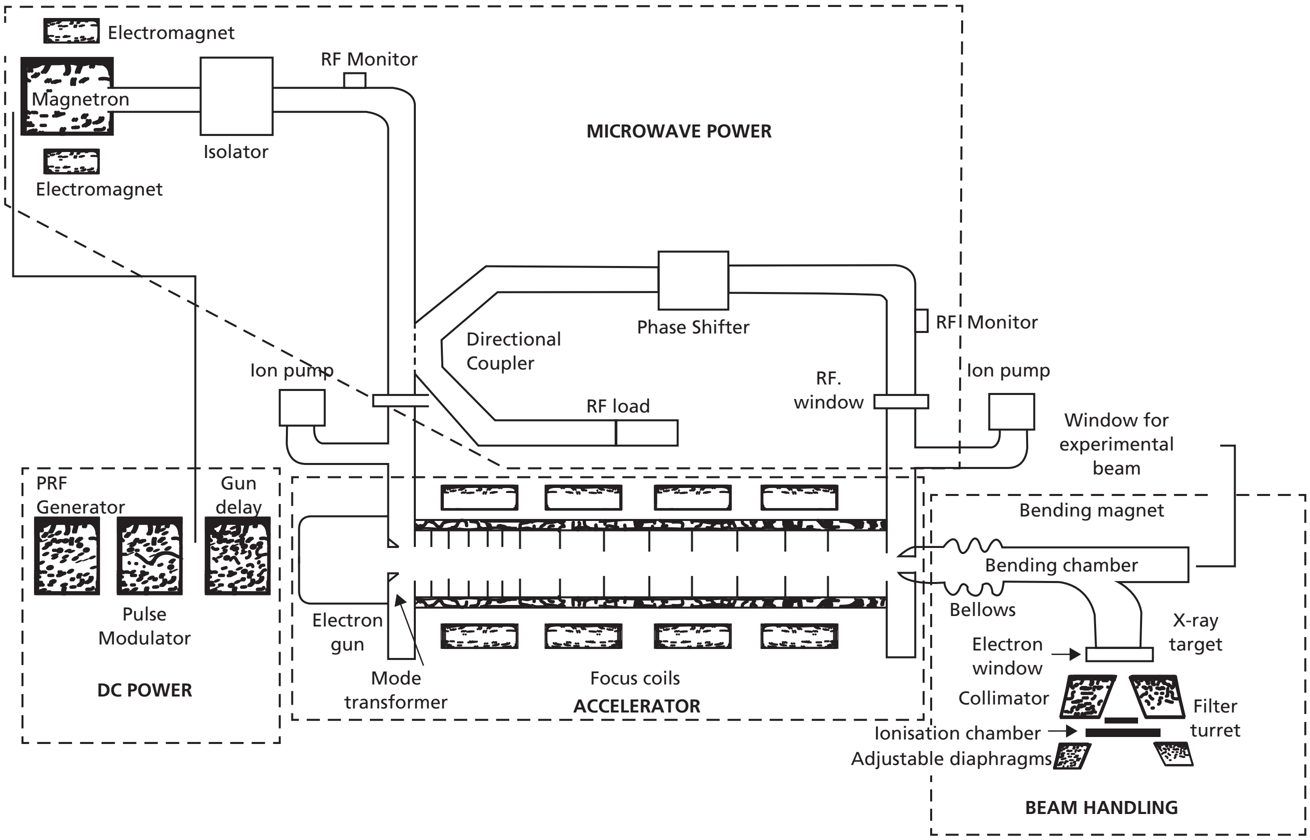

Figure 4-16 RF power-handling equipment for a linear accelerator.

where v is the velocity of the 10 MeV electron (99.88% of the speed of light) and m is the mass of the electron; m is related to m_0, the rest mass of the electron (9.11×10^{-31} kg) by:

$$m = \frac{m_0}{\sqrt{1 - \frac{v^2}{c^2}}}$$

Therefore, $m = 1.86 \times 10^{-29}$ kg. The intensity of the magnetic field $B = 0.7$ Weber/m^2 and the charge of an electron $q = 1.6 \times 10^{-19}$ C. Therefore:

$$r = \frac{(1.86 \times 10^{-29}\ \text{kg})(3 \times 10^{8}\ \text{m/s})}{(1.6 \times 10^{-19}\ C)(0.7\ \text{weber/m}^2)}$$

$$r = 0.05\ \text{m or 5 cm}$$

The Weber is a unit of magnetic flux and describes the integral of the magnetic field strength over a surface. Magnetic field

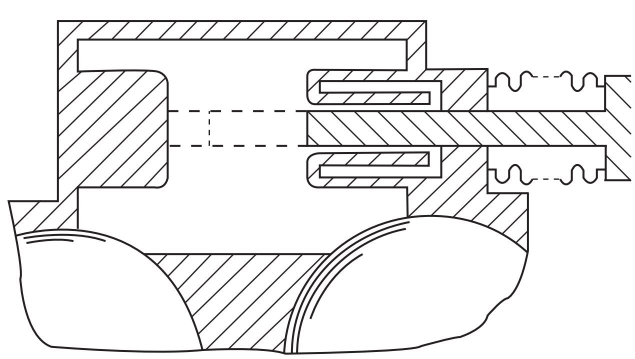

Figure 4-17 A noncontact-type microwave energy switch for a standing-wave accelerator guide.
Source: Karzmark et al. 1993.[23]

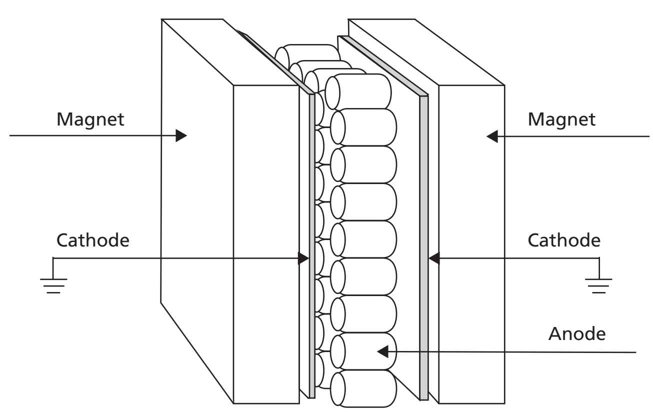

Figure 4-18 A representative sputter-ion pump.

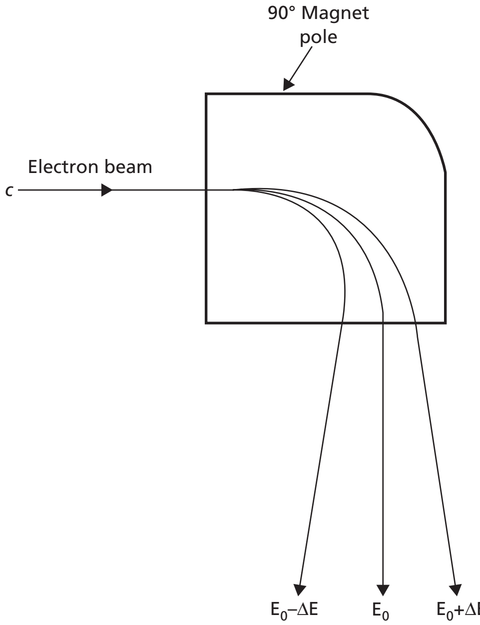

Figure 4-19 A simple 90° bending magnet, showing the paths of electrons of three energies.

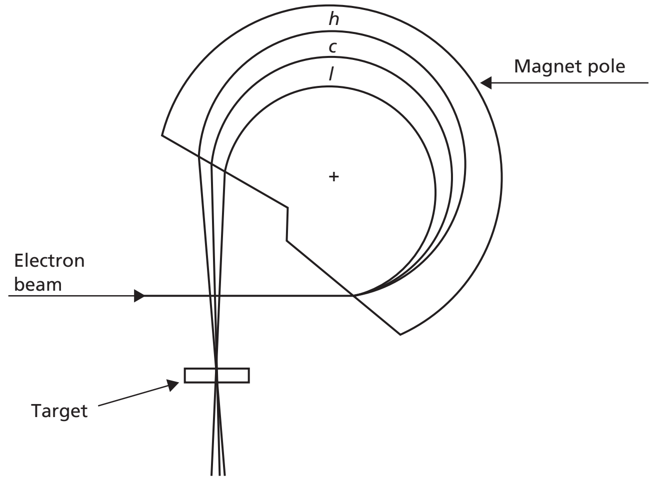

Figure 4-20 A modern 270° *achromatic* bending magnet, showing the paths of electrons of three energies.

strength is defined by the force exerted by the field on a charged particle:

$$F = qv \times B$$

B therefore is expressed in units of $\frac{\text{newton}}{\text{coulomb (m/s)}}$, which has been given the special name Weber/meter^2: 1 Weber/meter^2 = 1 Tesla = 10^4 Gauss. The Earth's magnetic field is about 0.000025–0.000065 Tesla or 0.25–0.65 Gauss.

For a chosen magnetic field strength, electrons with higher energy (higher values of mv) are bent through a larger radius than are lower-energy electrons. As shown in Figure 4-19 of a 90° bending magnet, lower-energy electrons strike the target at a different point than the higher-energy component of the beam. However, in a 270° achromatic bending magnet, as shown in Figure 4-20, the low-energy and high-energy components of the beam converge at a point called the *triple focus*. A target positioned at this point intercepts all electrons emerging from the bending magnet. Many bending magnets are equipped with an energy-defining slit consisting of barriers that intercept electrons whose energies vary from the desired energy by more than a selected amount.

The energy-defining slit of a modern linear accelerator consists of a mechanical barrier placed near the midpoint of the bending magnet. A window in the barrier allows electrons of the selected energy range (correct radius of bend) to pass through. Electrons outside the range are stopped. The choice of barrier material is important to minimize the production of x rays. Nevertheless, the bending magnet is a major source of leakage radiation. If an accelerator is not tuned properly, it can steer a large fraction of the electron beam into the energy slits. The resulting leakage radiation may exceed regulatory limits.

X-ray target

When an x-ray beam is desired, a *target* of an appropriate material is moved into the path of the electrons. The material usually has a high atomic number (e.g., tungsten) in low-energy linacs, but may be of intermediate atomic number (e.g., copper) in high-energy units. In contrast to conventional x-ray tubes, an accelerator target is generally a *transmission* target, meaning that the generated x rays are transmitted through the target material to reach the patient. The thickness of the target is a compromise between one that ensures every electron interacts and one that absorbs the fewest x rays. The ideal thickness is related to the *radiation length*, the thickness in which $1/e$ of the electron beam is absorbed. Bremsstrahlung x rays leave the location of production at an angle relative to the direction of incident electrons. At low electron energies, the mean angle of emission (ϕ) is large. As the energy, E, of the incident electrons increases, ϕ decreases so that at megavoltage energies the x rays are emitted predominantly in the forward direction. Consequently, a reflection target of the type used in lower-energy x-ray tubes is not used in a linear accelerator. Instead, a thin transmission target is used.

Flattening filter and scattering foil

The x-ray beam from a linac is frequently strongly *forward peaked*. The mean scattering angle subtended by the beam is related to the electron energy by the *Rossi–Griesan equation*:

$$(\phi) = \frac{15}{E_0}\sqrt{\frac{X}{X_0}}$$

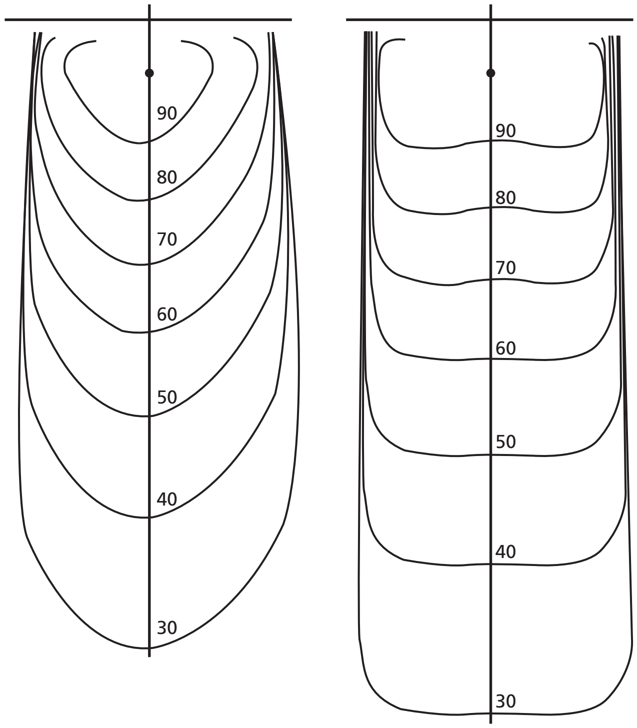

Figure 4-21 Isodose distributions for a 10 MV x-ray beam without (left) and with (right) the beam-flattening filter in place. Lateral horns of the distribution are apparent near the surface for the distribution obtained with the beam-flattening filter.

where ϕ is the mean scattering angle in steradians, E_0 is the energy in MeV of the incident electrons, X is the target thickness, and X_0 is the radiation length (a characteristic of a material, related to the energy loss of the electrons with the material). For example, a 15 MeV electron beam incident on a target whose thickness is equal to the radiation length generates a photon beam whose mean scattering angle is only 1 steradian.

A *flattening filter* is used to create a beam of sufficient area and uniformity for clinical use. The effect of a flattening filter on the *raw lobe* of a 10 MV beam is shown in Figure 4-21. Today's medical linac uses a flattening filter so that the x-ray beam reaching the patient is as uniform as possible. Treatment techniques recently introduced, such as intensity-modulated radiation therapy (IMRT), do not require a uniform beam and may allow the production of machines without flattening filters and the *flattening filter free* (FFF) mode is now used clinically. The raw lobe beams are starting to be used for very high output treatments, which is one of the main benefits of FFF mode. However, one should always use FFF beam in conjunction with CT-based treatment planning as the non-flat beams will produce significantly different dose distributions than conventional flat beams. A summary of the strengths and weaknesses of FFF beams is provided by Georg et al.[24]

The electron beam exiting from the accelerator waveguide or bending magnet is often no more than 1 or 2 mm in diameter. When the electron beam is to be used for treatment, a *scattering foil* is employed to provide a uniform beam of dimensions suitable for treatment. Modern scattering foils are of complex design, to scatter the beam without generating an unacceptable quantity of Bremsstrahlung radiation or degrading the beam energy by too great an amount.

Monitor ionization chamber

In contrast to ^{60}Co units, in which the source decays predictably to furnish a beam of slowly decreasing intensity, the dose rate of an accelerator beam may vary unpredictably or by design. Consequently, it is not possible to rely on the elapsed time to control the dose delivered to a patient. Instead the radiation leaving the target or scattering foil passes through a *monitor ionization chamber* (Figure 4-22), where it produces an ionization current that is proportional to the beam intensity. The ionization current is conducted to the control panel, where it is converted to a digital display of *monitor units* (MUs). The dose delivered to the patient is controlled by programming the accelerator to deliver a prescribed number of MUs.

Collimator

The final control over the beam, before it is delivered to the patient, is exerted by the collimator. In contrast to ^{60}Co sources, which are often several centimeters in diameter, the source of x rays in a linac is only 1 or 2 mm in diameter. As a result, the collimator can be of simpler design because the geometric penumbra is smaller. For x-ray beams, the collimator consists of jaws made of a high atomic number material, such as lead or tungsten. In most cases, the jaws are adjusted under motor control to create rectangular beams of almost any size. Most modern accelerators can deliver beams of up to 40×40 cm^2. A modern linear accelerator is shown in Figure 4-23.

Because electrons can be scattered easily by the intervening air between the scattering foil and the patient, the final stage of collimation of an electron beam must be close to the skin surface. An *electron applicator* or *cone*, such as those shown in Figure 4-24, is used to shape electron beams. Although the cone is rectangular, a slot is often provided to place a shaped insert to customize the field shape to the patient's target volume.

Treatment couch

To support the patient during treatment, a treatment couch is provided as part of a linear accelerator. The couch is usually mounted so that it pivots about an axis that passes through the gantry isocenter. Sometimes an eccentric axis of rotation is provided to provide greater convenience for patients or to permit extended lateral movement of the couch. Typical weight limits for a treatment couch are about 400 lb but can be greater for new robotic-style couches. One should always confirm with the vendor before loading a couch with more than 300 lb.

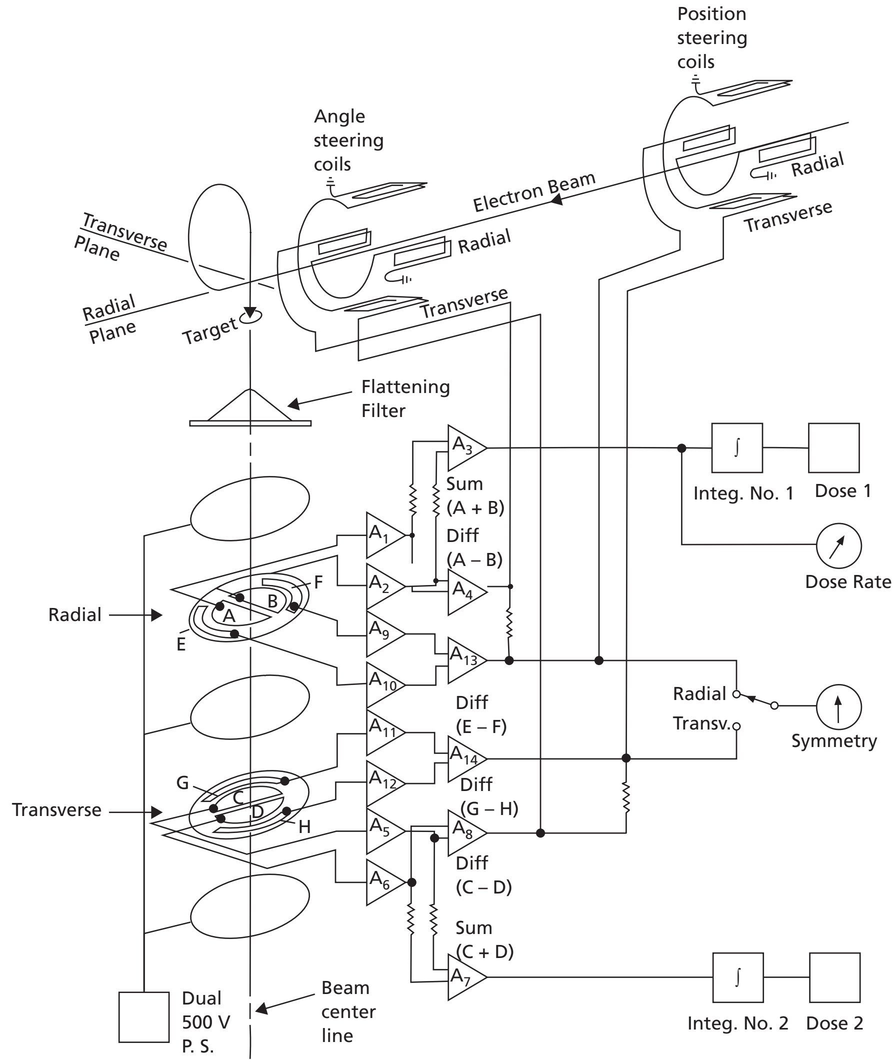

Figure 4-22 A modern multisegment monitor ion chamber showing the role of each segment in measuring the beam flatness and symmetry as well as the dose rate.
Source: Karzmark et al. 1993.[23]

Figure 4-23 A modern dual-energy linear accelerator for radiation therapy.

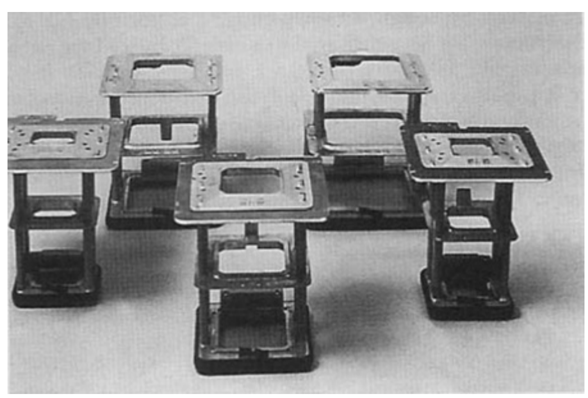

Figure 4-24 Representative electron applicators.
Source: Courtesy of Varian Associates, Inc.

Other medical accelerators

The first particle accelerator developed by Van de Graaff was described in 1931.[25] Van de Graaff generators used in research provide beams of positively charged particles with energies of 20 MeV and higher. In radiation therapy, Van de Graaff generators have been used to accelerate electrons to energies of up to 3 MeV but are no longer used.

Betatrons have largely been replaced in the clinic by linear accelerators, but for several decades after their introduction they enjoyed considerable popularity. However, presently, there are not any in clinical operation in the United States. The first betatron, constructed by Kerst in 1941, accelerated electrons to an energy of 2 MeV.[26] In later years, electrons and x rays with energies of up to 45 MeV were available from betatrons, but, primarily because of their relatively low dose rates, betatrons are no longer routinely used for patient treatments.

Cyclotrons

The first cyclotron, developed by Lawrence and Livingston in 1932, provided the background for modern orbital accelerators.[27] Electrons are not accelerated in cyclotrons, but, proton facilities almost exclusively use cyclotrons for accelerating protons. Cyclotrons are also used extensively to produce radioactive nuclides, including positron emitters that are useful in nuclear imaging and medical research.

The operation of a cyclotron is outlined in Figure 4-25. Two hollow, semicircular electrodes, or *dees*, are mounted between the poles of an electromagnet and separated from each other by a gap of 2–5 cm. The electromagnet is energized by direct current and furnishes a magnetic field of constant intensity across the dees. An alternating voltage applied to the dees oscillates with a frequency that is chosen with consideration for the intensity of the magnetic field and the type of particle being accelerated. For most cyclotrons available commercially, the frequency is between 10 and 40 MHz.

Positive ions (e.g., ^{1}H+, ^{2}H+, ^{3}He^2+, or ^{4}He^2+) are released by a cathode-arc source in the gap between the dees and are accelerated in bunches toward the negative dee. After the ions enter the dee, they are shielded from the electric field. The magnetic field forces the particles into a circular path, which the particles follow with constant speed. Just as the polarity reverses across the dees, the ions emerge from the first dee and accelerate across the gap toward the second dee. The ions follow a circular orbit in the second dee and emerge as the polarity reverses again. In this manner, the particles are accelerated each time they cross the dee aperture until they attain the desired energy.

The radius of the orbit followed by ions within a dee is described by:

$$r = \frac{mv}{Bq}$$

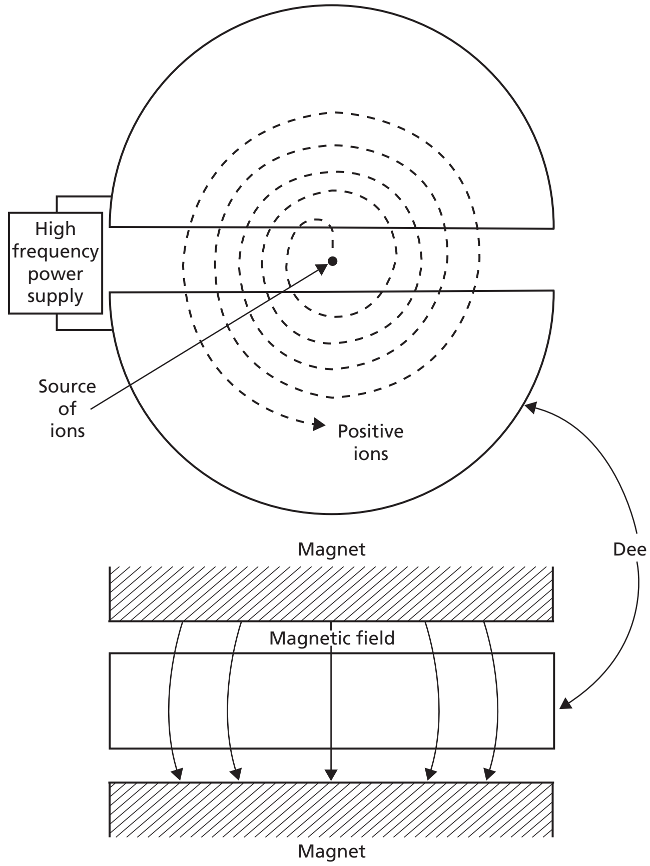

Figure 4-25 Conventional two-dee cyclotron. The path of positively charged particles is denoted by the *dashed curve*. The particles are accelerated each time they cross the gap between the dees.

where r is the radius, m is the mass of the accelerated particles, v is the velocity of the particles, B is the magnetic field intensity, and q is the charge of the particles. This equation is identical to the equation that describes the equilibrium orbit of electrons in a betatron and in a bending magnet. If the magnetic field intensity, B, and the mass, m, are constant, the radius of the orbit increases linearly with the velocity of the accelerated particles.

The time, T, for a half-revolution of the particles in a dee is:

$$T = \frac{\pi m}{Bq}$$

If the mass of the particles remains constant, the time for a half-revolution is also constant, and the emergence of the particles from the dees is synchronized easily with changes in the dee polarity. However, the relativistic mass, m, of a particle moving with velocity, v, is:

$$m = \frac{m_0}{\sqrt{1 - \frac{v^2}{c^2}}}$$

When the velocity of particles in a cyclotron reaches about 0.2c, the increased mass of the particles disturbs the synchronization between the emergence of particles from the dees and the changing polarity across the dees. For example, deuterons may be accelerated in a cyclotron to a maximum energy of about 35 MeV. A deuteron is a combination of a proton and a neutron. Above 35 MeV, the relativistic increase in mass causes the deuterons to emerge from the dees out of phase with the changing polarity of the dees. Electrons reach the limiting velocity of $0.2c$ when their kinetic energy is only about 10 keV. Consequently, electrons are not accelerated in cyclotrons.

To maintain the synchronization between particle emergence from the dees and changing polarity of the dees, the frequency of the alternating voltage applied to the dees may be reduced as the particles gain energy. This approach is used in the synchrocyclotron, and deuterons have been accelerated to energies of up to 200 MeV in these machines.

For experimental studies, a beam of heavy particles may be extracted from a cyclotron. Additionally, radioactive nuclides may be produced in a cyclotron by directing accelerated particles onto a target. An increasing number of medical institutions are using a cyclotron to produce short-lived, positron-emitting radioactive nuclides (^{11}C, ^{13}N, ^{15}O, ^{18}F) useful in nuclear medicine. These nuclides are used in combination with *positron emission tomography* (PET).

Microtrons

With the availability of microwave accelerator cavities came the development of another device for accelerating electrons in circular or elliptical orbits. This device, called the *microtron*, combines the static magnetic field of the cyclotron with the accelerating cavity of the linear accelerator.[28] In contrast to the linear accelerator, electrons pass through the accelerating cavity multiple times, gaining energy each time. The path of the electron bunch is bent by the static magnetic field in a circular or racetrack shape, bringing the electrons back to the cavity.

As electrons reach velocities close to the speed of light, they can be considered to be traveling at constant velocity during the entire acceleration process. As their energy increases, however, their momentum increases proportionately, and their bending radius increases as well.

In a circular microtron, the electron's energy increases in equal increments, and therefore the circumference of the path increases by corresponding increments. Consequently, the electron bunch arrives at the accelerating cavity at the correct moment to be accelerated again.

The maximum energy of electrons in a microtron is essentially limited only by the dimensions and strength of the static magnetic field. In practice, microtrons can accelerate electrons to energies of up to 50 MeV.[28] Difficulties with the stability of operation have made microtrons unreliable in the clinic, and pursuit of their clinical application has been discontinued.