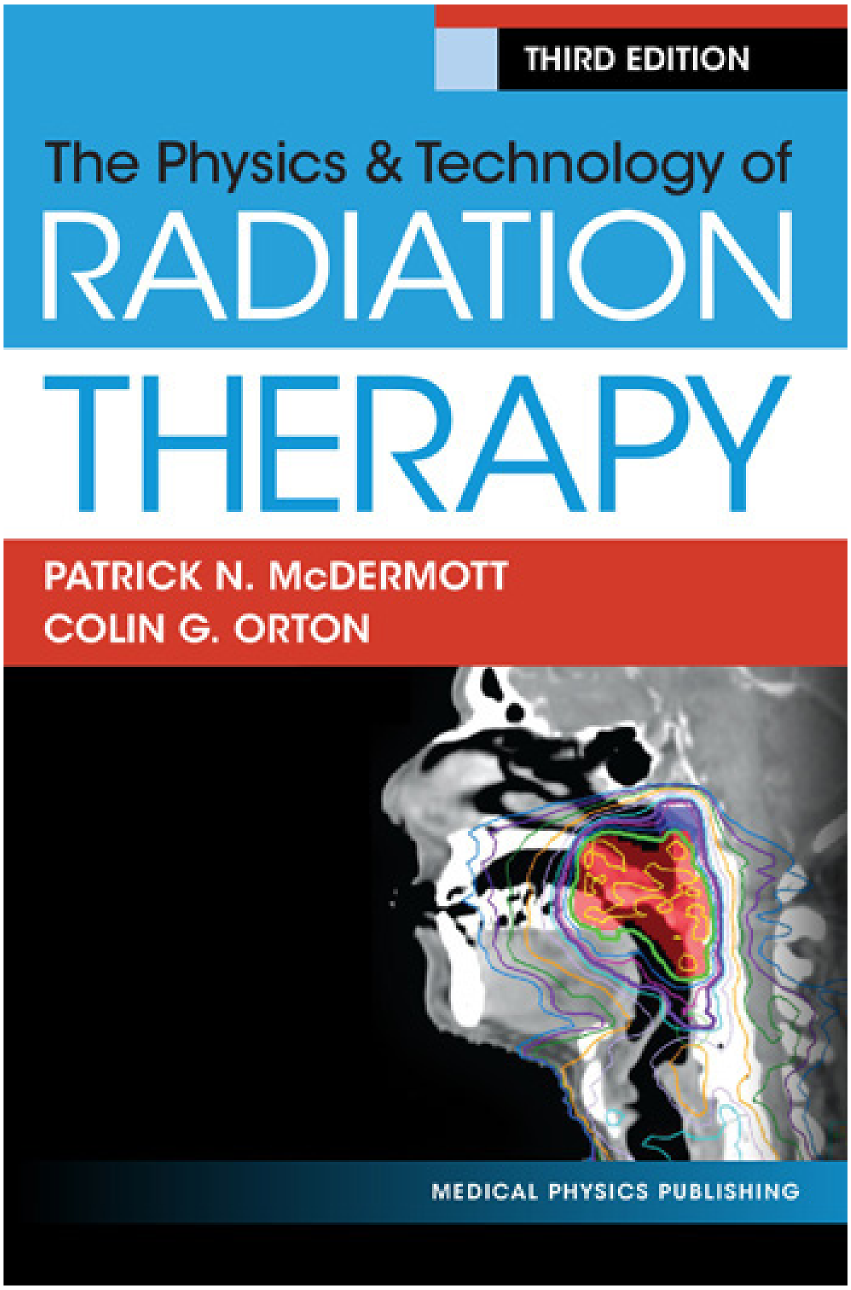

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9 Medical Linear Accelerators

9.1 Introduction

Throughout the history of radiation therapy, a variety of machines have been used to produce beams of radiation. Radiation therapy delivered with an external beam is sometimes referred to as *teletherapy*. There are two major classes of external beam treatment units: those that use radioactive isotopes and those machines called accelerators, which employ electric fields to accelerate charged particles. We will confine our discussion here to megavoltage (MV) beams and, therefore, we will not discuss superficial or orthovoltage x-ray units in this chapter (see chapter 4). Isotope machines have used cesium-137 (Cs-137) and cobalt-60 (Co-60). Cesium is not used any more. Co-60 units have almost completely disappeared, at least in the United States. Co-60 radiation has a relatively low penetrating power, a large penumbra, and a low dose rate. Specialized external beam units—such as robotic linacs and gamma stereotactic units—are discussed in chapter 20. The linacs discussed in this chapter, and shown in Figures 9.1–9.3, are sometimes referred to as “C-arm” linacs. The reason for this terminology is because, if you look at the gantry from the side, the shape vaguely resembles the letter “C”. Quality assurance tests for linear accelerators are discussed in chapter 21. Imaging devices attached to linacs are covered in chapter 19.

To obtain the benefits of deeply penetrating radiation and skin sparing, megavoltage photons are required. Such photons can be produced by accelerating electrons to high energy and directing them against a metallic target. As the electrons lose energy in the target, they produce x-rays (as well as heat) via the bremsstrahlung mechanism. For some types of therapeutic treatment, it is desirable to use the high-energy electron beam directly. In this case, the metal target is removed, and the electron beam is allowed to enter the patient.

There are two main types of accelerators: those that accelerate charged particles in a straight line—called *linear accelerators* or *linacs* (see Figures 9.1 and 9.2)—and those that accelerate them in a circular or approximately circular fashion. There are a variety of circular machines: microtrons, cyclotrons, synchrotrons, and betatrons. Circular accelerators for radiation therapy are relatively rare and are found in only a handful of centers. Cyclotrons and synchrotrons are discussed in chapter 22 in the context of proton therapy.

Linear accelerators were developed after World War II for research in high-energy elementary particle physics and later adapted to medical use. Medical electron linear accelerators were first introduced in Britain in the 1950s. They first appeared in the United States in the 1960s, and their widespread use began in the 1970s. They have almost completely replaced every other type of external beam treatment machine. It is estimated that there are about 4500 medical linacs in the United States. There are now only two manufacturers of medical electron linear accelerators: Varian and Elekta (formerly Philips; see Figures 9.1, 9.2, and 9.3).

An electron linear accelerator accelerates electrons along the length of an evacuated tube (called a *waveguide*) to almost the speed of light. As an example of this, if electrons are accelerated through the equivalent of 20 million volts (20 MV), they will travel at a speed that is within 0.03% of the speed of light! The electrons are accelerated by microwaves that travel down the waveguide. Microwaves are electromagnetic waves, and it is the electric field associated with these waves that accelerates the electrons (see section 2.4).

The electron beam can be used directly to treat patients, or it can be aimed at a metallic “target” (see Figure 9.4). The target absorbs the energy of the electrons and converts some of it to very energetic, highly penetrating x-rays via the bremsstrahlung process discussed in chapter 5. The radiation passes through the “collimator,” whose field-defining jaws determine the cross-sectional size of the beam (see Figure 9.4). Modern linacs allow selection of two or three photon beam energies and five or six electron beam energies.

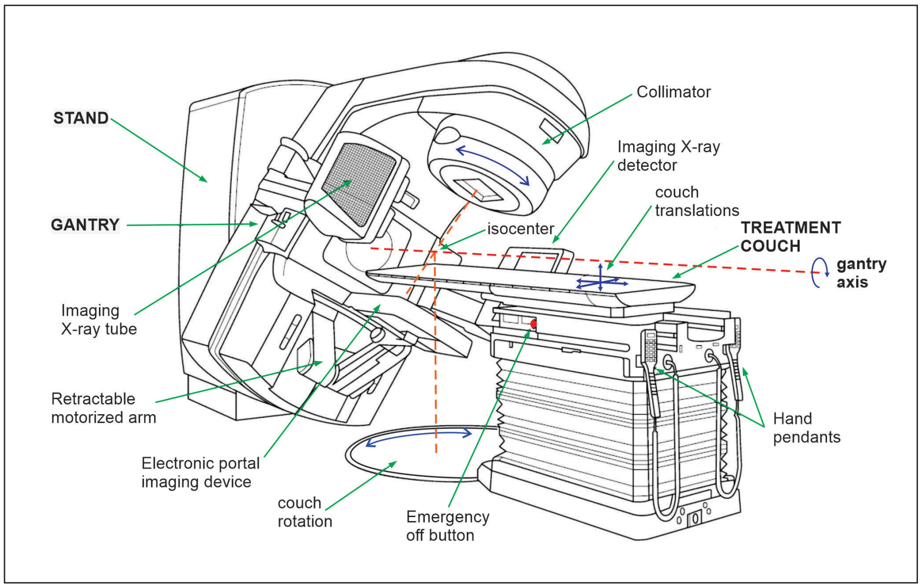


Figure 9.1 A linear accelerator with a gantry stand. The major parts of the linac are the gantry, the gantry stand, and the couch. This diagram illustrates the large variety of mechanical motions possible. Motion is controlled by use of the hand controls, which are sometimes called hand pendants. The radiation beam is directed along the beam central axis (not labeled). The gantry can rotate around an axis (labeled gantry axis), thus allowing rotation around the patient on the couch. The collimator can rotate around the beam central axis. The treatment couch or table can move up or down, in or out toward the gantry (longitudinal motion), or from side to side (lateral motion). The couch can also rotate around a vertical axis. The axes of rotation of the couch, the gantry, and the beam central axis meet at a point in space called the isocenter. (Adapted from "Technical specifications of radiotherapy equipment for cancer treatment." Geneva: World Health Organization; 2021. Licence: CC BY-NC-SA 3.0 IGO.)

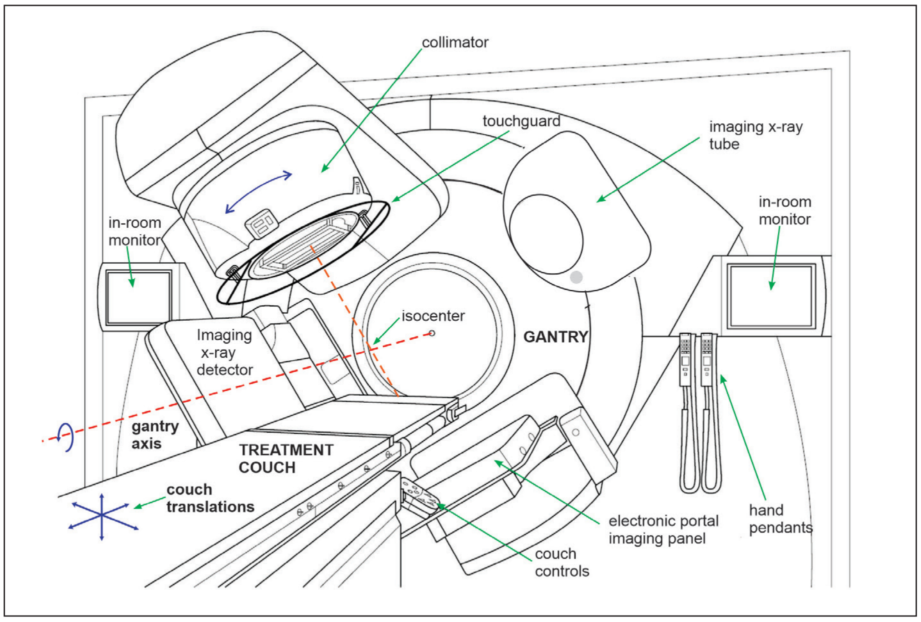


Figure 9.2 A linac with a drum gantry (no gantry stand). The mechanical motions are the same as the linac with a gantry stand as shown in Figure 9.1. (Adapted from World Health Organization, 2021.)

Figure 9.3 Varian (left) and Elekta (right) linear accelerators. Notice that the Elekta linac does not have a gantry stand. Compare this figure with Figures 9.1 and 9.2. (Courtesy of Varian Medical Systems and Elekta AB).

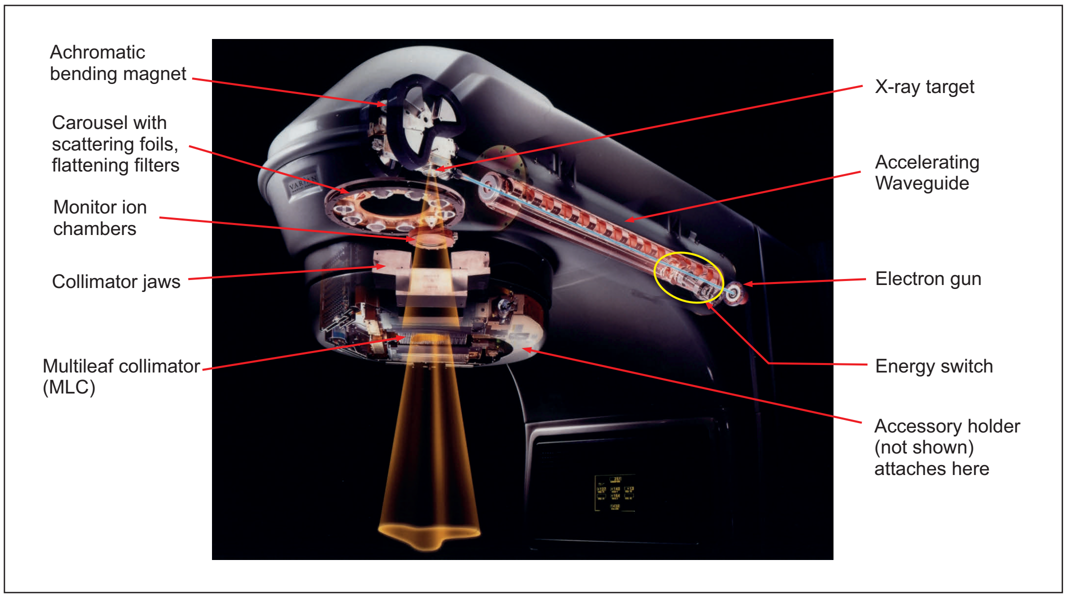


Figure 9.4 The waveguide and the treatment head of a modern dual photon energy linear accelerator. Electrons injected by the electron gun are accelerated down the waveguide. An electromagnet at the end of the waveguide deflects the electron beam downward so that it strikes the x-ray target. The collimator jaws and the MLC define the cross-sectional shape of the beam. Not shown are lead plates, positioned in the treatment head, that are used to shield against stray “leakage” radiation. (Image courtesy of Varian Medical Systems.)

The major components of the linac are the gantry, the gantry stand, the treatment couch, and the treatment console. The console is outside the treatment room and is used to control the machine during the time that the beam is on. The gantry and the couch can move in numerous directions, allowing radiation beams to enter a patient from almost any angle (see Figures 9.1 and 9.2).

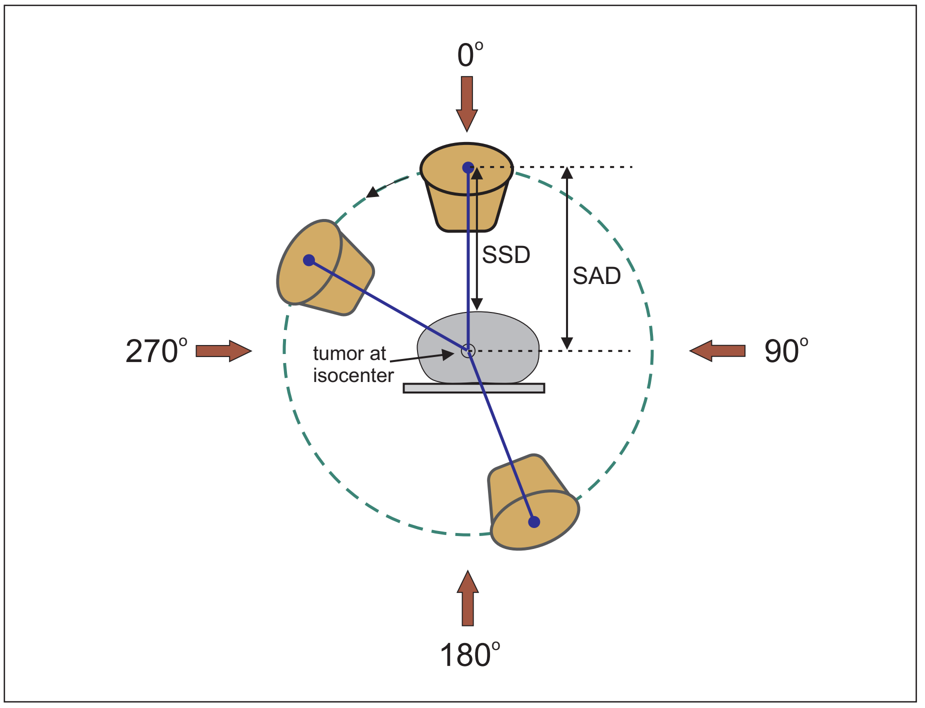


Figure 9.5 Gantry rotation around the patient. Patients are often positioned so that the isocenter is at or near the center of the volume to be treated. This is called an SAD or isocentric treatment. As the gantry rotates, the beam always points at the isocenter. It is possible to treat through the table. The SSD is the distance from the source to the patient surface. The SSD changes with changing gantry angle. The IEC (International Electrotechnical Commission) scale is shown for gantry angles (IEC 61217 & IEC 60601-2-1) as seen facing the gantry. The arrows show the beam direction at each of the cardinal angles.

The gantry can rotate around the patient. The rotation axes of the gantry, the collimator (same as beam central axis), and the couch meet at a common point in space called the *isocenter*.

Digital position indicators show the gantry and collimator angle, as well as the collimator jaw settings. In Figure 9.2 these would be displayed on the in-room monitor.

The source-to-axis distance, or SAD, is the distance from the source of radiation to the isocenter. In all modern linacs this distance is 100 cm. The SAD always remains the same; it is not adjustable.

The couch top is made of relatively thin carbon fiber for patient support. When the gantry head is underneath the table, it is possible to treat through the carbon fiber couch top.

There is extensive safety circuitry to ensure that a linac is not run in a dangerous configuration. If the status of the linac is not safe for the machine settings, then an “interlock” will prevent the beam from turning on.

Figure 9.5 shows a so-called SAD patient treatment. The isocenter is positioned at or near the center of the patient’s tumor. With this arrangement, the beam is always pointed directly at the tumor for all gantry angles. For a given gantry angle, the distance from the radiation source to the patient skin surface is called the source-to-surface distance, or SSD. The SSD will change when the gantry is rotated to a new angle. Figure 9.5 also shows the IEC angulation scale (IEC 61217 and IEC 60601-2-1) for gantry angle.

9.2 Accelerating Waveguides

When electrons (charge Q) are accelerated through a potential difference V, they acquire a kinetic energy $T = Q\,V$ (this was discussed in chapter 2). If electrons are accelerated through a potential difference of 1 million volts, they will acquire a kinetic energy of 1 MeV. As an electron penetrates the x-ray target, it undergoes bremsstrahlung interactions in which photons are radiated. The electrons lose an amount of energy equal to the energy of the photon produced. Usually, an electron will only lose a fraction of its kinetic energy in any single interaction. The photons that emerge from the target will, therefore, have a range of energies from close to zero up to a maximum value that is equal to the initial kinetic energy of the electron. As an example, if the electrons are accelerated to an energy of 6 MeV, the *maximum* photon energy produced in the target will be 6 MeV. The photons will have a broad range of energies from 0 to 6 MeV. It is, therefore, not correct to refer to the emerging x-ray beam as a 6 MeV beam. In fact, the average energy of the photons in such a beam is approximately 6/3 MeV = 2 MeV. The nomenclature that is used to refer to such a beam is based on the fact that the electrons were effectively accelerated through a potential difference of 6 million volts or 6 MV. The proper way to refer to the "quality" or energy of this x-ray beam is to describe it as a 6 MV beam. The average energy of the photons in a linear accelerator beam in MeV is approximately numerically equal to MV/3. The electron beam itself is nearly monoenergetic, with an energy of 6 MeV. It is, therefore, correct to refer to the electron beam as a 6 MeV beam.

Common x-ray beam energies for medical linear accelerators range from 4 MV to 18 MV. Most linacs have dual photon energies; 6 MV and 15 MV or 6 MV and 18 MV are common. Three photon energies are possible. Linacs also have multiple electron energies, ranging from as low as 4 MeV up to 22 MeV. A modern, dual-energy linac costs about $3 million, depending on options.

Let us think about how we might build an accelerator that would accelerate electrons to an energy of 10 MeV. Our first naive thought might be to take a pair of copper plates or disks and place a potential difference across them of 10 million volts (see Figure 9.6a). We will place a small hole in the positive electrode so that the electrons may exit through it. When an electron is placed between the plates it will be accelerated through a potential difference of 10 million volts and, thus, gain kinetic energy of 10 MeV. This would work in principle, but unfortunately it is not practical. Such large potential differences are very difficult to create. Let us imagine that we have a dial that will allow us to turn up the voltage. As we turn up the voltage and reach potential differences of thousands or tens of thousands of volts, the charge on the plates will begin to leak off into the surrounding air, frustrating our effort to increase the voltage. Eventually, when the potential difference becomes sufficiently large, there will be a spark that will reduce the potential difference to zero.

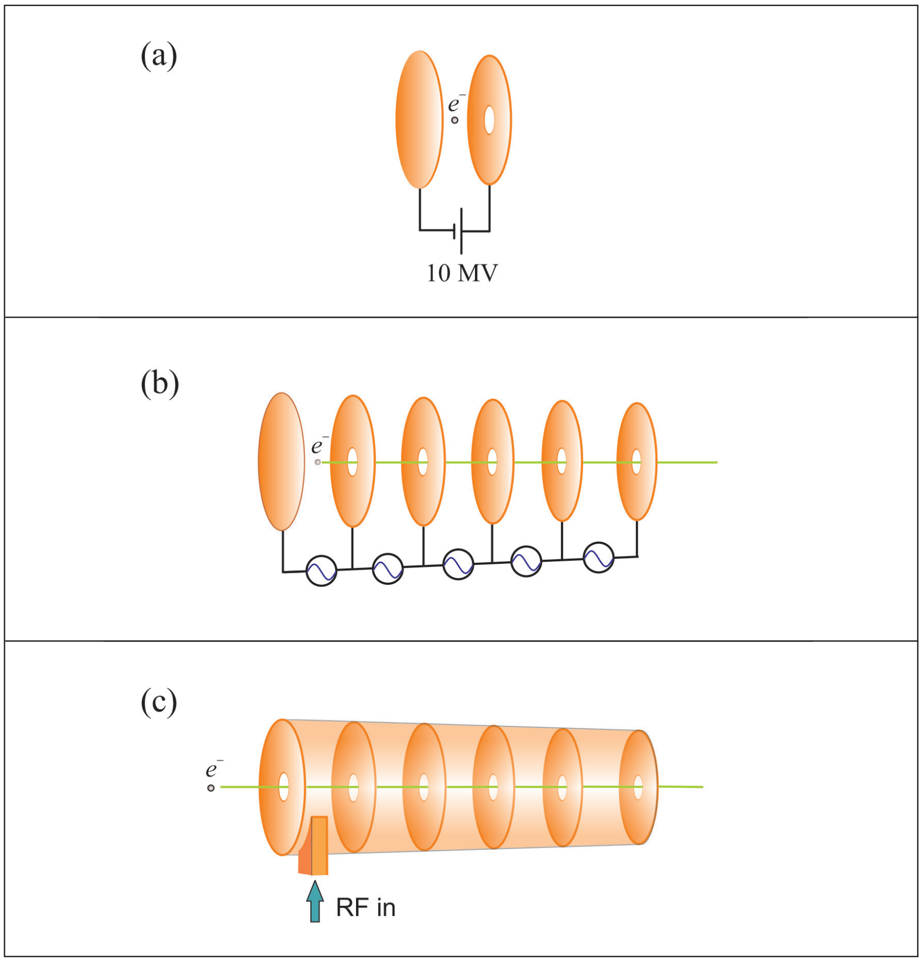


Figure 9.6 Progression of ideas for accelerating electrons. (a) Two copper plates with a potential difference of 10 million volts between the anode and the cathode. This method of accelerating electrons to high energy is not practical for reasons explained in the text. (b) A more sophisticated approach using a series of plates with a smaller potential difference between adjacent plates. The applied voltage must be oscillatory so that when an electron is in the gap between any two adjacent plates, the plate on the left always has negative polarity—ensuring that the electron is always accelerated toward the right. (c) Place the arrangement in (b) inside a cylindrical pipe, remove all the air, and replace the oscillators with microwave RF input; thus we have an accelerating waveguide. See the text for more detail.

We might now be a little more clever. If we cannot produce a potential difference that is large enough between two plates, how about using a series of plates, each with a relatively small potential difference between them? Instead of accelerating the electron once through a large potential difference, let us accelerate it many times through a small potential difference. This is the "trick" that is used in all high-energy accelerators.

Instead of a single set of plates, let us use a series of plates, each with a more modest potential difference between them. This is illustrated in Figure 9.6b. We now need an oscillating potential difference between the plates to ensure that whenever an electron is between a specific set of two plates, the plate on the left is negative and the one on the right is positive. The sign of the potential difference must switch completely during the time that the electron travels from the center of the gap between one set of plates and the next adjacent set of plates.

We want to accelerate the electron to 10 MeV. We know that by the time the electron gains about 0.5 MeV (rest mass of the electron) it will be highly relativistic (see section 2.5). This will happen quickly in the so-called buncher section of the linac, perhaps after traversing a few plates. Let us ignore the problem of initial acceleration for now and assume that the electrons are already relativistic. This means that they are traveling at close to the speed of light. Under these circumstances, the speed of an electron hardly increases as the energy increases. Let us estimate how rapidly the polarity of the potential difference between adjacent plates must switch back and forth. Assume that the distance between plates is 5.0 cm. If the electrons are traveling at close to the speed of light, they will travel from the center of the gap between one set of plates to the center between the adjacent set of plates in a time equal to $(0.05\text{m}) / (3 \times 10^{8}\text{m/s}) = 1.66 \times 10^{-10}\text{s}$. The voltage between the plates must return to its original polarity every 3.33×10^{-10}s. Let us compute the frequency that corresponds to this: $\nu = 1 / (3.33 \times 10^{-10}\text{s}) = 3 \times 10^{9}$ Hz or 3000 MHz = 3 GHz. This frequency is in the "S-band" microwave portion of the electromagnetic spectrum (see section 2.4). The microwave power is sometimes referred to as RF (radio frequency).

In Figure 9.6b we show wires carrying the oscillatory potential difference to the plates. The frequency of the oscillatory voltage is very high, and this will cause the wires to act like a broadcast antenna and radiate microwaves. Instead of using wires, we shall use a rectangular copper pipe called a *transmission waveguide* to bring the microwave power to the accelerating structure. In addition, we need to accelerate the electrons in a vacuum, otherwise they will collide with molecules of oxygen and nitrogen in the air, thus interfering with acceleration. We can place the disks shown in Figure 9.6b in a pipe (see Figure 9.6c) and use a vacuum pump to evacuate all the air. This pipe with the disks in it is called an *accelerating waveguide*. Do not confuse the rectangular transmission waveguide that carries microwave power with the cylindrical accelerating waveguide in which the electrons are accelerated.

The accelerator waveguide is a highly evacuated copper "pipe" in which electrons are accelerated. This is one of the most expensive components of a linear accelerator. Its replacement cost is on the order of $200,000. There are two major types of accelerating waveguides: (1) traveling wave and (2) standing wave. Both of these designs use electromagnetic waves to accelerate electrons. As we have seen, these waves are in the microwave region of the spectrum: $\nu \approx 3000$ MHz (S-band radar) for conventional medical linacs (wavelength $\lambda = 10$ cm in free space). In both designs, electrons are accelerated in bunches and, therefore, the radiation output is pulsed. The pipe must be under high vacuum so that the electrons do not collide with air molecules and lose energy.

Electrons travel in bunches down the waveguide. In a traveling wave linac, the electrons "surf" on a traveling electromagnetic wave (see Figure 9.7). The

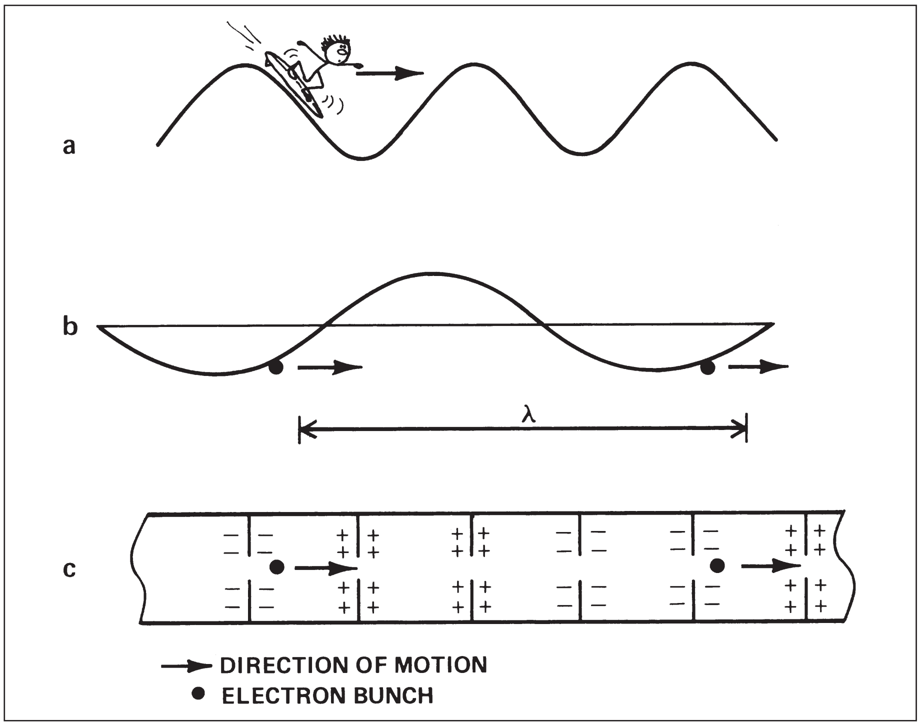


Figure 9.7 Traveling wave acceleration. In (a) a surfer is pushed along by a wave. In a similar way, electron bunches (b) are pushed along by an electromagnetic wave in a traveling wave accelerator waveguide. A snapshot of a waveguide is shown in (c) illustrating the charge distribution inside the waveguide at that instant which accelerates the electron bunches shown toward the right. (Reprinted from Karzmark, C. J., and R. Morton, *A Primer on Theory and Operation of Linear Accelerators in Radiation Therapy*, 3rd Edition, Fig. 28. © 2017, with permission from Medical Physics Publishing.)

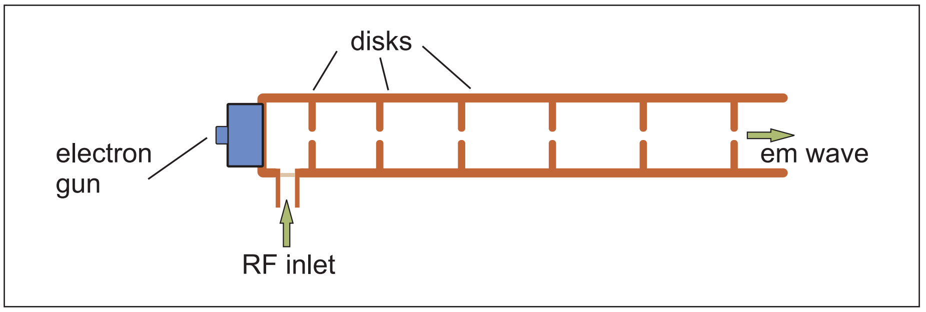


Figure 9.8 A cross section of an accelerating waveguide for a traveling wave linear accelerator. The waveguide is disk loaded to slow down the microwave electromagnetic waves so that bunches of electrons may "surf" on these waves down the guide from left to right. The electron gun injects electrons into the waveguide.

electrons must travel at the same speed as the electromagnetic wave to "surf" along. In a vacuum with no conductors nearby, electromagnetic waves travel at speed c; electrons are prohibited from traveling at this speed by the special theory of relativity. Unless the waves are slowed down, they will roll over the electrons. The waves are slowed down by the disks in the waveguide. Such a waveguide is called a "disk loaded" waveguide (see Figure 9.8).

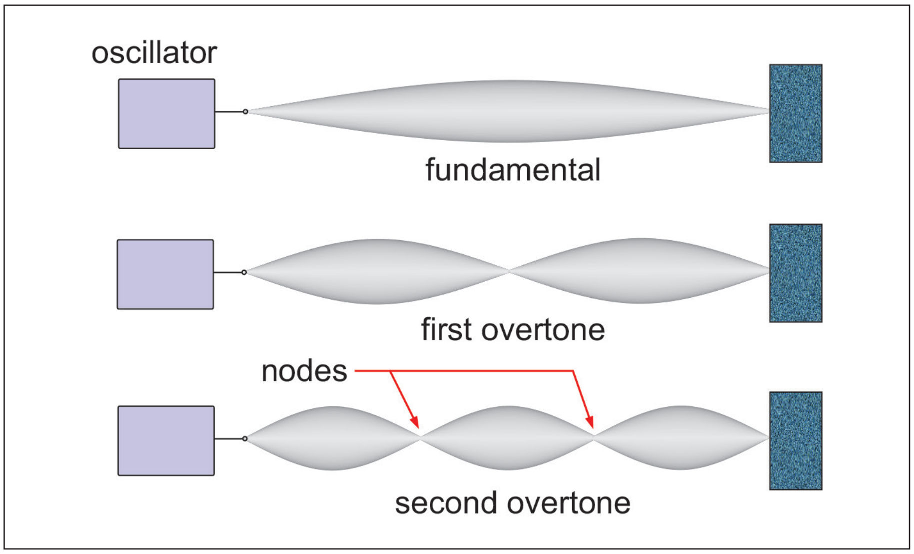


Figure 9.9 Time-lapse image of standing waves on a string. The string is tied to the wall on the right. The "fundamental" vibration is shown at the top. Increasing the frequency of the oscillator produces successive overtones.

Toward the beginning of the waveguide, the disks are spaced closer together. The disks are spaced equally apart once the electrons reach (nearly) the speed of light. The effect of this is to increase the speed of the electromagnetic traveling wave. If this is done in just the right way, the electron bunches will be able to continue to surf on the wave as they speed up and, in this manner, they will gain energy.

The other major type of waveguide is found in a standing wave linac. You can produce standing waves on a string by tying it to a nail on a wall, pulling the string taut, and then plucking it. Such waves are produced in the strings of musical instruments. Standing waves are illustrated in Figure 9.9. A standing wave is formed when a traveling wave moving down the string arrives at the wall and is then reflected back. This leads to two traveling waves moving in opposite directions. The two traveling waves add to produce a standing wave.

How can a standing wave accelerate electrons? The electrons cannot surf along the wave. The acceleration process is illustrated in Figure 9.10, which shows snapshots of a standing wave. This figure shows a graph of the electric field as a function of position at three instants in time. The electric field is proportional to the height of the graph above the horizontal axis. If the force on electron bunches due to this electric field always acts in the same direction, then the electrons will gain kinetic energy as they move down the waveguide. In the first instant shown, the force on the electron bunch is toward the right. In the middle instant, the electric field has momentarily become zero. There is no force on the electron bunch, but it will continue to move toward the right owing to its velocity in that direction. In the third frame, the electric field has returned, and in the meantime the electrons have moved to a new position; however, the electrons are in the same relative position with respect to the wave as they were in the first instant and, therefore, they again experience a

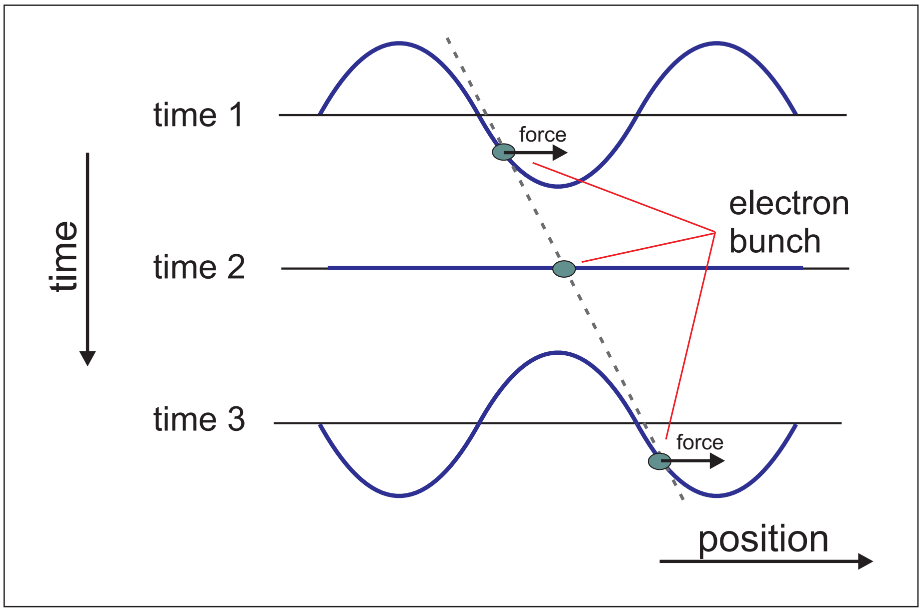


Figure 9.10 An electron bunch undergoing acceleration by a standing wave in a waveguide. The height of the wave represents the strength of the electric field. This shows three different instants in time starting at the top. At "time 1" the electric field seen by the negatively charged electron bunch is negative and, therefore, there is a force toward the right. At "time 2" the electric field is momentarily zero; it then reverses at the bottom. As the electric field oscillates in strength, the electron bunch will experience a force that is constantly toward the right.

force toward the right. If the electron bunch can maintain its relative position with respect to the standing wave, it will always be accelerated forward.

Among the two major linac manufacturers, Varian machines are standing wave linacs and Elekta uses a traveling wave. The microwave power is introduced through the side of the waveguide (see Figure 9.11). Standing wave linacs require a device called a circulator, which prevents microwaves from reflecting back into the klystron (discussed later in this chapter). Traveling wave linacs require a "terminating" or "dummy" load to absorb the residual microwave energy. It is necessary to prevent a backward-reflected wave to avoid a standing wave from forming. For some Elekta linacs, the residual microwave power is fed back (called RF feedback) to the input to increase efficiency. In a standing wave linac, every other waveguide cavity can be moved off to the side, as shown in Figure 9.11. This is called *side cavity coupling*. The side cavities serve only to couple electromagnetic energy from one centerline cavity to the next. This reduces the length of the waveguide by almost a factor of two, a big advantage. The electrons gain roughly 2 MeV energy per cavity.

The orientation of the accelerating waveguide of a conventional linac depends on the design energy. This is illustrated in Figure 9.12. For low, single-energy standing wave linacs—in the range 4 to 6 MV—the waveguide is short enough (perhaps 30 cm) so that it can be oriented vertically. For intermediate-energy standing wave linacs, in the range 6 to 25 MV, the waveguide is too long to orient vertically. Instead, the waveguide is oriented horizontally and a "bending magnet" is employed to "bend" or deflect the electron

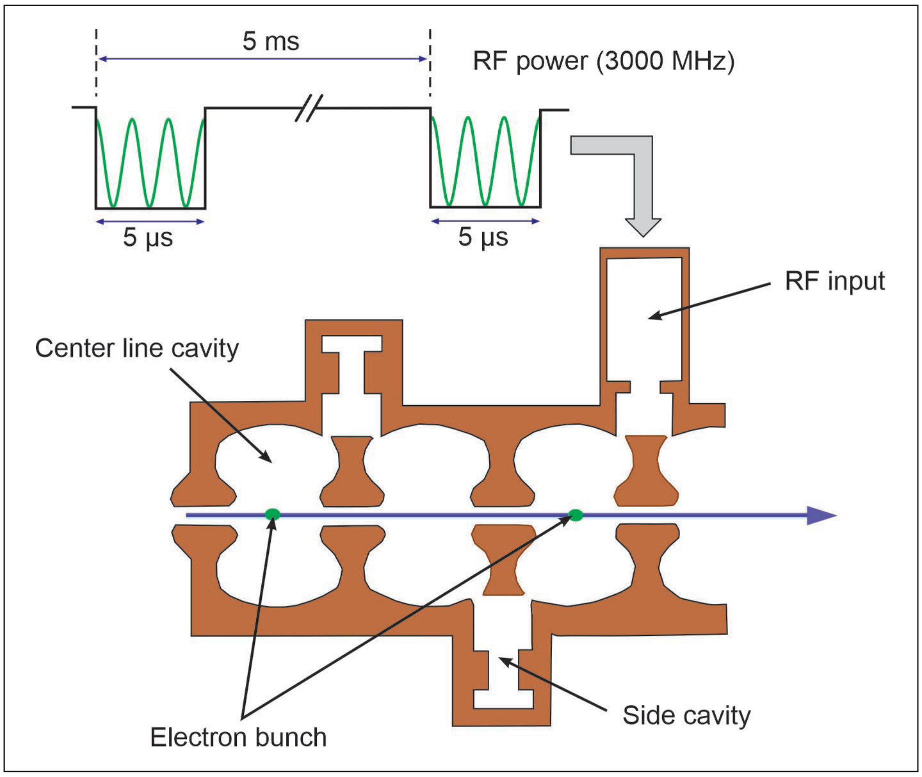


Figure 9.11 A cross-sectional view of a side-coupled standing wave accelerating waveguide. The length of the waveguide is reduced by a factor of two by introducing side cavities. The side cavities couple electromagnetic power from one center line cavity to the next. The center line cavities have been shaped to reduce power losses in the copper walls. The microwave radiofrequency (RF) power is fed into the waveguide through the RF input in pulses. Each pulse is about 5 microseconds in duration and pulses repeat approximately every 5 milliseconds. The electrons are accelerated in bunches.

beam in a downward direction toward the patient. In the highest-energy standing wave waveguides (25 to 35 MeV) or in traveling wave waveguides, the guide may need to be oblique. A dual-energy traveling wave waveguide is about 2.5 m in length. Dual-energy standing wave waveguides are approximately 1.3 m in length. A long waveguide could be oriented vertically, but this would require both a tall ceiling and a pit in the floor so that the gantry could rotate underneath the patient couch.

9.3 Bending Magnets

The bending electromagnet in a linac must change the direction of the electron beam from horizontal or oblique to vertical (see Figure 9.12 and 9.13). The bending magnet is made from coils of wire that must be supplied with a high current to produce the magnetic field necessary to deflect the electron beam. Standing wave linacs use an “achromatic” bending magnet that deflects the electron beam through 270° rather than 90° (see Figure 9.13). When the electrons enter the region between the bending magnet poles, they have a small spread in energy. If a 90° bending magnet were used, the paths of electrons

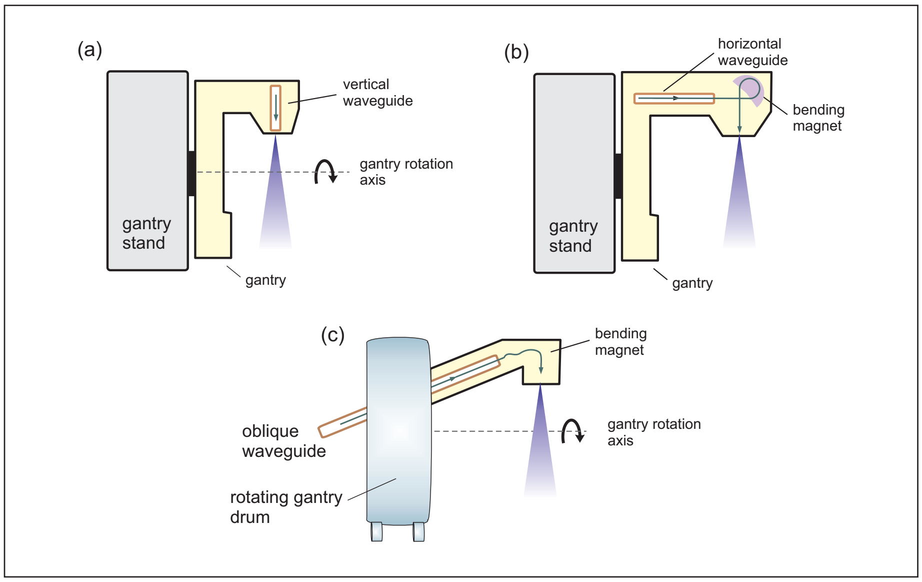


Figure 9.12 Orientation of the accelerating waveguide in linacs of different types. In standing wave low single-energy machines (4 to 6 MV), the waveguide is short enough that it can be mounted vertically as in (a). In higher-energy machines, the waveguide is too long to be mounted vertically and instead is mounted horizontally. This requires a bending magnet to redirect the beam down toward the patient. The addition of the bending magnet makes the linac considerably more complex. Most modern, dual-energy standing wave linacs are oriented as in (b). Traveling wave linacs and the highest-energy standing wave linacs require an oblique waveguide as in (c).

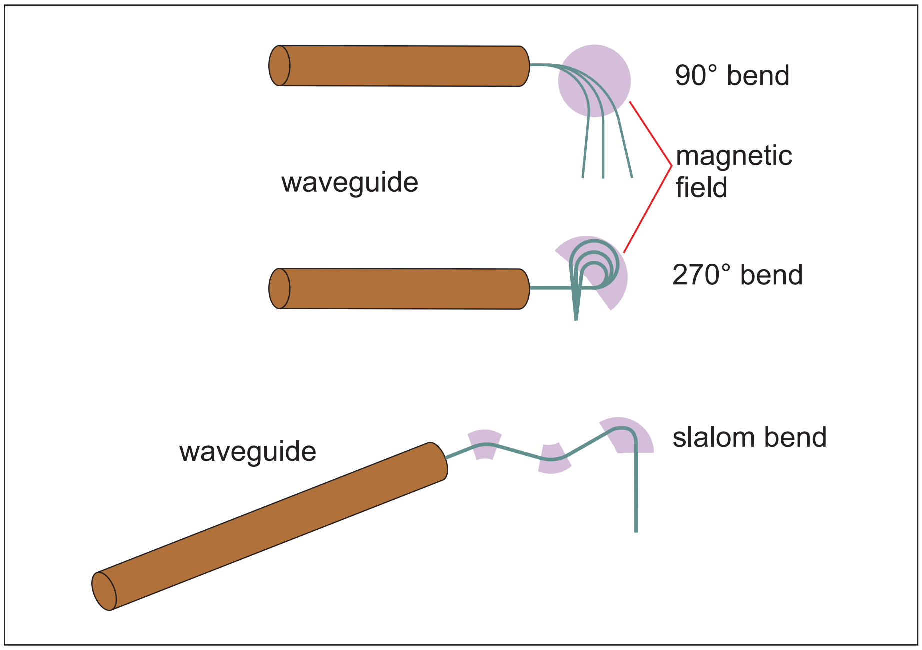


Figure 9.13 Bending magnet arrangements. The magnetic field is confined to the purple-shaded areas. Electrons emerge from the accelerating waveguide with a slight distribution in energy. If a 90° bending magnet is used, the electrons will be spread out by the magnetic field. The path of more energetic electrons will bend less than those with lower energy. By using a 270° bending magnet, all the electrons can be made to come together at the target. An actual achromatic bending magnet system is somewhat more complicated than this. Elekta linacs use a slalom bending magnet system like the one shown at the bottom of the diagram.

with slightly different energies would diverge. A 270° bending magnet causes all the electrons with various energies to converge at the focal spot. This is the arrangement employed in Varian linacs; Elekta machines use a more elaborate "slalom" design. When the selection of the beam energy is changed, the bending magnet current must change—this takes a short while.

9.4 Sources of Microwave Power

Linear accelerators require high-power microwaves to accelerate electrons. There are two different devices that can supply the necessary power: magnetrons and klystrons.

A magnetron generates high-power microwaves (see Figure 9.14). You may even own one without realizing it. A magnetron supplies the microwaves for your microwave oven. Your home oven operates at a frequency of 2450 MHz and is capable of producing about 1 kW of power. By comparison, the magnetron in a linac produces pulses of microwaves with a frequency of approximately 3000 MHz and peak power of about 2.5 to 5.0 MW. This is why a magnetron for a linac costs about $30,000. The magnetron was invented during World War II, and it is widely used for radar applications (see the box at the end of this chapter entitled "The Invention of the Cavity Magnetron"). The lifetime of a magnetron is about 2000 hours of operation (perhaps two to four years).

In a magnetron, electrons move past cavities (see Figure 9.15) that have a resonant or natural frequency in the microwave part of the spectrum. The electrons induce electromagnetic oscillations in the cavities. This is somewhat like blowing air over the top of a soda bottle to produce a loud sound. If conditions are right, the airflow over the top of the bottle induces oscillations of the air inside the bottle, producing a loud sound.

While a magnetron generates high-power microwaves, a klystron amplifies them. Today, klystrons are capable of producing higher power levels than magnetrons: up to about 8 MW peak power. Magnetrons are used in

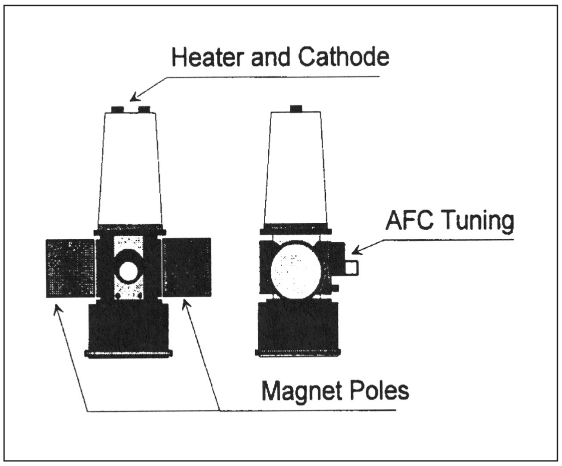


Figure 9.14
Two views of the external appearance of a magnetron. The magnets supply the magnetic field in which the electrons spiral. The AFC (automatic frequency control) plunger continually adjusts the frequency of the microwaves to maintain optimum accelerating conditions for the waveguide. The microwaves emerge from the bottom. (Courtesy of Siemens Medical Solutions USA, Inc.)

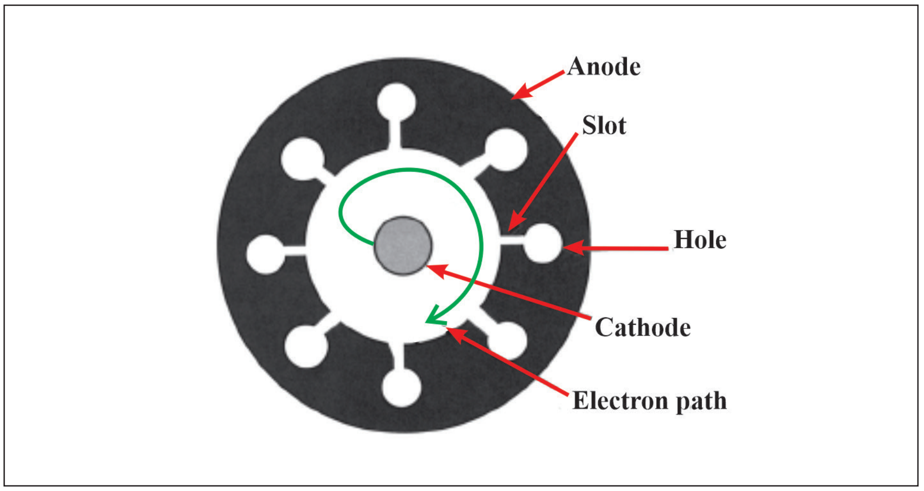


Figure 9.15 A cross section through a magnetron illustrating the principle of operation. Electrons emitted by the cathode spiral toward the anode in a magnetic field that is perpendicular to the page. As the electrons pass the cavities consisting of the holes and the slots, they induce oscillations at microwave frequencies. This is analogous to the sound waves that are produced by blowing air across the top of a soda bottle. (Adapted from Stanton, R., and D. Stinson, *Applied Physics for Radiation Oncology*, Fig. 9.6. © 1996, with permission from Medical Physics Publishing.)

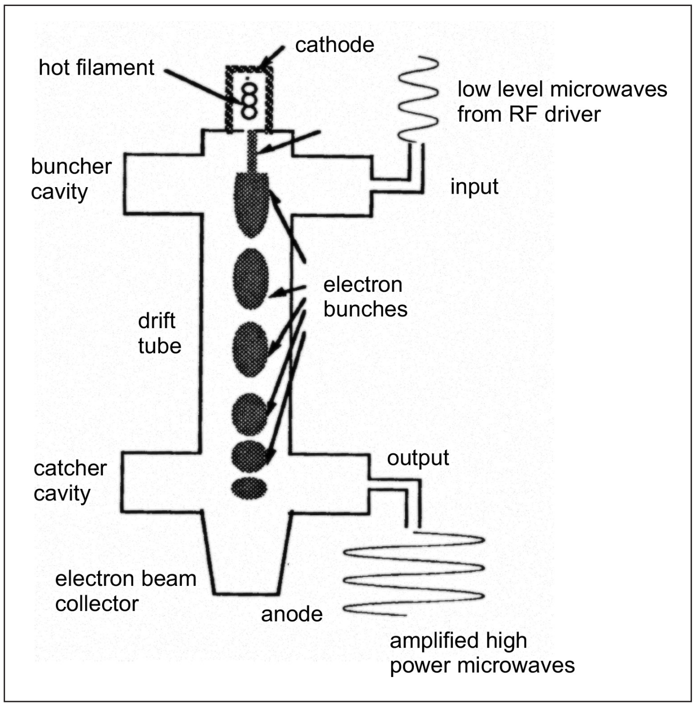


Figure 9.16 The operation of a klystron. Electrons are injected by the cathode at the top, and they are accelerated toward the anode at the bottom. Low-energy microwaves from the RF driver cause the electrons to break up into bunches. The electron bunches move past the catcher cavity and induce high-power microwave oscillations. (Reprinted from Stanton, R., and D. Stinson, *Applied Physics for Radiation Oncology, Fig. 9.7* © 1996, with permission from Medical Physics Publishing.)

low-energy standing wave linacs and in traveling wave linacs. Klystrons are used in standing wave linacs with energies above about 12 MV. Linear accelerators with klystrons (see Figures 9.16 and 9.17) require a low-energy source of microwaves called an "RF driver." The output pulse power from the RF driver only needs to be about 100 W. Klystrons are bulkier than magnetrons, and

Figure 9.17 A klystron inside the gantry stand of a high-energy standing wave linac. The tank of insulating oil can be seen at the bottom.

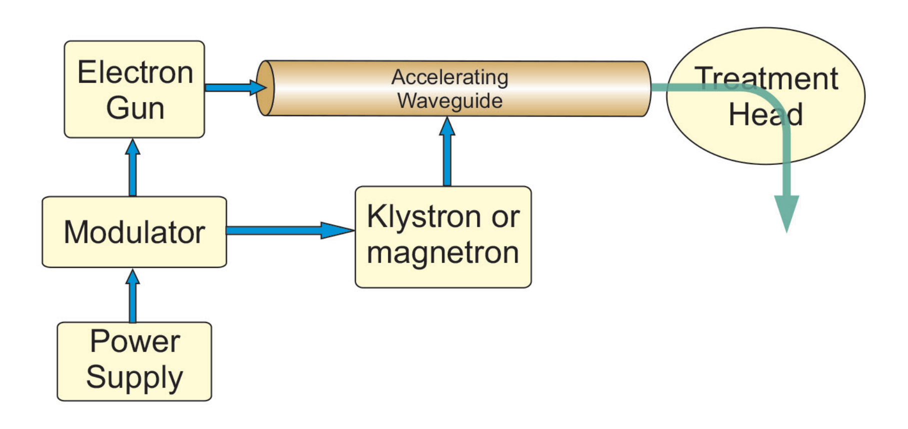


Figure 9.18 A block diagram showing the major components of a linear accelerator.

they sit inside a tank of insulating oil (see Figure 9.17). This precludes gantry mounting and, therefore, the microwave power must be transmitted farther to reach the accelerating waveguide. Klystrons are positioned in the gantry stand and the microwaves are sent via a transmission waveguide to the accelerating waveguide in the gantry. A klystron costs about $70,000, considerably more than a magnetron. The operating lifetime of a klystron is about four to seven years.

Let us now consider the entire operation of a linear accelerator as a system. This is illustrated in the block diagram of Figure 9.18. Microwave power is supplied to the accelerating waveguide in short (5 μs) pulses. A power supply furnishes high-power DC to the modulator. The modulator contains the so-called "pulse forming network" (pfn). The high-power pulses are delivered simultaneously to the klystron (or magnetron) and the electron gun. The pulses are triggered by a vacuum tube called a *thyratron* that acts as a switch and is capable of handling high current. The electron gun injects a pulse of electrons into the accelerating waveguide. The electrons are accelerated down the waveguide and emerge as a narrow beam about 3 mm in diameter.

In standing wave linacs, the photon beam energy can be changed with the use of an energy switch (see Figure 9.4). The switch has the effect of reducing the electric field in a portion of the waveguide, thus reducing the final energy of the electrons. For the high-energy photon beam, the electric field is at full strength over the entire length of the guide. In a traveling wave linac, the radiofrequency (RF) power delivered to the guide is adjusted to change the beam energy.

9.5 The Treatment Head

Inside the cover of the treatment head of a linear accelerator is a thick shell of shielding material (not shown in Figure 9.4) designed to reduce the amount of radiation that escapes through the head (leakage radiation). Radiation escaping from the head contributes dose to the entire body of the patient, and it requires thicker walls to shield personnel outside the room. We only want radiation to emerge from the machine as part of the useful beam. Head leakage will be discussed further in chapter 18 on radiation protection. There are nine major components attached to, or inside the head (the first seven of these are shown in Figure 9.19 and the last two in Figure 9.22):

1. x-ray target; used for photon beams only
2. scattering foils; used for electron beams only
3. flattening filter; used in x-ray mode only (but see section 9.11)
4. monitor ion chambers
5. fixed (primary) and movable (adjustable) collimators
6. multi-leaf collimator (MLC)
7. electron applicator (or cone) for electrons beams only
8. light localizing system (or field defining light)
9. optical distance indicator (ODI) or rangefinder.

9.5.1 X-ray Therapy Mode

In x-ray therapy mode a target is placed in front of the narrow electron beam. The electrons enter the target, and some of their energy is converted via the bremsstrahlung mechanism into x-rays. Linear accelerators use transmission targets, unlike a diagnostic or superficial therapy x-ray tube (see Figure 9.20). The target is made out of a high-Z material, and it must be water-cooled; otherwise it would quickly melt. Bremsstrahlung x-ray production is strongly peaked in the forward direction at high energies (see Figure 5.10), which is why transmission targets are used in linacs. As a result of this, the radiation intensity emitted by the target is not uniform across the beam. To make the radiation field have uniform intensity, a flattening filter is (usually) placed in

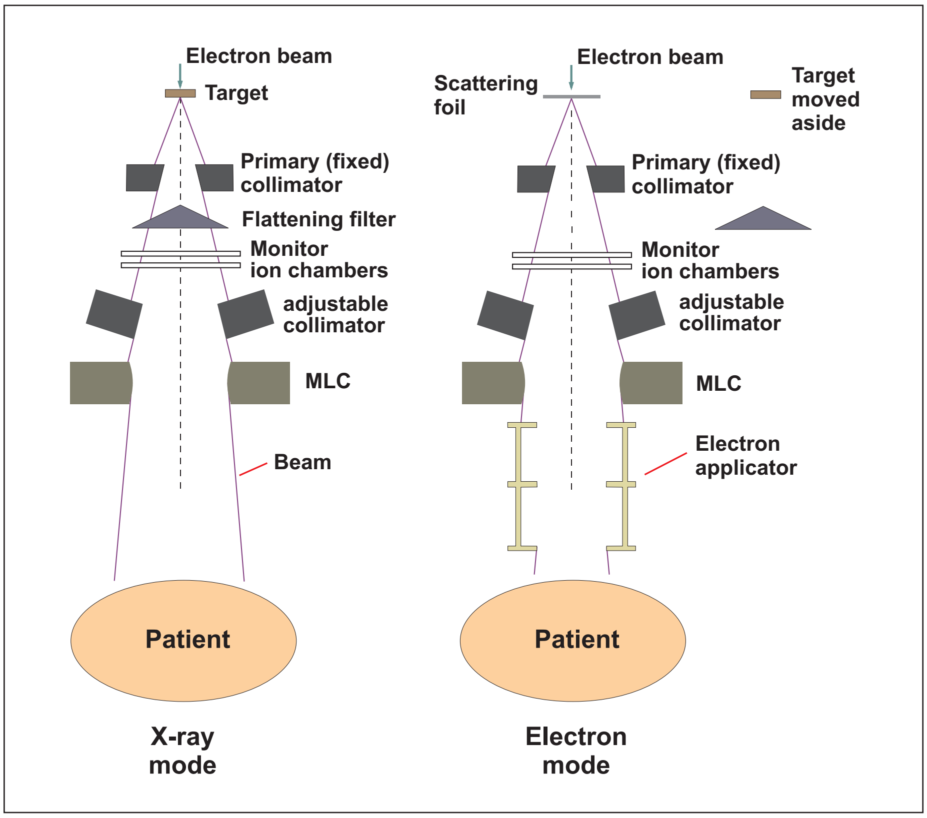


Figure 9.19 Schematic diagram of the components of a linear accelerator treatment head. The two major modes of operation are illustrated: x-ray mode on the left and electron mode on the right. In x-ray mode, a metallic target is placed in the electron beam. In electron mode, the target and flattening filter are moved aside, and a scattering foil is placed in the beam to spread out the narrow incident electron beam. In electron mode, an electron applicator is added to collimate the beam down close to the patient's skin surface.

the beam, which is thicker in the middle (and thus attenuates more) than at the edges. The flattening filter is shaped like an inverted cone. If the flattening filter is designed correctly, the radiation field will be of uniform intensity across the beam (see section 9.10 for a discussion of beam flatness). Flattening filters are made of a high-*Z* material. The flattening filter must be carefully centered on the beam central axis. If it is not precisely positioned, the beam will not be flat or symmetric (see section 9.10). The flattening filter is designed to flatten the beam at a depth of 10 cm. This is a compromise, because it is not possible to make the beam flat at all depths. Some linacs allow the option of moving the flattening filter out of the beam. These beams are referred to as flattening filter free (FFF, see section 9.11).

The dose delivered to a given location in a patient by a linac depends on: the depth in the patient, the beam energy, the collimator jaw setting, the distance of the patient from the source of radiation, etc. It is therefore not possible to set a treatment machine directly for dose. The intensity of a linac beam may fluctuate slightly from moment to moment and therefore we cannot set a timer for the amount of time the beam should remain on. Instead, we set

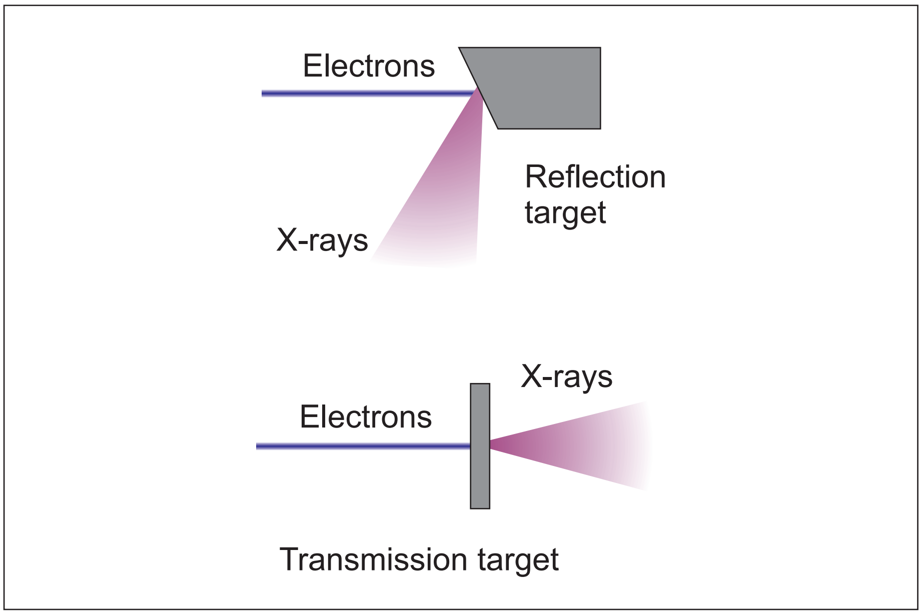


Figure 9.20 Linear accelerators use a transmission target rather than a reflection target[1] like those used for low-energy x-ray generation because x-ray production at high energy is strongly peaked in the forward direction. This is shown in chapter 5, Figure 5.10.

monitor units, or *MU*. The number of monitor units is a measure of the cumulative amount of radiation passing through the head of the linac. It is part of the treatment planning process to determine, for a particular patient beam, the relationship between the MU and the desired dose to be given to the patient.

The monitor ion chambers shown in Figure 9.19 and Figure 9.4 serve three purposes. They provide feedback to enable the accelerator to maintain a constant dose rate. Typical "dose" rates range between 200 and 800 MU/min. On some linacs the dose rate can be selected. The ion chambers also track the total or integrated dose; that is, the total MU delivered. This is related to the total amount of radiation passing through them. There are two sets of ion chambers that monitor this. The second is a backup. The output from the backup is sometimes referred to as MU2, and the primary is MU1. A safety interlock will shut off the beam if MU2 exceeds MU1 by a certain amount (typically 10%). If MU1 and MU2 both fail to shut off the beam, then a backup timer will do so. This timer will shut off the machine after a certain period of time. If there is a power failure, a backup counter keeps track of the number of MU actually delivered. The third purpose that the ion chambers serve is to monitor beam flatness and symmetry (see section 9.10). If the beam flatness or symmetry depart from expectation by more than a specified amount, then an interlock shuts off the beam. Some monitor chambers are unsealed. In this case, transducers in the treatment head must measure the atmospheric pressure and the

[1] The term "reflection target" is somewhat of a misnomer. The low energy x-rays are not reflected, rather they are emitted at large angles with respect to the direction of the incident electron beam. See Figure 5.10.

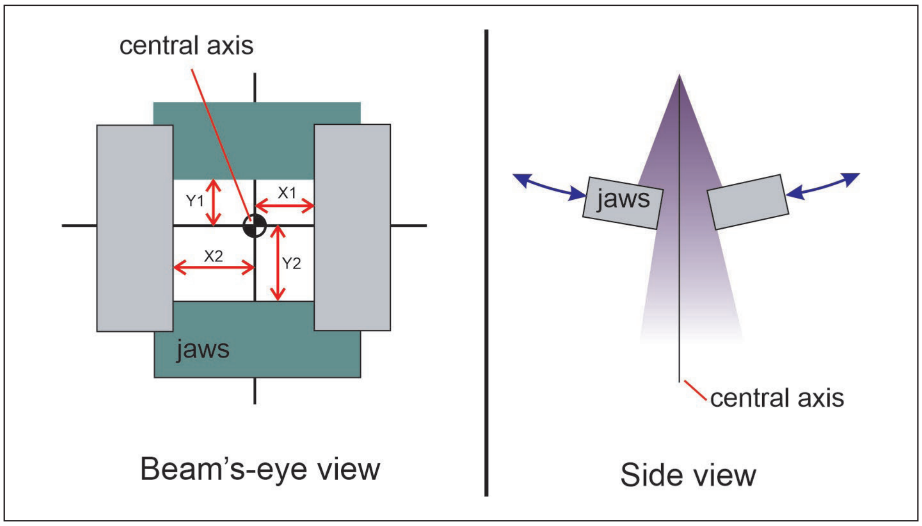


Figure 9.21 A "beam's-eye view" and a side view of a radiation field established by dual asymmetric jaws. Each of the four field edges is defined by movable jaws that can move independently of each another.

ambient temperature and apply a correction to the output from the chamber (see section 8.3.2). Sealed chambers require no correction. If a sealed chamber should develop a leak. however, the output from the linac may become erratic.

The adjustable or secondary collimator jaws produce rectangular field sizes ranging from 0×0 cm^2 to 40×40 cm^2 defined at the SAD (usually 100 cm). These jaws are usually made of tungsten. Modern linacs have asymmetric jaw capability (see Figures 9.19 and 9.21). The amount of radiation transmitted through these jaws is typically 0.5% or less.

Later in this section we will discuss multileaf collimators (MLCs). An MLC is used to shape the beam cross section so that it corresponds to the projected view of the target to be treated. In chapter 19 we shall discuss electronic portal imaging devices (EPIDs) and cone beam CT (CBCT), which are used to verify targeting of the radiation beam.

The light localizing system (also called the "field defining light," see Figure 9.22) projects a light field down onto the patient that is congruent with the radiation field. The total path length for the light field is the same as for the radiation field and, therefore, the divergence is also the same. The light localizing system shows where the radiation is going to go. The light localizing system must be checked periodically to ensure that the light field is congruent with the radiation field (see chapter 21 for further discussion). If the light field and radiation field are not congruent, the tilt of the reflecting mirror (not labeled in Figure 9.22) can be adjusted to shift the light field, or the light source can be moved to change the size of the light field. Figure 9.22 also shows the optical distance indicator (ODI), which is sometimes called the "range finder." The ODI projects a scale down onto the patient surface, which indicates the value of the SSD.

An accessory holder (not shown in Figure 9.19) can be attached to the treatment head. The accessory holder is used to insert beam modifiers into the beam such as cast blocks, wedges, and compensators.

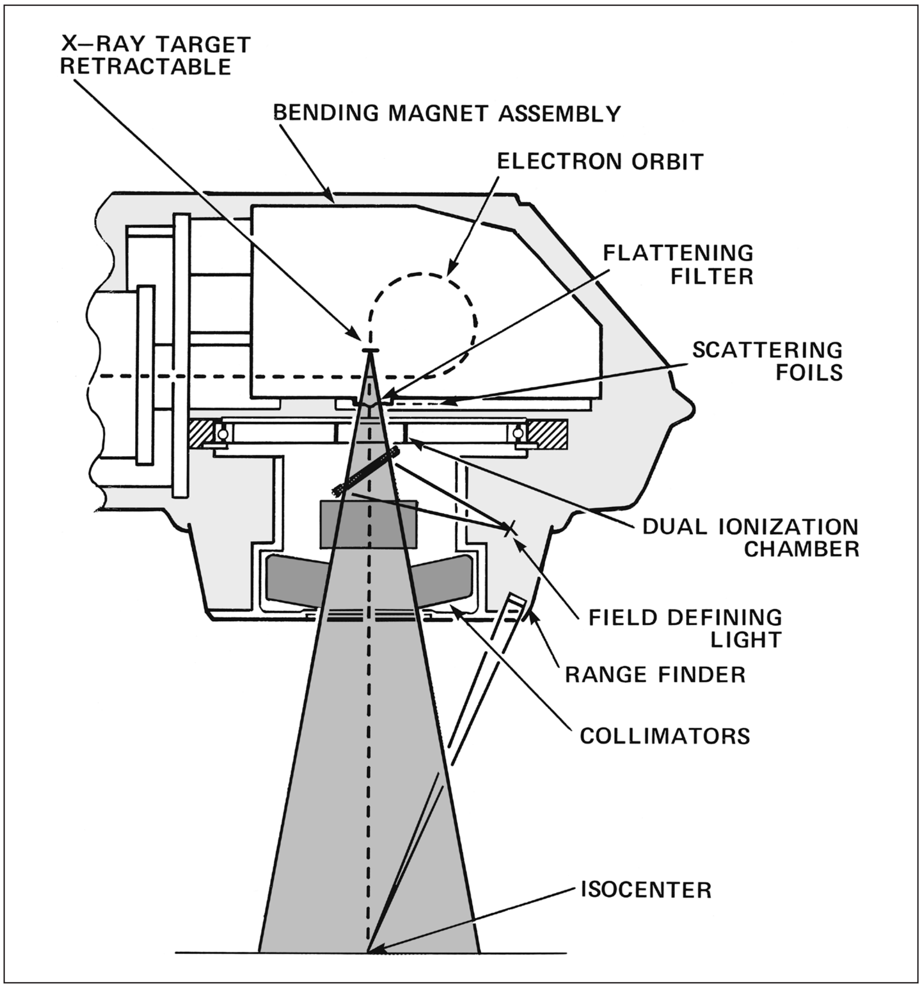


Figure 9.22 The head of a linear accelerator showing the field-defining light. The light source is off to the side of the beam and projects up to a mirror that reflects it downward. The total light path length is the same as for the radiation and, therefore, the divergence is the same. The optical distance indicator (ODI) or range finder is also shown. The ODI projects a scale downward onto the surface below. The projected scale indicates the distance of the surface from the source of the radiation. This is called the source-to-surface distance (SSD). (Reprinted from Karzmark, C. J., and R. Morton, *A Primer on Theory and Operation of Linear Accelerators in Radiation Therapy*, 3rd Edition, Fig. 38. © 2017, with permission from Medical Physics Publishing. Image courtesy of Varian Medical Systems.)

9.5.2 Electron Therapy Mode

When a linac is run in electron therapy mode, the target and the flattening filter must be moved out of the way (see Figure 9.19). The flattening filter is sometimes moved with the use of a carousel or “lazy Susan” type of arrangement (see Figure 9.4). The “raw” electron beam cannot be used to treat a patient because it is a very narrow beam only about 3 mm in diameter. The use of a scattering foil (or multiple foils) is the most common method of spreading the beam. A scattering foil is a thin foil made of a high-*Z* material (copper or lead) that easily scatters electrons. This spreads out the beam. The foil should be just thick enough to spread out the beam, but not so thick as to cause significant

Figure 9.23 A therapist prepares to attach an electron applicator to the collimator of a linac. The applicator provides extra collimation. (Courtesy of Elekta AB.)

bremsstrahlung production, which would contaminate the electron beam with x-rays (see chapter 16 on electron beams).

Electrons are easily scattered, even by air. To prevent the electrons from straying outside the desired field, the electron beam must be collimated all the way down to the skin surface or as closely as possible. This is accomplished with the use of an electron applicator (sometimes called an "electron cone," see Figures 9.19 and 9.23). Without the applicator, electrons would scatter out of the beam, and the beam would not be flat (see section 9.10). An interlock prevents the linac from running in electron mode unless an applicator is inserted in the collimator by the operator. Applicators come in a variety of field sizes: 10×10 cm^2, 15×15 cm^2 (14×14 cm^2 for Elekta), 20×20 cm^2 and 25×25 cm^2 are common. Smaller cones having a circular aperture of about 5 cm diameter are also common. When an electron cone is inserted, the adjustable collimator jaws must be set to a specific field size that is larger than the cone aperture. This field size may depend on the electron beam energy. The linac automatically sets the jaws to the correct field size. An interlock prevents operation with an incorrect setting. If the jaws were set incorrectly, the radiation output would be incorrect, and the radiation beam might not be flat.

When the linac is in x-ray mode, the electron beam current inside the waveguide is up to 1000 times higher than when the linac is running in electron mode. If the linac is run in a configuration in which the beam current is this high but there is no target or flattening filter in the beam, then the rate of beam delivery will be several hundred thousand MU per minute. This may be fatal to the patient (see chapter 21, section 21.2 on therapy accidents). A dose rate interlock prevents this.

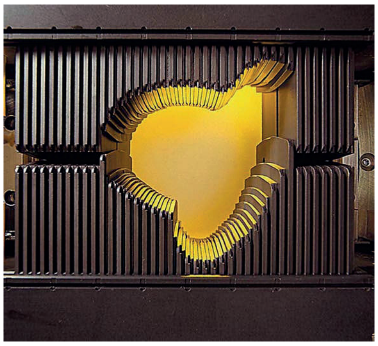


Figure 9.24 A close-up photo of MLC leaves looking up into the collimator head. (Copyright 2010, Varian Medical Systems, Inc. All rights reserved.)

9.6 Field Shaping

To reduce the amount of normal tissue that is exposed to radiation, it is useful to shape the beam aperture so that it conforms closely to the volume containing disease and to block the volume outside this region. The shape of the beam aperture as seen from the point of view of the radiation source is called a "beam's-eye view."

Methods that can be used to accomplish field shaping are: asymmetric jaws (see Figure 9.21), hand blocks, cast blocks, and multileaf collimators (MLCs). Hand blocks and cast blocks are no longer used and therefore they will not be discussed here. In this chapter we shall concentrate on field shaping for photon beams. Field shaping for electron beams will be discussed in chapter 16.

The Multileaf Collimator (MLC)

The purpose of an MLC is to define the cross-sectional (beam aperture) shape of radiation beams. The MLC consists of a series of motorized tungsten leaves (see Figures 9.24–9.29) organized in two opposing leaf banks. The leaves move under computer control. Each leaf has a motor attached that moves the leaf in and out. These motors occasionally fail and the leaf becomes "stuck."

MLC beam shapes are described in terms of the coordinates of the leaf ends (see Figures 9.25 and 9.26). Beam shapes are specified in a beam's-eye view portal outline with treatment planning software. The treatment planning computer calculates the positions of the leaf ends and stores the data in a file for that patient beam. The collimator is sometimes rotated to get an optimum fit to the desired beam outline by reducing the scalloping (see Figure 9.25). The file containing the leaf positions is loaded into the linac control computer, which will then drive the leaves to the required locations. The position of each leaf must be monitored by the linac to ensure that it is where it is supposed to be. For Elekta linacs this is accomplished with the use of a camera system that monitors light reflected by a reflector on the top of each leaf.

Figure 9.26 shows a perspective view of a single leaf, along with the nomenclature used to describe the leaf. Figure 9.27 shows a side view of an

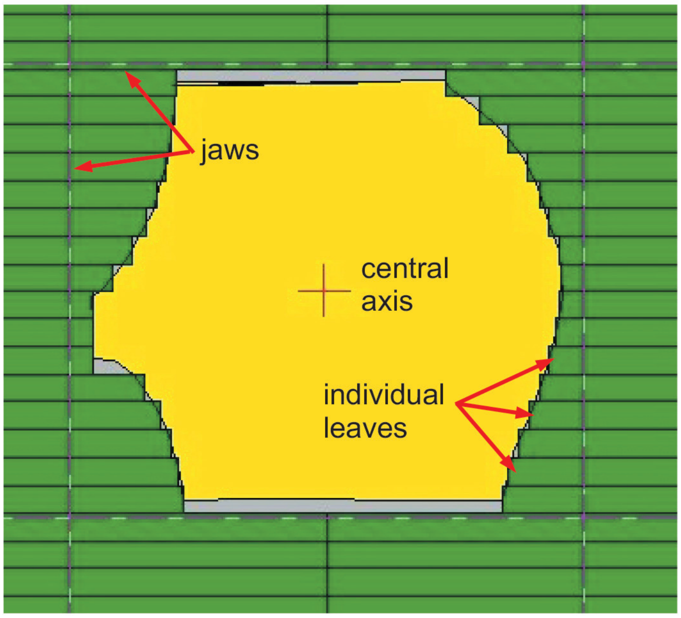


Figure 9.25 An MLC portal outline (beam's-eye view) as shown on a computer display. This MLC is a tertiary system. The position of the secondary jaws is shown. There are locations where the leaves overlap the desired treatment outline and other spots where they underlap. This produces a "scalloped" contour. The leaf width is 1.0 cm projected to isocenter. Varian SHAPER program software. (Copyright 2010, Varian Medical Systems, Inc. All rights reserved.)

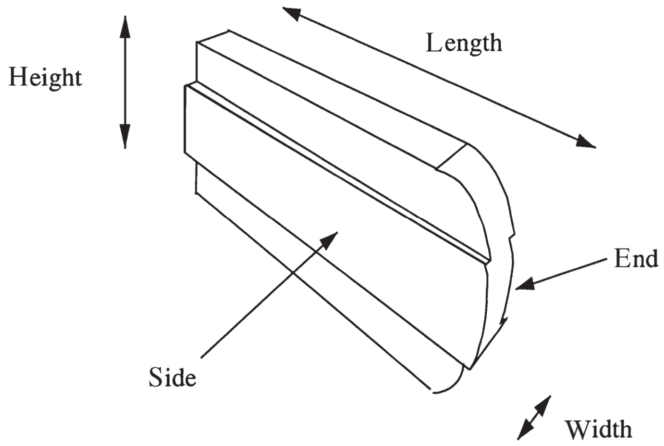


Figure 9.26 A perspective view of an MLC leaf with the nomenclature used to describe the leaves. This leaf has a rounded edge, and it has a "tongue" on the side. (Reprinted from AAPM Report No. 72, © 2001, with permission from American Association of Physicists in Medicine (AAPM).)

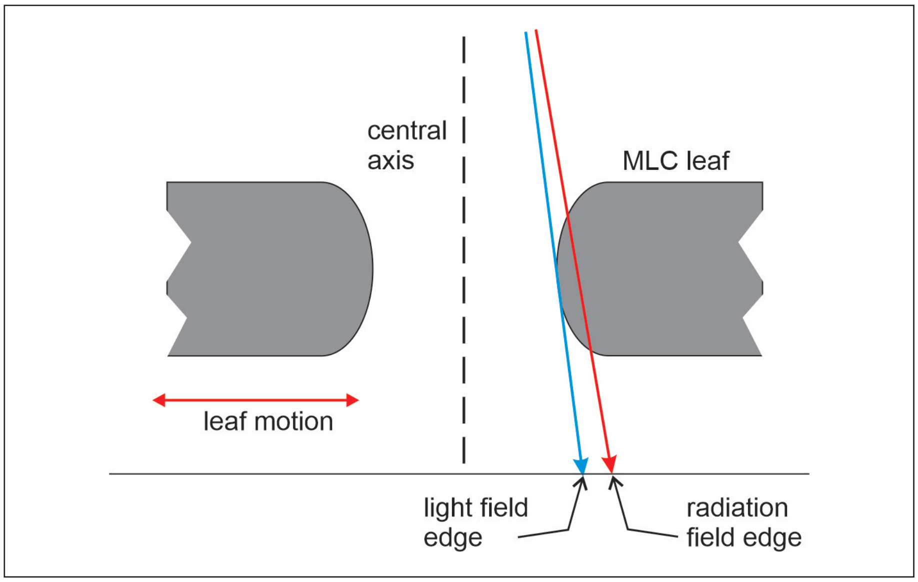


Figure 9.27 A side view of opposing MLC leaves, showing partial transmission (red arrow) through a rounded leaf edge, leading to a larger penumbra than if the leaves were focused. There is an offset between the light field edge and the edge of the radiation field.

opposed leaf pair with rounded ends. The penumbra is larger than it would be for focused collimation because of partial transmission through the rounded leaf edges. In a single focus leaf design, the leaf sides are tilted to align with beam divergence (see Figure 9.29). The ends of the leaves are rounded (Figures 9.26 and 9.27) so that the penumbra remains relatively constant with respect to the distance of the leaf edge from the central axis. There is a slight offset between the edge of the light field and the edge of the radiation field.

Depending on the manufacturer, the MLC either replaces one set of jaws or is used as a tertiary field-shaping device (see Figure 9.28). If the MLC is described as a tertiary field-shaping device, it is below the jaws. The term "tertiary" is used because the field is shaped by: (1) the primary collimator, (2) movable secondary jaws and, finally, by (3) the MLC. A tertiary MLC reduces the available clearance between the head of the linac and the isocenter, decreasing the collision-free zone for patient treatment. The Varian MLC is a tertiary field-shaping device, whereas Elekta uses the MLC to replace a set of jaws. For a tertiary MLC, the leaves can be withdrawn entirely, and the jaws can be used to define rectangular fields.

The number of leaves ranges from approximately 52 to 160. They are arranged in banks of opposing pairs as shown in Figure 9.28. The leaf width, as projected to the isocenter, ranges from 2.5 mm to 1.0 cm. A common leaf width is 0.5 cm. Smaller leaf width gives more conformal coverage (less scalloping at the edges, see Figure 9.25) but more leaves are required.

For some field-shaping applications, it is necessary for leaves to travel across the midplane. This is sometimes called "over-travel." One manufacturer has an over-travel of 10 cm. If over-travel is small, it will not be possible to form small apertures far from the central axis.

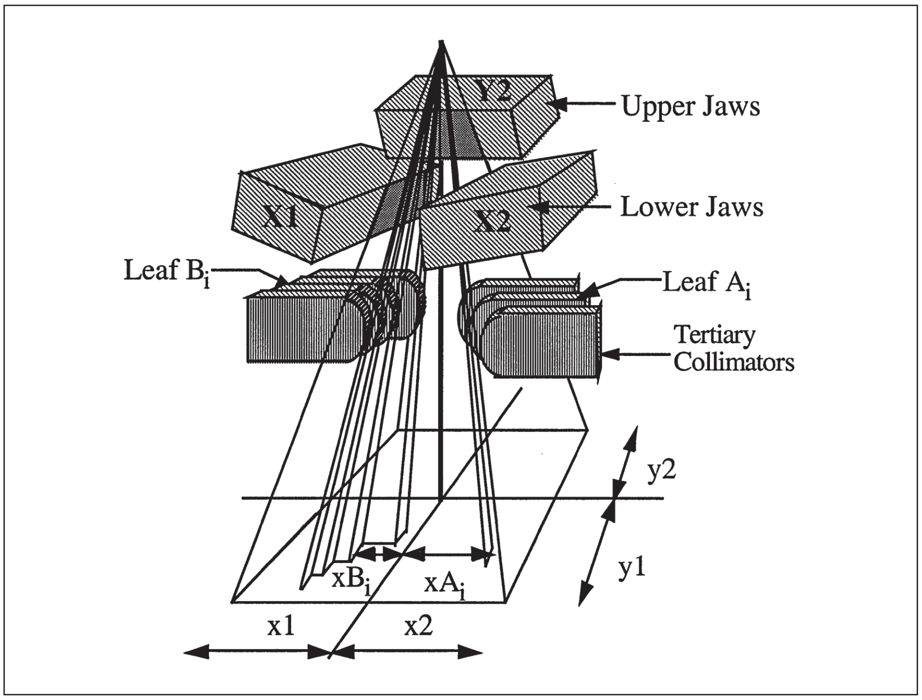


Figure 9.28 An MLC that provides tertiary beam shaping. There are two banks of leaves, A and B. The primary collimator is not shown. The secondary collimation is provided by the jaws. For clarity, jaw Y1 is not shown. The MLC provides final field shaping. (Reprinted from AAPM Report No. 70, © 2001, with permission from AAPM and Medical Physics Publishing.)

The transmission through a leaf (intraleaf transmission) of a tertiary MLC is approximately 1%, varying somewhat with energy. Transmission between leaves (interleaf transmission) is larger—on the order of 1.5%. Interleaf transmission can be reduced by introducing steps or a tongue-and-groove arrangement, as shown in Figure 9.29.

The Elekta Agility has two leaf-banks with 80 tungsten alloy leaves each for a total of 160 leaves. The leaf width is 0.5 cm projected to isocenter. Leaf over-travel beyond the central axis is 15 cm. The interleaf transmission is less than 0.5%.

A recent MLC leaf design is found in the Varian Halcyon/Ethos™ linacs that use two sets of vertically stacked MLC leaf banks as shown in Figure 9.30. In this design there are no moveable secondary jaws. The leaves in the upper and lower leaf banks are offset from one another by one half of the leaf width. The advantages of this arrangement are reduced interleaf leakage and an effective leaf width (5 mm) that is one half the physical leaf width (10 mm).

9.7 Auxiliary Subsystems

Linear accelerators are complex machines with a variety of subsystems necessary to make all of the major components operate. They have a vacuum system, a water-cooling system, a dielectric gas system, and some have a compressed air system.

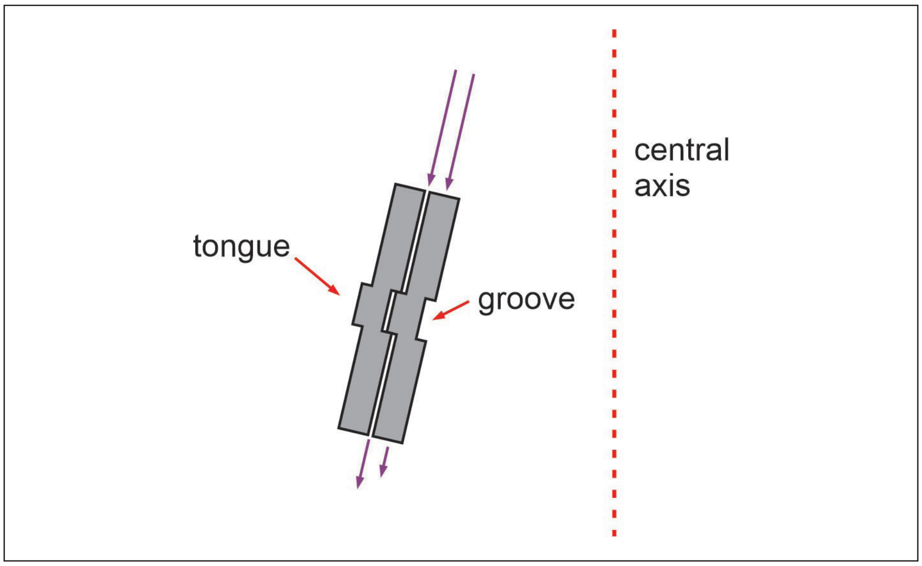


Figure 9.29 Cross section showing two adjacent MLC leaves. Leaf motion is perpendicular to the page in this diagram. The leaves are tilted to provide focusing in the left-right direction. The tongue-and-groove design reduces leakage (interleaf) between leaves.

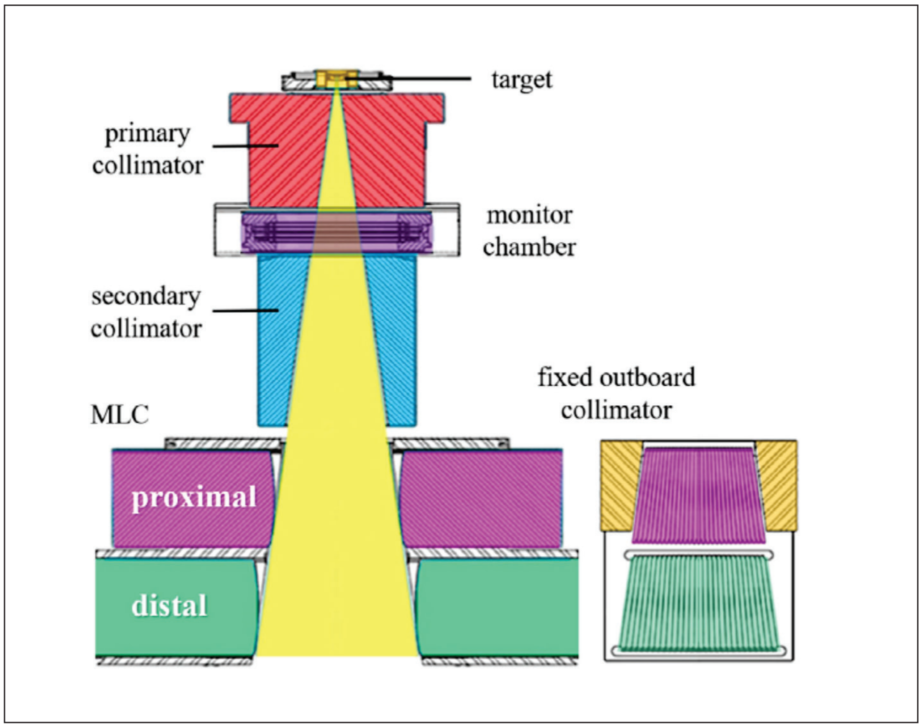


Figure 9.30 MLC design of the Varian Halcyon™ linac. There are two vertically stacked sets of MLC leaf banks and no adjustable secondary jaws. Two views of the leaves are shown: a side view and an end-on view (bottom right). The individual leaves in the upper and lower leaf banks are offset horizontally by one-half leaf width. This arrangement reduces interleaf leakage and reduces the effective leaf width to one half the physical leaf width. (From Miyasaka, R. et al. "Investigation of Halcyon multi-leaf collimator model in Eclipse treatment planning system: A focus on the VMAT dose calculation with the Acuros XB algorithm," *J Appl Clin Med Phys*. 2022;23:e13519. 2022)

The klystron, the electron gun, and the accelerating waveguide need to operate under a high vacuum. The high vacuum serves two purposes. It prevents arcing, and it prevents collisions between the accelerated electrons and air molecules, which would scatter and decelerate the electrons. A special type of vacuum pump called a "vacuum ion pump" or "vacion pump" is used to provide the necessary high vacuum.

Linacs are not usually completely powered down at the end of a treatment day. Instead, they are put into "standby" mode. In standby mode some of the subsystems, such as the vacuum system, continue to operate. If a linac were to lose part of its vacuum, it might take an excessively long time to reestablish a sufficient vacuum to produce a beam.

Circulating cooling water is necessary for two purposes. The first of these is the obvious one: to carry excess heat away from critical components, such as the accelerator waveguide, the microwave power source, the x-ray target, and the bending electromagnet. High-energy microwaves heat the accelerating waveguide. The target is heated by the electron beam, which strikes it. The bending magnet is heated by the high electrical currents that travel through the wire coils. The second function of the cooling system is to keep the accelerating waveguide at a nearly constant temperature (usually between 30°C and 40°C). This is necessary because the propagation properties of microwaves in the accelerating waveguide are exquisitely sensitive to the dimensions of the copper waveguide, and these dimensions are affected by the temperature of the metal.

The water-cooling system employs a heat exchanger. The water that is used internally to cool the parts of the linac directly is in a closed-loop system. The external cooling water is either brought in directly from the municipal water supply or from a device that first cools it, called a "chiller." The temperature and the flow rate of the external cooling water must meet certain requirements in order to provide adequate cooling. The internal water enters the heat exchanger, which has a radiator-like structure (designed for maximum heat exchange) immersed in a tank of external cooling water to allow for maximum heat exchange with the hot internal water. The internal cooling water is distilled water. The water level should be checked daily, and distilled water should be added when necessary. There is a demineralizing cartridge in the system to absorb minerals picked up by the internal cooling water. This cartridge needs to be changed from time to time.

Some linacs require a supply of compressed air. Most hospitals have a hospital-wide system of compressed air, and if not, a small compressor can be used. Compressed air is sometimes used to move the target out of the beam (the default position is target in), to operate the energy switch, and to operate the locking pin on the carousel. A pressure of 45 to 50 psi (pounds per square inch) is common.

Transmission waveguides carry the microwaves from their source (magnetron or klystron) to the accelerating waveguide. These are usually large copper "pipes" of rectangular cross section. The electric field strengths associated with the microwaves are very high, and they would cause arcing in air at normal atmospheric pressure. This is the reason that you are warned against putting any metallic objects in your microwave oven—because dramatic arcing can occur. To prevent arcing, the transmission waveguide is filled with a pressurized nonconducting (dielectric) gas. The gas used for this purpose is sulfur hexafluoride (SF_6). A nonconducting window (ceramic or quartz) is used where the transmission waveguide joins the accelerating waveguide. This window permits microwave passage but does not allow the SF_6 to enter the accelerating waveguide. There is usually a tank of SF_6 inside the gantry stand with a regulator. As the SF_6 slowly leaks out of the transmission waveguide, it can be replenished from the tank (the required pressure is 25 to 32 psi, or about 2 atmospheres for Varian machines and 0.8 atmosphere for Elekta).

9.8 Interlocks and Safety Systems

Medical linear accelerators have an interlock system that prevents or terminates operation under unsafe conditions. Interlocks are intended to safeguard the staff, the patient, and the machine from potential harm. Safety interlocks are designed to protect patients and staff from mechanical, electrical, and radiation hazards. Interlock conditions appear on the machine console.

Machine interlocks are designed to protect the machine from operation under conditions which may cause damage. After the machine is turned on, it must be warmed up before it can produce a beam. The cathodes of the thyratron and the klystron (or magnetron) require a heating period of 5 to 15 min to come up to operating temperature. Operation under cold conditions could cause very expensive damage to one of these components. This is why linacs have a warm-up timer and cannot be run until warm-up is complete. Any attempt to bypass this is asking for costly trouble. There are interlocks for the vacuum system, cooling water flow and temperature, bending magnet current, and SF_6 pressure. There are limit switches that prevent mechanical movement beyond the design limits. These apply to collimator rotation, gantry rotation, and couch vertical motion.

The door interlock is designed to safeguard the staff. If the door to the treatment room is not closed, the beam cannot be turned on. If the beam is on and the door is opened, the beam will be automatically shut off. The door interlock is usually set up as a double interlock: two separate and independent switches must be closed for the beam to come on and to remain on.

There are a variety of safety systems and interlocks to protect patients from harm and to ensure that they receive the intended treatment. In the United States, many states require an audiovisual system. A closed-circuit TV system is used so that the patient can be viewed at all times. There is also a two-way intercom system. There are emergency off buttons on the console and the treatment couch, and there may be others that are wall mounted. During beam operation, the monitor chambers check beam flatness and symmetry and inspect the dose rate. If the dose rate becomes excessive, the beam will turn off. A collision avoidance system is necessary to prevent contact between the patient and the linac head. Some linacs employ a mechanical touch guard system on the treatment head (see the ring in Figures 9.2 and 9.23). If the head of the gantry should touch the patient, switches are activated which stop all motion. Other linacs use a laser or optical system. Electron cones also have a collision detection system built into the bottom end. This is necessary because the bottom of the cone is designed to be close to the patient. Any undue pressure on the bottom of the electron applicator causes an interlock, preventing any further motion.

There are a number of interlocks to help ensure that a patient receives the intended treatment. Beam-modifying devices—such as cast blocks, compensators, and wedges—must be explicitly chosen when programming the treatment console. Each of these devices is coded. When these devices are inserted into the accessory holder, the machine "knows" which one is inserted. If the beam modifier chosen at the console differs from that actually inserted into the machine, an interlock will appear until the discrepancy is resolved. There are limits on the field size for many wedges. When a particular wedge is inserted into the collimator, the machine will trigger an interlock if the field size is too large for that particular wedge. There are also interlocks for electron applicators. The machine will not produce an electron beam unless an electron applicator is inserted. Electron applicators are also coded. The size of the electron applicator must be entered at the console, and it must match the applicator actually mounted. The collimator jaws must be set to the correct opening for the applicator mounted and for the electron energy set.

9.9 Patient Support Assembly

The patient support assembly, sometimes called the "pedestal," is a fancy name for the patient couch. Important considerations for couch design are safety, positional accuracy, and rigidity. The couch support is offset from the isocenter so that the end of the couch can extend out over the isocenter, permitting the gantry to rotate underneath it (see Figures 9.1 and 9.2). The couch can move vertically, laterally (from side to side), and longitudinally (in and out). In addition, the couch can rotate about a vertical axis through the isocenter (see Figure 9.1). This is sometimes referred to as a pedestal rotation or "couch kick." Couch movement controls are found on both sides of the couch and on a pendant. Couch motion requires that a "deadman" switch be depressed while activating motion. Emergency off buttons are also located on the couch and the pendant. Couches will also operate in a free-float mode, in which the operator can freely move the couch in both a lateral and a longitudinal direction to position a patient or phantom.

The couch top is constructed from carbon fiber and is light, flat, and rigid for setup reproducibility. One portion of the couch top consists of a mesh, sometimes with a clear plastic placed over it. This allows for maximum radiation transmission for treatment through the couch top. A portion of the couch may have a dense spine or metal side rails. Caution must be used to ensure that radiation beams are not obstructed by these obstacles. The patient load limit ranges from 200 to 250 kg (441 to 551 lb).

Some linacs now have couches with "six degrees of freedom." They move longitudinally, laterally, and vertically. In addition, rotational motions of the table top alone are possible to correct for patient rotation, thus ensuring that the patient is in the same position as originally intended. Pitch, yaw, and roll rotations of a few degrees are possible.

9.10 Photon Beam Characteristics

A radiation beam spreads or diverges with increasing distance from the source. The width of the beam becomes larger. The field size displayed at the linac console is the length and the width of the beam cross section, as measured at the isocenter. We will denote this field size as f, where f can be numerically equal to either the length or the width. Closer to the source the field size will be smaller, and farther from the source it will be larger.

Figure 9.31 shows the geometry of a diverging radiation beam. It is useful to be able to calculate the field size at any distance r other than $r = \text{SAD}$. The distance r is the distance from the source of radiation, not the distance from the isocenter. We will denote the field size at distance r as f_r. In Figure 9.31, triangle ADE is similar to triangle ABC and, therefore, $f_r/f = r/SAD$ and, hence:

$$f_r = f \times \frac{r}{\text{SAD}}. \tag{9.1}$$

Although Figure 9.31 shows the case in which $r > \text{SAD}$, Equation (9.1) is also applicable when $r < \text{SAD}$.

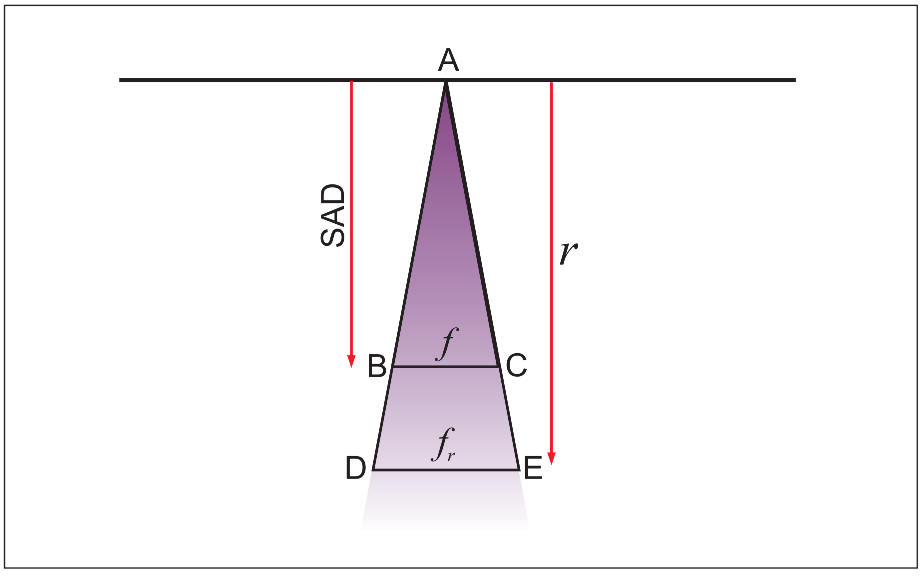


Figure 9.31 A radiation beam diverges with increasing distance from the source. The numerical field size displayed on a linac console is the field size measured at a distance equal to the SAD of the unit.

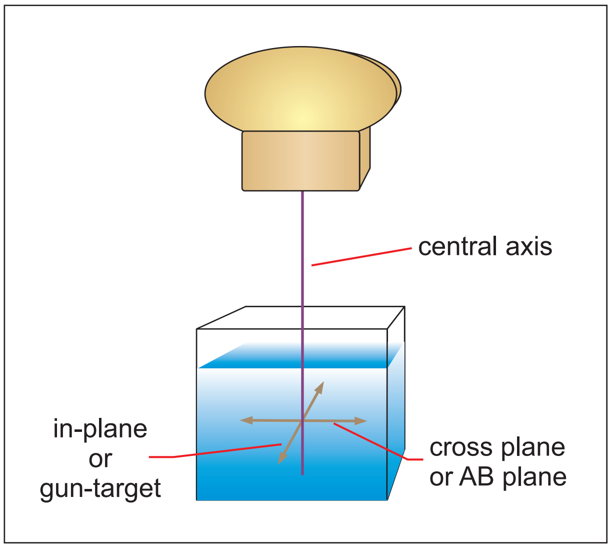


Figure 9.32
A computerized water phantom beam scanner consists of a tank of water with a radiation detector that can scan either vertically (along the central axis), in the cross-plane direction (right to left in the plane of this page), or in the in-plane direction (in and out of this page).

The properties of radiation therapy beams are measured using a scanning water phantom. This is a tank of water (like a large fish tank, see Figures 9.32 and 9.33), which can be up to 60 cm × 60 cm × 50 cm deep. A radiation detector mounted on a carriage moves through the water. The radiation detector is driven by stepper motors that are computer controlled, and it can be positioned very precisely. The radiation detector is usually an ion chamber, but a diode can be used in some circumstances. The signal from the radiation detector is fed into an electrometer, which is connected to a computer. As shown in Figure 9.33, the computer can drive the detector up and down, from side to side (called cross plane, transverse plane, or AB plane) and in and out (called in-plane, gun-target, or radial plane). For the in-plane direction, the gun side is the side toward the electron gun, and the target side is the side toward the target. The computer records the output from the detector and graphically displays the signal as a function of detector position. When the detector scans a photon beam vertically, the resulting graph is called a *depth dose curve* (see Figure 7.4). The detector can be set at a fixed depth and scanned in either the cross plane or in-plane direction. The resulting graphs are called *beam profiles* (see Figure 9.34).

A typical beam profile is shown in Figure 9.34 for an 18 MV, 20 cm × 20 cm (jaw setting) beam measured at a depth of 10 cm in the cross-plane direction. The dose at the central axis is set to 100%. Earlier we defined the field size as the width of the beam cross section. A more precise definition is that the field size is the distance between the 50% levels, as illustrated in Figure 9.34.

It is usually desirable to have the intensity of the beam uniform over the irradiated area and to drop abruptly to zero at the edges of the irradiated area, as shown in Figure 9.35. The graph of the beam intensity in Figure 9.35 is completely flat over the central portion of the beam. Real radiation beams deviate from this ideal. At the beam edges the profile is rounded (see Figure 9.34). It is not realistic to expect beams to be flat near the beam edge and, therefore, the degree of flatness is evaluated over the inner 80% of the beam usually at

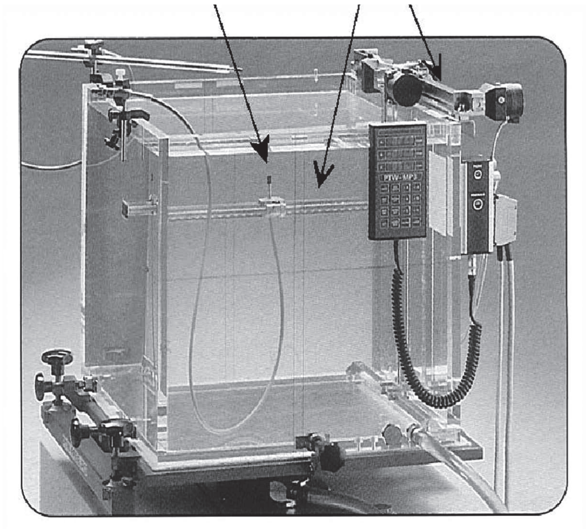


Figure 9.33 A computerized scanning water phantom consists of a tank about 60 x 60 x 50 cm^3 which can be filled with water. A radiation detector in the tank can be moved in any direction: up/down, left/right, or in/out. (Photograph courtesy of PTW.)

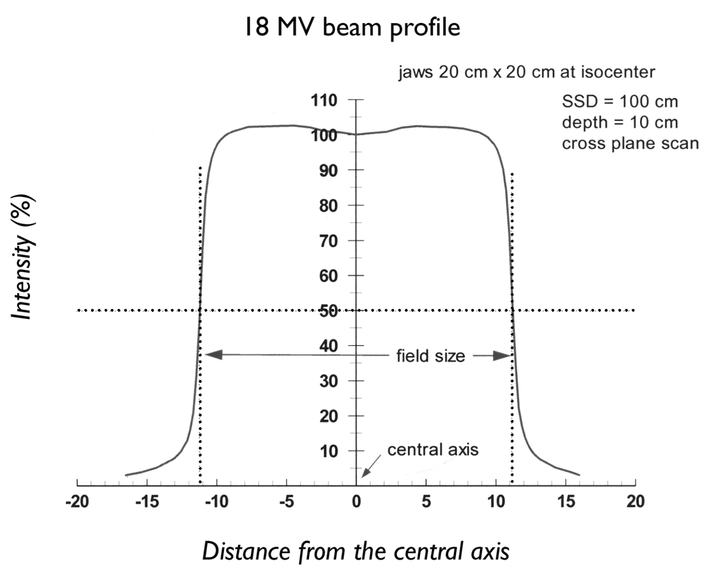


Figure 9.34 A beam profile measured with an ion chamber in a water phantom. This profile was measured by moving the ion chamber in the cross-plane direction at a depth of 10 cm. The intensity at the central axis is set to 100%. The field size is defined as the distance between the 50% intensity points. In this instance, the ion chamber is at a distance of 110 cm (100 cm SSD plus 10 cm deep), and the field size at this distance is expected to be 20 cm × (110/100) = 22 cm [see Equation (9.1)]. The measured field size corresponds closely to this.

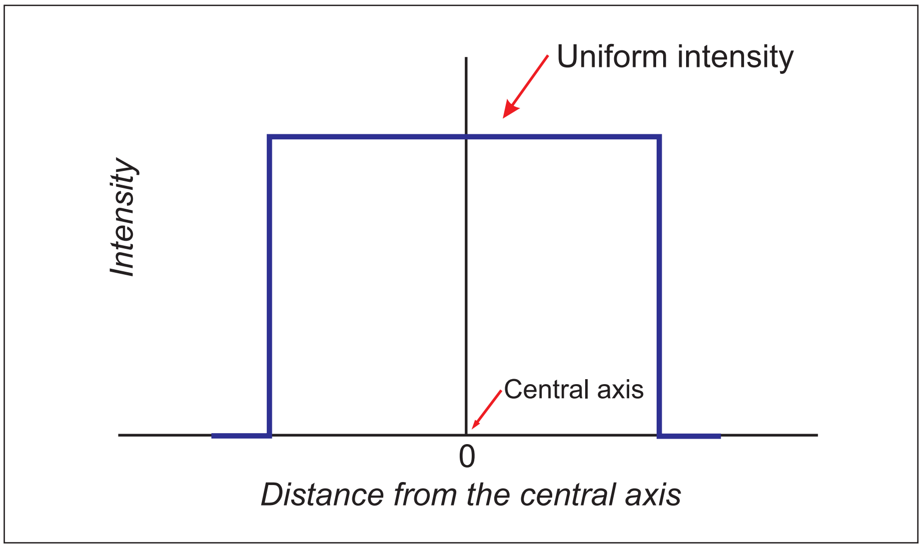


Figure 9.35 An ideal beam profile would be of uniform intensity inside the field and zero intensity outside the field. In the real world, such radiation beams are not possible because of the non-zero source size and scattering in the medium. Both of these effects lead to a penumbra region.

a depth of 10 cm. The field size of the beam in Figure 9.34 is 22 cm; 80% of this is 17.6 cm. This beam profile is shown again in Figure 9.36. A measure of beam flatness is defined as:

$$F = \frac{M - m}{M + m} \times 100\%, \tag{9.2}$$

where M is the maximum intensity value and m is the minimum value measured over the inner 80% of the field.[2] The flatness depends on the depth at which the profile is scanned. The flatness is usually specified at 10 cm deep, and sometimes also at d_m. The flatness also depends on the field size, so this too must be specified. The flatness may differ for cross-plane and in-plane profiles. Therefore, it should be evaluated for both. For beam flatness measured at a depth of 10 cm, the tolerance is usually 3% when the flatness is defined as in Equation (9.2). The flattening filter is designed to flatten the beam at a depth of 10 cm. At shallower depths, such as d_m, beam profiles may exhibit "horns." This is an increase in the dose away from the central axis. The horns are more pronounced for low energy and large field sizes.

Examining Figure 9.36 we see that $M = 103$ and $m = 100$. Substituting these values into Equation (9.2) we find that $F = 1.5\%$. This is well within the 3% tolerance.

Another measure of beam shape is beam symmetry about the central axis. There are many different ways to define this. One definition is that the area under the curve on either side of the central axis should be the same within some tolerance. If the beam symmetry does not meet tolerance, it may mean that the axis of the beam is not centered on the flattening filter. In this case,

[2] Beware: there are many definitions of flatness and symmetry.

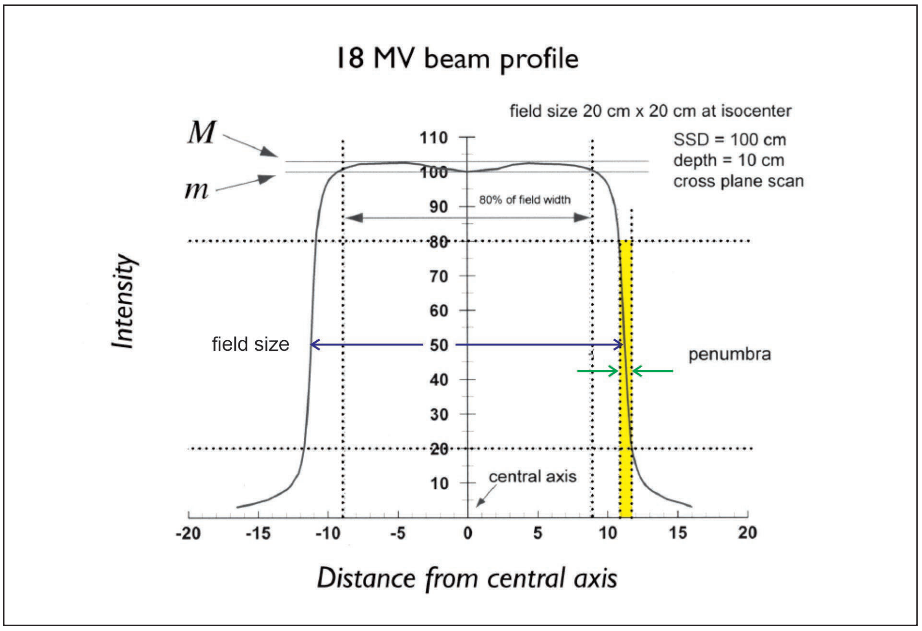


Figure 9.36 Penumbra and flatness for an 18 MV cross-plane beam profile with field size set to 20×20 cm^2 (at isocenter) and a depth of 10 cm. The field size at 10 cm deep is 22 cm. A common definition of the penumbra is that it is the distance between the 80% and 20% intensity levels (the intensity is set to 100% at the central axis). The penumbra is approximately 1.0 cm (shown in yellow). Flatness is evaluated over the central 80% of the field, which is 0.80×22 cm = 17.6 cm in this case. *M* is the maximum intensity value over this range and *m* is the minimum.

the beam must either be "steered" electronically, or the flattening filter must be shifted.

Figure 9.37 shows some circumstances under which there are departures from flatness and symmetry. In (a) the electron beam is perpendicularly incident on the target and is centered on the flattening filter. The resulting beam profile is flat and symmetric. In (b) the electron beam is still perpendicularly incident but has been shifted slightly away from the center of the flattening filter. This results in a departure from beam symmetry. In scenario (c) the beam is centered on the flattening filter but it is tilted, again resulting in a departure from symmetry. In (d) the electron beam is centered and perpendicularly incident but the energy is incorrect. This beam is symmetric but it is not flat.

The penumbra is the region at the edge of the beam over which the beam intensity drops sharply (see Figure 9.36). Whenever a source of light or radiation is not a point source (and, of course, a point source is an idealization, see section 5.5.1), the source of radiation will not cast a sharp shadow of an object. In the case of visible light, the grey area between the illuminated region and the dark shadow is called the *penumbra*. For a radiation beam, the penumbra is the region at the edge of the beam over which the dose rate drops sharply. It is caused by three factors:

1. non-point source: geometric penumbra
2. transmission penumbra: collimator jaws or MLC

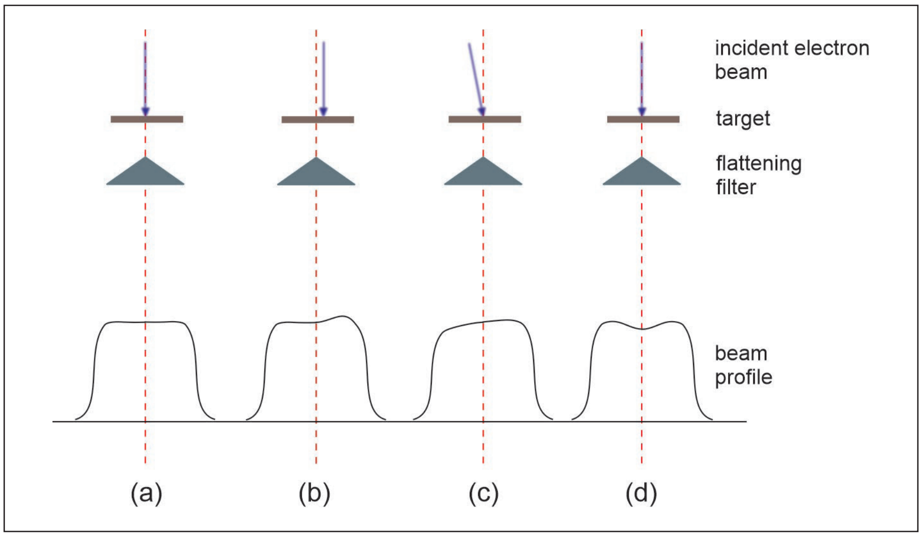


Figure 9.37 Shows causes of departures from flatness and symmetry. In (a) the electron beam incident on the target is perpendicular to the target and centered on the flattening filter resulting in a flat beam profile. In (b) and (c) the electron beam is shifted or tilted resulting in beam profiles that are not symmetric. In (d) the flattening filter is centered but the energy of the incident electron beam is incorrect, leading to a beam profile that is symmetric but not flat.

3. scattering of photons and secondary electrons: scattered photons and secondary electrons "smear" the edge of the beam. We will discuss each of these contributions below.

The physical penumbra is the measured penumbra. It encompasses all possible causes, geometric penumbra, scattering, etc. Unfortunately, there is no universally agreed upon quantitative definition of penumbra. A common definition is that it is the lateral distance between the 80% and 20% intensity levels measured at a depth of 10 cm and at either the SAD (usually 100 cm) or for SSD = 100 cm, for a field size of 10×10 cm^2. This is illustrated in Figure 9.36 (although the jaws are set to 20×20 cm^2 in this figure). The Varian Corporation TrueBeam linac has a penumbra (80% to 20%) of about 7.1 mm for 6 MV and 7.7 mm for 15 MV.[3] This is for a 10×10 cm^2 field at a depth of 10 cm in water (water surface at 100 cm SSD) although it is not clear if this is for cross-plane or in-plane.

Geometric penumbra results from the fact that sources of radiation are not point objects. A radiation beam from an extended source, defined by collimator jaws, exhibits geometric penumbra as shown in Figure 9.38.

It is possible to derive a quantitative relationship for the size of the geometric penumbra. The size of the penumbra depends on the source size, the distance to the collimator (sometimes called the "diaphragm"—either jaws or

[3] Beyer, G. (2013) "Commissioning measurements for photon beam data on three TrueBeam linear accelerators and comparison with Trilogy and Clinac 2100 linear accelerators." *JACMP* 14(1), 273.

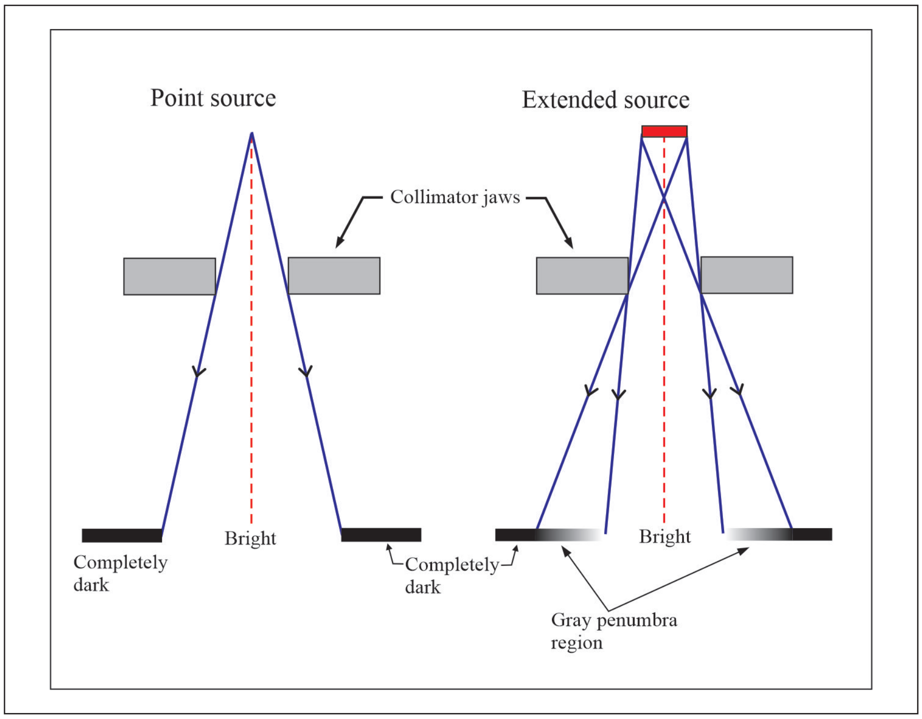


Figure 9.38 Two sources of collimated radiation: a point source on the left and an extended source on the right. The point source casts a very sharp shadow of the collimator jaws. The extended source does not cast a sharp shadow. There is a grey area between the "bright" central region and the region that is completely dark. The grey region is called the penumbra region.

MLC), and the total distance from the source. (The beam may be defined by hand blocks, cast blocks, or an MLC rather than the jaws.)

We will refer to Figure 9.39 for the derivation. The symbols in the diagram have the following meaning:

SSD = source-to-surface distance, where the surface is the surface of the patient or phantom.
SDD = source-to-diaphragm distance. This is the distance to the bottom of the jaws or MLC, whichever is defining the field.
s = the source diameter.
d = depth below the surface.
P = width of geometric penumbra.

Note that triangle ABC is similar to triangle DEC and, therefore:

$$\frac{P}{s} = \frac{\text{SSD} + d - \text{SDD}}{\text{SDD}}. \tag{9.3}$$

Notice that SSD + d – SDD is simply the distance from the bottom of the collimator jaw to the position at which the penumbra is measured. Equation (9.3) can be rearranged slightly to yield:

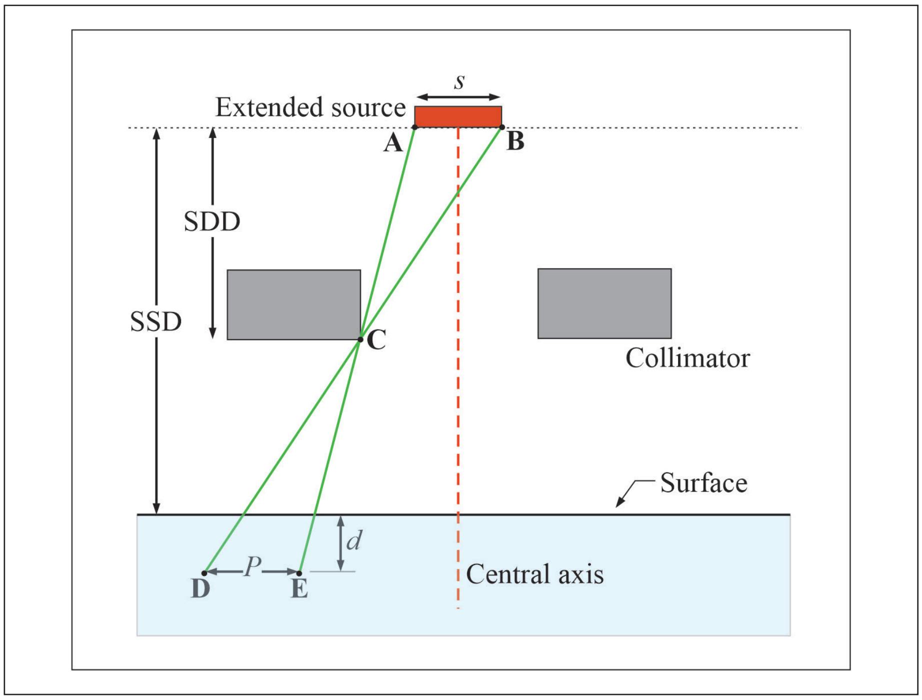


Figure 9.39 Geometry for calculation of geometric penumbra. An extended source of size s produces a geometric penumbra of width P. If you place your eye along the line joining point D and E you will only be able to see a portion of the source. If you place your eye to the left of point D you will not be able to see any of the source. If you place your eye just to the right of point E you will be able to see the entire source.

$$P = s\frac{\mathrm{SSD} + d - \mathrm{SDD}}{\mathrm{SDD}}. \tag{9.4}$$

The x-ray source diameter for a linear accelerator may only be 3 to 5 mm (although the flattening filter may effectively enlarge this), whereas a Co-60 source may be up to 2 cm in size. The consequence of this is that the geometric penumbra of a linac is less than for a Co-60 unit because the Co-60 source is considerably larger. This is one of the many advantages of a linac.

It is useful to understand systematic changes in the geometric penumbra. When the SSD or depth goes up, the penumbra goes up. When the SDD goes up, the penumbra goes down. This can be demonstrated with a flashlight. Punch a hole in an opaque piece of paper with a hole punch. In a darkened room, project the beam from the flashlight through the hole onto a wall. You will see a grey area surrounding a bright circle. The grey area is the penumbra. The SDD is the distance from the flashlight bulb to the paper. If the flashlight is held still and the paper is moved toward the wall, the bright image of the hole will become sharper because the penumbra becomes smaller. If the flashlight and the paper are together moved farther from the wall, the penumbra will go up. This is analogous to increasing the SSD.

Another cause of physical penumbra is transmission penumbra. This is produced when the beam cuts through the edge of a square collimator jaw, as shown in Figure 9.40. This can be eliminated by angling the face of the jaws, as shown on the right in Figure 9.40. Linear accelerators have such "focused"

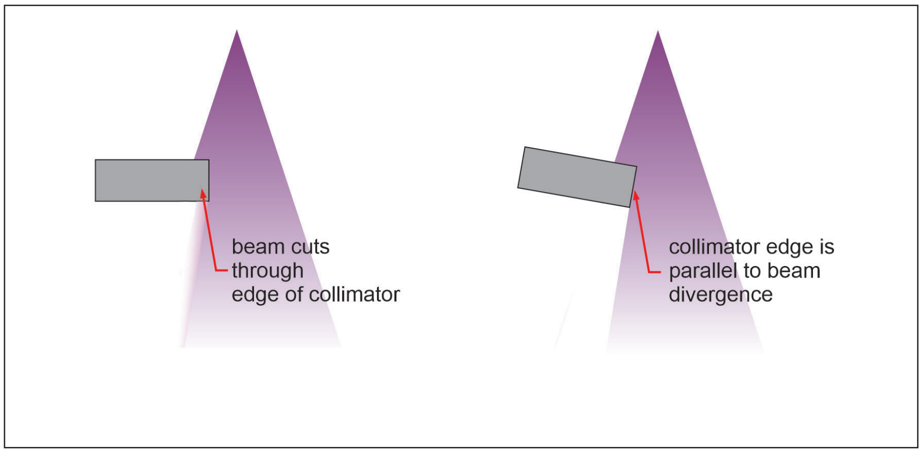


Figure 9.40 On the left, the beam cuts through the edge of the square collimator jaw. Where the beam cuts through the edge, it will be partially transmitted. The partial transmission will result in a penumbra region, even for a point source of radiation. In the diagram on the right, the edge of the collimator jaw is parallel to the beam edge. This type of collimator is called a focused collimator, and it reduces the transmission penumbra.

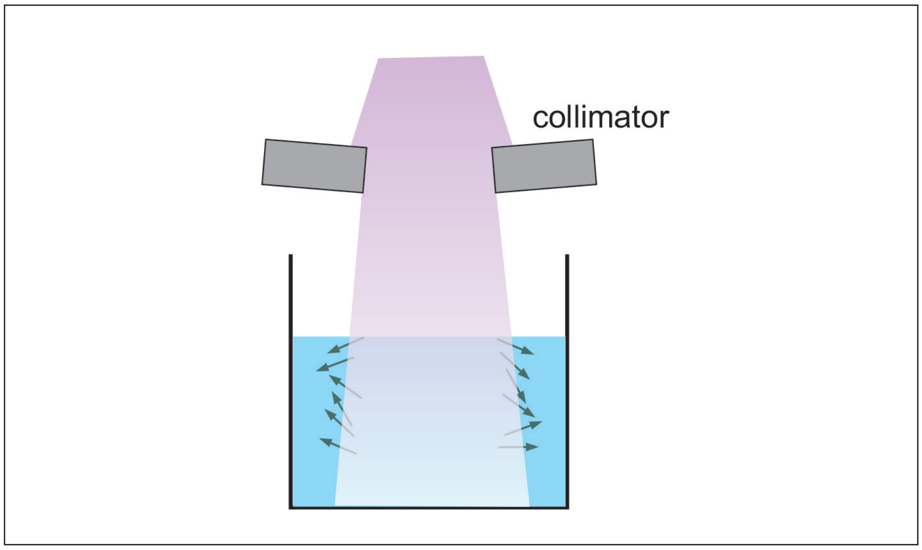


Figure 9.41 Scattered photons and the electron tracks (green arrows) at the edge of a radiation beam "smear" out the dose, even if the edge of the beam is very sharply defined by the collimator.

jaws. The jaws are designed so that they remain focused even when the opening (field size) changes.

The third source of physical penumbra results from an unavoidable characteristic of radiation: scattering. This is illustrated in Figure 9.41. Photon scattering and the tracks of secondary electrons smear out the beam edge no matter how sharply it is initially defined.

Linac x-ray beams inevitably include some electron contamination. The electrons are produced by Compton interactions as the x-ray beam traverses the head of the linac. These electrons raise the surface dose and are thus unwanted (see section 13.8 for a further discussion). In addition to electron contamination, neutrons are produced in the head of the linac by photonuclear reactions for beam energies of 10 MV and above. These neutrons do not contribute significantly to patient dose, but they do present a radiation safety hazard (see section 18.9.3 for further information).

9.11 Flattening Filter Free Linacs

In the early days of radiation therapy, it was thought desirable to have beams with flat profiles. As we have seen, this requires the use of a flattening filter. There are two reasons for this: to deliver uniform doses to targets and for ease of dose calculations, especially manual calculations. With the rise of IMRT and VMAT (see chapter 15), the need for flat beams becomes less essential. For IMRT or VMAT the beam intensities are going to be modulated anyway. In addition to this, for small fields, the variation in beam intensity over the field is relatively small, even with the flattening filter removed.

When the flattening filter is removed, the beam is referred to as flattening filter free (FFF). There are a number of advantages to be gained by eliminating the flattening filter. The biggest and most obvious advantage is dose rate. The Varian TrueBeam is capable of dose rates of 1400 MU/min for 6 MV and 2400 MU/min at 10 MV. This is a gain in dose rate of approximately a factor of 2–4. This decreases treatment time and increases efficiency. Shorter treatment time means that patients have less opportunity to move during treatment. Another advantage is a reduction in head leakage (see chapter 18) and a reduction in neutron contamination of the beam (see section 6.1.5 and 18.9.3). Electron contamination in the beam is also reduced. For FFF beams a metal plate is used to filter electrons and low-energy photons generated in the target. The removal of the flattening filter affects beam profiles, the energy spectrum, and head scatter. A flattening filter hardens the spectrum of the radiation emerging from the linac head. Removal of the filter, *by itself*, therefore results in a decrease in the average energy of the photons and this, in turn, implies a reduction in the depth dose (see chapter 10 for a definition of depth dose). Elekta linacs, however, are "energy compensated," to maintain the same depth dose (see chapter 10) at a depth of 10 cm.[4] Patient treatment with FFF beams has been found to be especially useful when the PTV is small and the dose per fraction is high (stereotactic radiosurgery and stereotactic body radiotherapy—see chapter 20).

9.12 X-Band Linacs

Conventional medical linear accelerators employ so-called "S-band" microwaves with a frequency of approximately 3 GHz. So-called "X-band" microwaves have frequencies ranging from 8 to 12 GHz. Linacs have been built that

[4] The energy spectrum of Varian and Elekta FFF beams is discussed in a paper by McDermott et al. (*Linac primary barrier transmission: Flattening filter free and field size dependence*, JACMP, 24(3) 2023.)

use X-band microwaves. The waveguide for such a linac is a factor of three or four times shorter than a comparable S-band linac. S-band linacs use widely available microwave parts. Construction requires exceptionally exacting dimensional tolerances. This is one of the reasons that accelerating waveguides are so expensive. The tolerances are even tighter for X-band linacs—they are even more difficult and expensive to fabricate. The CyberKnife (Accuray) is a 6 MV linac operating at 9.3 GHz and mounted on a robotic arm that is used for stereotactic radiosurgery (see chapter 20). The Mobetron intraoperative radiation therapy unit uses a mobile X-band linac that can produce electron beams with energies of 6, 9, and 12 MeV.

The Invention of the Cavity Magnetron

Before the invention of the cavity magnetron, radar was unreliable and limited to short range. The shortcomings of early radar were due to the fact that it was not possible to generate microwaves of sufficiently high power and short wavelength to have the needed range and spatial resolution. The cavity magnetron was invented by physicists Randall and Boot at the University of Birmingham, England in early 1940. It was one of the single most important technical developments of World War II. Even early models produced hundreds of times more microwave power output than any other type of microwave-generating device. As the physicist Luis Alvarez commented in the 1980s, "If automobiles had been similarly improved, modern cars would cost about a dollar and go a thousand miles on a gallon of gas." Furthermore, the magnetron is a compact device that can be mounted inside airplanes.

The original Randall and Boot cavity magnetron (courtesy of Wikipedia).

The devotees of radar have made a compelling argument that it played a decisive role in WWII. It has been argued that it played a more important role than the atomic bomb, although certainly less dramatic. Admirers of radar like to say that radar ended the war and that the atomic bomb finished it. The invention of the cavity magnetron gave the Allies a distinct advantage, not only for detecting airplanes at long distance, but also for detecting German U-boats. Toward the end of the war, a U-boat could hardly surface without being pounced upon by Allied aircraft with on-board radar.

At first there was no clear or detailed theoretical understanding of the mechanism of operation of the magnetron. When the device arrived in the United States, a prominent group of theoretical physicists gathered around to look at it. "It's simple," the physicist I. I. Rabi declared. "It's just kind of a whistle." "Okay, Rabi," said E. U. Condon, "How does a whistle work?" The klystron was developed prior to the invention of the cavity magnetron by brothers Sigurd and Russell Varian at Stanford University. Initially, it was unable to produce the high-power output that the cavity magnetron was capable of. Today klystrons are capable of higher power output than magnetrons. Magnetrons are now widely used in police radar and in microwave ovens.

Further Reading:

The Invention that Changed the World by Robert Buderi. New York: Simon and Schuster, 1996.
Winning the Radar War by Jack Nissen. New York: St. Martin's Press, 1987.

Chapter Summary

- **Accelerators** use electric fields to accelerate charged particles; linear accelerators (linacs); circular accelerators: microtrons, cyclotrons, synchrotrons and betatrons, etc.
- **Linacs** accelerate electrons down an evacuated tube (waveguide) to almost the speed of light. The electrons can be used to treat directly or they can be directed onto a metallic target and produce x-rays via bremsstrahlung emission.
- **Isocenter** of a linac is the (ideal) point in space where the gantry rotation axis, collimator rotation axis, and couch rotation axis meet. The location of this point is fixed.
- **Source-to-axis distance (SAD):** Distance from radiation source to isocenter (usually 100 cm).
- **Source-to-surface distance (SSD):** Distance from the radiation source to the surface of the patient. This distance will generally vary with gantry angle.
- **Beam Energy:** Photon beam energy stated in terms of MV (nominal accelerating potential) not MeV; energies range from 4 MV to 25 MV, average photon energy (in MeV) is about (1/3) of stated energy in MV. The nominal accelerating potential is the effective potential difference through which the electrons are accelerated.
- Cannot set beam time on but rather monitor units (MU): MU1 primary setting, MU2 is backup.
- **Accelerating waveguide:** Essentially a copper pipe under high vacuum; electrons are accelerated down waveguide by 3000 MHz microwaves (S-band radar). Two major types: standing wave (Varian) and traveling wave (Elekta). In low-energy machines (4 to 6 MV) waveguide can be mounted vertically. In higher-energy machines the waveguide is longer and is mounted either horizontally or obliquely; therefore, a bending magnet is required. “Side cavity coupling” allows a significant decrease in the length of standing wave waveguides.
- **Bending magnet:** standing wave linacs use “achromatic” 270º bending magnet—electrons with varying energies all converge on one spot.
- **Source of microwave power**: (1) magnetron: generates high-power microwaves, which are fed into waveguide; (2) klystron amplifies low-energy microwaves supplied by “RF driver.” Klystron can produce higher power than magnetron; generally used in high-energy standing wave linacs.
- **Electron gun:** Injects pulses of electrons into waveguide. Electron beam current is 100–1000 times greater in x-ray mode than in electron mode.

- **Modulator**: Supplies high-power pulses to the klystron (or magnetron) and to electron gun; pulses are triggered by a vacuum tube called a thyratron that acts like a switch.

- **Treatment head:**
  1. X-ray target: Used for photon beams only, electron beam strikes target, x-rays produced, transmission target.
  2. Scattering foils: Used for electron beams only, spreads beam out, makes beam "flat."
  3. Flattening filter: Used in x-ray mode only, shaped like an inverted cone, flattens beam at depth of 10 cm. Must be carefully centered on beam central axis.
  4. Monitor ion chambers: Determine MU1 and MU2 (backup); beam symmetry, flatness, and dose rate are monitored. There are two of them for redundancy.
  5. Fixed (primary) and movable (adjustable jaws) collimators: often independent (asymmetric) jaws, usually up to 40 cm × 40 cm field size. Most linacs have multileaf collimator (MLC) for field shaping.
  6. MLC
  7. Electron applicator (or cone) for electrons beams only, electrons are easily scattered by air.
  8. Light localizing system (or field defining light).
  9. Optical distance indicator (ODI) or rangefinder, used to read SSD.

- **MLC**: Series of motorized tungsten leaves used to shape the beam.
  – Number of leaves ranges from 52–160, each leaf has a motor.
  – Leaf width at isocenter ranges from 2.5 mm to 10 mm (5 mm is common).
  – Intraleaf transmission is approximately 1%, interleaf transmission is larger, reduced by stepped leaves or tongue-and-groove arrangement.

- **Field size:** Distance between 50% levels (central axis 100%). If the field size at the isocenter is f, the field size at a nonstandard distance r is given by $f_r = f\dfrac{r}{\text{SAD}}$, where SAD is usually 100 cm.

- **Penumbra**: Region at edge of radiation beam over which dose drops sharply. No universal quantitative definition. One definition: Lateral distance between the 80% and 20% level measured at a depth of 10 cm with SSD = 100 cm and field size 10×10 cm^2. For this definition, penumbra 0.7–0.8 cm (Varian). Caused by: (1) non-point source (called geometric penumbra); (2) transmission through collimator jaws or blocks, and (3) scattering of photons and electrons in the medium.

- **Geometric penumbra:** $P = s\dfrac{\text{SSD} + d - \text{SDD}}{\text{SDD}}$,

  where s is the source size, SSD is the source-to-surface distance, SDD is the source-to-diaphragm distance (distance to distal end of jaws or

MLC), and d is the depth in the patient. Typical values of the source size are several millimeters for the x-ray source spot size of a linear accelerator.

Systematic behavior: $P \uparrow$ when SSD or $d \uparrow$,

$P \downarrow$ when SDD $\uparrow$.

- **Beam flatness**: Usually evaluated at 10 cm depth, must be flat to ±3% or less, measured over central 80% of the beam.
- **Beam symmetry**: A comparison of dose on one side of the central axis to the dose on the opposite side. Many ways to define, e.g., area under the beam profile on either side of central axis should be the same to within some tolerance.
- **Flattening Filter Free (FFF)**: Remove flattening filter from beam.
  - Advantages: dose rate goes up by factor of 2–4. Less head leakage, less neutron and electron beam contamination.
  - Beam energy is reduced slightly (unless energy compensated–Elekta).

Problems

1. Define the term isocenter and source-to-axis distance (SAD).

2. An electromagnetic wave (S-band microwaves) has a frequency of 3000 MHz.
   a. Calculate the wavelength in free space. Express your answer in cm.
   b. If each cavity in an accelerator waveguide is 1/2 of a wavelength long, then how long is each cavity in cm?
   c. If the energy gain is 0.6 MeV per cavity and the maximum electron energy is 18 MeV, how long is this waveguide?

3. Explain the major differences between traveling wave and standing wave linacs.

4. a. How do the features of klystrons and magnetrons differ?
   b. Why are klystrons used in some accelerators and magnetrons in others?

5. What is the meaning of the stated x-ray beam energy in MV of a linac?

6. An 18 MeV electron beam strikes the target in a linear accelerator. What is the maximum energy, average energy, and minimum energy of the photons produced? What is the mechanism by which x-rays are produced in the target of a linac?

7. Why is a transmission target used for a linac and a reflection target for a low-energy diagnostic x-ray machine?

8. What is an achromatic bending magnet?

9. How much does the beam current for a linac change in going from x-ray mode to electron mode?

10. If the dose at the center of a patient's square treatment field is 2 Gy at a depth of 10 cm, what is the dose at the field edge?

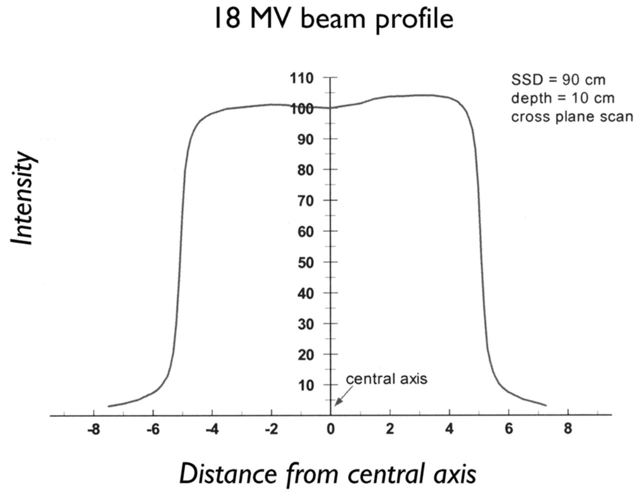
11. A linac beam has cross-sectional dimensions of 40 cm by 40 cm at the isocenter (distance of 100 cm from the source). What is the field length at a distance of 3 m from the isocenter?

12. For a particular brand of linac, the source-to-diaphragm distance is 60 cm.
    a. Calculate the geometric penumbra at a distance of 110 cm from the source assuming the source size is 3 mm. Express the answer in units of mm.
    b. What is a typical value for the physical (total) penumbra of a linac?

13. The bottom of the lower jaws on a linac (see Figure 9.4) is at a distance of 44 cm from the source, and the bottom of the upper jaws is at 36 cm. Assuming a source size of 3 mm, compute the geometric penumbra for both sets of jaws at the isocenter (100 cm). Express the answer in mm.

14. How does a flattening filter free (FFF) beam compare to a beam with a flattening filter in terms of the dose rate, depth dose, head leakage, and neutron production?

15. For the beam profile shown use a ruler to make careful measurements:
    a. measure the field size.
    b. measure the flatness (as defined in this text).
    c. does the beam meet typical specifications for flatness?
    d. what is likely to be the cause of this asymmetry?



16. What are the differences between X-band and S-band linacs?

17. If you have access to a linac, ask a medical physicist to give you a tour. You should look at the console, the information displayed on the console, and how to program the console for simple beam delivery. Note the backup MU counter and the door interlock. Inside the room, locate the emergency off switches. Observe gantry and collimator rotation and the pendant used to control these functions. Observe couch movements (lateral, longitudinal, vertical, and rotation). Observe operation of the light field and the ODI. Rotate the gantry to 180°. Use a flashlight to look into the head of the machine. Identify the upper and lower jaws and the MLC. Examine an electron applicator and see how it is attached to the collimator. Check the collision avoidance on the applicator. For a Varian linac, examine a wedge, and see how wedges are mounted on the collimator. Examine the magnetron or the klystron (and the RF driver, if present) and the thyratron, if accessible. Look at the heat exchanger, water level indicator, water temperature gauge, demineralizing cartridge, and the SF_6 supply and gauge.

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Online linac videos:

http://www.youtube.com/watch?v=hy9atKAqAf4

http://www.youtube.com/watch?v=jSgnWfbEx1A