made of lead, although tungsten, uranium, steel, and aluminum or a combination have also been used or suggested. The choice of target and flattening filter materials has been discussed by Podgorsak *et al.* (4).

E. Beam Collimation and Monitoring

The treatment beam is first collimated by a *fixed primary collimator* located immediately beyond the x-ray target or scattering foil. In the case of x-rays, the collimated beam then passes through the flattening filter. In the electron mode, the filter is moved out of the way (Fig. 4.6B).

The flattened x-ray beam or the electron beam is incident on the *dose monitoring chambers*. The monitoring system consists of several ion chambers or a single chamber with multiple plates. Although the chambers are usually transmission type, *i.e.* flat parallel plate chambers to cover the entire beam, cylindrical metal sleeve thimble chambers have also been used in some linacs.

The function of the ion chamber is to monitor dose rate, integrated dose, and field symmetry. Since the chambers are in a high intensity radiation field and the beam is pulsed, it is important to make sure that the ion collection efficiency of the chambers remains unchanged with changes in the dose rate. Bias voltages in the range of 300 to 1000 V are applied across the chamber electrodes, depending on the chamber design. opposite in direction Contrary to the beam calibration chambers, the monitor chambers in the treatment head are sealed so that their response is not influenced by temperature and pressure of the outside air. However, these chambers have to be periodically checked for leaks.

After passing through the ion chambers, the beam is further collimated by a continuously *movable x-ray collimator*. This collimator consists of two pairs of lead or tungsten blocks (jaws) which provide a rectangular opening from 0 × 0 to the maximum field size (40 × 40 cm or a little less) projected at a standard distance such as 100 cm from the x-ray source (focal spot on the target). The collimator blocks are constrained to move so that the block edge is always along a radial line passing through the target.

The field size definition is provided by a *light localizing system* in the treatment head. A combination of mirror and a light source located in the space between the chambers and the jaws projects a light beam as if emitting from the x-ray focal spot. Thus, the light field is congruent with the radiation field. Frequent checks are required to ensure this important requirement of field alignment.

Whereas the x-ray collimation systems of most medical linacs are similar, the *electron collimation systems* vary widely. Since electrons scatter readily in air, the beam collimation must be achieved close to the skin surface of the patient. There is a considerable scattering of electrons from the collimator surfaces including the movable jaws. Dose rate can change by a factor of 2 or 3 as the collimator jaws are opened to maximum field size limits. If the electrons are collimated by the same jaws as for x-rays, there will be an extremely stringent requirement on the accuracy of the jaw opening since output is so critically dependent on the surface area of the collimator. This problem has been solved by keeping the x-ray collimator wide open and attaching an auxillary collimator for electrons in the form of trimmers extended down to the skin surface. In other systems, the auxillary electron collimator consists of a set of attachable cones of various sizes.

It should be mentioned that, due to the electron scattering, the dose distribution in an electron field is significantly influenced by the collimation system provided with the machine.

F. Gantry

Most of the linear accelerators currently produced are constructed so that the source of radiation can rotate about a horizontal axis (Fig. 4.7). As the gantry rotates, the collimator axis (supposedly coincident with the central axis of the beam) moves in a vertical plane. The point of intersection of the collimator axis and the axis of rotation of the gantry is known as the *isocenter.*

The isocentric mounting of the radiation machines has advantages over the units that move only up and down. The latter units are not suitable for

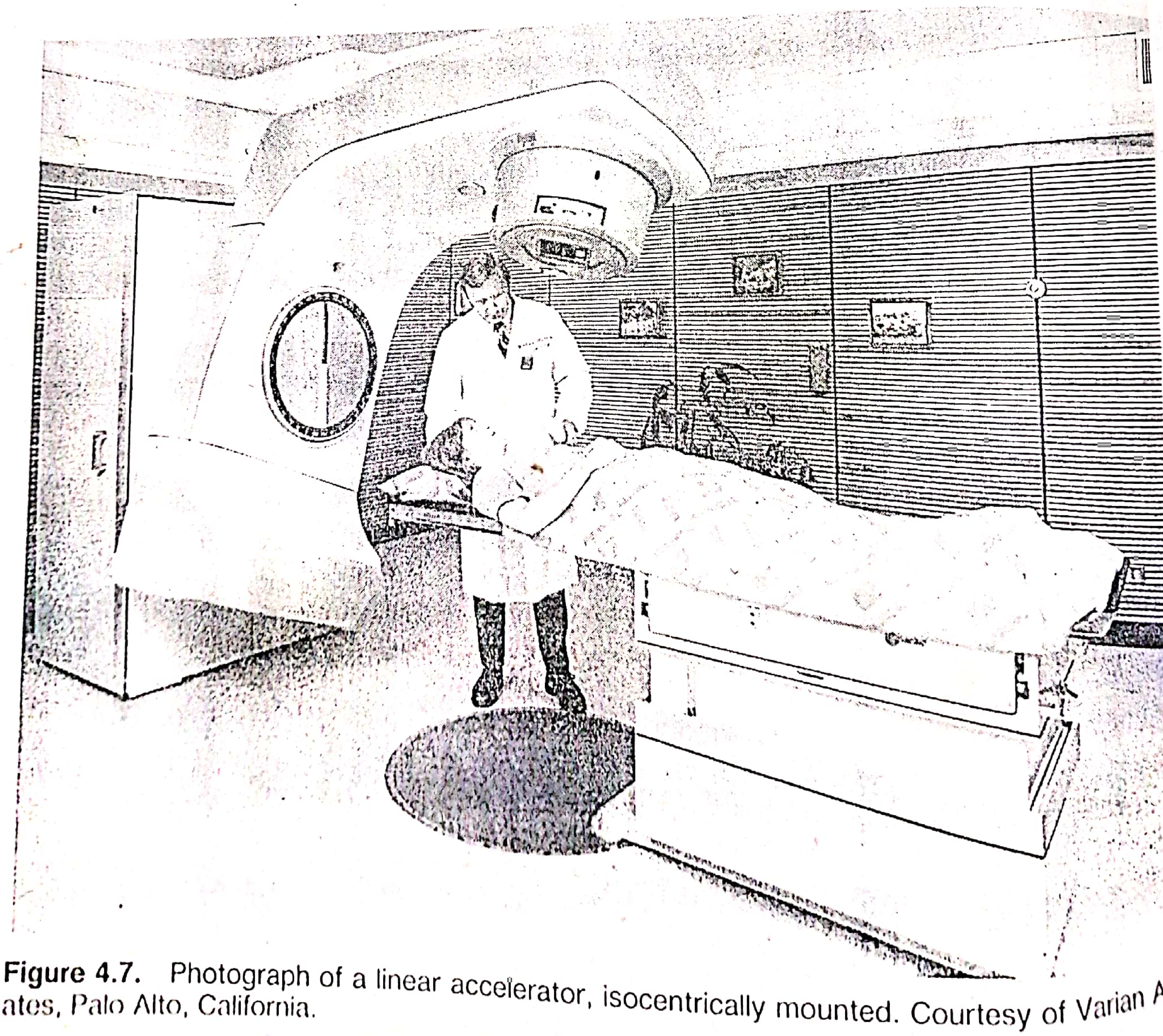

Figure 4.7. Photograph of a linear accelerator, isocentrically mounted. Courtesy of Varian Associates, Palo Alto, California.

isocentric treatment techniques in which beams are directed from different directions but intersect at the same point, the isocenter, placed inside the patient. However, the nonisocentric units are usually swivel mounted, *i.e.* the treatment head can be swiveled or rotated in any direction while the gantry can move only upward or downward. Although these units are not as flexible, they are mechanically simpler, more reliable, and less expensive than the isocentric models.

4.4 BETATRON

The operation of the betatron is based on the principle that an electron in a changing magnetic field experiences acceleration in a circular orbit. Figure 4.8 shows a schematic drawing of the machine. The accelerating tube is shaped like a hollow doughnut and is placed between the poles of an alternating current magnet. A pulse of electrons is introduced into this evacuated doughnut by an injector at the instant that the alternating current cycle begins. As the magnetic field rises, the electrons experience acceleration continuously and spin with increasing velocity around the tube. By the end of the first quarter cycle of the alternating magnetic field, the electrons have made several thousand revolutions and achieved maximum energy. At this instant or earlier, depending upon the energy desired, the electrons are made to spiral out of the orbit by an additional attractive force. The high energy electrons then strike a target to produce x-rays or a scattering foil to produce a broad beam of electrons.

Betatrons were first used for radiotherapy in the early 1950s. They preceded the introduction of linear accelerators by a few years. Although the betatrons can provide x-ray and electron therapy beams over a wide range of energies, from less than 6 to more than 40 MeV, they are inherently low electron beam current devices. The x-ray dose rates and field size capabilities of medical betatrons are low compared to medical linacs and even modern cobalt units. However, in the electron therapy mode, the beam current is adequate to provide a high dose rate. The reason for this difference between x-ray and

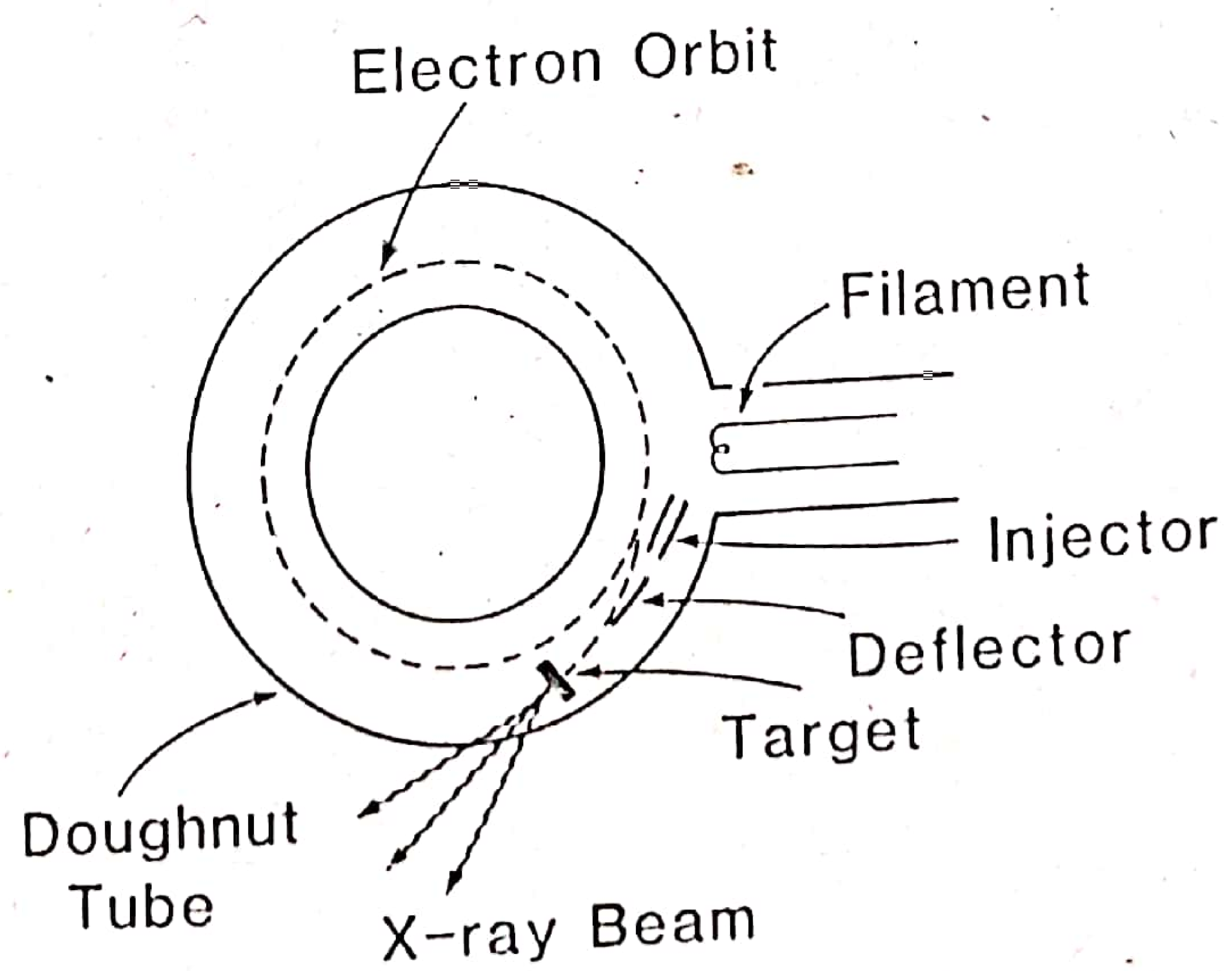

Figure 4.8. Diagram illustrating the operation of a betatron.