energies through a linear tube. The high energy electron beam itself can be used for treating superficial tumors or it can be made to strike a target to produce x-rays for treating deep seated tumors.

There are several types of linear accelerator designs but the ones used in radiotherapy accelerate electrons either by traveling or stationary elecromagnetic waves of frequency in the microwave region (~3000 megacycles/sec). difference The difference between traveling wave and stationary wave accelerators is the design of the accelerator structure. Functionally, the traveling wave structures require a terminating or "dummy" load in order to absorb the residual power at the end of the structure, thus preventing a backward reflected wave. On the other hand, the standing wave structures provide maximum reflection of the waves at both ends of the structure so that the combination of forward and reverse traveling waves will give rise to stationary waves. In the standing wave design, the microwave power is coupled into the structure via side coupling cavities rather than through the beam aperture. Such a design tends to be more efficient than the traveling wave designs since axial, beam transport cavities and the side cavities can be independently optimized (3). However, it is more expensive and requires installation of a *circulator* (or isolator) between the power source and the structure to prevent reflections from reaching the power source. For further details on this subject and linear accelerator operation the reader is referred to a review article by Karzmark and Pering (3).

Figure 4.5 is a block diagram of a medical linear accelerator showing major components and auxillary systems. A *power supply* provides DC power to the *modulator* which includes the *pulse forming network* and a switch tube known as *hydrogen thyraton.* High voltage pulses from the modulator section are flat topped DC pulses of a few microseconds in duration. These pulses are delivered to the *magnetron* or *klystron*[2] and simultaneously to the electron gun. Pulsed microwaves produced in the magnetron or klystron are injected into the accelerator tube or structure via a *waveguide* system. At the proper instant electrons, produced by an *electron gun*, are also pulse-injected into the *accelerator structure.*

The accelerator structure consists of a copper tube with its interior divided by copper discs or diaphragms of varying aperture and spacing. This section is evacuated to a high vacuum. As the electrons are injected into the accelerator structure with an initial energy of about 50 keV, the electrons interact with the electromagnetic field of the microwaves. The electrons gain energy from the sinusoidal electric field by an acceleration process analogous to that of a surf rider.

As the high energy electrons emerge from the exit window of the acclerator structure, they are in the form of a pencil beam of about 3 mm in diameter. In the low energy linacs (up to 6 MV) with relatively short accelerator tube, the

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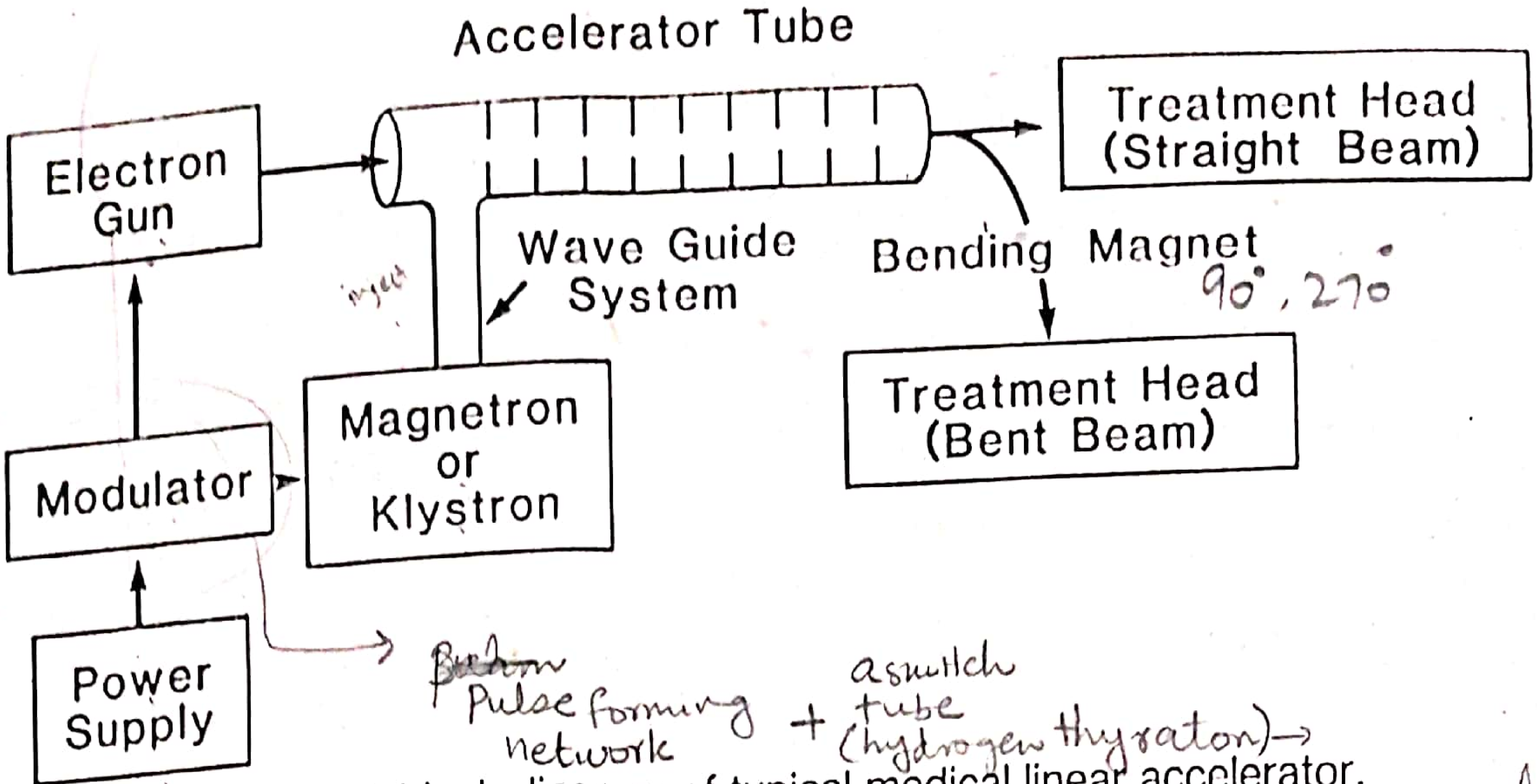
[2] Magnetron and klystron are both devices for producing microwaves. Whereas magnetrons are generally less expensive than klystrons, the latter have a long life span. In addition, klystrons are capable of delivering higher power levels required for high energy accelerators and are preferred as the beam energy approaches 20 MeV or higher.



Pulse forming network + a switch tube (hydrogen thyraton)→

Figure 4.5. A block diagram of typical medical linear accelerator.

Thyraton is a type of gas filled-tube used as a high power electrical switch and controlled rectifier.

electrons are allowed to proceed straight on and strike a target for x-ray production. In the higher energy linacs, however, the accelerator structure is too long and, therefore, is placed horizontally or at an angle with respect to the horizontal. The electrons are then bent through a suitable angle (usually about 90° or 270°) between the accelerator structure and the target. The precision bending of the electron beam is accomplished by the *beam transport system* consisting of bending magnets, focusing coils, and other components.

A. The Linac X-ray Beam

Bremsstrahlung x-rays are produced when the electrons are incident on a target of a high Z material such as tungsten. The target is water cooled and is thick enough to absorb most of the incident electrons. As a result of bremsstrahlung type interactions (Section 3.4A), the electron energy is converted into a spectrum of x-ray energies with maximum energy equal to the incident electron energy. The average photon energy of the beam is approximately one-third of the maximum energy.

It is customary for some of the manufacturers to designate their linear accelerators that have both electron and x-ray treatment capabilities by the maximum energy of the electron beam available. For example, the Varian Clinac 18 unit produces electron beams of energy 6, 9, 12, 15, and 18 MeV and x-rays of energy 10 MV. It should be noted that the electron beam is designated by million electron volts because it is almost monoenergetic before incidence on the patient surface. The x-ray beam, on the other hand, is heterogeneous in energy and is designated by megavolts, as if the beam were produced by applying that voltage across an x-ray tube.

B. The Electron Beam

As mentioned earlier, the electron beam, as it exits the window of the accelerator tube is a narrow pencil of about 3-mm diameter. In the electron mode of linac operation, this beam, instead of striking the target, is made to strike an electron *scattering foil* in order to spread the beam as well as get a uniform electron fluence across the treatment field. The scattering foil consists of a thin metallic foil, usually of lead. The thickness of the foil is such that most of the electrons are scattered instead of suffering bremsstrahlung. However, small fraction of the total energy is still converted into bremsstrahlung and appears as x-ray contamination of the electron beam.

In some linacs, the broadening of the electron beam is accomplished by electromagnetic scanning of the electron pencil beam over a large area. Although this minimizes the x-ray contamination, some x-rays are still produced by electrons striking the collimator walls or other high atomic number materials in the electron collimation system.

C. Treatment Head

The treatment head (Fig. 4.5) consists of a thick shell of high density shielding material such as lead, tungsten, or lead-tungsten alloy. It contains an x-ray target, scattering foil, flattening filter, ion chamber, fixed and movable collimator, and light localizer system. The head provides sufficient shielding against leakage radiation in accordance with radiation protection guidelines (Section 16.7B).

D. Target and Flattening Filter

In Section 3.4A, we discussed the angular distribution of x-rays produced by electrons of various energies incident on a target. Since linear accelerators produce electrons in the megavoltage range, the x-ray intensity is peaked in the forward direction. In order to make the beam intensity uniform across the field, a *flattening filter* is inserted in the beam (Fig. 4.6A). This filter is usually

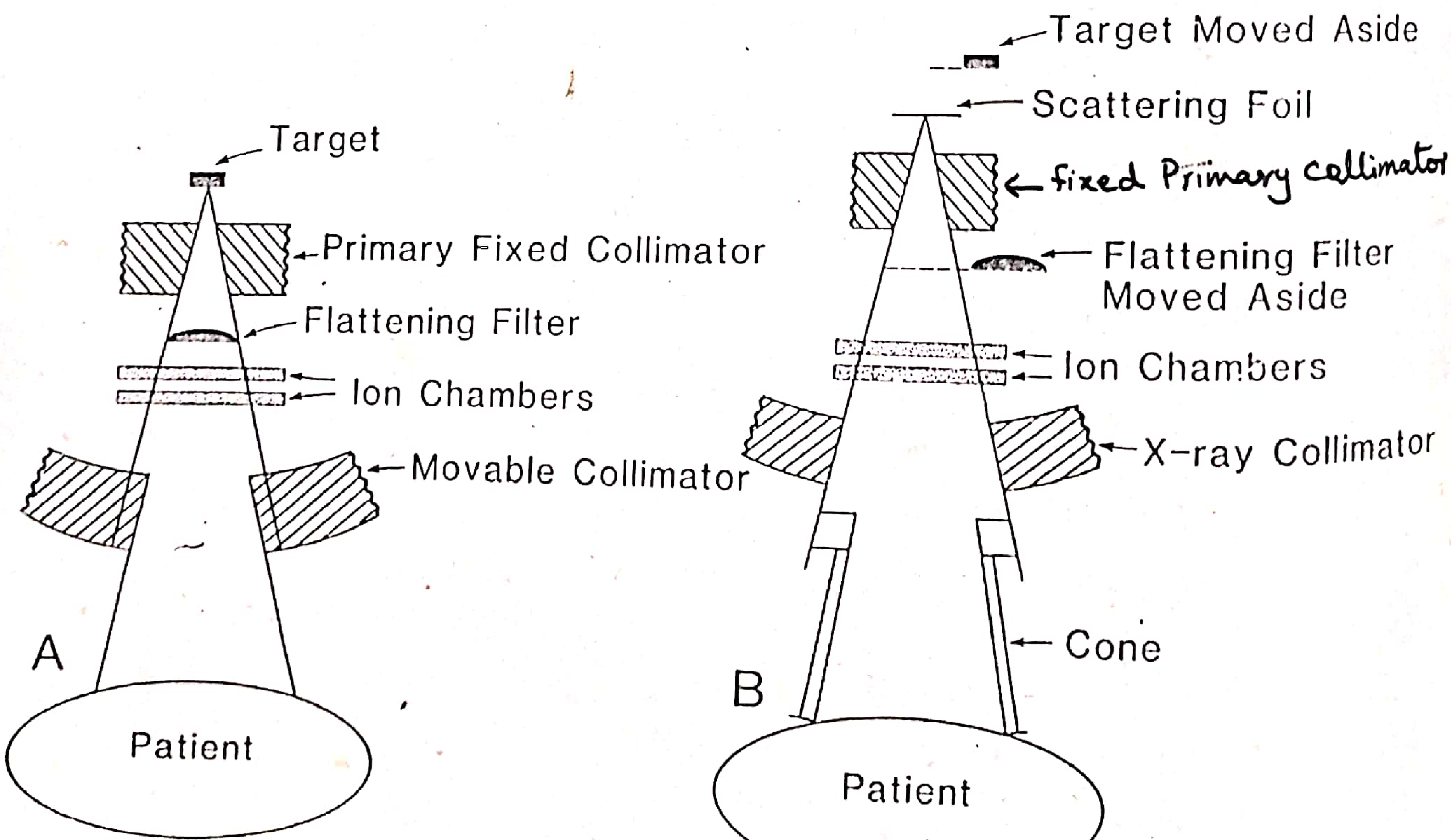


Figure 4.6. Components of treatment head. (A) X-ray therapy mode; (B) electron therapy mode