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AN ELECTRON LINAC FOR INTRAOPERATIVE RADIATION THERAPY

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Abstract

Several medical linear accelerators for intraoperative electron therapy have been built by Siemens. The first one, at the University of Texas M.D. Anderson Cancer Center, has been in clinical use for more than one year. The linac is isocentrically mounted in a gantry that can rotate ±115° from the vertical. The energy of the electron beam is defined and controlled through the use of a 270° achromatic bending magnet. The source to rotation axis distance is 100cm. The radiation field size is controlled by the inserted size of the annulus fixture and the applicator cone. Dose rate settings of 300, 600, and 900 cGY/min are available. Power is normally supplied by a 2.9 MW magnetron. Currently a machine is being commissioned that is powered by a klystron amplifier system capable of as much as 7.5 MW. The latter will permit operation from 6 MeV to 21 MeV. On the other hand, the magnetron driven machines can operate up to 18 MeV.

Introduction

The number of institutions performing intraoperative therapy has been increasing in recent years.[1] Usually such procedures are performed by doing the surgery in the operating room and then transporting the patient to the radiation therapy room where the irradiation is performed. It can be done in this way because many radiation therapy machines have both X-ray and electron treatment modes for improved flexibility. The machine to be described here is permanently located in the operating room, thus eliminating the need for transporting the patient back and forth between two rooms. Siemens has developed an accelerator system with only electron beams to fill the above need. The electron mode is appropriate and adequate for this application because surface treatment is usually desired due to the fact that the tumor is exposed during the operation.

The machine is wall mounted so as not to waste space in the operating room. All components requiring service, such as the RF power, cooling, and control systems, are outside of the operating room. RF power is fed through the wall and then through a rotary joint, into the gantry and into the accelerator. Thus no cooling fans and high voltage electrical systems are located in the operating room.

In many instances, intraoperative procedures are performed with electron energies of less than 18 MeV. This need was met by Siemens with the Mevatron ME, which makes use of an RF system containing a magnetron. Recently, a machine was built and designated the KE. This machine is capable of electron energy up to 21 MeV using the same accelerator but a klystron driven RF system.

RF System

The heart of the Mevatron is the electron linac and its associated RF system. A block diagram of the major components used is shown in Fig. 1.

As in all Siemens Mevatrons, and in fact as in most commercial radiation therapy machines, the accelerator consists of a side-coupled cavity structure. The foundations for this technology were laid at Los Alamos during the 1960s.[2] The accelerator operates in the standing wave mode at a frequency of 2998.5 MHz. Table I lists operating parameters at the maximum energy available from the system.

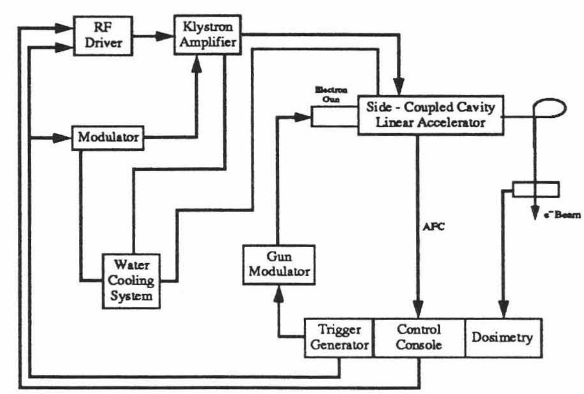

Fig. 1. Functional block diagram

TABLE I
Representative Operating Parameters

Operating Frequency	2998.5 MHz
Electron Energy	21 MeV
RF Power	4 MW
Accelerating Gradient	15 MV/m
Peak Surface Gradient	~30 MV/m
Kilpatrick Gradient	45 MV/m
Peak Injected Current	10 mA
Capture Ratio	<10%
Beam Loading	~1%
Beam Pulse Length	3 microsec
Beam PRF	135 Hz
Radiation Output	900cGy/min

A photograph of the accelerator is shown in Fig. 2.

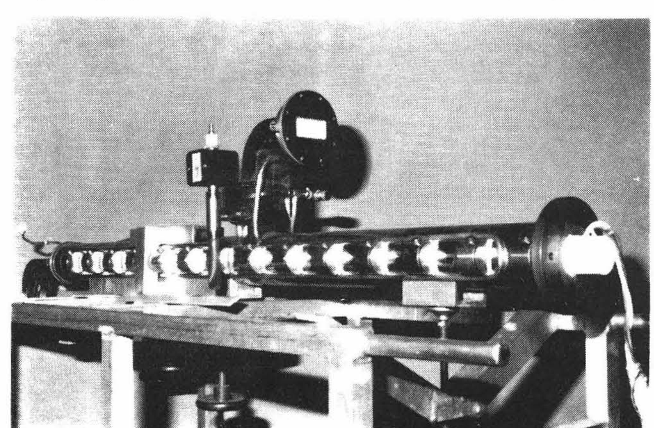

Fig. 2. Side-coupled cavity linear accelerator

A 2 l/s ion pump is provided for monitoring vacuum quality in the accelerator. Figure 3 shows many of the RF components in the gantry and in the rotating structure.

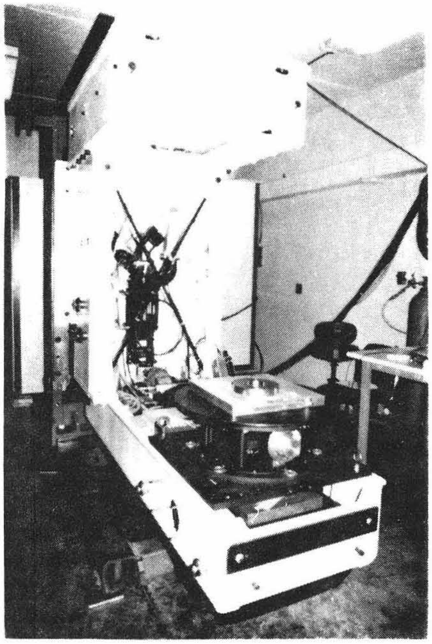

Fig. 3. Intraoperative radiation therapy machine assembly

Note that the current needed for this application (900cGy/min max.) is only 10mA, resulting in very light beam loading. A relatively low capture efficiency is also acceptable. The dose rate required for treatment is obtained by adjusting the repetition frequency.

When the beam leaves the accelerator it passes through a 270° achromatic bending magnet. This magnet serves as an energy filter and as a beam focusing element. Primary and secondary dose chambers monitor the beam, as well as flatness and symmetry, dose rate and dose per pulse for beam control purposes.

RF power is supplied by a Thomson CSF TH2066U klystron capable of 7.5 MW peak power. A circulator in the waveguide run insures that the tube always works into a well-matched load. The operating frequency of the accelerator is tracked by an AFC loop which controls the RF driver. The klystron modulator is a conventional line-type modulator[3] with a pulse transformer. The maximum operating voltage required by the tube is 160KV. Individual pulse energy is controlled by a command charging system consisting of a pass tube. In that way the length of time during which the thyratron has to hold off high voltage is minimized. Also, the energy per pulse can be maintained constant over a wide range of repetition frequency. A photograph of the modulator and klystron is shown in Fig. 4. Component cooling is provided by a closed loop water circulation system containing a heat exchanger to industrial water. The cooling system operates at ~40° C.

Description of Machine

The electron source to the rotation axis distance is 100cm, which is common in the industry. The isocenter height above the floor is adjusted at installation between 112 and 135cm. The gantry rotation range is ±115° from the vertical. For simplicity, and to insure reliability, all adjustments needed in the gantry to set up the system for treatment, except gantry rotation, are manually made. The primary scattering foils that must be in place for each energy setup are manually inserted by rotating a carousel. An interlock prevents a mismatch between the energy selected at the console and the foil chosen. A shaped secondary scattering foil is used to flatten the beam.[4]

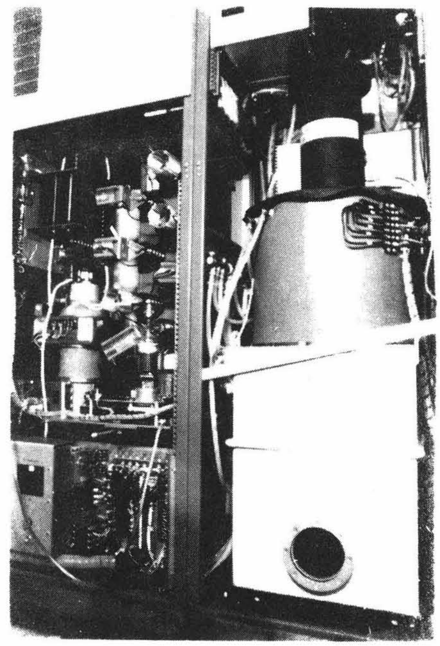

Fig. 4. RF system showing klystron and modulator.

Figure 5 shows a cross-sectional view of the entire machine including the gantry.

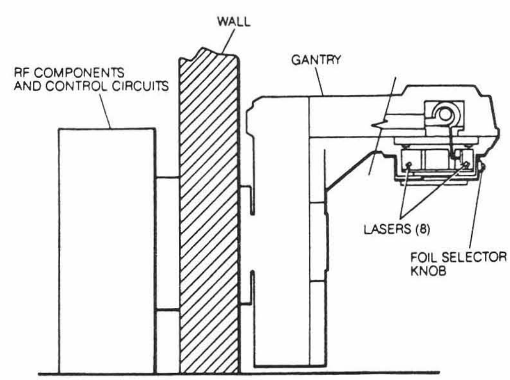

Fig. 5. Mevatron ME and control cabinets as installed

In Fig. 6, a view of the treatment head and the applicator cone may be seen.

The beam is collimated in two steps by round collimators attached to the annuli fixture and inserted into the accessory holder. The size of the precollimators selected depends on the field size needed for treatment. This machine uses a laser light alignment system so that there is no electrical or mechanical contact to the patient. This minimizes the chance for patient injury. The electron applicator cone is attached to the table side rails and can be prepositioned when the operating table is away from the machine or under it. The eight laser lights project dots to the surface of a round plate mounted on the applicator cone. When the tip of the cone is at isocenter and the electron beam axis and cone axis coincide, the eight dots coalesce to four dots equally spaced on a ring engraved on the surface of the alignment plate.

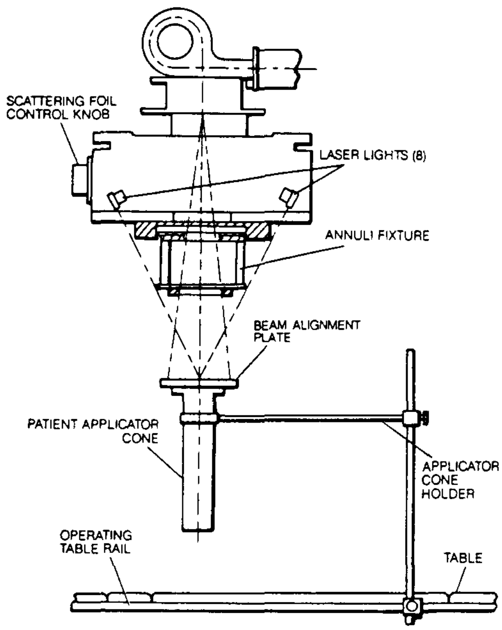

Fig. 6. Cross section of treatment head and patient applicator cone system

Operation of Machine

The operating conditions for the machine are shown in Fig. 7, where electron energy is plotted as a function of RF power. Note that the required energy range of 6 to 21 MeV is covered by varying the klystron output power from 1MW to 4MW. The klystron output power is adjusted by controlling the klystron beam voltage and current. The bending magnet current setting automatically corresponds to the energy selected.

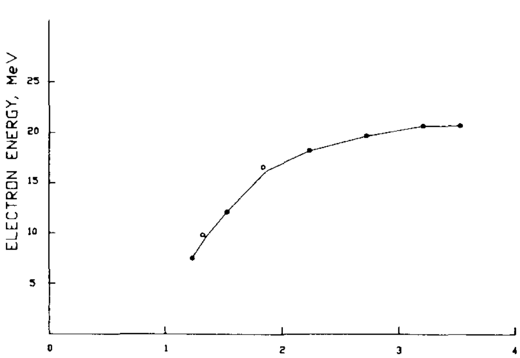

Fig. 7. Accelerator energy vs. RF power

For intraoperative radiation therapy procedures, smaller fields than are used in regular applications are sufficient. The maximum fields needed can be obtained from a beam flattened in a 20cm diameter field. Because the field requirements are smaller for this machine, thinner scattering foils can be used. The thinner scattering foils produce less X-ray contamination and contribute less energy spread in the beam at isocenter compared to thicker foils. The small energy spread results in a steeper drop-off of the depth dose curve. Figure 8 shows typical depth dose curves.

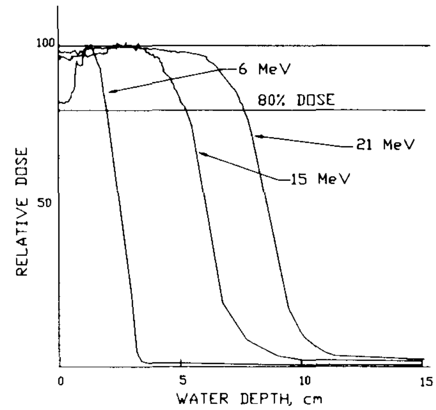

Fig. 8. Depth dose curves from water phantom

Some dose profiles for 15 MeV at a depth where the maximum dose occurs are shown in Fig. 9. These data were taken in a solid phantom and cylindrical applicator cones of the size indicated were used.

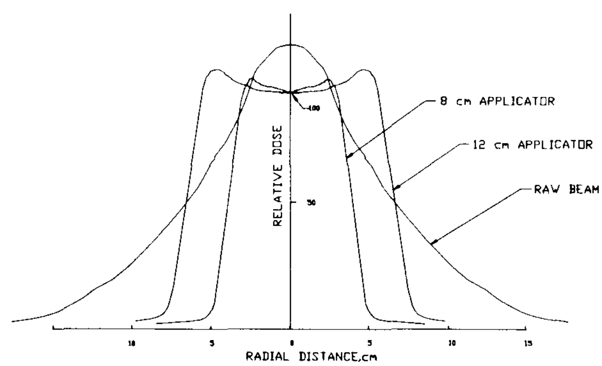

Fig. 9. Dose distribution with 15 MeV beam

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