ADVANCES IN MEDICAL ELECTRON ACCELERATOR THECHNOLOGY

Eiji Tanabe and John Ford
Oncology Systems, Varian Associates, Inc.
911 Hansen Way, Palo Alto, CA 95303

ABSTRACT

The election linear accelerator has become the dominant device for radiation therapy since it was first introduced in the early 1950s. Currently, there are more than 4000 medical electron liner accelerators in operation in the world, treating over 100,000 patients per day. This paper reviews the current status of radiation technology, and the anticipated requirements for radiation therapy in the foreseeable future.

INTRODUCTION

Cancer is the major cause of death in most developed countries. Approximately one in every four persons will develop a cancer at sometime in his on her lifetime. About half of all cancer patients in developed countries will receive radiation therapy, as definitive therapy, for palliation, or as an adjunct to surgery. Of those patients presenting with local disease, 56 percent will be cured. Although it is not generally recognized, the value to modern society of the linear accelerator technology is immense.

MEDICAL ELECTRON ACCELERATORS

The side-coupled standing-wave accelerator structure is most commonly used in medical accelerators, since it offers higher shunt impedance (as high as 120MΩ/m at S-band frequencies). Consequently, the accelerator length can be as short as 20 cm for 6 MeV energy gain (30 MeV/m gradient).

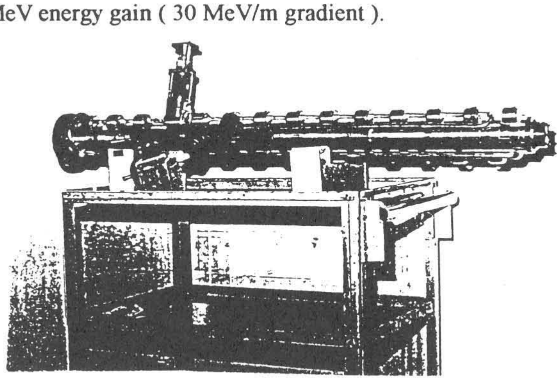

Fig. 1 Varian Clinac 2300 C/D Accelerator

Fig. 1 shows the Varian Clinac 2300 C/D accelerator with an "energy switch",which allows for the generation of high-output stable X-ray modes at widely separated energies, thus providing the full range of treatment capabilities.

Table I summarizes the accelerator design parameters. The variable coupling side cavity "energy switch" is located between the RF input coupler cavity and the adjacent accelerating cavity. This provides a means for varying the output accelerating gradient, while maintaining a constant accelerating field in the buncher section and tightly bunching the electrons, regardless of the output energy. Accelerators fully utilize contemporary computer technologies which have allowed radiation to be delivered to the patient as a function of time and space. Real time digital beam control has thus become very important. Optimum use of these technologies requires higher beam current (high dose rate) and transient pulse train stability, i.e. rapid stabilization of the beam after 'beam-on '.

Fig. 2 shows the architecture of a computer controlled accelerator system, employing multiple asynchronous parallel processing, which precisely coordinates all accelerator functions. Modern medical accelerators provide a variety of new capabilities that require greater performance from the beam generation system. For example, medical accelerators can now be equipped with high energy imaging devices that allow the clinician to visualize the anatomy through which the treatment beam has actually passed. Such imaging can be done in real, or

Table I

CLINAC 2300 C/D DUAL ENERGY ACCELETOR DESIGN PARAMETERS

STRUCTURE :	STANDING-WAVE SIDE-COUPLED	
ACCELERATOR LENGTH :		1.45m
NUMBER OF ACCELERATING CAVITIES :		28 + 1/2
FREQUENCY :		2856 MHz
EFFECTIVE SHUNT IMPEDANCE :		102 MΩ/m
Q_0 :		15000
COUPLING FACTOR K_1 :		0.04
ENERGY (MeV)	6	20
RF POWER (MW)	1.2	3.8
LOAD LINE (MeV)	9.8-30I	23-70I

pseudo - real time. Beam characteristics, and beam transport, to allow for the generation of small, circular X -ray focal spots have become much more important to insure optimum and uniform image resolution.

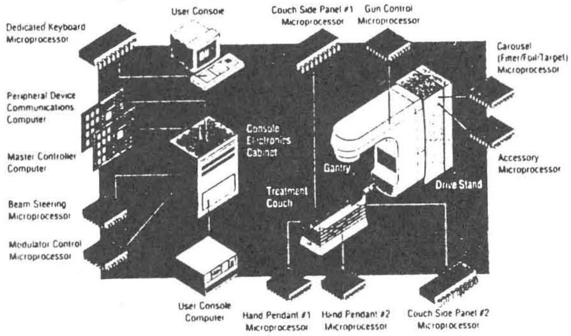

Fig. 2 Architecture of Computer Controlled Accelerator System

Modern medical accelerators utilize digital control electronics, which has allowed radiation to be delivered to the patient as a function of time, dose, or as a function of the mechanical configuration of the accelerator gantry, collimators and patient support systems. One example of time dependent dose delivery is 'physiological gating' wherein the beam is produced only over certain time intervals governed by some physiological indication such as respiration, in order to account for internal patient movement. Since only a small time interval may be available for beam delivery, high output is very helpful. Also, since the irradiation may be broken down into a series of short pulse trains, it is important that the accelerator stabilize itself rapidly, i.e. have good transient pulse train stability.

Even in medical accelerators using highly achromatic bending magnets to transport the electron beam from the accelerator guide to the X-ray target, the beam energy, or positional effects, can affect the X-ray field symmetry, flatness, and/or output. Ideally, one would like each rf power pulse to produce an X-ray burst with exactly the correct energy, field symmetry and flatness. Failing that, one would like the beam to stabilize after as small a number of pulses as possible. In conventional radiotherapy the first few seconds of 'beam-on' operation may constitute only a small fraction of the total radiation delivered. In dynamic beam delivery the treatment may be a series of short bursts of irradiation under different machine or patient configurations and beam-on conditions, transient effects can be significant.

DYNAMIC THERAPY AND MULTILEAF COLLIMATORS

The fundamental problem of radiotherapy is the delivery of a very high tumorcidal dose to the treatment volume with acceptable dose to normal tissue. This requires that radiation distribution be concentrated and shaped through the use of very sophisticated beam control techniques. A central strategy in radiotherapy is to use multiple beam directions so that the beams intersect to concentrate the radiation in a particular volume, while sparing the overlying tissue. For this reason, all modern medical linacs are so-called 'isocentric machines', which have rotating gantries that can guarantee that all beam axes pass through a common point in the plane of rotation of the X-ray target, called the 'isocenter'. The isocenter is typically positioned within the prescribed treatment volume.

Heretofore, it has been possible to concentrate the dose in prescribed volume, but it has been difficult to shape that volume. Computer-controlled dose delivery as a function of other parameters such as time or gantry position , so-called 'dynamic therapy', and 'conformal therapy' using such computerized beam shaping devices as the multi-leaf collimator, are now becoming commonplace in radiotherapy. This in turn puts new demands on the fundamental accelerator technology.

The delivery of radiation while the accelerator gantry rotates is a rudimentary form of dynamic therapy, traditionally called 'arc therapy', which has been done with analog accelerators for many years. A common problem in radiotherapy is 'hot spots' created by the oblique incidence of the beam on the patient surface, or created where two beams intersect. Traditionally, these problems have been handled by placing a wedge-shaped absorber in the beam to produce a wedge-shaped radiation field. Under computer control, a collimator block can be swept across an open field to produce an exquisitely shaped 'wedged' field. Such 'dynamic wedge' treatments are routine at many institutions.

The multileaf collimator, MLC, utilizes a number of beam absorbing bars, or leaves, to produce an irregular beam shape in the plane normal to the beam axis in order to 'conform' the treatment beam X-ray field to the 'beam's-eye view', a. two-dimensional projection of the prescribed three-dimensional treatment volume. The MLC contrasts with conventional collimators, which produce rectangular fields. This technology was pioneered in Japan in the pre-computer control era. Very clever designs were developed to change the beam shape as a function of beam direction using mechanical cams. Modern MLC's have drive motors for each leaf which operate under computer control. With such a device, the medical linac can deliver a series of exposures with a static conformal beam, or can operate in a dynamic conformal mode in which the beam shape changes as a function of other mechanical factors such as gantry rotation.

The MLC has been rapidly accepted in Europe and the United States as a labor saving replacement for customized beam blocks that are used to shield certain critical anatomical structures for static fields. But clearly, the MLC has great value in more sophisticated applications. Not only can the MLC be used to shape the beam as a function of beam direction, but it can also be used to modulate the beam intensity in the plane normal to the beam axis in a given beam direction . Using this approach one can, for example, account for patient surface irregularities to produce the desired dose distribution at the depth where the diseased tissue may be located. Through computer modeling, a series of two-dimensional intensity modulated beams can be determined that intersect to produce a particular three-dimensional dose distribution. The MLC has the advantage that a broad area can be irradiated at once, but has the disadvantage of a jagged field boundary due to the finite number of leaves.

In principle, smooth-boundary fields can be generated by sweeping fields made with the conventional collimator blocks under computer control. Such techniques generally demand much higher X-ray output . In general, dynamic therapy consumes time because extra time is required for the machine to 'reconfigure' itself during treatment. In some cases these reconfiguration intervals must take place while the beam is off, making the treatment less time efficient. In other cases, the beam may remain on but motor speeds or dose gradient requirements made also lead to time inefficiencies. Higher dose gradients. Here again we see the need for greater beam currents.

And again, we come back to the need for transient pulse train stability. For fine control, the accelerator should be able to produce a very small increment of radiation that meets all the symmetry, flatness, energy (depth dose) and instantaneous dose rate requirements in a precise and fully reproducible way, independent of accelerator orientation. This minimum increment of radiation must be available on demand by the controlling computer for any given accelerator orientation, such that the specifications for any sum of radiation increments do not differ.

This dynamic therapy requirement has profound implications for fundamental accelerator design; otherwise, serious compromises have to be made. For example, for accelerators that take a relatively long time to stabilize once irradiation is initiated, situations arise where the beam must be reduced in output to approximate a beam-off condition, which defeats the basic objective of dynamic beam delivery. Minimum dose increments, on demand, that meet the above criteria with the highest possible maximum instantaneous dose rate capability, provide the maximum flexibility.

MEDICAL ACCELERATORS IN THE FUTURE

The goal of curative radio therapy is to sterilize the primary tumor cells without damaging surrounding normal cells. The margin for error in dose can be quite small and the therapist need to be able to deliver a dose distribution with an accuracy of a few percent.

Meanwhile, more than 40,000 of medical accelerators will be needed in the future to treat a world-wide population of 5 billion people. In order to achive this goal, the medical accelerators must be not only efficient and lower in cost but also flexible, stable, reliable and precise.

Table II summarizes the major design criteria for future medical accelerators. Higher dose rates (1000 rad/min) with a wider range of beam energy (2 to 25 MeV) will be available gated on in synchronism with low velocity intervals for periods in organ motion while maintaining acceptable treatment times. Also, high spacial precision, typically within 0.5 mm, with higher dose is mandatory for stereotactic radiosurgery. However, accurate positioning of patient and accelerator may not be sufficient to provide accurate dose delivery at the target volume, since most of organs are dynamic and do not remain stationary during a finite treatment period. In order to accurately deliver dose at the target volume, it will be necessary to monitor the position and motion of the target in real time, and feed back this information to control the accelerator system. Thus real time imaging and treatment planning will be needed in the near future in order to obtain true dynamic treatment or stereotactic radiosurgery.

Table II

CRITERIA FOR FUTURE MEDICAL ACCELERATORS

ENERGY	2 MeV to 25 MeV
DOSE RATE	1000 MU/min
BEAM SPOT SIZE	1 mm
SPACIAL PRECISION	0.1 mm
FIELD SIZE	1X1 mm to 40X40 cm
FLATNESS & SYMMETRY	1 %

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