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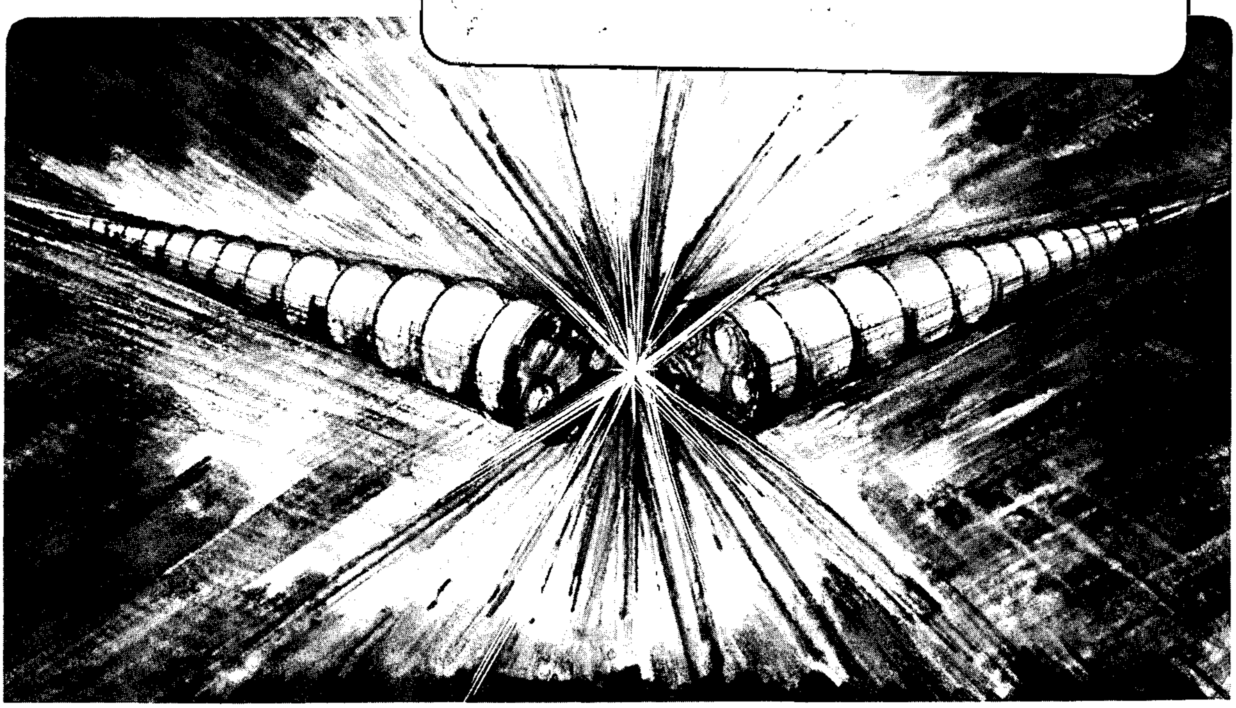

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Abstract

The advantageous physical characteristics of slowing-down and stopping charged particle ion beams have been demonstrated to be highly desirable for application to radiation therapy. Specifically, the prospect of concentrating the dose delivered into a sharply-defined treatment volume while keeping to a minimum the total dose to tissues outside this volume is most appealing, offering very significant improvements over what is possible with established radiation therapy techniques. Key to achieving this physical dose distribution in an actual treatment setting is the technique used for delivering the beam into the patient. Magnetically scanned beams are emerging as the technique of choice, but daunting problems remain still in achieving the utmost theoretically possible dose distributions.

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Introduction

The use of charged particles in radiation therapy offers the potential for very significant improvements in the quality of treatments. Properly utilized, the superb dose localization which is possible with beams of protons or heavy ions can lead to treatment techniques and clinical results which for many treatment sites are potentially far superior to what is possible today with photon beams. The main physical property of these charged particles which makes this possible is their stopping characteristics; the rate of energy deposition as the particle slows down increases dramatically as it nears the end of its range. This energy loss property, summarized in what is known as the Bragg Curve, indicates that it is possible by stopping particles inside a target volume to deposit most of the radiation dose directly in the target volume, with minimal damage to tissue traversed on the way in, and incidentally with little or no dose beyond the target volume, since there is no primary particle flux in this region. ("Tail" dose beyond the Bragg Peak arises only from fragments of nuclear reactions of the primary beam, which are essentially nonexistent for protons, but can contribute significantly for the heaviest ions at longest ranges, such as 30 cm argon.)

Unfortunately, placing stopping particles at all points in the target volume is a tremendously difficult problem if one wants also to stop none of the beam in adjacent normal tissue. Difficult, but as we shall see not impossible. The crux of the problem lies in that target volumes are in general irregular, three-dimensional objects.

Beam Delivery Considerations

Beams from accelerators are generally very small, and quite monoenergetic, so that with no added devices the volume in which particles stop will be very small, typically about one cubic centimeter or less. (The size of this volume depends on the depth of treatment and the multiple scattering and range straggling properties of the ions used; the characteristic volume, denoted as a "voxel", for heavy ions such as neon will be of the order of ten times smaller than that for protons.)

Tailoring the beam stopping volume to that of the target volume requires spreading the beam out laterally, and adjusting its energy to vary the penetration depth. Lateral control of the beam can be done either with scattering of the beam, which makes the whole beam spot larger, or by magnetic deflection. Range adjustment is done with a variable absorber, or by varying the energy of the beam emerging from the accelerator. To place stopping particles in an arbitrary volume, independent control of the stopping point and lateral coordinates is required.

Passive Beam Delivery

One largely-used technique for performing large field treatments is with scattering foils and ridge filters [1,2]. Scattering spreads the beam into a Gaussian shape, not uniform enough for treatments in of itself. This distribution can be flattened very effectively by the use of blocking rings placed in the path of the beam downstream of the scattering foil. (Fig 1a) Circular fields up to 20 to 30 cm in diameter can be formed by this technique. Conformation of the lateral extent of the beam with the target volume is accomplished with collimators shaped to the target outline. While effective, this technique has some drawbacks. Beam utilization is low, typically only 10 to 25% of the beam actually reaches the suitably flat treatment field, the rest is lost on the blocking rings or is outside the flat area and must be collimated out. This loss of efficiency reduces the available dose rate, and also increases neutron background in the treatment room.

Depth modulation employs ridge filters, material of non-uniform thickness which the beam must penetrate on its way to the patient. Particles traversing thinner regions of the filter lose less energy and reach farther into the treatment volume, those going through thicker regions stop sooner. Ridge filters can be fabricated from brass, with a properly shaped groove machined into the surface, or lucite propellers with wedge-shaped blades. They are placed far enough upstream, and are rotated constantly so that all areas of the treatment field see the full energy modulation of the beam. The depth of the groove is adjusted for the desired depth modulation, typically a whole family of such devices will be available to cover different target thicknesses. Additionally, the shape of the groove must be carefully calculated to ensure uniform dose delivery to all depths of the treatment field.

This delivery system, used with protons [1,3], and helium ions [2], is termed a Passive system, as the filling of the treatment volume is done entirely with elements placed in the beam and requires no active control over the beam position or energy. This can be a great advantage, the inherent simplicity of the system and lack of need for sophisticated beam position and intensity controls enhances confidence in the accuracy and reliability of treatments. The disadvantage of the system is that the treatment volume must of necessity have cylindrical symmetry. There can be no correlation between lateral position {x,y} and depth modulation {z}, all elements of the treatment volume will see the same modulation of stopping points. Thus to treat an average irregular target by enclosing it into a cylindrical volume will mean including much normal tissue in the treatment volume. An irregularly shaped collimator can contour the outer edge of the field to correspond to the boundary of the target, and a machined bolus placed on the surface of the patient can adjust the deepest stopping point of the beam to conform to the distal edge of the treatment volume (see Fig. 2), but the maximum thickness of the target still determines the range modulation function across the entire treatment volume and treating thinner parts of the target must involve the irradiation of normal tissue.

Active Beam Delivery Systems, Overview

Overcoming this limitation requires abandoning the purely passive delivery system in favor of one where more active control of beam parameters is possible. The most flexible system would have totally independent control of lateral coordinates, as well as depth. Lateral control is accomplished with magnetic deflection, two dipole magnets placed in the beamline with orthogonal magnetic fields capable of deflecting the beam to any {x,y} location. Such systems are generically termed Scanning Systems, specific examples are shown in Figure 1b-e. In the Wobbler system [4] (Fig 1b) the magnets are driven sinusoidally and 90° out of phase so the beam describes a circular pattern on the treatment field. Adjusting the amplitude of the driving sine wave controls the radius of the circle. The

Lissajous scanner [5] (Fig 1c) is a variant of the Wobbler, the magnets are driven with saw-tooth waves of unrelated frequencies. The Raster Scanner [6] (Fig 1d) has a fast sweep saw-tooth drive and a slow sweep single pass for each spill from the accelerator. The Pixel Scanner [7] (Fig 1e) dwells at one spot delivering the desired dose at that spot before moving to the next spot. These systems all provide great flexibility in treatment delivery options, with prospects of good improvements in the quality of treatments which can be delivered. However, the added complexity of these systems raises issues of safety and control. Exquisite control over magnet currents and beam intensity is needed, in the ultimate case well beyond the state of the art, while safety is an issue because if a failure of the magnets occurs and the beam does not move off of one coordinate a truly significant overdose can be delivered to that point.

Very fortunately, development of scanning system technology can proceed in an incremental fashion, one does not have to solve all of the problems to begin benefitting from the positive attributes of such systems, such as increased beam utilization and reduction of background radiations. The yardstick for this development is the spot size of the beam which is scanned and modulated by the delivery system. Ultimately, one would wish to use the smallest possible voxel, namely the intrinsic size of the beam in the target volume, but by using a larger spot size many of the problems become simpler. These problems can be placed into two categories: lateral uniformity, and depth control necessary for achieving 3-dimensional treatment volumes.

Lateral uniformity

Lateral uniformity relates to matching the edges of voxels at each depth of the treatment volume. If edges were sharp, any misalignment would cause dosage at the edge to be either double or zero, both unacceptable. One way of avoiding the edge-matching problem is to sweep the beam continuously across the field. Systems using this technique are the Wobbler [4], the Lissajous scanner [5] and the Raster scanner [6], in which the entire field is painted over many times, a small fraction of the dose being delivered on each pass. In such systems the edge-matching problem is reduced to ensuring uniformity of beam spill during the sweep time. Microstructure in the beam can be washed out if the spot size is large and the sweep speed is slow, and if many passes of the field are made. Care should be taken, though that the sweep rate is non-synchronous with any beam structure, to prevent reinforcement of intensity variations. A good analysis of this is given in Renner and Chu's paper [4].

The principle behind a pixel scanning system [7,8] is to dwell at a given set of coordinates until the desired dose is delivered there, then to move the beam spot (the pixel element) to the next set of coordinates. In this case, matching edges is critical. As given in the analysis of Chu and Kuenning [9], such edge matching is determined by the falloff of the beam intensity at the edge of the scanned spot; which should be set by the precision and stability with which the coordinates of the spot can be set. A fuzzy spot would seem to be indicated, but the edges of the overall treatment volume are determined by this same falloff, and making this falloff too large negates much of the advantages of this modality of therapy.

One issue with pixel scanning is the dose rate and control problem. Analysis [7] of scanning with the smallest possible spot size indicates that very high instantaneous dose rates, possibly as high as 10^4 Gy/second, as well as very short dwell times at each voxel, around 100 μsec, will be necessary to cover the treatment field in about 1 minute. Not only are precision of dosimetry and overall control and magnet slewing rates critical problems, but even the biological effects of such high instantaneous dose rates are an important unknown. By scanning with a larger spot, however, dose rates and slewing speeds drop dramatically, as the number of voxels drops as the cube of the edge length. Again, though, the larger the spot size the more is lost in edge-definition for the total treatment field, so ultimately it will be important to address the issues associated with the smallest spot sizes.

Strategies for 3-dimensional treatment fields

As stated above, a 3-D treatment field requires independent control of {x,y} coordinates as well as depth {z}. The pixel scanning system achieves this. Each arbitrarily-shaped layer of the treatment volume is irradiated with adjoining voxels, then the range of the beam is incremented to treat the next layer with its different outline. Such a system has been built and tested at NIRS [10] with a 70 MeV proton beam. Voxels of 1 cm square sides and about 2 mm depth (small owing to the limited range of 70 MeV protons) have been scanned to irradiate irregular test shapes, with very good overall uniformity and control.

Another technique which doesn't share the edge-matching problems utilizes a multi-leaf variable collimator along with a beam-spreading system and a controlled range-shifter [11]. If the field size can be adjusted easily during the treatment, the shape of each layer of the treatment volume can be set, this layer painted, then the range adjusted for the next layer, and the process repeated till the entire volume is treated. Achieving uniformity of treatment to the whole volume can be done as shown in Figure 2. A bolus shapes the distal edge of the volume to conform to the largest field size. After treating this the range is shortened and the field size is reduced to treat the next shallowest depth. By this strategy each area of a given layer will have the same ratio of slowing down ions as stopping ions. In contrast, dividing a volume into right circular disks will see the central portion of a middle disk having dose contributions from slowing-down particles while the outer edges will have not received any dose. Placing stopping particles uniformly into such a disk will produce a hotter spot in the center. (Note that this can be overcome by a non-uniform irradiation, moving the beam spot more rapidly over the central region so less dose is delivered there.)

Achieving fine enough shape-gradations for the treatment volume contours requires a large number of independently-controlled leaves, and a high bulk and cost for such a device, but these problems are perhaps somewhat more tractable than those of the pixel-scanning systems. If the above-mentioned bolusing technique is used, then any of the simpler uniform-field generation systems can be used, such as the Wobbler, Raster scanner, or even a scattering system.

A possible compromise system which may simplify the engineering problems of the multi-leaf collimator is known as the Line-Scanning system [12]. A single scanning magnet sweeps the beam back and forth in the {x} direction across a defining aperture of fixed {Δy} whose limits {x(min), x(max)} are adjustable (with a single-leaf collimator) according to the field width at that value of {y}. When the correct dose is given the {y} coordinate is incremented by {Δy}, probably by translating the patient, new {x} limits are set, and the next slice is dosed. When the entire layer is completed, the beam range is incremented and the process is repeated for the next layer. Although simpler from an engineering standpoint, treatment times will probably be longer because of all the motions involved. Nevertheless, the goal of a 3-dimensional treatment volume will have been achieved.

Medical Beam-Scanning Research Centers

Several laboratories are actively engaged in developing the techniques described above, and by way of acknowledgement the highlights of the programs at these centers will be briefly described. At the Harvard Cyclotron Laboratory, (160 MeV proton synchrocyclotron, Cambridge, MA) Koehler and Gottschalk have been most active and successful in developing the scattering foil techniques [1,3]. Similar development work at LBL, Berkeley CA was undertaken by Lyman et al [2] at the 184" synchrocyclotron, and by Alonso and Criswell at the Bevalac [20]. Pioneering work on pixel scanning systems has been going on at NIRS (70 MeV proton cyclotron, Chiba, Japan) under the leadership of Kawachi and Kanai [8,10], who have also developed manually controlled multi-leaf collimators. Larsson and Brahme at the Gustaf Werner Institute (185 MeV proton synchrocyclotron, and now 250 MeV proton synchrotron, Uppsala, Sweden) have developed the Lissajous scanning technique [5], and have developed algorithms for effective pixel scanning techniques [13]. Pi meson delivery techniques have been developed at four different laboratories. A novel "Pion concentrator" developed first at Stanford, CA [14], then at SIN [15](now PSI/SIN, Villigen, Switzerland) takes up to 60 channels emanating radially from the production target and directs these pions through a cylindrical system of magnets to a convergence point in the patient. The pion channel at LAMPF (Los Alamos, NM) [16] utilized a simplified version of the line-scanning concept. TRIUMF (Vancouver, Canada) [17] has developed a full pixel-scanning system, with 2.5 cm pixel size. At the LBL Bevalac

(2.1 GeV/amu heavy ion synchrotron), Chu et al have developed the Wobbler system [4], and the Raster Scanner [6].

Research activities continue at Harvard, LBL, Uppsala, and NIRS, while new initiatives are being undertaken at other centers around the world. SIN is developing a pixel scanning system for a new proton therapy beamline from their cyclotron [18], and for the Loma Linda University Medical Center (Southern California), Coutrakon [19] is designing a raster scanning system for installation in at least three of the treatment rooms in their medical accelerator (250 MeV protons) complex scheduled for completion in 1990. According to Hirao and Kawachi [21], HIMAC, a medical heavy ion synchrotron facility at NIRS scheduled to be operational in 1993 is planning on installing Wobblers and eventually pixel scanners in their three treatment rooms with both horizontal and vertical beams.

With all of this activity, and with the driving force of new facilities coming on line which will benefit from developments in beam delivery systems, it is certain that progress in the field will be rapid, and that the full potential of charged-particle therapy will be clearly realized in the not-too-distant future.

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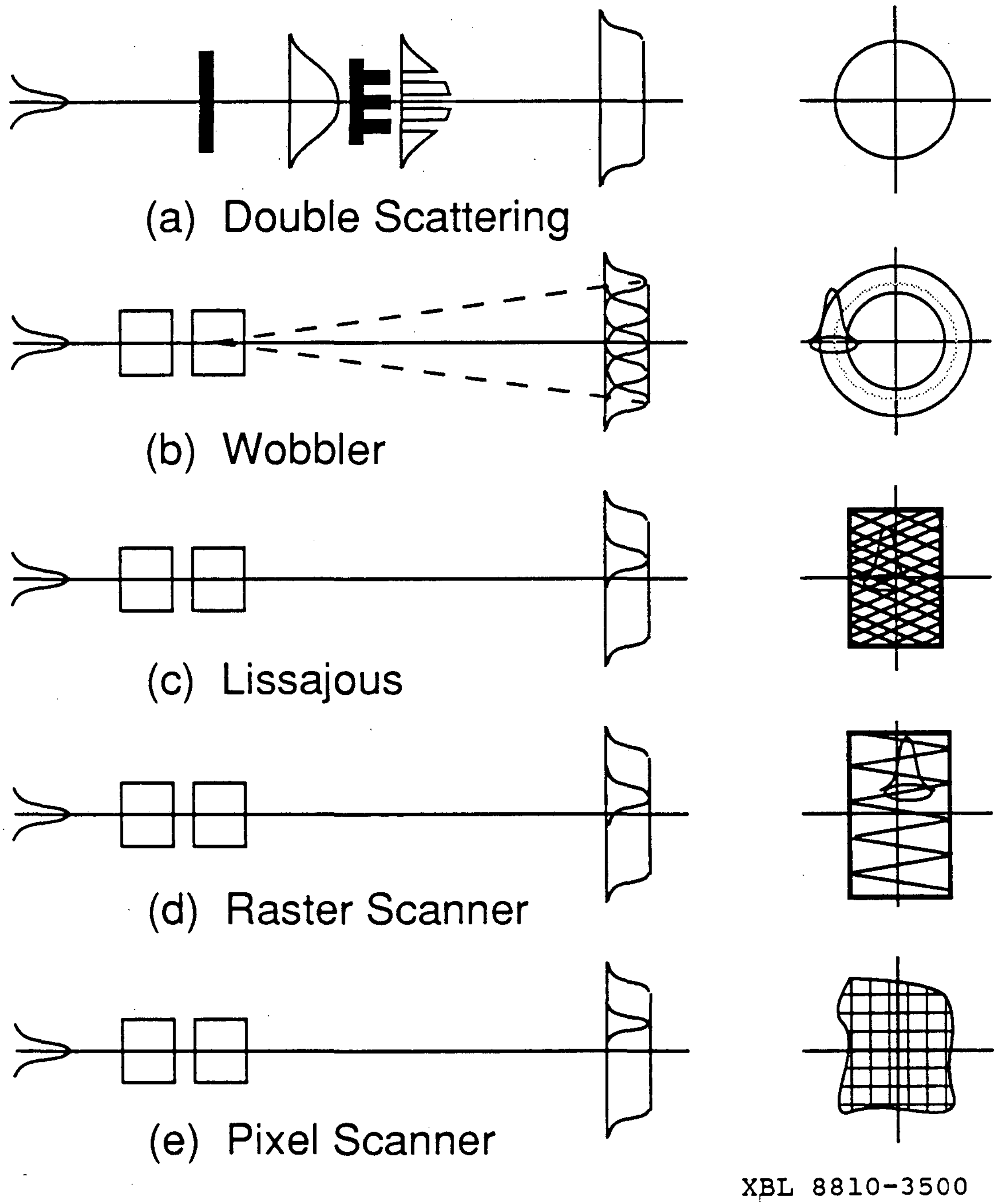

Figure 1. Schematic diagrams showing the principles of five different beam spreading systems: (a) Double scattering, (b) Wobbler, (c) Lissajous scanning, (d) Raster scanning, (e) Pixel scanning.

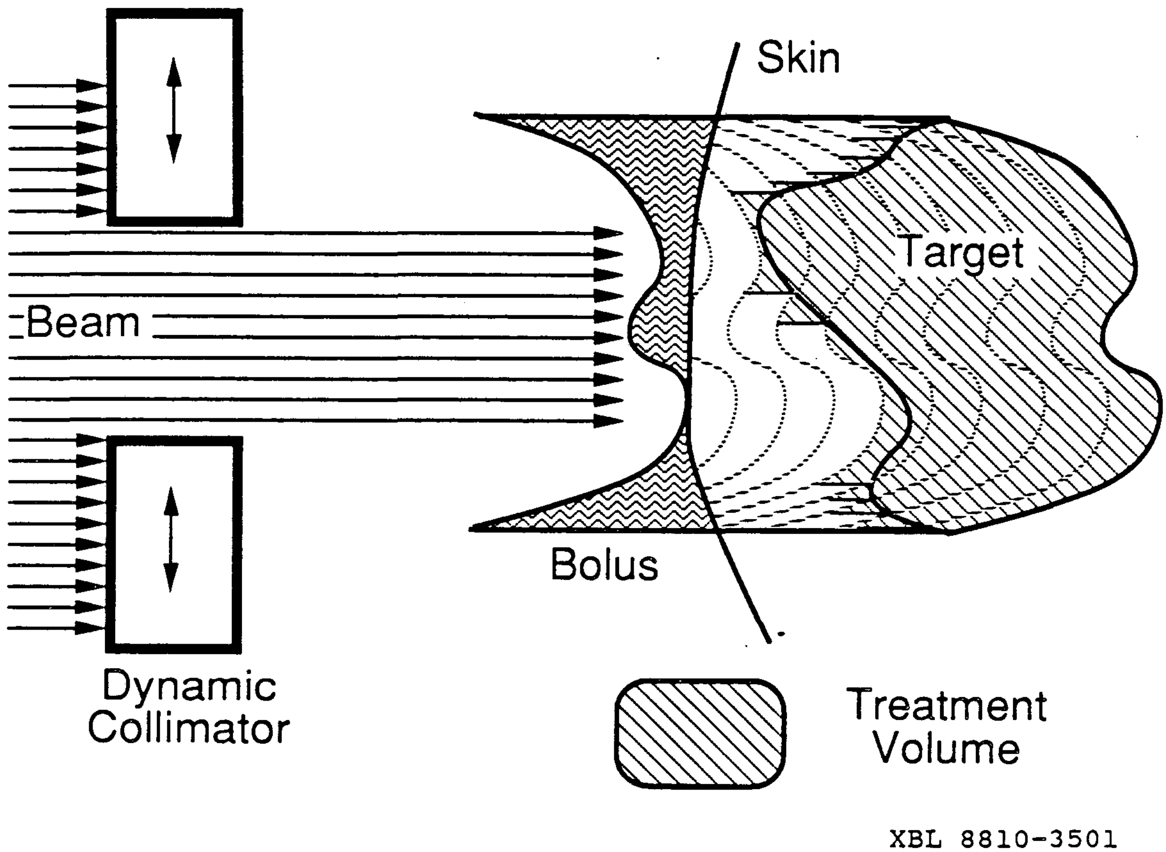

Figure 2. Section of a 3-dimensional treatment volume obtained with a uniform beam spreading system with a stepped range-modulator (constant-depth contours), a bolus compensator on the patient surface, and a dynamic multi-leaf collimator. Very little normal tissue is included in this treatment volume.

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