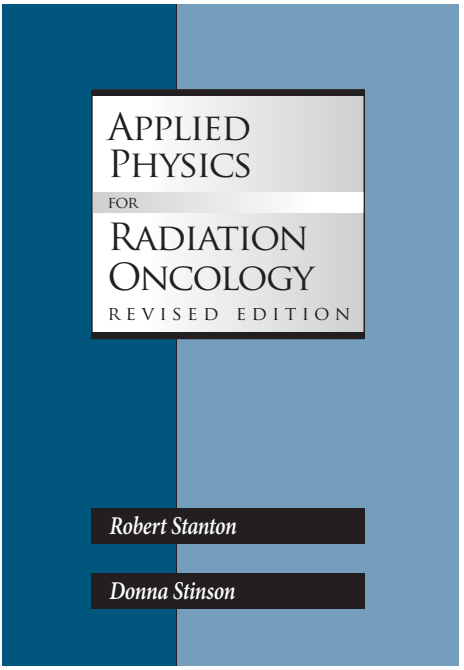

Here is a sample chapter from this book.

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Preface

In this edition of *Applied Physics for Radiation Oncology*, the authors have tried to maintain the basic character of our earlier text as an introduction for radiation therapists. Most of the basic physics chapters remain unchanged, as the basic principles have not changed, and because of our belief that therapists should be taught physics principles to help them understand the technologies they apply to patients.

We have both expanded several chapters to add new techniques and removed some sections of only historic interest in others.

The collaborative team that produced this book includes those who helped produce previous versions including Rodger Holst, Joyce Keil, and Michael Nunno and a new contributor, Daniel Januseski, who updated the radiation safety chapter. We, the authors, guided the entire project and hope that our years of experience helped produce a textbook that is easy to read and helps students to learn.

Robert Stanton
Donna Stinson
September 2009

14

Treatment Planning

- **Tumor Targeting Vocabulary**
- **Aims of Treatment Planning**
- **What Treatment Planning Includes**
- **Patient Alignment Devices**
- **Patient Positioning Aids**
- **Body Contours**
- **Isodose Distributions**
- **Oblique Incidence Corrections to Isodose Distributions**
- **Isodose Summations**
- **Treatment Techniques**
- **Stationary or Fixed Beam Treatment**
- **Moving Fields Treatment**
- **Tissue Inhomogeneities**
- **Tissue Compensation**
- **Wedge Filters**
- **Standard Treatment Calculation**
- **Beam On Time Calculations (Timer Settings)**
- **Monitor Unit Calculations**

Student Objectives

1. List five types of target volumes and explain the importance of each.
2. Identify three aims of treatment planning.
3. Describe the process of: visualization, localization, and field selection and placement.
4. Describe field verification and documentation.
5. State the equations for timer and monitor unit calculations, and identify the type of equipment used for each.

Treatment planning is the process of determining the best method of treating a tumor with radiation. The major objective of treatment planning is to ensure that the tumor receives a uniform radiation dose while healthy tissue and critical structures are protected. Other important objectives of treatment planning are to develop reproducible setups and maintain patient comfort.

Treatment planning includes determining the volume to be treated and then designing appropriate radiation fields to treat that volume. It begins before the first radiation treatment and continues throughout a course of therapy to ensure that the intended plan is being implemented. A treatment plan may sometimes be changed during a course of therapy to compensate for changes in a patient's condition.

Professionals from several areas work together in developing the best treatment plan for each patient. These professionals include radiation oncologists, physicists, treatment planners (dosimetrists), and radiation therapists.

TUMOR TARGETING VOCABULARY

All of the individuals involved need to use the same vocabulary when discussing and planning patient treatment. **Tumor dose** (**TD,** or **prescribed dose**, $\mathbf{D_{Rx}}$) is the dose to be delivered to the disease being treated. The patient "work-up," defines the **gross tumor volume** (**GTV**), the palpable or visible extent of tissue that makes up the tumor. The **clinical target volume** (**CTV**) is the gross tumor plus a margin to include any areas of direct but subclinical disease spread. During the process of treatment planning, the **planning target volume** (**PTV**) is determined. It includes the clinical target volume plus the needed margin to ensure delivery of dose to the target when motion, setup variations, etc. are taken into account. Determining the tumor and target volumes are two of the key clinical decisions of the radiation oncologist. Once the treatment plan is developed, two other volumes can be determined. **Treated volume** (**TV**) is the volume of tissue that actually receives the tumor dose; it may be larger but should not be smaller than the clinical target volume. The **irradiated volume** is the entire volume of tissue hit by any portion of the radiation aimed at the tumor that receives a significant absorbed dose. See figure 14.1 for illustration.

AIMS OF TREATMENT PLANNING

Treatment planning is often a challenging process during which teamwork is required to achieve optimal and reproducible radiation treatments.

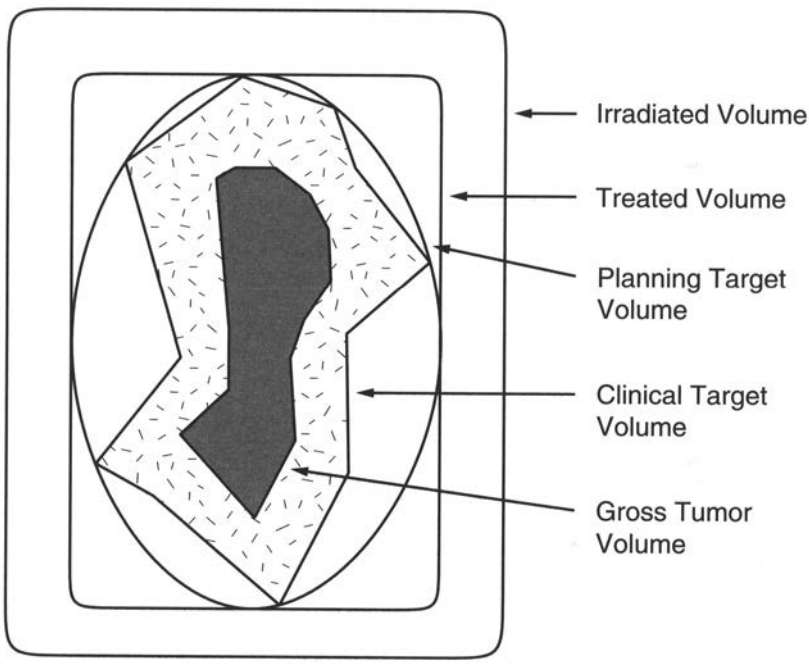


Figure 14.1
Illustration of various volumes according to ICRU 50.

There are three aims of treatment planning:

- Develop a plan that treats the tumor volume. This plan should give as homogeneous a dose distribution as possible throughout the clinical target volume.
- Minimize radiation dose to healthy structures. Areas outside the target volume should receive as little radiation as possible. Limiting dose to healthy tissues requires knowledge of how much radiation the tissues in the treatment field can tolerate. Dose levels to critical structures such as the spinal cord, kidneys, and the lens of the eyes must be minimized.
- Provide a permanent record of dose calculations and distributions so that others may, in the future, understand the treatment plan. This permanent record is not just the computer printout of a dose distribution; it includes the prescription, diagrams of the treatment field, tattoo identification, simulation and portal films, as well as other computations performed before, during, and even after each course of treatment. This record may become important if the patient needs to receive more radiation in the future.

WHAT TREATMENT PLANNING INCLUDES

Treatment planning includes visualization, localization, field selection and placement, and verification.

Visualization is the determination of the location and extent of the tumor, particularly with respect to anatomical landmarks. This process uses every reasonable method of examination, including palpation, radiography, or radiographic imaging (CT, PET-CT, MR, ultrasound, mammograms, etc.) and, when possible, direct visual examination. All the information gathered in these various examinations is correlated by the radiation oncologist, who uses it to define the volume to be treated in each patient. **Fusion** techniques may be used to combine images from more than one modality. When technology is used to overlay a diagnostic image, such as a PET-CT, with a treatment planning image, the images are described as fused.

Localization, or **simulation,** is the radiographic determination of the field borders required to encompass the clinical target volume. Ideally, localization is performed using a **simulator**—radiographic (imaging) unit that simulates all the movements of the linear accelerator or Cobalt-60 treatment unit and matches its geometry, distances (SSD, SDD, etc.), beam divergence, and field size. A simulator produces high-quality images of the treatment field.

During simulation, the patient is placed in as comfortable a position as is consistent with good beam placement. Accurate, reproducible positioning is critical since positioning during localization is almost always the position the patient must maintain throughout the course of treatment. During traditional radiographic localization, **orthogonal images** (those taken at right angles to each other) are frequently taken so that all three dimensions (x, y, and z planes) of a patient's anatomy can be documented. Sometimes these orthogonal beams are actually used for treatment. Other times they are used solely as a step in tumor localization.

CT-simulation has largely replaced radiographic simulation. **Digitally reconstructed images (DRR)** are produced by using data acquired by the CT in the axial plane and translated to another plane (i.e., sagittal plane).

Cone beam CT (CBCT) is a technique that uses electronic portal imaging associated with modern linear accelerators. The megavoltage energy of the accelerator, along with an amorphous silicon flat panel, is used to reconstruct an axial image. These images are used to document accuracy of treatment plans at the initiation of therapy and/or as a method to monitor anatomy, including tumor size, throughout a course of therapy.

The next step is **field selection and placement.** The treatment planner selects various field sizes and beam placements to irradiate the specific tumor volumes. The dose contributions to the target as well as other organs are evaluated

by the planner and used to choose the best combinations of beams. The chosen plan is then documented, listing the dose contributions to all important structures.

The final step in treatment planning is verification. The goal of **verification** is to ensure that the volumes planned for treatment are actually the areas being treated. This takes place on the actual treatment unit. Prior to the first treatment, radiographic documentation of the treatment field is necessary. Radiographic films, known as **portal images**, are made with electronic methods, or with film and/or cassettes designed for high-energy exposure. Film-holding cassettes often include a metal screen that produces electrons through high-energy photon interaction. These compton electrons help produce the radiographic image. Although portal films are of poorer overall image quality than other radiographic images, they contain sufficient information to determine the accuracy of beam placement.

Verification images are often repeated on a weekly basis during each patient's treatment course to ensure continued correct field placement. New electronic portal imaging devices, similar in function to x-ray fluoroscopes, are also used for this purpose.

Documentation is critical throughout every step of treatment planning. All the data derived from visualization, localization, field selection and placement, and dosimetry calculations must be precisely recorded. Complete and accurate documentation of the details of each treatment is required so that accuracy and consistency, which directly influence the success of every patient's course of therapy, are maintained.

PATIENT ALIGNMENT DEVICES

To achieve accurate alignment of the radiation fields on the patient several devices are used.

Field lights are the most essential beam alignment devices, because the light field on a therapy machine corresponds geometrically to the radiation field to an accuracy of ± 3 mm. Tattoos and field edge marks on the patient are aligned with this light field before each treatment.

The **protractor**, or gantry angle indicator, is another alignment device. Just as every beam has a particular size that is defined by the field light, there is a particular gantry angle that is read on the machine protractor. This is read either mechanically or from a digital readout.

Lasers are commonly used as positioning aids but are not actually part of the treatment machine. There should be at least one laser on each side of the patient (referred to as lateral or side lasers) and one pointing down at the machine isocenter from the ceiling (referred to as an overhead laser). A sagittal laser also

projects a straight line and points toward the isocenter along the longitudinal axis of the patient or table. Lasers are used in daily patient setup, and their alignment should be checked daily as a part of the quality assurance program. Daily alignment of lasers ensures reproducible and accurate positioning.

Back pointers are mechanical devices that, when mounted on the gantry, point to the exit point of the beam's central ray; that is, they indicate the position at which the radiation will exit the patient. Historically, back pointers were particularly useful in positioning parallel opposed tangential fields, e.g., for breast treatments, but have been replaced by laser positioning technology in the modern treatment environment.

PATIENT POSITIONING AIDS

Once the location of the tumor is ensured and the patient has undergone the simulation process, it is necessary to position the patient accurately every day in the same treatment position. The use of patient positioning aids helps the patient maintain the proper treatment position without sacrificing comfort.

Many such devices are available. One of the most simple is adhesive tape used in conjunction with specially contoured head holders for head and neck treatments.

Better precision in positioning is achieved with individual **bite blocks** (custom mouthpieces). These are devices with calibrated bars and support arms attached to the patient using a custom mouthpiece, as shown in figure 14.2. When the patient is positioned, he or she bites down on the bite block, positioning his or her head and neck rigidly in the correct position.

There are several other patient positioning devices used. Specially cast patient supports made of plaster, thermal plastic materials, or cast Styrofoam materials can be used to make rigid supports for the patient. See figure 14.3. For most accurate treatment, these positioning devices should be fabricated before simulation so that the patient is in an identical position throughout the whole planning and treatment process.

BODY CONTOURS

A **body contour** is a precise outline of a patient, usually in a transverse plane that includes the region of the tumor. A contour is required both for computing and for displaying the dose distribution in the treatment volume. Frequently done at the time of localization, a contour is usually taken at the central axis plane of the fields but may be taken at other levels as well to evaluate the dose distribution.

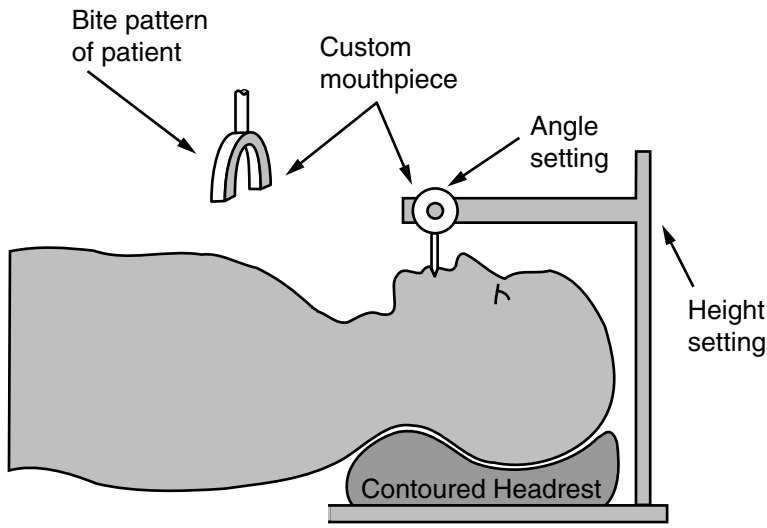


Figure 14.2

Bite block device (custom mouthpiece). A contoured headrest is fixed in position in a calibrated and adjustable holder. Each patient has an individual mouthpiece that fits into this device. When all of the settings on the device are matched with those determined at simulation, the patient position exactly reproduces that held during simulation.

Figure 14.3

Patient in Alpha Cradle immobilization device. The patient, once placed in this custom formed device, is held in a reproducible position for daily treatments. Photograph reproduced with permission from Smithers Medical Products, Inc., North Canton, Ohio.

There are several ways of **contouring** a patient, i.e., determining the patient's shape quantitatively. The best method to use is a CT scan, because it provides richly detailed information about external and internal anatomy. This method gives a three-dimensional rendering of the patient outline, allowing three-dimensional treatment planning with much freedom in the choice of radiation beam orientations. However, due to the small opening of most CT scanners (which usually ranges from 60 to 80 cm), CT scans are not always helpful in indicating the actual treatment position (especially in a breast setup). Large bore CT scanners help to resolve this issue.

A less commonly used and less expensive method of contouring is plaster of Paris bandage material. A strip of bandage at least four layers thick is wet completely, squeezed out, and then placed over the patient in all areas through the central axis plane of all the radiation fields to be used. While the plaster dries (which takes several minutes), the exact locations of portals and other important information such as name, date, and patient dimensions through beam centers are recorded on the contour itself and/or on an accompanying record.

Other means of contouring a patient include using ultrasound devices, lead solder wire, and special devices constructed of movable rods that take on the shape of the object against which they are pressed.

Contour types include **sagittal**, **coronal**, and **transverse** plane; figure 14.4 illustrates these planes. These three types of contours provide the three dimensions of the patient's body.

The diameters of the patient are documented. These diameters are called **interfield distances (IFD),** or **separations**.

Regardless of the method selected, precise contouring is a critical part of good treatment planning.

After contouring, the shape is transferred to paper and a cross section of the **planning target volume** as described earlier is drawn on the paper by either the radiation oncologist or the dosimetrist from films marked by the radiation oncologist. This method of transferring data, however, is being replaced by the direct reading of CT images into the treatment planning system. As CT becomes more widely available in radiation oncology centers, the direct reading of CT images into the treatment planning system is becoming more common.

Contour placement on paper includes carefully measured patient dimensions, marks corresponding to simulation film field centers, and comments on patient positioning. The accuracy of the patient contour is critical, because misregistering of the beam center could lead to errors in the positioning of internal organs or the beams themselves. The treatment planner must clearly understand the physician's prescription. A 1 cm error in depth could result in a 75% error in dose using electron beams, a 15% error in dose with low-energy photon beams, and a 5% error in dose with megavoltage photon beams.

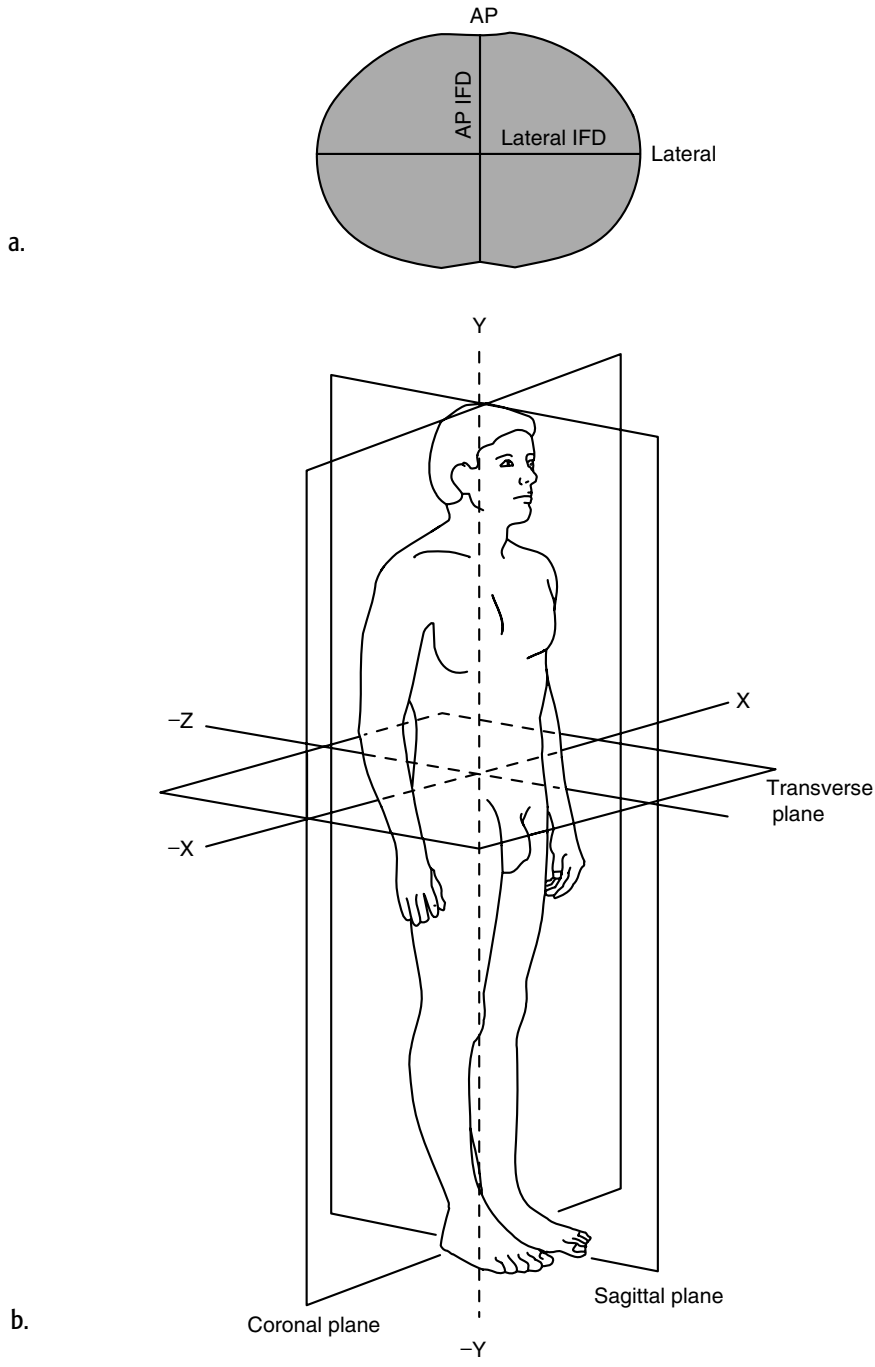


Figure 14.4

a) Transverse contour of abdomen.
b) The three planes in which contours are taken and their positions on the body.

ISODOSE DISTRIBUTIONS

Isodose charts are families of isodose curves, usually drawn for equal increments of percent depth dose (%DD or PDD). The curves represent the lines of equal DD for a particular field size and SSD at a specific plane in tissue. They are different for different beams, showing the effect on dose distribution due to energy, source size, SSD, and beam attenuation.

Figure 14.5 shows a 6 MV beam isodose distribution. Two points are shown on the distribution, P and Q. The point P corresponds to a dose of 90%, and the point Q to a dose of about 75%. Note that the depths on the central ray of various DD values, e.g., 90%, 80%, etc., increase as the percent DD decreases. Also, note that as you move to the edge of the beam at a specific depth, the DD drops. In other words, the isodose curves bend toward the skin surface as you move to the edge of the beam. This drop in DD values at the edge of the beam is due to the fact that there are fewer scattered photons at the edge than at the center. In the center of the beam, scatter can come from both sides; at the edge, it can come only from one side.

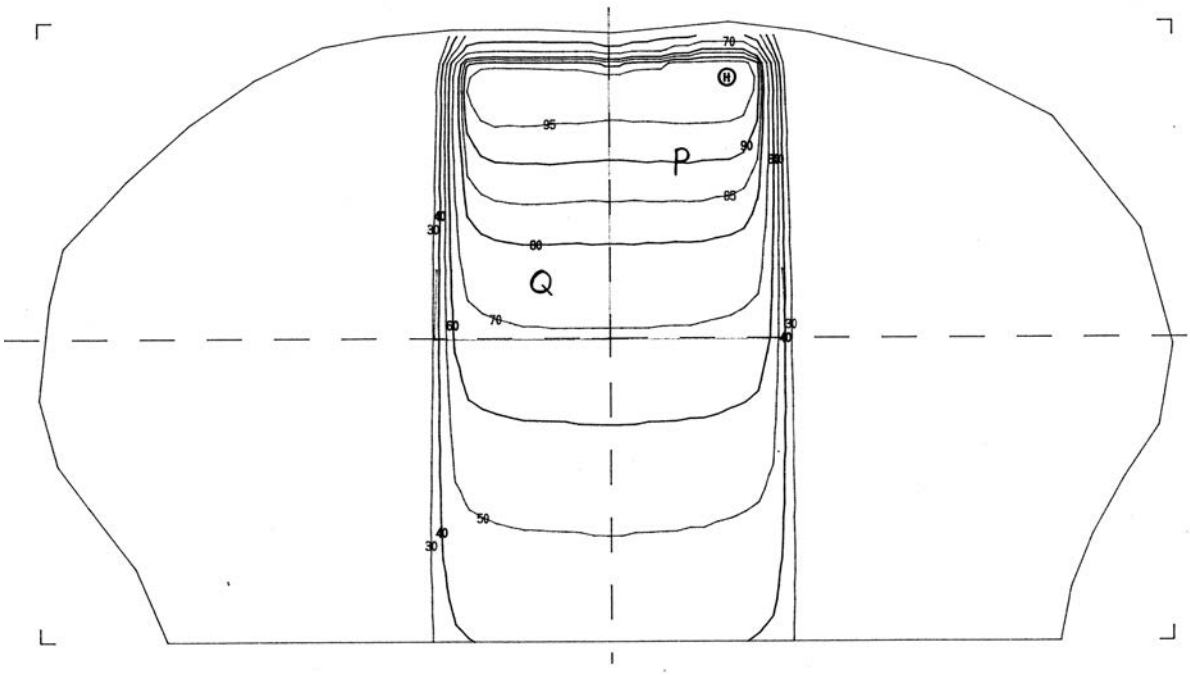


Figure 14.5

Isodose curves for a 6 MV linear accelerator beam. Note the points labeled P and Q. They are located at percent depth dose levels of 90% and 75% respectively.

In figure 14.6, the distribution on the left is for a 200 kV beam, 1.5 mm Cu HVL, a 50 cm SSD, and a 5 cm × 7 cm field. The middle distribution is for a 6 cm × 6 cm field of a ^{60}Co beam at 80 cm SSD. The distribution on the right is for a 25 MV beam with a 6 cm × 6 cm field size at 100 cm SSD. Note that the depth of a particular isodose line increases with increasing photon energy. The shape of the isodose lines at the edge of the beam is also noteworthy. The ^{60}Co beam, which has a larger source size, has a much larger penumbra than the megavoltage x-ray beam. This is apparent when you compare the greater separation of the isodose lines at the beam edges.

OBLIQUE INCIDENCE CORRECTIONS TO ISODOSE DISTRIBUTIONS

In the previous discussion it was assumed that the radiation fields struck the patient perpendicular to the skin surface and that the patient's tissue was homogeneous.

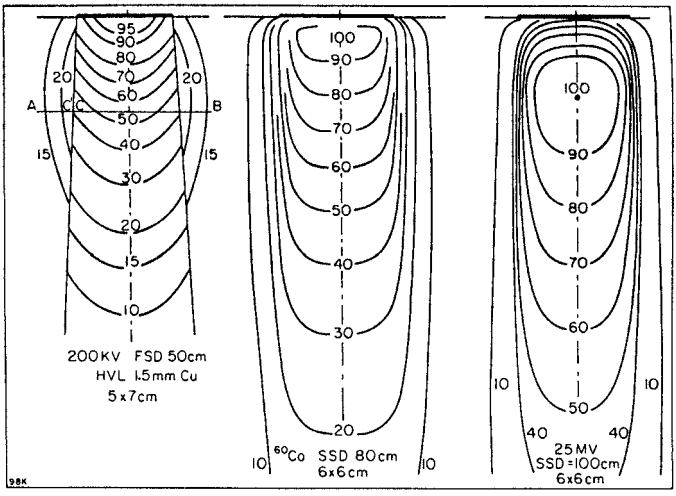


Figure 14.6

Comparison of isodose lines for various energies. The left set of curves is an orthovoltage beam. Each curve has rounded edges and the low percentiles bow outward. The middle set of curves is for a ^{60}Co beam. Note that its isodose lines are flatter and penetrate more deeply. The right set of curves is for a high-energy photon accelerator beam. Note the deep penetration of the isodose lines and the tightly packed nature of the lines at the beam's edge. (Reprinted from Johns & Cunningham, *The Physics of Radiology*, 4th edition, 1983. Courtesy of Charles C Thomas Publisher, Ltd., Springfield, Illinois.)

These conditions are reflected in standard isodose lines, which are measured under the ideal conditions of the radiation beam striking a flat water surface at right angles.

Of course, patients have curved surfaces and tissue **inhomogeneities** like bones and lungs. Most treatment planning systems correct for these inhomogeneities using calculation algorithms such as tissue-air ratio corrections and effective attenuation corrections.

These corrections are made by computer; however, in order to appreciate the benefits of computerized corrections, we will demonstrate a manual calculation method. In this approach, standard isodose lines are modified in order to approximate the clinical situation. This is known as the **isodose shift,** or 2/3 or 1/2 shift technique. It is a useful method for compensating for too little or too much tissue along the patient contour; see figure 14.7.

Shifting corrects the standard isodose lines by applying correction factors in the form of moving the isodose lines by a fixed portion of the tissue difference. (The standard isodose lines are obtained with the radiation beam entering perpendicular to a flat surface.) The portion chosen for the shift factors is a function of the beam energy and the machine SAD. The values of shift are shown in table 14.1.

In table 14.1, the shift factor decreases with increasing energy because of two effects: inverse square law and tissue attenuation. High-energy machines have larger SADs than low-energy machines (e.g., ^{60}Co and orthovoltage units). The larger distances reduce the drop-off of dose with distance. In addition, the higher the energy, the less attenuation by tissue. Figure 14.7 shows the curved surface of a patient contour with a straight line corresponding to the SSD at the central axis. A thick line shows a standard isodose curve. The corrected isodose line for the actual contour is formed by moving the standard curve by the appropriate fraction of the distance between the actual contour and the straight line. The direction of the "shift" is away from the tissue deficit and toward the tissue surplus. In figure 14.7 we see the effect of skin contour on an isodose line of a single field. This means that the level of d_{max} follows the skin surface.

Table 14.1 Isodose Shift Factors for Curved Skin Surfaces

Machine Energy	*Isodose Shift Factor*
150 kV-1 MV	0.8
1-5 MV	0.7
Cobalt-60	0.75
5-15 MV	0.6
15-30 MV	0.5
30 MV	0.4

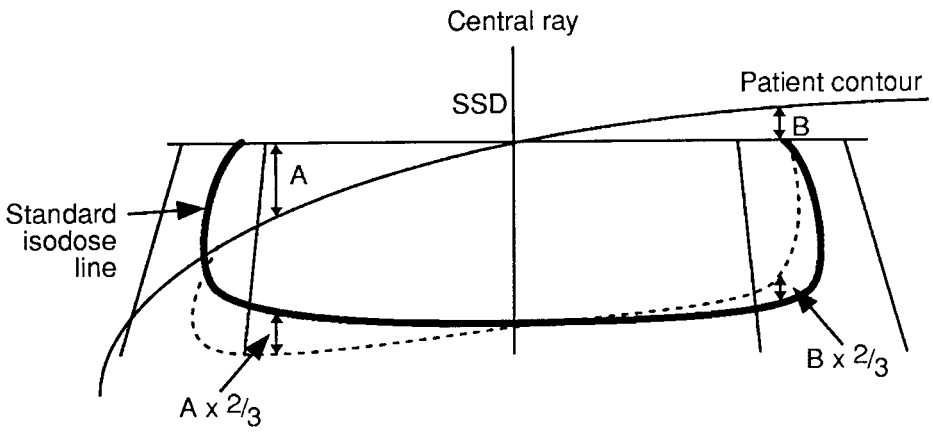


Figure 14.7

Example of isodose shifting of a 4 MV beam. The thick line is a standard curve. On the left, the missing tissue length A is corrected for by shifting the isodose curve down by a distance 2/3 × A in 2/3the same direction as the tissue deficit. On the right, the extra tissue in length B is corrected for by shifting the isodose line up toward the tissue surface by a distance of 2/3 B.

One important practical effect of non-perpendicular beam incidence is on skin sparing. In some clinical situations the beam strikes the skin surface tangentially (e.g., breast tangents). In this case, the d_{max} depth is reduced significantly, reducing skin sparing and producing more skin erythema and other skin reactions as well.

ISODOSE SUMMATIONS

When more than one field is used, the isodose curves of all fields must be added together to produce an **isodose summation**. Summations are usually produced using computers but can also be done (tediously) by hand. The manual summation process will be discussed using the example of two parallel opposed fields as illustrated later in figure 14.8.

The first step in summing two parallel opposed beams is to trace the isodose curves for each single field separately onto the contour. It helps to use two colors. When the lines of two isodose sets intersect, the percentages of those crossing isodose lines are added together. This result is called the **summated percentage**. This process is repeated at enough overlapping points to get a collection of summated percentages that sufficiently describe the **summated distribution**.

Computers do this process as a geometrical array of points or grid in the calculation plane thousands of times faster than the manual process. See reference 1 in chapter 6 for more information on the manual process. Chapter 15 will describe the alternative ways for describing the resulting distributions.

Figure 14.8 shows the summated isodose distribution of two parallel opposing fields. Note the hourglass shape of the highest isodose line; the hourglass shape is reduced as the beam energy increases.

TREATMENT TECHNIQUES

There are many ways of using external beam therapy to treat tumors. In **stationary,** or **fixed beam,** treatment the patient is positioned on the treatment couch and then irradiated by fields directed at the treatment volume at one specific gantry angle. These fields can be set up as either **SSD** or **SAD** (**isocentric**).

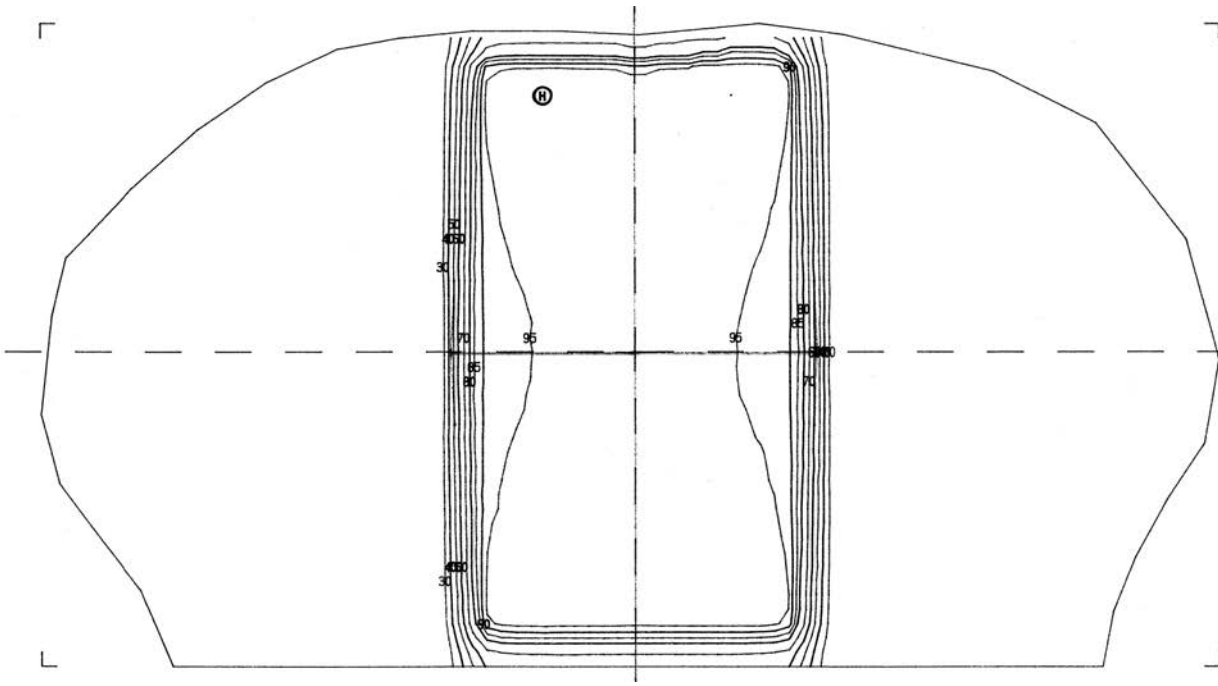


Figure 14.8

Parallel opposed x-ray fields used to treat a centrally located lesion. The hourglass shape of the high-percentage isodose curve is due to the effects of penumbra and scatter of each beam.

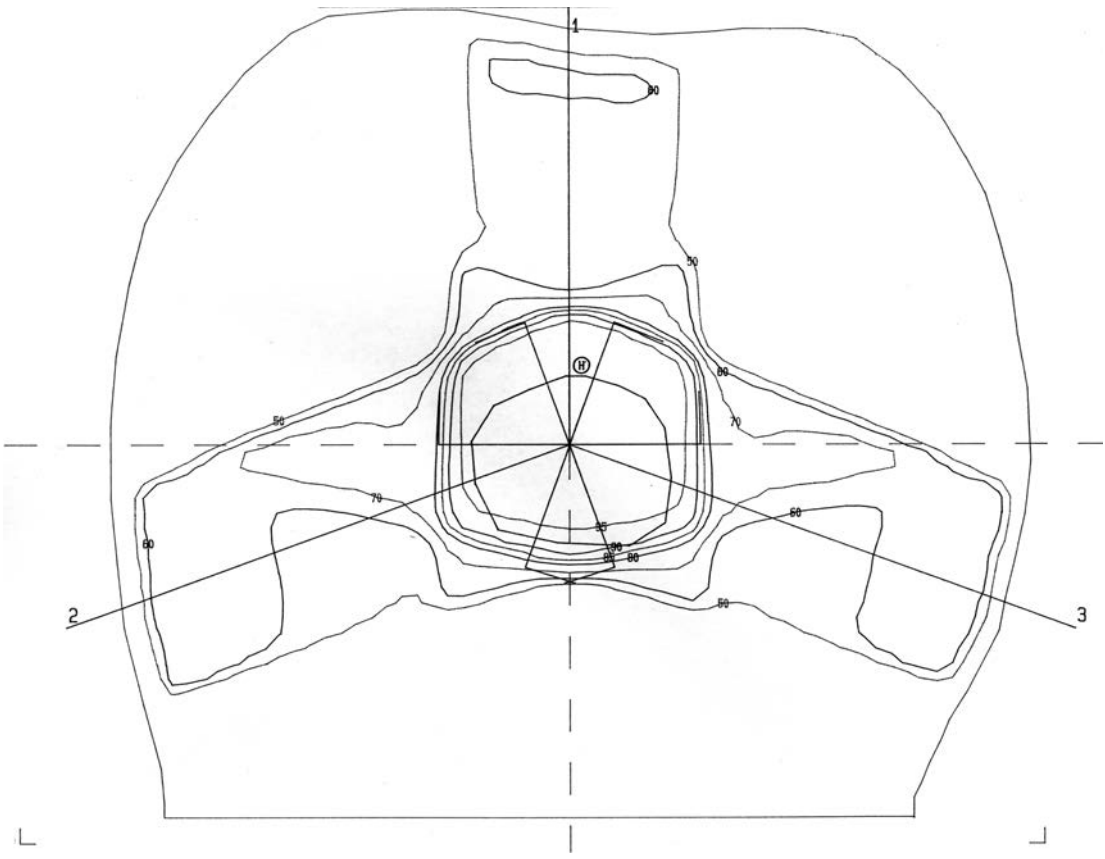


Figure 14.9

Three-field isodose distribution. Note in this distribution of three 6 MV beams with 100 cm SAD that the 100% dose value is centrally located (at the isocenter), and the 90% line surrounds the treatment volume. With lower energy beams there are often "hot spots," areas of high dose (90%) at the entrance region of the beams.

STATIONARY OR FIXED BEAM TREATMENT

Figure 14.8 illustrates the isodose distribution for the **parallel opposed field** (**POF**) beam configuration. Note the characteristic hourglass shape of the highest intensity isodose line, where it pinches in or narrows at the midline. This application of fields is suited to treatment of midline structures. A uniform dose (±5-10%) is given to a very large volume from d_{max} on the entrance of the AP field all the way to the d_{max} on the entrance of the PA field. While good for large lesions or central tumors including more superficial nodal areas, this distribution is not optimum

when midline structures are treated behind sensitive structures like the spinal cord.

The addition of other beams can reduce the relative dose to superficial tissues while maintaining a high central dose. Whenever another beam is added, the isodose lines are pulled in the direction of that new beam. Figure 14.9 shows the effect of a **three-field** plan on the isodose distribution.

One very common treatment plan for bladder, rectum, and prostate cancer uses two sets of parallel opposed fields, four fields at right angles aimed at the same center. The resulting isodose distribution is box-like; this treatment approach is therefore called a **four-field box** technique. Such an approach is shown in figure 14.10 for a 18 MV machine at 100 cm SAD. The summation of the four fields' dose lines produces a uniform distribution in the center and results in lower entrance doses with the box technique than with parallel opposing fields. In effect, the more fields used, the higher the central dose relative to the entrance doses.

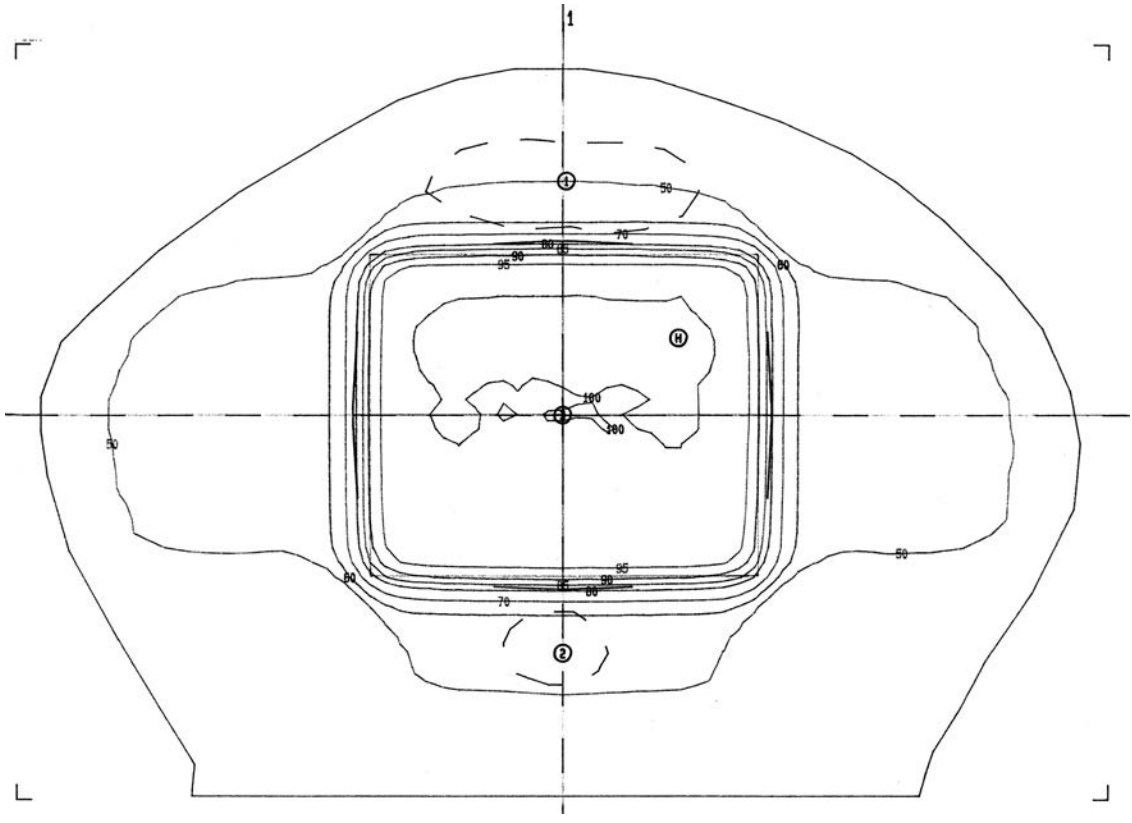


Figure 14.10

Four-field box technique using four 18 MV photon fields. The box shape of the high-dose region gives the technique its name. Note the high level of dose concentration in the center, leaving the peripheral tissues in low-dose regions.

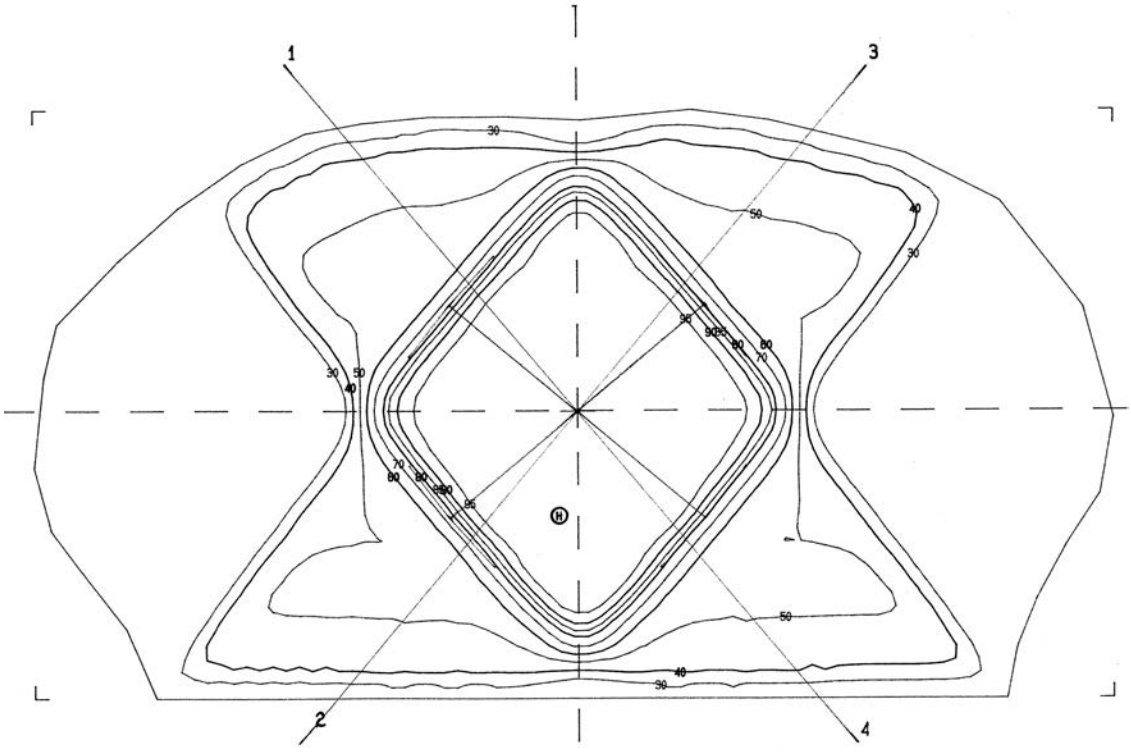


Figure 14.11
Diamond technique. This technique is sometimes used in order to minimize dose to tissues lateral to and in front of the target volume.

A variation of this technique is sometimes used. The **four-field diamond** is shown in figure 14.11. This configuration of beams lowers the dose at the sides of the patient. In pelvis treatments, for example, this means that the femoral head and rectal doses are less. The choice of beam combinations is dependent on the shape of the target volume, the type of tissues adjacent to that volume, the patient's size, and the beam energies available.

MOVING FIELDS TREATMENT

The ultimate approach to adding fields that limit superficial doses is moving the beam during treatment. Such fields are called **moving,** or **dynamic,** fields. In this type of treatment, the patient lies at rest on the linear accelerator couch, and

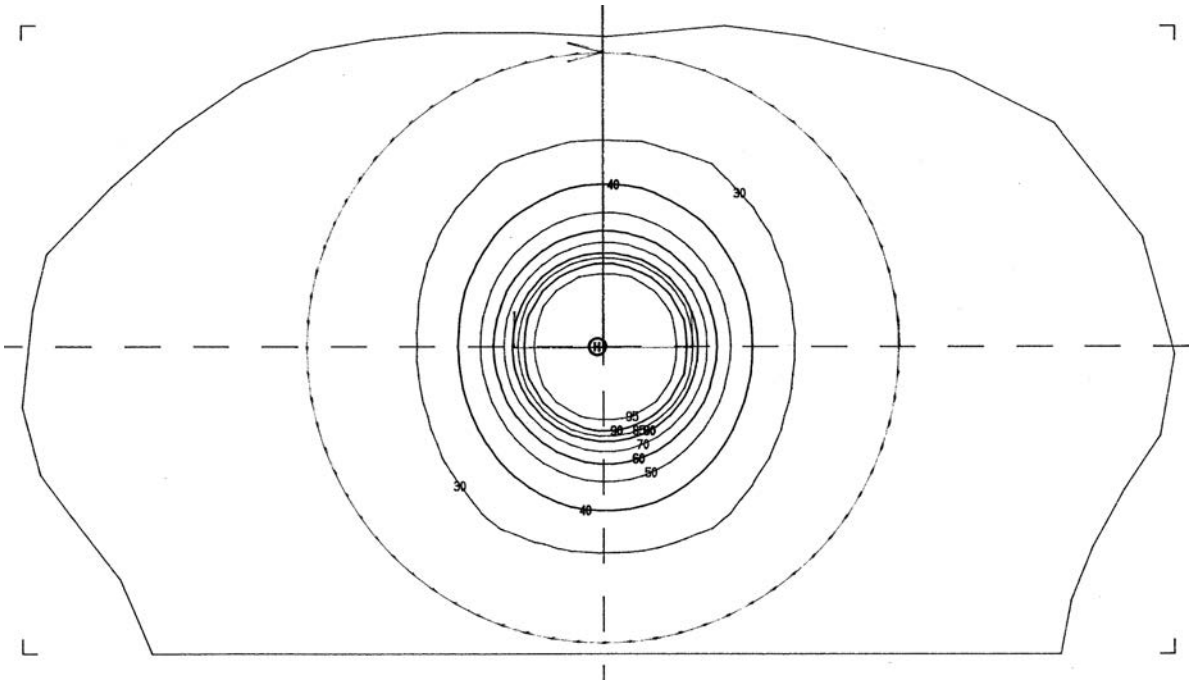


Figure 14.12
Full rotation (360°) technique. The complete rotation of the machine around the patient produces a circular or oval isodose distribution, since the patient is not perfectly round.

the machine irradiates the patient while the gantry rotates around the patient. **Rotational therapy** is when the beam makes a complete 360° rotation around the patient. If the beam makes less than a full rotation, it is called **arc therapy**. The word **rotation** is often used for any moving field technique, either full rotation or arc therapy. Double or multiple arc therapy is when two or more arcs are used. **Skip arc** therapy is when the beam is turned off for specific segments of the rotation. In most rotational therapy, the center of the tumor is placed at the isocenter of the machine. Therefore, the SSD varies continuously as the machine rotates.

Figure 14.12 illustrates the isodose distribution for a 360° rotation of a 7 cm × 9 cm field size at 80 cm SAD with a 4 MV beam. Note that the isodose lines are circular in shape at the high percentiles, while the lower percentiles are more oval in shape with the long axis of the oval at right angles to the width of the patient. If the patient were shaped like a circle, all of the isodose lines would be circles. Also note that, with the full rotation technique, the isodose lines are centered on the machine isocenter.

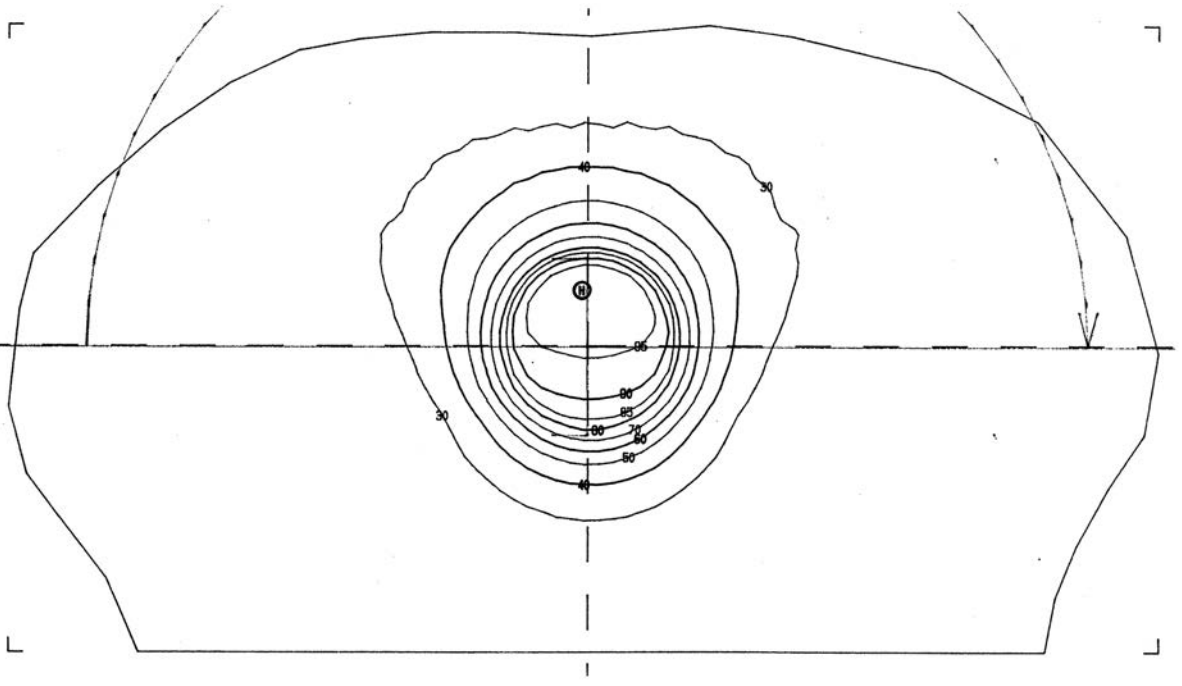


Figure 14.13
Partial rotation, past pointing. In this example, the machine arcs above the patient; therefore, the high-dose region lies above, i.e., in front of the machine isocenter.

When partial rotations are used, the maximum dose always occurs towards the machine head. That is, if the radiation beam arcs over the anterior part of the patient, the dose will lie mostly in the anterior portion of the patient. Figure 14.13 demonstrates the isodose distribution for a 180° anterior arc. It shows that in a partial rotation treatment, the isocenter should be placed a few centimeters posterior to the tumor center. This is called **past pointing**. In this technique, the isocenter of the machine is placed behind the target volume and the beam is aimed from in front of the lesion. By placing the isocenter of rotation past the tumor center, the dose distribution is effectively pulled in a manner that covers the tumor volume as shown in figure 14.13.

TISSUE INHOMOGENEITIES

Tissue inhomogeneities are volumes within the patient that have non-uniform tissue densities. For example, while the density of most soft tissue is about 1, that

Table 14.2 Isodose Shift Factors for Tissue

Inhomogeneity	*Isodose Shift Factor*⋆
air cavity	0.6
lung	0.4
hard bone	−0.5
spongy bone	−0.25

⋆Positive—shift isodose curves deeper in the body.
Negative—shift isodose curves toward the skin.

of lung is much lower, bone somewhat above 1, and metal plates used in bone repair much greater than 1. Tissue inhomogeneities alter the dose distribution from the standard curves due to attenuation and scatter. Calculations to account for these inhomogeneities, **heterogeneity corrections**, inside the patient are handled in a similar fashion to oblique incidence (discussed earlier). Instead of shifting 1/2 or 2/3 in the direction of tissue deficit, the curves shift away from the patient surface (deeper) with low-density inhomogeneities (e.g., air cavities) or toward the patient surface with high-density inhomogeneities (e.g., bones). Table 14.2 illustrates factors used for various tissue corrections.

Other approaches to tissue inhomogeneities are also possible. One method for correcting tissue inhomogeneities is the **effective TAR method**. It is a point-by-point calculation technique instead of a line-by-line technique and is therefore appropriate only for computerized point calculations. For each point in the patient (actually each point of interest) the equivalent water path length or depth is calculated using electron densities of overlying tissues. Imagine that the patient contour is filled with a matrix of calculation points and that the computer calculates doses at each of the matrix points. See figure 14.14.

Figure 14.15 illustrates the effective TAR computation process using only three points. The three points on the right of the phantom (P_1, P_2, and P_3) are traversed by the x-ray beam from the left. To reach the three points, however, the radiation beam has to travel through three different materials—air, water, and bone.

In order to calculate the appropriate TAR for each material, the **water equivalent path length** must be determined. This is the thickness of water that will provide the identical attenuation of the actual beam path.

The calculation for P_1 is as follows:

$$\begin{aligned}\text{Path length } P_1 &= 5 \text{ cm water} + 6 \text{ cm air} + 5 \text{ cm water}\\ &= 10 \text{ cm water} + 6 \text{ cm air}\\ &= 10 \text{ cm water}\end{aligned}$$

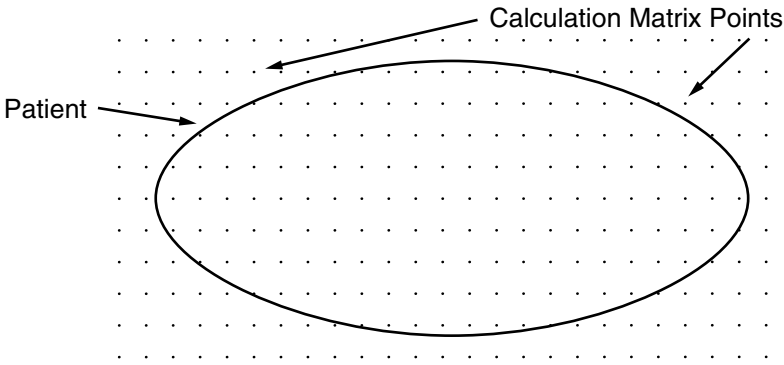


Figure 14.14

Computer view of a patient is actually a set of points in space surrounded by attenuating material. Each point lies in a rectangular array of points, a matrix of calculation points at which individually computed doses are produced.

Because of its low density, 6 cm of air is equivalent to less than 1 mm of water and is ignored. Therefore, the equivalent length is just 10 cm.

The calculation for P_2 is handled similarly:

$$\text{Path length } P_2 = 5 \text{ cm water} + 6 \text{ cm water} + 5 \text{ cm water}$$
$$= 16 \text{ cm water}$$

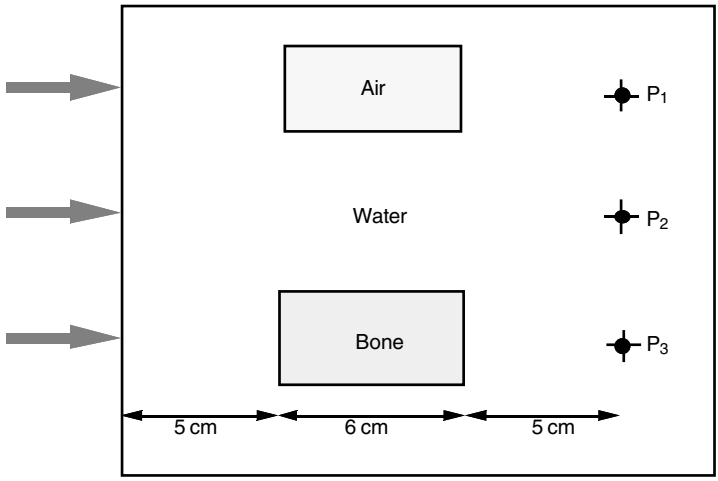


Figure 14.15

Effective TAR example. This figure illustrates the TAR computation process involved, using only three points. Points P_1, P_2, and P_3 are behind three different materials—air, water, and bone. The air supplies less attenuation than the water, while the bone supplies more. Note that the air cavity and bone are surrounded by water.

and for P_3,

$$\begin{aligned}\text{Path length } P_3 &= 5\text{ cm water} + 6\text{ cm bone} + 5\text{ cm water}\\ &= 10\text{ cm water} + 6\text{ cm bone}\left(\frac{\text{bone density}}{\text{water density}}\right)\\ &= 10\text{ cm water} + 6\text{ cm bone}\left(\frac{1.5\text{ g/cm}^3}{1.0\text{ g/cm}^3}\right)\\ &= 19\text{ cm water.}\end{aligned}$$

Thus the dose at P_1 > dose at P_2 > dose at P_3. In this equation, bone density (1.5 g/cm^3, i.e., 1.5 times the density of water) is used to convert the bone thickness to water thickness. Once the path length for a point is calculated, it is used as the depth to determine the TAR (by looking it up in the TAR tables) for the point in question.

Both equivalent path length and isodose shift are only approximation methods. They do not account for scatter from different nearby structures, but give answers good to within a few percent. They have been presented here to demonstrate some of the considerations of isodose construction.

TISSUE COMPENSATION

Another way of correcting doses in cases of tissue deficit is to make the surface flat using tissue equivalent material called **bolus** (see figure 14.16). Materials used for bolus include rice, sugar pellets, flour, sodium bicarbonate, and other various mixtures.

Bolus may also be used to bring d_{max} to the patient's skin surface. In these cases, materials such as wet gauze, Vaseline gauze, or other malleable tissue equivalent materials are used in a thickness equal or close to that required for the production of electron equilibrium.

When bolus is used, SSD in both patient setup and treatment planning must be consistent. If the dosimetrist calculates using 100 cm SSD to the bolus (101.5 cm SSD to the surface with 1.5 cm d_{max} for a 6 MV photon beam) and if the therapist uses 100 cm SSD to the skin (98.5 SSD to the bolus), an error as large as 3% can be introduced into the patient treatment.

The placement of bolus on the skin surface reduces skin sparing in megavoltage photon beams as described in chapter 8. Custom-shaped **compensating filters** (see figure 14.17) are sometimes used to avoid this bolus effect. The distance between the compensator and the skin surface is large enough (at least

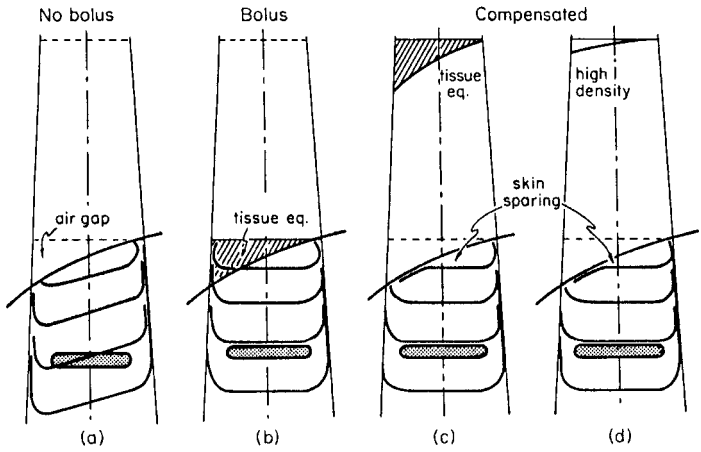


Figure 14.16

Tissue deficit compensation. In this example, the hatched area at depth is to be treated. The air gap shown on the left (a) causes an uneven dose distribution over the tumor and can be compensated for by using either of two approaches. The first is bolus (b), where the compensation material is in contact with the patient. One unfortunate consequence of bolus is the loss of skin sparing. Effectively moving the bolus from the patient surface toward the machine source (blocking tray position) retains the compensation of the bolus while re-establishing skin sparing (c and d). (Reprinted from Johns & Cunningham, *The Physics of Radiology*, 4th edition, 1983. Courtesy of Charles C Thomas Publisher, Ltd., Springfield, Illinois.)

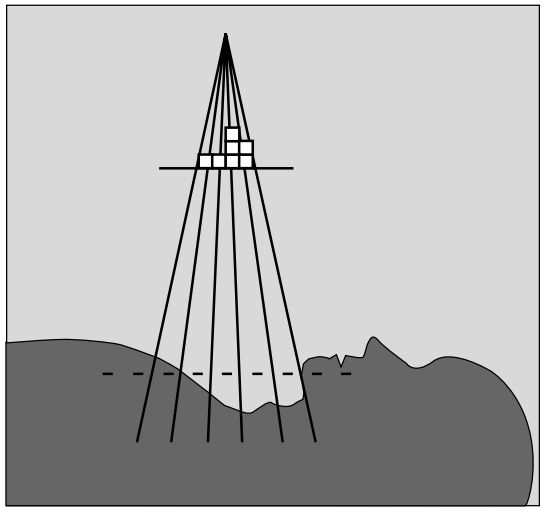


Figure 14.17

Geometry of tissue compensating filters. The compensators are placed in a holder near the photon source and mimic the shape of the patient's inhomogeneities. Because of beam divergence, the compensator is smaller than the bolus it replaces.

15 cm) to disperse electron and photon scatter produced in the filter, thereby re-establishing skin sparing. These filters are made of brass (only used with low-energy photons), aluminum, lead, leaded plastics, or Cerrobend and are shaped to have the same effect at depth in the patient as the more simple bolus. Because of the loss of side scatter from compensators, they are generally thinner in mass-length than the tissues they replace. Recently, computer-controlled milling machines have been adapted to create customized tissue compensators out of composite lead and plastic mixtures and other materials. Compensators are used most often in lung, head, and neck treatments.

WEDGE FILTERS

A common, non-customized compensating device is the **wedge filter,** or **wedge,** shown in figure 14.18. A wedge is used to distort the isodose distribution by tilting isodose lines through a specific angle (figure 14.19). Wedges are sometimes used to compensate for sloping body surfaces. They are also used to good effect for areas of great obliquity such as neck, breast tangents, and other anatomic areas where **hot spots** (regions of maximum dose occurrence) could occur (shown later in figure 14.21). The thick end of the wedge is called the **heel** (which transmits less of the initial beam) and the thin end the **toe** (which transmits more of the initial beam).

The angle to which a specific isodose line is bent is also the name of the wedge and is called the **wedge angle**. Lower energy machines use a 50% or 60% isodose line, while higher energy machines use 70% or 80%. Wedges are available for all megavoltage treatment machines at specific angles, usually 15°, 30°, 45°, and 60°.

A slightly different approach is used by some manufacturers who use one single wedge permanently mounted in a motorized moving tray within the machine head. By varying the proportion of the mu setting that the wedge is in the beam, you can vary the **effective wedge angle**. Some manufacturers call this a **universal wedge.**[1] This approach can also be used with conventional wedge sets in order to create effective wedge angles of any angle by mixing open and wedged beams.

1. In the 1950s through the 1970s, the term universal wedge was used in a different context, referring to a wedge of a specific angle designed for use for all field sizes (with its center always on the beam axis) instead of for a specific field size. In these individualized wedges, the toe was always aligned with the beam edge. This usage of the word "universal" is no longer common.

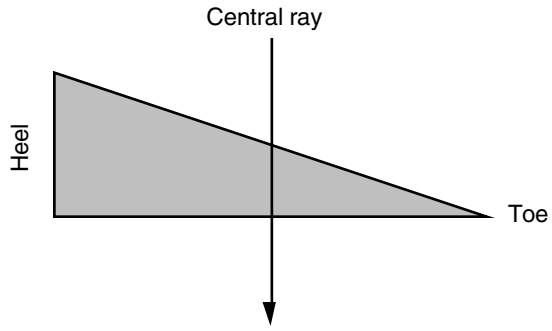


Figure 14.18
Wedge vocabulary. Wedges are usually made of high Z materials to minimize space needed in the machine head or above the blocking tray holder.

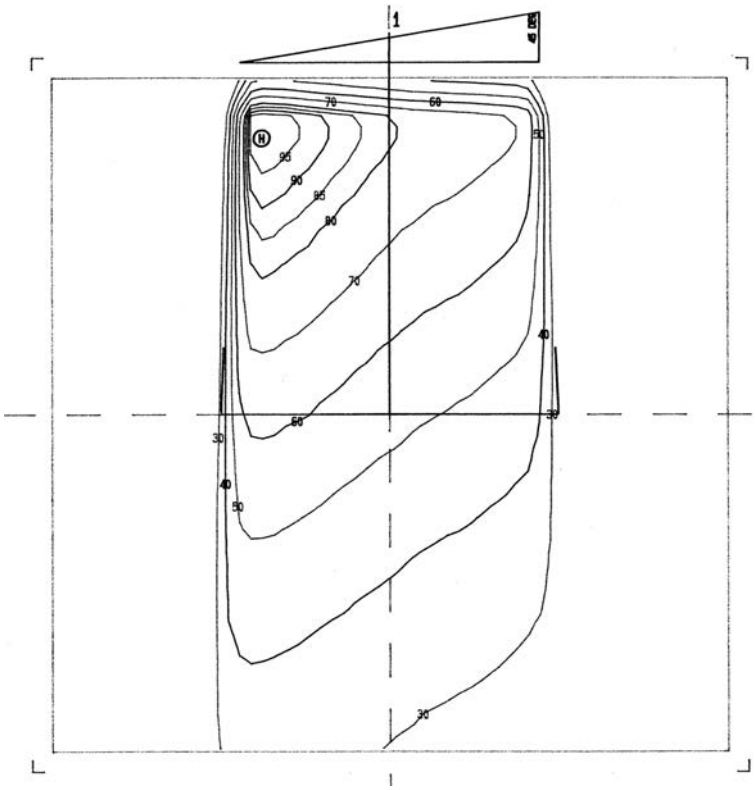


Figure 14.19
An isodose chart for a 10 cm × 10 cm field size, 6 MV x-ray beam, used with a 45° wedge.

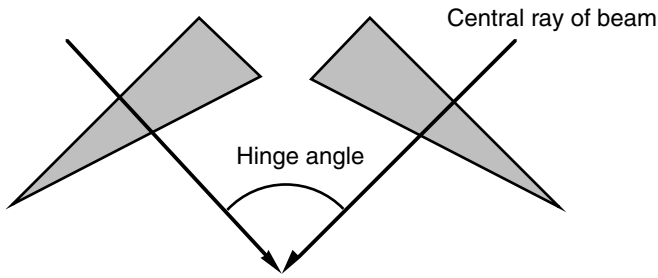


Figure 14.20
The hinge angle is the angle between the central rays of the two beams.

Another approach is used by most new linear accelerators. In these units, the independently variable jaws are moved during the therapy treatment, producing the effective dose distribution of a standard wedge but without the use of any wedge filter. This approach is sometimes called **dynamic** or **virtual wedge.**[2]

2. Before moving jaw dynamic wedges became common, the internal wedges described above were sometimes called dynamic wedges.

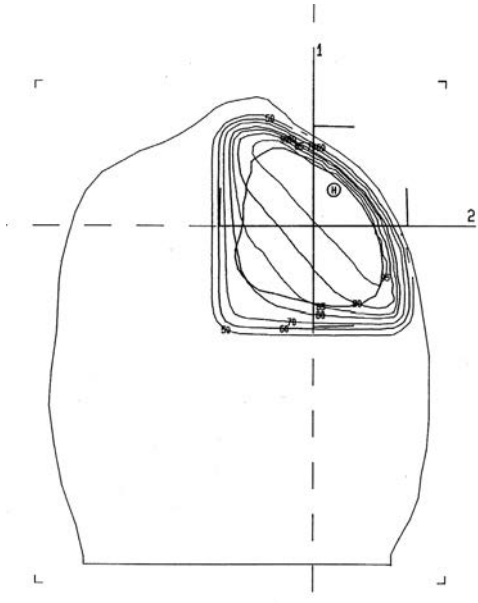


Figure 14.21
A pair of unwedged fields at 90° angles. Note the hot spot (H) where the beams overlap in the area of tissue deficit.

Wedged fields are often used in pairs, called **wedge pairs**. The angle between the two fields is dictated by patient anatomy and is called the **hinge angle** (figure 14.20). Then the following simple formula is used to predict the appropriate wedge angle to use for the optimal dose distribution:

$$\text{Wedge angle} = 90° - (0.5 \times \text{hinge angle})$$

or:

$$\text{Hinge angle} = 180° - (2 \times \text{wedge angle})$$

Thus, 45° wedged fields usually have a 90° hinge angle. Similarly, 60° wedged fields usually have a 60° hinge angle.

Figures 14.21 and 14.22 demonstrate the use of wedges on a maxillary sinus tumor. The first figure shows two beams aimed at a structure at right angles without wedges. With the insertion of 45° wedges with their heels toward each other, the dose uniformity over the tumor is markedly improved.

Wedges are used frequently in tangential breast, three-field pelvic, and numerous other field irradiation arrangements. Some of these will be discussed in greater detail in chapter 15.

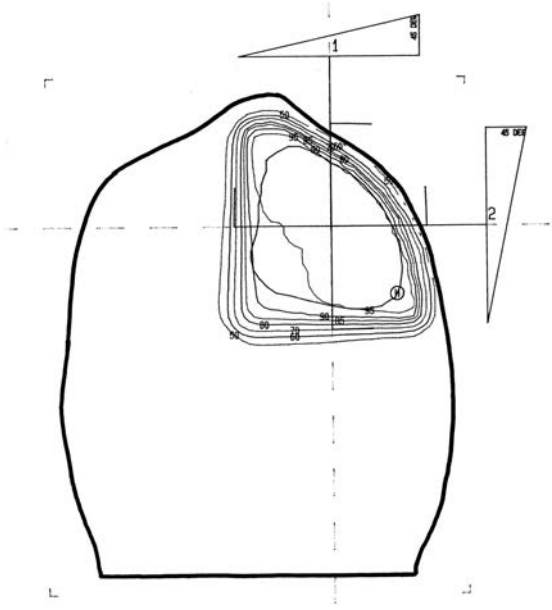


Figure 14.22

Wedged pair distribution. Note that the hot region shown in figure 14.21 is eliminated by attenuation of the beam by the thick end (heel) of the wedge.

STANDARD TREATMENT CALCULATION

The final step in any treatment plan is the calculation of treatment time or monitor unit setting for the energy being used. The basic physics of this process is discussed in chapter 12 along with specific equations. This section demonstrates some of these techniques.

BEAM ON TIME CALCULATIONS (TIMER SETTINGS)

Machines such as ^{60}Co units, orthovoltage, and superficial x-ray machines are controlled using a timer setting. The technique for time calculation will be demonstrated in two steps. First, the general approach to the problem will be outlined. Second, a numerical orthovoltage example will be presented.

Surface lesions are often treated using fixed cones that are larger than the tumor to encompass both the lesion and an adequate margin. In these cases, lead masks or cutouts are used to shield all but the desired area. The thickness of lead necessary is usually 1/8" (3.2 mm) or less. For example, the skin of a patient is treated to the area shown in figure 14.23. The rectangular area has an equivalent square field size of 2 cm × 3 cm or 2.4 cm^2 as determined from the equivalent squares in table 12.1.

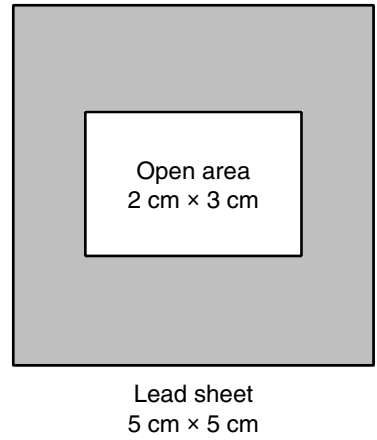


Figure 14.23

Lead cutout in a 5 cm × 5 cm field. The hatched area is covered by lead, leaving an open area with an effective field size of 2 cm × 3 cm.

Table 14.3 Backscatter Factors

Field Size (cm)	*HVL (mm Cu)*			
	0.5	*1*	*2*	*3*
0	1.00	1.00	1.00	1.00
4	1.21	1.18	1.14	1.12
6	1.28	1.25	1.20	1.16
8	1.33	1.31	1.25	1.20
10	1.37	1.36	1.29	1.24
15	1.44	1.43	1.36	1.30
20	1.49	1.49	1.42	1.34

There are three steps involved in calculating the timer setting for a treatment. The first step is the determination of **dose rate**. Its value is determined from the results of the machine calibration combined with other data on the beam as follows:

$$\text{Dose rate} = \text{Machine output} \times f_{med} \times \text{BSF}$$

Here, **machine output** is obtained at the time of annual calibrations, usually determined for the unblocked *cone* when using superficial and orthovoltage beams, and/or the open *field* when using ^{60}Co beams. The f_{med} value (R to cGy conversion factor) is determined from tables, such as table 7.3, and the value of BSF is obtained from table 14.3 for the specific quality of the beam, which is measured at the time of machine calibration, or appendix 5 for ^{60}Co. Second, once the dose rate is determined, the **beam on time** can be obtained using the simple equation:

$$\text{Beam on time (min)} = \frac{\text{Dose per fraction}}{\text{Dose rate}}$$

where the numerator indicates the dose per fraction in cGy and the denominator indicates the rate, which is the actual number of cGy per minute (or for orthovoltage units, R/min) at the skin.

Third, we measure the timer error (discussed in chapter 10), which is determined at the time of machine calibration. While timer errors for superficial beams are usually negligible, orthovoltage beams and ^{60}Co machines may have significant timer errors. Beam on time is determined using the simple equation:

$$\text{Timer setting} = \text{Beam on time (min)} + \text{Timer error}$$

As a numerical example, we are given the following calibration information:

Output of 5 cm square cone:	90.3 R/min
kVp:	220
Filter:	0.75 mm Cu
HVL:	1 mm Cu
Timer Error:	+ 1.5 seconds
Prescription:	300 cGy per fraction

Now we are asked to determine the timer setting to deliver 300 cGy to the patient's skin. From table 12.1, the effective field size is 2.4 cm. That means the dose rate at the skin is as follows:

$$\begin{aligned}\text{Dose Rate} &= \text{Machine output (open cone)} \times f_{med} \times \text{BSF (blocked field)} \\ &= 90.3 \text{ R/min} \times 0.95 \text{ cGy/R} \times 1.11 \\ &= 95.2 \text{ cGy/min.}\end{aligned}$$

Note that the BSF is determined for the open area of 2 cm × 3 cm (2.4 cm equivalent square field size). The value 1.11 is interpolated from table 14.3). The beam on time necessary for the treatment is:

$$\begin{aligned}\text{Beam on time (min)} &= \frac{\text{Dose per fraction}}{\text{Dose rate}} \\ &= \frac{300 \text{ cGy}}{95.2 \text{ cGy/min}} \\ &= 3.15 \text{ min.}\end{aligned}$$

Finally, to determine timer setting, we first convert the decimal minutes to seconds before adding in the timer error as follows.

$$\begin{aligned}\text{Timer Setting} &= \text{Beam on time} + \text{Timer error} \\ &= 3' 9'' + 1.5'' \\ &= 3' 11''\end{aligned}$$

This example demonstrates the technique for use in a roughly rectangular field. If the field is more circular, the equivalent field size can be determined using techniques outlined in BJR Supplement 25 (reference 3). This same technique is also used for superficial x-ray machine treatments and is similar to that for ^{60}Co machines.

MONITOR UNIT CALCULATIONS

In linear accelerators, dose delivery is controlled by selecting the number of monitor units (mus). Therefore, there is no timer error and the formulas for dose calculations are different than those for machines that use timer setting. Two examples will illustrate the approach.

Example 1: A patient is treated to the sacrum through a single PA field to deliver 300 cGy at a depth of 6 cm using a 4 MV linac. SSD = 80 cm and the field size is 10 cm × 15 cm using a small corner block on a plastic tray. As described in chapter 12, the equation is:

$$N_{mu} = \frac{\text{Tumor dose}}{C_{cal} \times C_{fs} \times C_{attn} \times DD}.$$

The appropriate values for these factors are determined as previously outlined. The 10 cm × 15 cm field size has an equivalent square of 11.9 cm^2. The field size factor is determined from figure 12.4 and is 1.02. The attenuation of the tray is 0.96. The depth dose is 0.783, which is determined from appendix 5, table 2. The resulting mu setting is therefore:

$$N_{mu} = \frac{300\,cGy}{1.0\ cGy/mu \times 1.02 \times 0.96 \times 0.783}$$

$$N_{mu} = 391.$$

Example 2: A more complicated treatment is a pair of parallel opposed 6 MV fields to treat a patient's pelvis. The field size is 20 cm × 12 cm, the patient thickness is 20 cm, and the patient is treated isocentrically at a SAD of 100 cm (SSD of 90 cm) to a dose of 200 cGy midplane, two fields per day. In this case, the treatment dose per field is 100 cGy, the depth of treatment is 10 cm (one-half of the patient thickness), and the machine SAD is 100 cm. The equation we use is:

$$N_{mu} = \frac{\text{Dose}}{C_{cal} \times C_{fs} \times C_{attn} \times TMR} \times \left(\frac{SSD + d_{tumor}}{SAD_{cal}} \right)^2$$

The field size yields an effective square field of 14.8 cm^2 for a field size factor of C_{fs} = 1.030 from appendix 5, table 8. The patient is treated by the SAD technique; therefore, the appropriate attenuation factor for tissue is the TMR. From appendix 5, table 7, the TMR value is 0.816. We know the calibration is performed at 100 cm SSD and, therefore, the SAD to the point of calibration is 101.5 cm. Substituting the values in the above equation, we obtain:

$$\begin{aligned} N_{mu} &= \frac{100}{1.0 \times 1.030 \times 0.816} \times \left(\frac{100}{101.5}\right)^2 \\ &= 119.0 \times 0.971 \\ &= 115.5. \end{aligned}$$

therefore, the mu setting is 116.

If a blocking tray is used in the beam, the tray transmission factor would be used in the denominator as C_{attn}. If a manual calculation is performed for a wedged field, the C_{attn} factor would be the wedge transmission factor. The calculation equation then becomes:

$$N_{mu} = \frac{\text{Tumor dose}}{C_{cal} \times C_{fs} \times C_{tray} \times C_{wedge} \times DD}.$$

Calculations for ^{60}Co beams are similar in principle to those described above except that the output is in cGy per minute and the answer is in units of time (i.e., minutes and seconds). Tables in appendix 5 provide sample beam data for several different beam energies.

PROBLEMS

Using the data tables in chapters 7 and 12 and appendix 5, perform the following calculations:

1. Determine the timer setting necessary for the following treatment:
   Skin Dose = 250 cGy
   Energy: Orthovoltage: f_{muscle} = 0.91
   Machine Calibration: 82.1 R/min at the cone tip, SSD = 50 cm
   Field Size: 2 cm × 8 cm: BSF = 1.25
   Tissue Treated: muscle equivalent
   Timer Error: 2 seconds (to be added)

2. Determine the mu setting necessary to deliver 100 cGy to a depth of 15 cm in the following field:

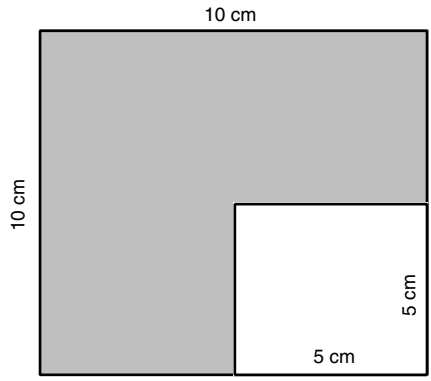
   Energy: 4 MV
   SSD: 80 cm
   Field Size: 12 cm × 15 cm

3. If 100 cGy is delivered in problem 2, what is the given dose?

4. If the depth in problem 2 is changed to 10 cm, what is the mu setting?

5. What is the timer setting for delivery of 300 cGy to a depth of 6 cm in a ^{60}Co beam with an SSD of 80 cm if the field size is 10 cm × 10 cm, the timer error is 0.01 min (to be added), and the machine output calibration is 82.1 R/min at 80 SAD?

6. What is the answer to problem 5 if the field size is not 10 cm × 10 cm but is instead blocked down from that to 5 cm × 5 cm (as in the following figure) and if the calculation is performed to deliver the dose at point P_1? (This concept was covered in chapter 12.)



7. In appendix 6, table 3, locate the d_{max} for a 6 MV beam.

8. What is the answer for problem 2 if a tray with a transmission factor of 0.98 is used to support a tiny block in the field? Assume that the small block has no affect on the depth dose.

9. What is the mu setting to deliver 150 cGy to a depth of 15 cm with a 6 MV 15 cm × 20 cm beam if the tumor is at the machine isocenter (100 cm SAD) and if the machine was calibrated at 100 SSD?

10. What is the output factor (cGy/mu) for a 10 cm × 10 cm field setting at 200 cm SSD if at 100 cm SSD the output factor is 1.000 cGy/mu for the 20 MV photon beam?

BIBLIOGRAPHY

1. Bentel, G.C. *Treatment Planning and Dose Calculation in Radiation Oncology,* 4th edition. McGraw-Hill Inc. US, New York, New York, 1989.
2. Bentel, G.C. "Positioning and immobilization device for patients receiving radiation therapy for carcinoma of the breast." *Medical Dosimetry* 15(1):3-6, 1990.
3. British Journal of Radiology Supplement 25. *Central Axis Depth Dose Data for Use in Radiotherapy*, British Institute of Radiology, London, 1996.
4. ICRU Report 24. *Determination of Absorbed Dose in a Patient Irradiated by Beams of X or Gamma Rays in Radiotherapy Procedures.* International Commission on Radiation Units and Measurements, Bethesda, Maryland, 1976.
5. ICRU Report 42. *Use of Computers in External Beam Radiotherapy Procedures with High-Energy Photons and Electrons*, International Commission on Radiation Units and Measurements, Bethesda, Maryland, 1977.
6. ICRU Report 50. *Prescribing, Recording and Reporting Photon Beam Therapy.* International Commission on Radiation Units and Measurements, Bethesda, Maryland, 1993.
7. ICRU Report 62. *Prescribing, Recording and Reporting Photon Beam Therapy (Supplement to ICRU Report 50).* International Commission on Radiation Units and Measurements, Bethesda, Maryland, 1999.
8. Johns, H.E. & Cunningham, J.R. *The Physics of Radiology,* 4th edition. Charles C Thomas, Springfield, Illinois, 1983.
9. Kahn, F.M. *The Physics of Radiation Therapy,* 4th edition. Lippincott Williams & Wilkins, Philadelphia, Pennsylvania, 1994.
10. Kartha, P.K.I. and Thompson, P. *Dosimetry Workbook*. Year Book Medical Publishers, Chicago, Illinois, 1982.
11. Selman, J. *The Basic Physics of Radiation Therapy*, 3rd edition. Charles C Thomas, Springfield, Illinois, 1990.
12. Zwicker, R.D., Shahabi, S., Wu, A., and Sternick, E.S. "Effective wedge angles for 6-MV wedges." *Medical Physics* 12(3):347-349, 1985.

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