PART II

CLASSICAL RADIATION THERAPY
It is seldom possible to measure dose distribution directly in patients treated with radiation. Data on dose distribution are almost entirely derived from measurements in phantoms—tissue equivalent materials, usually large enough in volume to provide full-scatter conditions for the given beam. These basic data are used in a dose calculation system devised to predict dose distribution in an actual patient. This chapter deals with various quantities and concepts that are useful for this purpose.

9.1. PHANTOMS

Basic dose distribution data are usually measured in a water phantom, which closely approximates the radiation absorption and scattering properties of muscle and other soft tissues. Another reason for the choice of water as a phantom material is that it is universally available with reproducible radiation properties. A water phantom, however, poses some practical problems when used in conjunction with ion chambers and other detectors that are affected by water, unless they are designed to be waterproof. In most cases, however, the detector is encased in a thin plastic (water equivalent) sleeve before immersion into the water phantom.

Since it is not always possible to put radiation detectors in water, solid dry phantoms have been developed as substitutes for water. Ideally, for a given material to be tissue or water equivalent, it must have the same effective atomic number, number of electrons per gram, and mass density. However, since the Compton effect is the most predominant mode of interaction for megavoltage photon beams in the clinical range, the necessary condition for water equivalence for such beams is the same electron density (number of electrons per cubic centimeter) as that of water.

The electron density ($\rho_e$) of a material may be calculated from its mass density ($\rho_m$) and its atomic composition according to the formula:

$$\rho_e = \rho_m \cdot N_A \cdot \left( \frac{Z}{A} \right)$$

(9.1)

where

$$\frac{Z}{A} = \sum_i a_i \cdot \left( \frac{Z_i}{A_i} \right)$$

(9.2)

$N_A$ is Avogadro's number and $a_i$ is the fraction by weight of the $i$th element of atomic number $Z_i$ and atomic weight $A_i$. Electron densities of various human tissues and body fluids have been calculated according to Equation 9.1 by Shrimpton (1). Values for some tissues of dosimetric interest are listed in Table 5.1.

Table 9.1 gives the properties of various phantoms that have been frequently used for radiation dosimetry. Of the commercially available phantom materials, Lucite and polysyrene are most frequently used as dosimetry phantoms. Although the mass density of these materials may vary depending on a given sample, the atomic composition and the number of electrons per gram of these materials are sufficiently constant to warrant their use for high-energy photon and electron dosimetry.
#### TABLE 9.1. PHYSICAL PROPERTIES OF VARIOUS PHANTOM MATERIALS

<table>
<thead>
<tr>
<th>Material</th>
<th>Chemical Composition</th>
<th>Mass Density (g/cm³)</th>
<th>Number of Electrons/g (×10²³)</th>
<th>Z&lt;br&gt;eff&lt;sup&gt;a&lt;/sup&gt; (Photoelectric)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>H₂O</td>
<td>1</td>
<td>3.34</td>
<td>7.42</td>
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<tr>
<td>Polystyrene</td>
<td>(C₅H₈)n</td>
<td>103-1.05</td>
<td>3.24</td>
<td>5.69</td>
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<tr>
<td>Plexiglas</td>
<td>(C₃H₅O₂H₃)n</td>
<td>1.16-1.20</td>
<td>3.24</td>
<td>6.48</td>
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<tr>
<td>(Perspex, Lucite)</td>
<td>Polyethylene</td>
<td>(CH₃)n</td>
<td>0.92</td>
<td>3.44</td>
</tr>
<tr>
<td>Paraffin</td>
<td>C₇H₂₃-2</td>
<td>0.87-0.91</td>
<td>3.44</td>
<td>5.42</td>
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<td>Mix D</td>
<td>Paraffin: 60.8</td>
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<td>Polyethylene: 30.4</td>
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<td>MgO: 6.4</td>
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<td>TiO₂: 2.4</td>
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<td>Paraffin: 100</td>
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<td>CaCO₃: 0.94</td>
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<tr>
<td>Solid water&lt;sup&gt;b&lt;/sup&gt;</td>
<td>Epoxy resin-based mixture</td>
<td>1.06</td>
<td>3.34</td>
<td>7.35</td>
</tr>
</tbody>
</table>

<sup>a</sup> Z<sub>eff</sub> for photoelectric effect is given by Eq. 6.4.

<sup>b</sup> Available from Radiation Measurements, Inc. (Middleton, Wisconsin).


In addition to the homogeneous phantoms, anthropomorphic phantoms are frequently used for clinical dosimetry. One such commercially available system, known as Alderson Rando Phantom, incorporates materials to simulate various body tissues—muscle, bone, lung, and air cavities. The phantom is shaped into a human torso (Fig. 9.1) and is sectioned transversely into slices for dosimetric applications.

White et al. (2) have developed extensive recipes for tissue substitutes. The method is based on adding particulate fillers to epoxy resins to form a mixture with radiation properties closely approximating that of a particular tissue. The most important radiation properties in this regard are the mass attenuation coefficient, the mass energy absorption coefficient, electron mass stopping, and angular scattering power ratios. A detailed tabulation of tissue substitutes and their properties for all the body tissues is included in a report by the International Commission on Radiation Units and Measurements (3).

Based on the previous method, Constantinou et al. (4) designed an epoxy resin-based solid substitute for water, called solid water. This material could be used as a dosimetric calibration phantom for photon and electron beams in the radiation therapy energy range. Solid water phantoms are now commercially available from Radiation Measurements, Inc. (Middleton, WI).

### 9.2. DEPTH DOSE DISTRIBUTION

As the beam is incident on a patient (or a phantom), the absorbed dose in the patient varies with depth. This variation depends on many conditions: beam energy, depth, field size, distance from source, and beam collimation system. Thus the calculation of dose in the patient involves considerations in regard to these parameters and others as they affect depth dose distribution.

An essential step in the dose calculation system is to establish depth dose variation along the central axis of the beam. A number of quantities have been defined for this

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9. Dose Distribution and Scatter Analysis

9.3. PERCENTAGE DEPTH DOSE

One way of characterizing the central axis dose distribution is to normalize dose at depth with respect to dose at a reference depth. The quantity percentage (or simply percent) depth dose may be defined as the quotient, expressed as a percentage, of the absorbed dose at any depth $d$ to the absorbed dose at a fixed reference depth $d_0$, along the central axis of the beam (Fig. 9.2). Percentage depth dose ($P$) is thus:

$$P = \frac{D_d}{D_{d_0}} \times 100$$

(9.3)

For orthovoltage (up to about 400 kVp) and lower-energy x-rays, the reference depth is usually the surface ($d_0 = 0$). For higher energies, the reference depth is taken at the position of the peak absorbed dose ($d_0 = d_m$).
II. Classical Radiation Therapy

Central Axis Surface Phantom

**FIG. 9.2.** Percentage depth dose is \( \left( \frac{D_d}{D_{do}} \right) \times 100 \), where \( d \) is any depth and \( d_o \) is reference depth of maximum dose.

In clinical practice, the peak absorbed dose on the central axis is sometimes called the maximum dose, the dose maximum, the given dose, or simply the \( D_{\text{max}} \). Thus,

\[
D_{\text{max}} = \frac{D_d}{P} \times 100
\]

(9.4)

A number of parameters affect the central axis depth dose distribution. These include beam quality or energy, depth, field size and shape, source to surface distance, and beam collimation. A discussion of these parameters will now be presented.

**A. Dependence on Beam Quality and Depth**

The percentage depth dose (beyond the depth of maximum dose) increases with beam energy. Higher-energy beams have greater penetrating power and thus deliver a higher percentage depth dose (Fig. 9.3). If the effects of inverse square law and scattering are not considered, the percentage depth-dose variation with depth is governed approximately by exponential attenuation. Thus the beam quality affects the percentage depth dose by virtue of the average attenuation coefficient \( \mu \). As the \( \mu \) decreases, the more penetrating the beam becomes, resulting in a higher percentage depth dose at any given depth beyond the build-up region.

A.1. Initial Dose Build-Up

As seen in Fig. 9.3, the percentage depth dose decreases with depth beyond the depth of maximum dose. However, there is an initial buildup of dose which becomes more and more pronounced as the energy is increased. In the case of the orthovoltage or lower-energy x-rays, the dose builds up to a maximum on or very close to the surface. But for higher-energy beams, the point of maximum dose lies deeper into the tissue or phantom. The region between the surface and the point of maximum dose is called the dose build-up region.

The dose build-up effect of the higher-energy beams gives rise to what is clinically known as the skin-sparing effect. For megavoltage beams such as cobalt-60 and higher energies the surface dose is much smaller than the \( D_{\text{max}} \). This offers a distinct advantage over the lower-energy beams for which the \( D_{\text{max}} \) occurs at the skin surface. Thus, in the case

\(^{2}\mu \) is the average attenuation coefficient for the heterogeneous beam.
of the higher energy photon beams, higher doses can be delivered to deep-seated tumors without exceeding the tolerance of the skin. This, of course, is possible because of both the higher percent depth dose at the tumor and the lower surface dose at the skin. This topic is discussed in greater detail in Chapter 13.

The physics of dose buildup may be explained as follows: (a) As the high-energy photon beam enters the patient or the phantom, high-speed electrons are ejected from the surface and the subsequent layers; (b) These electrons deposit their energy a significant distance away from their site of origin; (c) Because of (a) and (b), the electron fluence and hence the absorbed dose increase with depth until they reach a maximum. However, the photon energy fluence continuously decreases with depth and, as a result, the production of electrons also decreases with depth. The net effect is that beyond a certain depth the dose eventually begins to decrease with depth.

It may be instructive to explain the buildup phenomenon in terms of absorbed dose and a quantity known as kerma (from kinetic energy released in the medium). As discussed in Chapter 8, the kerma \( K \) may be defined as "the quotient of \( dE_r \) by \( dm \), where \( dE_r \) is the sum of the initial kinetic energies of all the charged ionizing particles (electrons) liberated by uncharged ionizing particles (photons) in a material of mass \( dm \)" (14).

\[
K = \frac{dE_r}{dm}
\]

Because kerma represents the energy transferred from photons to directly ionizing electrons, the kerma is maximum at the surface and decreases with depth because of the decrease in the photon energy fluence (Fig. 9.4). The absorbed dose, on the other hand, first increases with depth as the high-speed electrons ejected at various depths travel downstream. As a result, there is an electronic build-up with depth. However, as the dose depends on the electron fluence, it reaches a maximum at a depth approximately equal to the range of electrons in the medium. Beyond this depth, the dose decreases as kerma continues to decrease, resulting in a decrease in secondary electron production and hence a net decrease in electron fluence. As seen in Fig. 9.4, the kerma curve is initially higher than the dose curve but falls below the dose curve beyond the build-up region. This effect is explained by the fact that the areas under the two curves taken to infinity must be the same.
II. Classical Radiation Therapy

B. Effect of Field Size and Shape

Field size may be specified either geometrically or dosimetrically. The geometrical field size is defined as "the projection, on a plane perpendicular to the beam axis, of the distal end of the collimator as seen from the front center of the source" (15). This definition usually corresponds to the field defined by the light localizer, arranged as if a point source of light were located at the center of the front surface of the radiation source. The dosimetric, or physical, field size is the distance intercepted by a given isodose curve (usually 50\% isodose) on a plane perpendicular to the beam axis at a stated distance from the source.

Unless stated otherwise, the term field size in this book will denote geometric field size. In addition, the field size will be defined at a predetermined distance such as the source-surface distance (SSD) or the source-axis distance (SAD). The latter term is the distance from the source to axis of gantry rotation known as the isocenter.

For a sufficiently small field one may assume that the depth dose at a point is effectively the result of the primary radiation, that is, the photons which have traversed the overlying medium without interacting. The contribution of the scattered photons to the depth dose in this case is negligibly small or 0. But as the field size is increased, the contribution of the scattered radiation to the absorbed dose increases. Because this increase in scattered dose is greater at larger depths than at the depth of \(D_{\text{max}}\), the percent depth dose increases with increasing field size.

The increase in percent depth dose caused by increase in field size depends on beam quality. Since the scattering probability or cross-section decreases with energy increase and the higher-energy photons are scattered more predominantly in the forward direction, the field size dependence of percent depth dose is less pronounced for the higher-energy than for the lower-energy beams.

Percent depth dose data for radiation therapy beams are usually tabulated for square fields. Since the majority of the treatments encountered in clinical practice require rectangular and irregularly shaped (blocked) fields, a system of equating square fields to different field shapes is required. Semiempirical methods have been developed to relate central axis depth dose data for square, rectangular, circular, and irregularly shaped fields. Although general methods (based on Clarkson's principle—to be discussed later in this chapter) are available, simpler methods have been developed specifically for interrelating square, rectangular, and circular field data.

Day (16) and others (17,18) have shown that, for central axis depth-dose distribution, a rectangular field may be approximated by an equivalent square or by an equivalent circle.
9. Dose Distribution and Scatter Analysis

### TABLE 9.2. EQUIVALENT SQUARES OF RECTANGULAR FIELDS

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<thead>
<tr>
<th>Long Axis (cm)</th>
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Data for equivalent squares, taken from Hospital Physicists' Association (5) are given in Table 9.2. As an example, consider a 10 x 20-cm field. From Table 9.2, the equivalent square is 13.0 x 13.0 cm. Thus the percent depth dose data for a 13 x 13-cm field (obtained from standard tables) may be applied as an approximation to the given 10 x 20-cm field.

A simple rule of thumb method has been developed by Sterling et al. (19) for equating rectangular and square fields. According to this rule, a rectangular field is equivalent to a square field if they have the same area/perimeter (A/P). For example, the 10 x 20-cm field has an A/P of 3.33. The square field which has the same A/P is 13.3 x 13.3 cm, a value very close to that given in Table 9.2.

The following formulas are useful for quick calculation of the equivalent field parameters: For rectangular fields,

\[
A/P = \frac{a \times b}{2(a + b)}
\]  

(9.5)

where \(a\) is field width and \(b\) is field length. For square fields, since \(a = b\),

\[
A/P = \frac{a}{4}
\]

(9.6)

where \(a\) is the side of the square. From Equations 9.5 and 9.6, it is evident that the side of an equivalent square of a rectangular field is \(4 \times A/P\). For example, a 10 x 15-cm field has an A/P of 3.0. Its equivalent square is 12 x 12 cm. This agrees closely with the value of 11.9 given in Table 9.2.

Although the concept of \(A/P\) is not based on sound physical principles, it is widely used in clinical practice and has been extended as a field parameter to apply to other quantities such as backscatter factors, tissue-air ratios, and even beam output in air or in phantom. The reader may, however, be cautioned against an indiscriminate use of \(A/P\). For example, the \(A/P\) parameter, as such, does not apply to circular or irregularly shaped fields, although radii of equivalent circles may be obtained by the relationship:

\[
r = \frac{4}{\sqrt{\pi}} \cdot A/P
\]

(9.7)

Equation 9.7 can be derived by assuming that the equivalent circle is the one that has the same area as the equivalent square. Validity of this approximation has been verified from the table of equivalent circles given in Hospital Physicists' Association (5).
C. Dependence on Source-Surface Distance

Photon fluence emitted by a point source of radiation varies inversely as a square of the distance from the source. Although the clinical source (isotopic source or focal spot) for external beam therapy has a finite size, the source-surface distance is usually chosen to be large (≥80 cm) so that the source dimensions become unimportant in relation to the variation of photon fluence with distance. In other words, the source can be considered as a point at large source-surface distances. Thus the exposure rate or "dose rate in free space" (Chapter 8) from such a source varies inversely as the square of the distance. Of course, the inverse square law dependence of dose rate assumes that we are dealing with a primary beam, without scatter. In a given clinical situation, however, collimation or other scattering material in the beam may cause deviation from the inverse square law.

Percent depth dose increases with SSD because of the effects of the inverse square law. Although the actual dose rate at a point decreases with increase in distance from the source, the percent depth dose, which is a relative dose with respect to a reference point, increases with SSD. This is illustrated in Fig. 9.5 in which relative dose rate from a point source of radiation is plotted as a function of distance from the source, following the inverse square law. The plot shows that the drop in dose rate between two points is much greater at smaller distances from the source than at large distances. This means that the percent depth dose, which represents depth dose relative to a reference point, decreases more rapidly near the source than far away from the source.

In clinical radiation therapy, SSD is a very important parameter. Because percent depth dose determines how much dose can be delivered at depth relative to the surface dose or \( D_{\text{max}} \), the SSD needs to be as large as possible. However, because dose rate decreases with distance, the SSD, in practice, is set at a distance which provides a compromise between dose rate and percent depth dose. For the treatment of deep-seated lesions with megavoltage beams, the minimum recommended SSD is 80 cm.

Tables of percent depth dose for clinical use are usually measured at a standard SSD (80 or 100 cm for megavoltage units). In a given clinical situation, however, the SSD set on a patient may be different from the standard SSD. For example, larger SSDs are required for treatment techniques that involve field sizes larger than the ones available at

![Figure 9.5](image-url)
9. Dose Distribution and Scatter Analysis

FIG. 9.6. Change of percent depth dose with SSD. Irradiation condition (a) has SSD = $f_1$ and condition (b) has SSD = $f_2$. For both conditions, field size on the phantom surface, $r \times c$ and depth $d$ are the same.

the standard SSDs. Thus the percent depth doses for a standard SSD must be converted to those applicable to the actual treatment SSD. Although more accurate methods are available (to be discussed later in this chapter), we discuss an approximate method in this section: the Mayneord $F$ factor (20). This method is based on a strict application of the inverse square law, without considering changes in scattering, as the SSD is changed.

Figure 9.6 shows two irradiation conditions, which differ only in regard to SSD. Let $P(d,r,f)$ be the percent depth dose at depth $d$ for SSD = $f$ and a field size $r$ (e.g., a square field of dimensions $r \times r$). Since the variation in dose with depth is governed by three effects—inverse square law, exponential attenuation, and scattering—

\[
P(d,r,f) = 100 \cdot \left( \frac{f_1 + d_m}{f_1 + d} \right)^2 \cdot e^{-\mu(d-d_m)} \cdot K_i
\]

where $\mu$ is the linear attenuation coefficient for the primary and $K_i$ is a function which accounts for the change in scattered dose. Ignoring the change in the value of $K_i$ from one SSD to another,

\[
P(d,r,f) = 100 \cdot \left( \frac{f_1 + d_m}{f_2 + d} \right)^2 \cdot e^{-\mu(d-d_m)} \cdot K_i
\]

Dividing Equation 9.9 by 9.8, we have:

\[
\frac{P(d,r,f_2)}{P(d,r,f_1)} = \left( \frac{f_2 + d_m}{f_1 + d_m} \right)^2 \left( \frac{f_1 + d}{f_2 + d} \right)^2
\]

The terms on the right-hand side of Equation 9.10 are called the Mayneord $F$ factor. Thus,

\[
F = \left( \frac{f_2 + d_m}{f_1 + d_m} \right)^2 \left( \frac{f_1 + d}{f_2 + d} \right)^2
\]

It can be shown that the $F$ factor is greater than 1 for $f_2 > f_1$ and less than 1 for $f_2 < f_1$. Thus it may be restated that the percent depth dose increases with increase in SSD.

Example 1

The percent depth dose for a 15 x 15 field size, 10-cm depth, and 80-cm SSD is 58.4 ($^{60}$Co beam). Find the percent depth dose for the same field size and depth for a 100-cm SSD.

From Equation 9.11, assuming $d_m = 0.5$ cm for $^{60}$Co $\gamma$ rays:

\[
F = \left( \frac{100 + 0.5}{80 + 0.5} \right)^2 \left( \frac{80 + 10}{100 + 10} \right)^2
\]

\[
= 1.043
\]
From Equation 9.10,

\[
\frac{P(10,15,100)}{P(10,15,80)} = 1.043
\]

Thus, the desired percent depth dose is:

\[
P(10,15,100) = P(10,15,80) \times 1.043
\]

\[= 58.4 \times 1.043
\]

\[= 60.9
\]

More accurate methods that take scattering change into account would yield a value close to 60.6.

The Mayneord \(F\) factor method works reasonably well for small fields since the scattering is minimal under these conditions. However, the method can give rise to significant errors under extreme conditions such as lower energy, large field, large depth, and large SSD change. For example, the error in dose at a 20-cm depth for a 30 \(\times\) 30-cm field and 160-cm SSD \((^{60}\text{Co beam})\) will be about 3% if the percent depth dose is calculated from the 80-cm SSD tables.

In general, the Mayneord \(F\) factor overestimates the increase in percent depth dose with increase in SSD. For example, for large fields and lower energy radiation where the proportion of scattered radiation is relatively greater, the factor \((1 + F)/2\) applies more accurately. Factors intermediate between \(F\) and \((1 + F)/2\) have also been used for certain conditions (20).

### 9.4. TISSUE-AIR RATIO

Tissue-air ratio (TAR) was first introduced by Johns et al. (6) in 1953 and was originally called the "tumor-air ratio." At that time, this quantity was intended specifically for rotation therapy calculations. In rotation therapy, the radiation source moves in a circle around the axis of rotation which is usually placed in the tumor. Although the SSD may vary depending on the shape of the surface contour, the source-axis distance remains constant.

Since the percent depth dose depends on the SSD (section 9.3C), the SSD correction to the percent depth dose will have to be applied to correct for the varying SSD—a procedure that becomes cumbersome to apply routinely in clinical practice. A simpler quantity—namely, TAR—has been defined to remove the SSD dependence. Since the time of its introduction, the concept of TAR has been refined to facilitate calculations not only for rotation therapy but also for stationary isocentric techniques as well as irregular fields.

Tissue-air ratio may be defined as the ratio of the dose \((D_d)\) at a given point in the phantom to the dose in free space \((D_f)\) at the same point. This is illustrated in Fig. 9.7.

![Diagram of tissue-air ratio](image)

**FIG. 9.7.** Illustration of the definition of tissue-air ratio. TAR \((d, f) = D_d/D_f\).
For a given quality beam, TAR depends on depth \( d \) and field size \( r_d \) at that depth:

\[
\text{TAR}(d, r_d) = \frac{D_d}{D_s}
\]  
(9.12)

**A. Effect of Distance**

One of the most important properties attributed to TAR is that it is independent of the distance from the source. This, however, is an approximation which is usually valid to an accuracy of better than 2% over the range of distances used clinically. This useful result can be deduced as follows.

Because TAR is the ratio of the two doses \( (D_d, D_s) \) at the same point, the distance dependence of the photon fluence is removed. Thus the TAR represents modification of the dose at a point owing only to attenuation and scattering of the beam in the phantom compared with the dose at the same point in the miniphantom (or equilibrium phantom) placed in free air. Since the primary beam is attenuated exponentially with depth, the TAR for the primary beam is only a function of depth, not of SSD. The case of the scatter component, however, is not obvious. Nevertheless, Johns et al. (21) have shown that the fractional scatter contribution to the depth dose is almost independent of the divergence of the beam and depends only on the depth and the field size at that depth. Hence the tissue-air ratio, which involves both the primary and scatter component of the depth dose, is independent of the source distance.

**B. Variation with Energy, Depth, and Field Size**

Tissue-air ratio varies with energy, depth, and field size very much like the percent depth dose. For the megavoltage beams, the tissue-air ratio builds up to a maximum at the depth of maximum dose \( (d_m) \) and then decreases with depth more or less exponentially. For a narrow beam or a \( 0 \times 0 \) field size\(^3\) in which scatter contribution to the dose is neglected, the TAR beyond \( d_m \) varies approximately exponentially with depth

\[
\text{TAR}(d, 0) = e^{-\mu(d - d_m)}
\]  
(9.13)

where \( \mu \) is the average attenuation coefficient of the beam for the given phantom. As the field size is increased, the scattered component of the dose increases and the variation of TAR with depth becomes more complex. However, for high-energy megavoltage beams, for which the scatter is minimal and is directed more or less in the forward direction, the TAR variation with depth can still be approximated by an exponential function, provided an effective attenuation coefficient \( (\mu_{eff}) \) for the given field size is used.

**B.1. Backscatter Factor**

The term backscatter factor (BSF) is simply the tissue-air ratio at the depth of maximum dose on central axis of the beam. It may be defined as the ratio of the dose on central axis at the depth of maximum dose to the dose at the same point in free space. Mathematically,

\[
\text{BSF} = \frac{D_{\text{max}}}{D_s}
\]  
(9.14)

or

\[
\text{BSF} = \text{TAR}(d_m, r_d)
\]  
(9.15)

where \( r_{d_m} \) is the field size at the depth \( d_m \) of maximum dose.

\(^3\)A \( 0 \times 0 \) field is a hypothetical field in which the depth dose is entirely due to primary photons.
II. Classical Radiation Therapy

Half-Value Layer (mm Cu)


The backscatter factor, like the tissue-air ratio, is independent of distance from the source and depends only on the beam quality and field size. Figure 9.8 shows backscatter factors for various quality beams and field areas. Whereas BSF increases with field size, its maximum value occurs for beams having a half-value layer between 0.6 and 0.8 mm Cu, depending on field size. Thus, for the orthovoltage beams with usual filtration, the backscatter factor can be as high as 1.5 for large field sizes. This amounts to a 50% increase in dose near the surface compared with the dose in free space or, in terms of exposure, 50% increase in exposure on the skin compared with the exposure in air.

For megavoltage beams (60Co and higher energies), the backscatter factor is much smaller. For example, BSF for a 10 x 10-cm field for 60Co is 1.036. This means that the D_{max} will be 3.6% higher than the dose in free space. This increase in dose is the result of radiation scatter reaching the point of D_{max} from the overlying and underlying tissues. As the beam energy is increased, the scatter is further reduced and so is the backscatter factor. Above about 8 MV, the scatter at the depth of D_{max} becomes negligibly small and the backscatter factor approaches its minimum value of unity.

C. Relationship between TAR and Percent Depth Dose

Tissue-air ratio and percent depth dose are interrelated. The relationship can be derived as follows: Considering Fig. 9.9A, let TAR(d,r) be the tissue-air ratio at point Q for a field size r at depth d. Let r be the field size at the surface, f be the SSD, and d_{ref} be the reference depth of maximum dose at point P. Let D_{pf}(P) and D_{pQ}(Q) be the doses in free space at points P and Q, respectively (Fig. 9.9B,C). D_{pf}(P) and D_{pQ}(Q) are related by inverse square law.

\[
\frac{D_{pQ}(Q)}{D_{pf}(P)} = \left( \frac{f + d_{ref}}{f + d} \right)^2
\] (9.16)
The field sizes \( r \) and \( r_d \) are related by:

\[
r_d = r \cdot \frac{f + d}{f}
\]  

(9.17)

By definition of TAR,

\[
\text{TAR}(d, r_d) = \frac{D_d(Q)}{D_g(Q)}
\]  

(9.18)

or

\[
D_d(Q) = \text{TAR}(d, r_d) \cdot D_g(Q)
\]  

(9.19)

Since

\[
D_{\text{max}}(P) = D_g(P) \cdot \text{BSF}(r)
\]  

(9.20)

and, by definition, the percent depth dose \( P(d, r, f) \) is given by:

\[
P(d, r, f) = \frac{D_d(Q)}{D_{\text{max}}(P)} \cdot 100
\]  

(9.21)

we have, from Equations 9.19, 9.20, and 9.21,

\[
P(d, r, f) = \text{TAR}(d, r_d) \cdot \frac{1}{\text{BSF}(r)} \cdot \frac{D_g(Q)}{D_g(P)} \cdot 100
\]  

(9.22)

From Equations 9.16 and 9.22,

\[
P(d, r, f) = \text{TAR}(d, r_d) \cdot \frac{1}{\text{BSF}(r)} \left( \frac{f + d_m}{f + d} \right)^2 \cdot 100
\]  

(9.23)

C.1. Conversion of Percent Depth Dose from One SSD to Another—the TAR Method

In section 9.3C, we discussed a method of converting percent depth dose from one SSD to another. That method used the Mayneord \( F \) factor which is derived solely from inverse square law considerations. A more accurate method is based on the interrelationship between percent depth dose and TAR. This TAR method can be derived from Equation 9.23 as follows.

Suppose \( f_1 \) is the SSD for which the percent depth dose is known and \( f_2 \) is the SSD for which the percent depth dose is to be determined. Let \( r \) be the field size at the surface.
and \( d \) be the depth, for both cases. Referring to Fig. 9.6, let \( r_{d,f_1} \) and \( r_{d,f_2} \) be the field sizes projected at depth \( d \) in Fig. 9.6A,B, respectively.

\[
r_{d,f_1} = r \cdot \frac{f_1 + d}{f_1} \tag{9.24}
\]

\[
r_{d,f_2} = r \cdot \frac{f_2 + d}{f_2} \tag{9.25}
\]

From Equation 9.23,

\[
P(d,r,f_1) = \text{TAR}(d,r_{d,f_1}) \cdot \frac{1}{\text{BSF}(r)} \cdot \left( \frac{f_1 + d_m}{f_1 + d} \right)^2 \cdot 100
\]

and

\[
P(d,r,f_2) = \frac{\text{TAR}(d,r_{d,f_2})}{\text{TAR}(d,r_{d,f_1})} \cdot \frac{1}{\text{BSF}(r)} \cdot \left( \frac{f_2 + d_m}{f_2 + d} \right)^2 \cdot 100
\]

From Equations 9.26 and 9.27, the conversion factor is given by:

\[
\frac{P(d,r,f_2)}{P(d,r,f_1)} = \frac{\left( \frac{f_1 + d}{f_1 + d} \right)^2 \cdot \left( \frac{f_2 + d_m}{f_2 + d} \right)^2}{\text{TAR}(d,r_{d,f_1})} \cdot \frac{\text{BSF}(r/\sqrt{F})}{\text{BSF}(r)} \cdot F
\]

The last term in the brackets is the Mayneord factor. Thus the TAR method corrects the Mayneord \( F \) factor by the ratio of TARs for the fields projected at depth for the two SSDs.

Burns (22) has developed the following equation to convert percent depth dose from one SSD to another:

\[
P(d,r,f_2) = P \left( d, \frac{r}{\sqrt{F}}, f_1 \right) \cdot \frac{\text{BSF}(r/\sqrt{F})}{\text{BSF}(r)} \cdot F
\]

where \( F \) is the Mayneord \( F \) factor given by

\[
\left( \frac{f_1 + d}{f_1 + d} \right)^2 \cdot \left( \frac{f_2 + d_m}{f_2 + d} \right)^2
\]

Equation 9.29 is based on the concept that TARs are independent of the source distance. Burns's equation may be used in a situation where TARs are not available but instead a percent depth dose table is available at a standard SSD along with the backscatter factors for various field sizes.

As mentioned earlier, for high-energy x-rays, that is, above 8 MV, the variation of percent depth dose with field size is small and the backscatter is negligible. Equations 9.28 and 9.29 then simplify to a use of Mayneord \( F \) factor.

**Practical Examples**

In this section, I will present examples of typical treatment calculations using the concepts of percent depth dose, backscatter factor, and tissue-air ratio. Although a more general system of dosimetric calculations will be presented in the next chapter, these examples are presented here to illustrate the concepts presented thus far.

**Example 2**

A patient is to be treated with an orthovoltage beam having a half-value layer of 3 mm Cu. Supposing that the machine is calibrated in terms of exposure rate in air, find the time required to deliver 200 cGy (rad) at 5 cm depth, given the following data: exposure rate = 100 R/min at 50 cm, field size = 8 x 8 cm, SSD = 50 cm, percent depth dose = 64.8, backscatter factor = 1.20, and rad/R = 0.95 (check these data in reference 5).
Dose rate in free space = exposure rate × rad/R factor × $A_{eq}$

$= 100 \times 0.95 \times 1.00$

$= 95 \text{ cGy/min}$

$D_{max}$ rate = dose rate free space × BSF

$= 95 \times 1.20$

$= 114 \text{ cGy/min}$

Tumor dose to be delivered = 200 cGy

$D_{max}$ to be delivered = \[ \frac{\text{tumor dose}}{\text{percent depth dose}} \times 100 \]

$= \frac{200}{64.8} \times 100$

$= 308.6 \text{ cGy}$

Treatment time = \[ \frac{D_{max} \text{ to be delivered}}{D_{max} \text{ rate}} \]

$= \frac{308.6}{114}$

$= 2.71 \text{ min}$

**Example 3**

A patient is to be treated with $^{60}$Co radiation. Supposing that the machine is calibrated in air in terms of dose rate free space, find the treatment time to deliver 200 cGy (rad) at a depth of 8 cm, given the following data: dose rate free space = 150 cGy/min at 80.5 cm for a field size of 10 × 10 cm, SSD = 80 cm, percent depth dose = 64.1, and backscatter factor = 1.036.

$D_{max}$ rate = 150 × 1.036 = 155.4 cGy/min

$D_{max} = \frac{200}{64.1} \times 100 = 312 \text{ cGy}$

Treatment time = \[ \frac{312}{155.4} \] = 2.01 min

**Example 4**

Determine the time required to deliver 200 cGy (rad) with a $^{60}$Co γ ray beam at the isocenter (a point of intersection of the collimator axis and the gantry axis of rotation) which is placed at a 10 cm depth in a patient, given the following data: SAD = 80 cm, field size = 6 × 12 cm (at the isocenter), dose rate free space at the SAD for this field = 120 cGy/min and TAR = 0.681.

$A/P$ for 6 × 12 cm field = \[ \frac{6 \times 12}{2(6 \times 12)} = 2 \]

Side of equivalent square field = 4 × $A/P$ = 8 cm

TAR(10, 8 × 8) = 0.681 (given)

$D_d = 200 \text{ cGy (given)}$

Since TAR = $D_d/D_f$

$D_f = \frac{200}{0.681} = 293.7 \text{ cGy}$
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D. Calculation of Dose in Rotation Therapy

The concept of tissue-air ratios is most useful for calculations involving isocentric techniques of irradiation. Rotation or arc therapy is a type of isocentric irradiation in which the source moves continuously around the axis of rotation.

The calculation of depth dose in rotation therapy involves the determination of average TAR at the isocenter. The contour of the patient is drawn in a plane containing the axis of rotation. The isocenter is then placed within the contour (usually in the middle of the tumor or a few centimeters beyond it) and radii are drawn from this point at selected angular intervals (e.g., 20 degrees) (Fig. 9.10). Each radius represents a depth for which TAR can be obtained from the TAR table, for the given beam energy and field size defined at the isocenter. The TARs are then summed and averaged to determine TAR, as illustrated in Table 9.3.

Example 5

For the data given in Table 9.3, determine the treatment time to deliver 200 cGy (rad) at the center of rotation, given the data: dose rate free space for 6 x 6-cm field at the SAD

\[ D_e \text{ rate} = 120 \text{ cGy/min (given)} \]
\[ \text{Treatment time} = \frac{293.7}{120} = 2.45 \text{ min} \]

TABLE 9.3. DETERMINATION OF AVERAGE TAR AT THE CENTER OF ROTATION

<table>
<thead>
<tr>
<th>Angle</th>
<th>Depth along Radius</th>
<th>TAR</th>
<th>Angle</th>
<th>Depth along Radius</th>
<th>TAR</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>16.5</td>
<td>0.444</td>
<td>180</td>
<td>16.2</td>
<td>0.450</td>
</tr>
<tr>
<td>20</td>
<td>16.0</td>
<td>0.456</td>
<td>200</td>
<td>16.2</td>
<td>0.450</td>
</tr>
<tr>
<td>40</td>
<td>14.8</td>
<td>0.499</td>
<td>220</td>
<td>14.6</td>
<td>0.499</td>
</tr>
<tr>
<td>60</td>
<td>11.0</td>
<td>0.614</td>
<td>240</td>
<td>12.4</td>
<td>0.563</td>
</tr>
<tr>
<td>80</td>
<td>9.0</td>
<td>0.691</td>
<td>260</td>
<td>11.2</td>
<td>0.606</td>
</tr>
<tr>
<td>100</td>
<td>9.4</td>
<td>0.681</td>
<td>280</td>
<td>11.0</td>
<td>0.614</td>
</tr>
<tr>
<td>120</td>
<td>11.4</td>
<td>0.597</td>
<td>300</td>
<td>12.0</td>
<td>0.580</td>
</tr>
<tr>
<td>140</td>
<td>14.0</td>
<td>0.515</td>
<td>320</td>
<td>14.2</td>
<td>0.507</td>
</tr>
<tr>
<td>160</td>
<td>15.6</td>
<td>0.470</td>
<td>340</td>
<td>16.0</td>
<td>0.456</td>
</tr>
</tbody>
</table>

\(^{60}\text{Co beam, field size at the isocenter = 6 x 6 cm. Average tissue-air ratio (TAR) = 9.692/18 = 0.538.}\)
9. Dose Distribution and Scatter Analysis

9.5. SCATTER-AIR RATIO

Scatter-air ratios are used for the purpose of calculating scattered dose in the medium. The computation of the primary and the scattered dose separately is particularly useful in the dosimetry of irregular fields.

Scatter-air ratio may be defined as the ratio of the scattered dose at a point in the phantom to the dose in free space at the same point. The scatter-air ratio like the tissue-air ratio is independent of the source-to-surface distance but depends on the beam energy, depth, and field size.

Because the scattered dose at a point in the phantom is equal to the total dose minus the primary dose at that point, scatter-air ratio is mathematically given by the difference between the TAR for the given field and the TAR for the 0 \times 0 field.

\[
\text{SAR}(d,r_d) = \text{TAR}(d,r_d) - \text{TAR}(d,0)
\]  

(9.30)

Here TAR\((d,0)\) represents the primary component of the beam.

Because SARs are primarily used in calculating scatter in a field of any shape, SARs are tabulated as functions of depth and radius of a circular field at that depth. Also, because SAR data are derived from TAR data for rectangular or square fields, radii of equivalent circles may be obtained from the table in reference 5 or by Equation 9.7.

A. Dose Calculation in Irregular Fields—Clarkson’s Method

Any field other than the rectangular, square, or circular field may be termed irregular. Irregularly shaped fields are encountered in radiation therapy when radiation sensitive structures are shielded from the primary beam or when the field extends beyond the irregularly shaped patient body contour. Examples of such fields are the mantle and inverted Y fields used for the treatment of Hodgkin’s disease. Since the basic data (percent depth dose, tissue-air ratios, or tissue-maximum ratios—to be discussed later) are available usually for rectangular fields, methods are required to use these data for general cases of irregularly shaped fields. One such method, originally proposed by Clarkson (23) and later developed by Cunningham (24,25), has proved to be the most general in its application.

Clarkson’s method is based on the principle that the scattered component of the depth dose, which depends on the field size and shape, can be calculated separately from the primary component which is independent of the field size and shape. A special quantity, SAR, is used to calculate the scattered dose. This method has been discussed in detail in the literature (26,27) and only a brief discussion will be presented here.

Let us consider an irregularly shaped field as shown in Fig. 9.11. Assume this field cross-section to be at depth \(d\) and perpendicular to the beam axis. Let \(Q\) be the point of calculation in the plane of the field cross-section. Radii are drawn from \(Q\) to divide the field into elementary sectors. Each sector is characterized by its radius and can be considered as part of a circular field of that radius. If we suppose the sector angle is 10 degrees, then the scatter contribution from this sector will be \(10^\circ / 360^\circ = 1/36\) of that contributed by a
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FIG. 9.11. Outline of mantle field in a plane perpendicular to the beam axis and at a specified depth. Radii are drawn from point Q, the point of calculation. Sector angle = 10 degrees. (Redrawn from American Association of Physicists in Medicine, Dosimetry workshop: Hodgkin's disease. Chicago, IL, M.D. Anderson Hospital, Houston, TX, Radiological Physics Center, 1970.)

A circular field of that radius and centered at Q. Thus the scatter contribution from all the sectors can be calculated and summed by considering each sector to be a part of its own circle the scatter-air ratio of which is already known and tabulated.

Using an SAR table for circular fields, the SAR values for the sectors are calculated and then summed to give the average scatter-air ratio (SAR) for the irregular field at point Q. For sectors passing through a blocked area, the net SAR is determined by subtracting the scatter contribution by the blocked part of sector. For example, net (SAR)_{QC} = (SAR)_{QC} - (SAR)_{QB} + (SAR)_{CQ}.

The computed SAR is converted to average tissue-air ratio (TAR) by the equation:

$$\overline{TAR} = TAR(0) + SAR$$

where $TAR(0)$ is the tissue-air ratio for $0 \times 0$ field, that is,

$$TAR(0) = e^{-\mu(d-d_a)}$$

where $\mu$ is the average linear attenuation coefficient for the beam and $d$ is the depth of point Q.

The percent depth dose ($%DD$) at $Q$ may be calculated relative to $D_{max}$ on the central axis using Equation 9.23.

$$%DD = 100 \times \overline{TAR} \times \left( \frac{f + d_m}{f + d} \right)^2 / \text{BSF}$$

where BSF is the backscatter factor for the irregular field and can be calculated by Clarkson's method. This involves determining $TAR$ at the depth $d_m$ on the central axis, using the field contour or radii projected at the depth $d_m$. 

In clinical practice, additional corrections are usually necessary such as for the variation of SSD within the field and the primary beam profile. The details of these corrections will be discussed in the next chapter.

REFERENCES

Several methods are available for calculating absorbed dose in a patient. Two of these methods using percent depth doses and tissue-air ratios (TARs) were discussed in Chapter 9. However, there are some limitations to these methods. For example, the dependence of percent depth dose on source-to-surface distance (SSD) makes this quantity unsuitable for isocentric techniques. Although tissue-air ratios (TAR) and scatter-air ratios (SAR) eliminate that problem, their application to beams of energy higher than those of $^{60}$Co has been seriously questioned (1-3) as they require measurement of dose in free space. As the beam energy increases, the size of the chamber build-up cap for in-air measurements has to be increased and it becomes increasingly difficult to calculate the dose in free space from such measurements. In addition, the material of the build-up cap is usually different from that of the phantom and this introduces a bias or uncertainty in the TAR measurements.

In order to overcome the limitations of the TAR, Katzmark et al. (1) introduced the concept of tissue-phantom ratio (TPR). This quantity retains the properties of the TAR but limits the measurements to the phantom rather than in air. A few years later, Holt et al. (4) introduced yet another quantity, tissue-maximum ratio (TMR), which also limits the measurements to the phantom.

In this chapter, I develop a dosimetric system based on the TMR concept, although a similar system can also be derived from the TPR concept (5).

### 10.1. DOSE CALCULATION PARAMETERS

The dose to a point in a medium may be analyzed into primary and scattered components. The primary dose is contributed by the initial or original photons emitted from the source and the scattered dose is the result of the scattered photons. The scattered dose can be further analyzed into collimator and phantom components, because the two can be varied independently by blocking. For example, blocking a portion of the field does not significantly change the output or exposure in the open portion of the beam (6,7) but may substantially reduce the phantom scatter.

The above analysis presents one practical difficulty, namely, the determination of primary dose in a phantom which excludes both the collimator and phantom scatter. However, for megavoltage photon beams, it is reasonably accurate to consider collimator scatter as part of the primary beam so that the phantom scatter could be calculated separately. We, therefore, define an effective primary dose as the dose due to the primary photons as well as those scattered from the collimating system. The effective primary in a phantom may be thought of as the dose at depth minus the phantom scatter. Alternatively, the effective primary dose may be defined as the depth dose expected in the field when scattering volume is reduced to zero while keeping the collimator opening constant.

Representation of primary dose by the dose in a $0 \times 0$ field poses conceptual problems because of the lack of lateral electronic equilibrium in narrow fields in megavoltage photon beams. This issue has been debated in the literature (8-10), but practical solutions are still not agreed on. Systems that use electron transport in the calculation of primary and scattered components of dose would be appropriate but are not as yet fully developed and
implemented. Until then, the concept of $0 \times 0$ field to represent primary beams with the implicit assumption that lateral electronic equilibrium exists at all points will continue to be used for routine dosimetry.

A. Collimator Scatter Factor

The beam output (exposure rate, dose rate in free space, or energy fluence rate) measured in air depends on the field size. As the field size is increased, the output increases because of the increased collimator scatter, which is added to the primary beam.

The collimator scatter factor ($S_c$) is commonly called the output factor and may be defined as the ratio of the output in air for a given field to that for a reference field (e.g., $10 \times 10$ cm). $S_c$ may be measured with an ion chamber with a build-up cap of a size large enough to provide maximum dose build-up for the given energy beam. The measurement setup is shown in Fig. 10.1A. Readings are plotted against field size (side of equivalent square or area/perimeter [$A/P$]) and the values are normalized to the reference field ($10 \times 10$ cm).

In the measurement of $S_c$, the field must fully cover the build-up cap for all field sizes if measurements are to reflect relative photon fluences. For small fields, one may take the measurements at large distances from the source so that the smallest field covers build-up cap. Normally, the collimator scatter factors are measured at the source-to-axis distance (SAD). However, larger distances can be used provided the field sizes are all defined at the SAD.

B. Phantom Scatter Factor

The phantom scatter factor ($S_p$) takes into account the change in scatter radiation originating in the phantom at a reference depth as the field size is changed. $S_p$ may be defined as the ratio of the dose rate for a given field at a reference depth (e.g., depth of maximum dose) to the dose rate at the same depth for the reference field size (e.g., $10 \times 10$ cm), with the same collimator opening. In this definition, it should be noted that $S_p$ is related to the changes in the volume of the phantom irradiated for a fixed collimator opening. Thus one

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1 Collimator scatter includes photons scattered by all components of the machine head in the path of the beam.
could determine $S_p$, at least in concept, by using a large field incident on phantoms of various cross-sectional sizes.

For photon beams for which backscatter factors can be accurately measured (e.g., $^{60}$Co and 4 MV), $S_p$ factor at the depth of maximum dose may be defined simply as the ratio of backscatter factor (BSF) for the given field to that for the reference field (see Appendix, section A). Mathematically,

$$S_p(r) = \frac{\text{BSF}(r)}{\text{BSF}(r_0)}$$

where $r_0$ is the side of the reference field size ($10 \times 10$ cm).

A more practical method of measuring $S_p$, which can be used for all beam energies, consists of indirect determination from the following equation (for derivation, see Appendix, section A):

$$S_p(r) = \frac{S_{sp}(r)}{S_{sc}(r)}$$

where $S_{sp}(r)$ is the total scatter factor defined as the dose rate at a reference depth for a given field size $r$ divided by the dose rate at the same point and depth for the reference field size ($10 \times 10$ cm) (Fig. 10.1B). Thus $S_{sp}(r)$ contains both the collimator and phantom scatter and when divided by $S_{sc}(r)$ yields $S_p(r)$.

Since $S_p$ and $S_{sp}$ are defined at the reference depth of $D_{max}$, actual measurement of these factors at this depth may create problems because of the possible influence of contaminant electrons incident on the phantom. This can be avoided by making measurements at a greater depth (e.g., 10 cm) and converting the readings to the reference depth of $D_{max}$ by using percent depth dose data, presumably measured with a small-diameter chamber. The rationale for this procedure is the same as for the recommended depths of calibration (11).

**C. Tissue-Phantom and Tissue-Maximum Ratios**

The TPR is defined as the ratio of the dose at a given point in phantom to the dose at the same point at a fixed reference depth, usually 5 cm. This is illustrated in Fig. 10.2. The corresponding quantity for the scattered dose calculation is called the scatter-phantom ratio (SPR), which is analogous in use to the scatter-air ratio discussed in the previous chapter. Details of the TPR and SPR concepts have been discussed in the literature (1,3,5).

TPR is a general function which may be normalized to any reference depth. But there is no general agreement concerning the depth to be used for this quantity, although a 5-cm depth is the usual choice for most beam energies. On the other hand, the point of central axis $D_{max}$ has a simplicity that is very desirable in dose computations. If adopted as a fixed
reference depth, the quantity TPR gives rise to the TMR. Thus TMR is a special case of TPR and may be defined as the ratio of the dose at a given point in phantom to the dose at the same point at the reference depth of maximum dose (Fig. 10.2).

For megavoltage beams in the range of 20 to 45 MV, the depth of maximum dose \( d_m \) has been found to depend significantly on field size \((12, 13)\) as well as on SSD \((14, 15)\). For the calculative functions to be independent of machine parameters, they should not involve measurements in the build-up region. Therefore, the reference depth must be equal to or greater than the largest \( d_m \). Since \( d_m \) tends to decrease with field size \((12)\) and increase with SSD \((14)\), one should choose \((d_m)\) for the smallest field and the largest SSD. In practice, one may plot \([\%DD \times (SSD + d)^2]\) as a function of depth \(d\) to find \(d_m\) \((15)\). This eliminates dependence on SSD. The maximum \(d_m\) can then be obtained by plotting \(d_m\) as a function of field size and extrapolating to \(0 \times 0\) field size.

The reference depth of maximum dose \(r_0\), as determined above, should be used for percent depth dose, TMR, and \(S_p\) factors, irrespective of field size and SSD.

C.1. Properties of TMR

The concept of tissue-maximum ratio is based on the assumption that the fractional scatter contribution to the depth dose at a point is independent of the divergence of the beam and depends only on the field size at the point and the depth of the overlying tissue. This has been shown to be essentially true by Johns et al. \((16)\). This principle, which also underlies TAR and TPR, makes all these functions practically independent of source-surface distance. Thus a single table of TMRs can be used for all SSDs for each radiation quality.

Figure 10.3 shows TMR data for 10-MV x-ray beams as an example. The curve for \(0 \times 0\) field size shows the steepest drop with depth and is caused entirely by the primary beam. For megavoltage beams, the primary beam attenuation can be approximately represented by:

\[
TMR(d, 0) = e^{-\mu(d - r_0)}
\]

where \(\mu\) is the effective linear attenuation coefficient and \(r_0\) is the reference depth of maximum dose. \(\mu\) can be determined from TMR data by plotting \(\mu\) as a function of field size (side of equivalent square) and extrapolating it back to \(0 \times 0\) field.

TMR and percent depth dose \(P\) are interrelated by the following equation (see Appendix, section B, for derivation).

\[
TMR(d, r_0) = \left(\frac{P(d, r, f)}{100}\right) \left(\frac{f + d}{f + r_0}\right)^2 \left(\frac{S_p(r_0)}{S_p(r_d)}\right)
\]

where

\[
f = SSD, \quad r_d = r \cdot \left(\frac{f + d}{f}\right)
\]

\[
r_0 = r \cdot \left(\frac{f + r_0}{f}\right)
\]

Here the percent depth dose is referenced against the dose at depth \(r_0\) so that \(P(r_0, r, f) = 100\) for all field sizes and SSDs.

Although TMRs can be measured directly, they can also be calculated from percent depth doses, as shown by Equation 10.4. For \(^{60}\)Co, Equations 10.2 and 10.4 can be used to calculate TMRs. In addition, TMRs can be derived from TAR data in those cases, such as \(^{60}\)Co, where TARs are accurately known:

\[
TMR(d, r_d) = \frac{TAR(d, r_d)}{BSF(r_d)}
\]
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**D. Scatter-Maximum Ratio**

The scatter-maximum ratio (SMR), like the SAR, is a quantity designed specifically for the calculation of scattered dose in a medium. It may be defined as the ratio of the scattered dose at a given point in phantom to the effective primary dose at the same point at the reference depth of maximum dose (5). Mathematically,

$$\text{SMR}(d, r_d) = \text{TMR}(d, r_d) \left( \frac{S_p(r_d)}{S_p(0)} \right) - \text{TMR}(d, 0)$$  \hspace{1cm} (10.6)

For derivation of the above equation, see Appendix, section C.

From Equations 10.1, 10.5, and 10.6, it can be shown that for $^{60}$Co $\gamma$ rays, SMRs are approximately the same as SARs. However, for higher energies, SMRs should be calculated from TMRs by using Equations 9.7 and 10.6.

Another interesting relationship can be obtained at the reference depth of maximum dose ($t_0$). Since TMR at depth $t_0$ is unity by definition, Equation 10.6 becomes:

$$\text{SMR}(t_0, r_{t_0}) = \frac{S_p(r_{t_0})}{S_p(0)} - 1$$  \hspace{1cm} (10.7)

This equation will be used in section 10.2C.
10.2. PRACTICAL APPLICATIONS

Radiotherapy institutions vary in their treatment techniques and calibration practices. For example, some rely exclusively on the SSD or the SAD (isocentric) type techniques, while others use both. Accordingly, units are calibrated in air or in phantom at a suitable reference depth. In addition, clinical fields, although basically rectangular or square, are often shaped irregularly to protect critical or normal regions of the body. Thus a calculation system must be generally applicable to the above practices, with acceptable accuracy and simplicity for routine use.

A. Accelerator Calculations

A.1. SSD Technique

Percent depth dose is a suitable quantity for calculations involving SSD techniques. Machines are usually calibrated to deliver 1 rad (10^{-2} Gy) per monitor unit (MU) at the reference depth \( t_0 \), for a reference field size \( 10 \times 10 \) cm and a source-to-calibration point distance of SCD. Assuming that the \( S_c \) factors relate to collimator field sizes defined at the SAD, the monitor units necessary to deliver a certain tumor dose (TD) at depth \( d \) for a field size \( r \) at the surface at any SSD are given by:

\[
MU = \frac{TD \times 100}{K \times (\%DD)_d \times S_c(r) \times S_p(r) \times (SSD \text{ factor})}
\]  

(10.8)

where \( K \) is 1 rad per MU, \( r_c \) is the collimator field size, given by:

\[ r_c = \frac{r_{\text{SAD}}}{r_{\text{SSD}}} \]

and:

\[
\text{SSD factor} = \left( \frac{\text{SCD}}{\text{SSD} + t_0} \right)^2
\]

It must be remembered that, whereas the field size for the \( S_c \) is defined at the SAD, \( S_p \) relates to the field irradiating the patient.

Example 1

A 4-MV linear accelerator is calibrated to give 1 rad (10^{-2} Gy) per MU in phantom at a reference depth of maximum dose of 1 cm, 100-cm SSD, and 10 × 10 cm field size. Determine the MU values to deliver 200 rads to a patient at 100-cm SSD, 10-cm depth, and 15 × 15 cm field size, given \( S_c(15 \times 15) = 1.020 \), \( S_p(15 \times 15) = 1.010 \), \%DD = 65.1. From Equation 10.8,

\[
MU = \frac{200 \times 100}{65.1 \times 1.020 \times 1.010 \times 1} = 298
\]

A form for treatment calculations is shown in Fig. 10.4 with the above calculations filled in.

Example 2

Determine the MU for the treatment conditions given in Example 1 except that the treatment SSD is 120 cm, given \( S_c(12.5 \times 12.5) = 1.010 \) and \%DD for the new SSD is 66.7.
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Field size projected at SAD(= 100 cm) = 15 \times \frac{100}{120} = 12.5 \text{ cm}

S_r(12.5 \times 12.5) \text{ is given as 1.010 and } S_p(15 \times 15) = 1.010

SSD factor = \left( \frac{100 + 1}{120 + 1} \right)^2 = 0.697

From Equation 10.8,

MU = \frac{200 \times 100}{66.7 \times 1.010 \times 1.010 \times 0.697} = 442

A.2. Isocentric Technique

TMR is the quantity of choice for dosimetric calculations involving isocentric techniques. Since the unit is calibrated to give 1 rad (10^{-2} \text{ Gy})/MU at the reference depth \( r_0 \), calibration distance SCD, and for the reference field (10 \times 10 \text{ cm}), then the monitor units necessary to deliver isocenter dose (ID) at depth \( d \) are given by:

\[
MU = \frac{\text{ID}}{K \times TMR(d, r_0) \times S_r(r_0) \times S_p(r_0) \times \text{SAD factor}}
\]  

where

\[
\text{SAD factor} = \left( \frac{\text{SCD}}{\text{SAD}} \right)^2
\]

Example 3

A tumor dose of 200 rads is to be delivered at the isocenter which is located at a depth of 8 cm, given 4-MV x-ray beam, field size at the isocenter = 6 \times 6 \text{ cm}, \( S_r(6 \times 6) = 0.970 \).
\( S_p(6 \times 6) = 0.990 \), machine calibrated at \( \text{SCD} = 100 \text{ cm} \), \( \text{TMR}(8, 6 \times 6) = 0.787 \).

Since the calibration point is at the SAD, SAD factor = 1. Thus, using Equation 10.9,

\[
\text{MU} = \frac{200}{0.787 \times 0.970 \times 0.990 \times 1} = 265
\]

**Example 4**

Calculate MU values for the case in Example 3, if the unit is calibrated nonisocentrically, i.e., source to calibration point distance = 101 cm.

\[
\text{SAD factor} = \left( \frac{101}{100} \right)^2 = 1.020
\]

Thus,

\[
\text{MU} = \frac{200}{0.787 \times 0.970 \times 0.99 \times 1.02} = 260
\]

**B. Cobalt-60 Calculations**

The above calculation system is sufficiently general that it can be applied to any radiation generator, including \(^{60}\)Co. In the latter case, the machine can be calibrated either in air or in phantom provided the following information is available: (a) dose rate \( D_0(\rho, \theta, \phi) \) in phantom at depth \( \rho \) of maximum dose for a reference field size \( \theta \) and standard SSD \( \phi \); (b) \( S_\rho \); (c) \( S_\theta \); (d) percent depth doses; and (e) TMR values. If universal depth dose data for \(^{60}\)Co (16) are used, then the \( S_p \) and TMRs can be obtained by using Equations 10.1 and 10.5. In addition, the SSD used in these calculations should be confined to a range for which the output in air obeys an inverse square law for a constant collimator opening.

A form for cobalt calculations is presented in Fig. 10.5.

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**Fig. 10.5.** Calculation sheet—cobalt-60.
**Example 5**

A tumor dose of 200 rads is to be delivered at a 8-cm depth, using 15 × 15-cm field size, 100-cm SSD, and penumbra trimmers up. The unit is calibrated to give 130 rads/min in phantom at a 0.5-cm depth for a 10 × 10-cm field with trimmers up and SSD = 80 cm. Determine the time of irradiation, given $S_r(12 \times 12) = 1.012$, $S_p(15 \times 15) = 1.014$, and %DD (8, 15 × 15, 100) = 68.7.

Field size projected at SAD(= 80 cm)

\[ = 15 \times \frac{80}{100} = 12 \text{ cm} \times 12 \text{ cm} \]

SSD factor

\[ = \left( \frac{80 + 0.5}{100 + 0.5} \right)^2 = 0.642 \]

Time

\[ = \frac{(TD \times 100)}{\left[ D_p(0.5, 10 \times 10, 80) \times \%DD(8, 15 \times 15, 100) \times S_r(12 \times 12) \right.} \times \left. S_p(15 \times 15) \times \text{SSD factor} \right] / 200 \times 100 \]

\[ = \frac{130 \times 68.7 \times 1.012 \times 1.014 \times 0.642}{3.40 \text{ min}} \]

C. Irregular Fields

Dosimetry of irregular fields using TMRs and SMRs is analogous to the method using TARs and SARs (section 9.5). Since the mathematical rationale of the method has been discussed in detail in the literature (5), only a brief outline will be presented here to illustrate the procedure.

An irregular field at depth $d$ may be divided into $n$ elementary sectors with radii emanating from point $Q$ of calculation (Fig. 9.10). A Clarkson type integration (Chapter 9) may be performed to give averaged scatter-maximum ratio ($\overline{\text{SMR}}(d, r_d)$) for the irregular field $r_d$:

\[
\overline{\text{SMR}}(d, r_d) = \frac{1}{n} \sum_{i=1}^{n} \text{SMR}(d, r_i)
\]

where $r_i$ is the radius of the $i$th sector at depth $d$ and $n$ is the total number of sectors ($n = 2\pi/\Delta\theta$, where $\Delta\theta$ is the sector angle).

The computed $\overline{\text{SMR}}(d, r_d)$ is then converted to $\overline{TMR}(d, r_d)$ by using Equation 10.6.

\[
\overline{TMR}(d, r_d) = [\text{TMR}(d, 0) + \overline{\text{SMR}}(d, r_d)] \times \frac{S_p(0)}{S_p(r_d)}
\]

where $\overline{S_p}$, $(r_d)$ is the averaged $S_p$ for the irregular field and $S_p(0)$ is the $S_p$ for the $0 \times 0$ area field. The above equation is strictly valid only for points along the central axis of a beam that is normally incident on an infinite phantom with flat surface. For off-axis points in a beam with nonuniform primary dose profile, one should write

\[
\overline{TMR}(d, r_d) = [K_p \cdot \text{TMR}(d, 0) + \overline{\text{SMR}}(d, r_d)] \times \frac{S_p(0)}{S_p(r_d)}
\]

where $K_p$ is the off-axis ratio representing primary dose at point $Q$ relative to that at the central axis.

$TMR(d, r_d)$ may be converted into percent depth dose $P(d, n, f)$ by using Equation 10.4.
From Equations 10.7 and 10.13 we get the final expression:

\[
P(d, r, f) = 100[K_p \cdot \text{TMR}(d, 0) + \text{SMR}(d, r_d)]
\]

\[
\times \frac{S_p(0)}{S_p(r_d)} \times \frac{S_p(r_a)}{S_p(r_b)} \times \left(\frac{f + t_0}{f + d}\right)^2
\]

(10.13)

Thus the calculation of percent depth dose for an irregular field requires a Clarkson integration over the function SMR both at the point of calculation Q as well as at the reference depth \(t_0\) at central axis.

C.1. SSD Variation Within the Field

The percent depth dose at Q is normalized with respect to the \(D_{\text{max}}\) on the central axis at depth \(t_0\). Let \(f_0\) be the nominal SSD along the central axis, \(g\) be the vertical gap distance, i.e., "gap" between skin surface over Q and the nominal SSD plane, and \(d\) be the depth of Q from skin surface. The percent depth dose is then given by:

\[
\%\text{DD} = 100 \times [K_p \cdot \text{TMR}(d, 0) + \text{SMR}(d, r_d)]
\]

\[
\times \frac{1}{1 + \text{SMR}(t_0, r_b)} \times \left(\frac{f + t_0}{f + g + d}\right)^2
\]

(10.15)

The sign of \(g\) should be set positive or negative, depending on if the SSD over Q is larger or smaller than the nominal SSD.

C.2. Computer Program

A computer algorithm embodying the Clarkson's principle and scatter-air ratios was developed by Cunningham et al. (17) at the Princess Margaret Hospital, Toronto, and was published in 1970. Another program, based on the same principle, was developed by Khan (18) at the University of Minnesota. It was originally written for the CDC-3300 computer using SARs and later rewritten for the Artronix PC-12 and PDP 11/34 computers. The latter versions use SMRs instead of SARs.

The following data are permanently stored in this computer program: (a) a table of SMRs as functions of radii of circular fields and (b) the off-axis ratios \(K_p\), derived from dose profiles at selected depths. These data are then stored in the form of a table of \(K_p\) as a function of \(I/L\) where \(I\) is the lateral distance of a point from the central axis and \(L\) is the distance along the same line to the geometric edge of the beam. Usually large fields are used for these measurements.

The following data are provided for a particular patient:

1. Contour points: the outline of the irregular field can be drawn from the port (field) film with actual blocks or markers in place to define the field. The field contour is then digitized and the coordinates stored in the computer.
2. The coordinates \((x, y)\) of the points of calculation are also entered, including the reference point, usually on the central axis, against which the percent depth doses are calculated.
3. Patient measurements: patient thickness at various points of interest, SSDs, and source-to-film distance are measured and recorded as shown in Fig. 10.6 for a mantle field as an example.

Figure 10.7 shows a daily table calculated by the computer for a typical mantle field. Such a table is useful in programming treatments so that the dose to various regions of the
field can be adjusted. The areas that receive the prescribed dose after a certain number of treatments are shielded for the remaining sessions.

D. Asymmetric Fields

Many of the modern linear accelerators are equipped with x-ray collimators (or jaws) that can be moved independently to allow asymmetric fields with field centers positioned away from the true central axis of the beam. For example, an independent jaw can be moved to block off half of the field along central axis to eliminate beam divergence. This feature is useful for matching adjacent fields. Although this function can also be performed by beam splitters or secondary blocking on a shadow tray, an independent jaw feature reduces the setup time and spares the therapist from handling heavy blocks.

The effect of asymmetric beam collimation on dose distribution has been discussed in the literature (19,20). When a field is collimated asymmetrically, one needs to take into account changes in the collimator scatter, phantom scatter, and off-axis beam quality. The
latter effect arises as a consequence of using beam-flattening filters (thicker in the middle and thinner in the periphery), which results in greater beam hardening close to the central axis compared with the periphery of the beam (21,22).

A dose calculation formalism for asymmetric fields has been developed which is described below.

For a point at the center of an asymmetric field and a lateral distance $x$ away from the beam central axis, the collimator scatter factor may be approximated to a symmetric field of the same collimator opening as that of the given asymmetric field. In other words, the $S$, will depend on the actual collimator opening, ignoring small changes in the scattered photon fluence that may result owing to the change in the angle of the asymmetric jaws relative to the beam. This approximation is reasonable as long as the point of dose calculation is centrally located, that is, away from field edges.

The phantom scatter can also be assumed to be the same for an asymmetric field as for a symmetric field of the same dimension and shape, provided the point of calculation is located away from field edges to avoid penumbral effects.

The primary dose distribution has been shown to vary with lateral distance from central axis because of the change in beam quality, as mentioned earlier. Therefore, the percent depth dose or TMR distribution along the central ray of an asymmetric field is not the same as along the central axis of a symmetric field of the same size and shape. In addition, the incident primary beam fluence at off-axis points varies as a function of distance from the central axis, depending on the flattening filter design. These effects are not emphasized in the dosimetry of symmetric fields, because target doses are usually specified at the beam central axis and the off-axis dose distributions are viewed from the isodose curves. In asymmetric fields, however, the target or the point of interest does not lie on the beam central axis; therefore, an off-axis dose correction may be required in the calculation of target dose. This correction will depend on the depth and the distance from the central axis of the point of interest.

Since beam flatness within the central 80% of the maximum field size is specified within ±3% at a 10-cm depth, ignoring off-axis dose correction in asymmetric fields will introduce errors of that magnitude under these conditions. Thus the off-axis dose
correction will follow changes in the primary beam flatness as a function of depth and
distance from central axis.

In view of the above discussion, the following equations are proposed for the calculation
of monitor units for asymmetric fields.

For SSD type of treatments, Equation 10.8 is modified to:

\[ \text{MU} = \frac{\text{TD} \times 100}{K \times (\%\text{DD})_d \times S_t(r_c) \times S_p(r) \times (\text{SSD factor}) \times \text{OAR}_d(x)} \]  
(10.16)

where \( \text{OAR}_d(x) \) is the primary off-axis ratio at depth \( d \), that is, ratio of primary dose at
the off-axis point of interest to the primary dose at the central axis at the same depth
for a symmetrically wide open field. Primary off-axis ratios may be extracted from depth
dose profiles of the largest field available by subtracting scatter. A direct method consists
of measuring transmitted dose profiles through different thicknesses of an absorber under
“good geometry” conditions (a narrow beam and a large detector-to-absorber distance).
Another direct but approximate method is to measure profiles as a function of depth
for a narrow elongated field (e.g., 5 x 40 cm). Since the primary dose profile is created by
the flattening filter, which has a radial symmetry, primary OAR data can be tabulated as a
function of depth and radial distance from central axis.

For isocentric type of treatments, Equation 10.9 is modified to:

\[ \text{MU} = \frac{\text{ID}}{K \times \text{TMR}(d, r_d) \times S_t(r_c) \times S_p(r_d) \times (\text{SAD factor}) \times \text{OAR}_d(x)} \]  
(10.17)

The above formalism is general and can be used for an off-axis point dose calculation
in symmetric or asymmetric fields generated by blocks or collimators, including multileaf
collimators. For irregularly shaped fields the parameter \( r_d \) is the equivalent field size deter-
mined by Clarkson's technique or geometric approximation (section 10.3). The parameter
\( r_c \) is the collimator opening size projected at the standard SSD.

10.3. OTHER PRACTICAL METHODS OF CALCULATING
DEPTH DOSE DISTRIBUTION

A. Irregular Fields

Clarkson's technique is a general method of calculating depth dose distribution in an
irregularly shaped field, but it is not practical for routine manual calculations. Even when
computerized, it is time-consuming since a considerable amount of input data is required
by the computer program. However, with the exception of mantle, inverted \( Y \) and a few
other complex fields, reasonably accurate calculations can be made for most blocked fields
using an approximate method (18), to be discussed.

Figure 10.8 shows a number of blocked fields encountered in radiotherapy. Approximi-
te rectangles may be drawn containing the point of calculation to include most of the
irradiated area surrounding the point and exclude only those areas that are remote to the
point. In so doing, a blocked area may be included in the rectangle, provided this area
is small and is remotely located relative to that point. The rectangle thus formed may be
called the effective field, while the unblocked field, defined by the collimator, may be called
the collimator field.

Once the effective field has been determined, one may proceed with the usual calcu-
lations as discussed in section 10.2. However, it is important to remember that, whereas
the \( S_t \) is related to the collimator field, the percent depth dose, TMR, or \( S_p \) corresponds
to the effective field.

B. Point Off-Axis

It is possible to calculate depth dose distributions at any point within the field or outside
the field using Clarkson's technique. However, as stated earlier, it is not practical for
manual calculations. Day (24) has proposed a particularly simple calculation method for
rectangular fields. In this method, percent depth dose can be calculated at any point within the medium using the central axis data.

To calculate dose at any point $Q$, the field is imagined to be divided into four sections (Fig. 10.9) and their contribution is computed separately. Thus the dose at depth $d$ along the axis through $Q$ is given by $\frac{1}{4}(\text{sum of central axis dose at depth } d \text{ for fields } 2a \times 2b, 2a \times 2c, 2d \times 2b, \text{ and } 2d \times 2c)$.

Suppose the dose in free space on the central axis through $P$ at SSD + $d_m$ is 100 cGy (rad) and its value at a corresponding point over $Q$ is $K_Q \times 100$, where $K_Q$ is the off-axis ratio determined in air from the primary beam profile. If the BSF and central axis %DD for rectangular fields are available, the dose at depth $d$ along the axis through $Q$ will be given by:

$$K_Q \times 100 \times \frac{1}{4}(\text{sum of BSF } \times \%\text{DD at depth } d \text{ for fields } 2a \times 2b, 2a \times 2c, 2d \times 2b, \text{ and } 2d \times 2c)$$

FIG. 10.9. Day's method of calculating dose at any point $Q$ in a rectangular field. (See text.)
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Since the $D_{\text{max}}$ at $P$ is $100 \times \text{BSF}[(a + d) \times (b + e)]$, the percent depth dose at depth $d$ along the axis through $Q$, relative to $D_{\text{max}}$ at $P$, will be given by:

$$K_Q = \frac{4 \times \text{BSF}[(a + d) \times (b + e)]}{\text{sum of BSF}} \times \% \text{DD at depth } d \text{ for fields } 2a \times 2b, 2a \times 2c, 2d \times 2b, \text{ and } 2d \times 2c$$

**Example 6**

Suppose in Fig. 10.9 that the overall field size is $15 \times 15$ cm. Find the percent depth dose at point $Q$ at 10 cm depth, given $a = 10$, $b = 5$, $c = 10$, and $d = 5$. Assume $^{60}\text{Co}$ beam with $K_Q = 0.98$ and SSD = 80 cm.

Using the above procedure and consulting Table A.9.1 in the Appendix to the book, the required percent depth dose is given by:

$$K_Q \left[ \frac{\text{BSF}(20 \times 10) \times \% \text{DD}(20 \times 10) + \text{BSF}(20 \times 20)}{4 \times \text{BSF}(15 \times 15)} \times \% \text{DD}(20 \times 20) + \text{BSF}(10 \times 10) \times \% \text{DD}(10 \times 10) + \text{BSF}(10 \times 20) \right]$$

or

$$\frac{0.98}{4 \times 1.052} \left[(1.043 \times 56.3) + (1.061 \times 60.2) + (1.036 \times 55.6) + (1.043 \times 56.3)\right] = 55.8$$

In the above example, if the primary beam profile were flat, that is, $K_Q = 1$, the percent depth dose at $Q$ would be 56.9, which is still less than 58.4, the percent depth dose at $P$. This off-axis decrease in dose is due to the reduced scatter at point $Q$ compared with point $P$. Similarly, it can be shown that the magnitude of the reduction in scatter depends on the distance of $Q$ from $P$ as well as depth. Thus the depth dose profile across the field is a function not only of the beam flatness in air but also the depth in the phantom.

For higher-energy beams ($\geq 8$ MV), the above procedure may be further simplified by assuming BSF = 1 for all field sizes. Also, Day's procedure can be adopted using $S_p$ values instead of BSF, since the two quantities are related by Equation 10.1.

**FIG. 10.10.** Calculation of depth dose outside a rectangular field.

(See text.)

C. Point Outside the Field

Day's method can be extended also to the case of determining dose distribution at points outside the field limits. In Fig. 10.10, a rectangular field of dimensions $a \times b$ is shown with the central axis passing through $P$. Suppose $Q$ is a point outside the field at a distance $c$ from the field border. Imagine a rectangle adjacent to the field such that it contains point $Q$ and has dimensions $2c \times b$. Place another rectangle of dimensions $a \times b$ on the other side of $Q$ such that the field on the right of $Q$ is a mirror image of the field on the left, as shown in the figure. The dose at point $Q$ at depth $d$ is then given by subtracting the depth dose at $Q$ for field $2c \times b$ from that for field $(2a + 2c) \times b$ and dividing by 2. The procedure is illustrated by the following example.
Example 7

Suppose it is required to determine percent depth dose at \( Q \) (relative to \( D_{\text{max}} \) at \( P \)) outside a 15 \( \times \) 10-cm field at a distance of 5 cm from the field border. In Fig. 10.10, then, \( a = 15 \), \( b = 10 \), and \( c = 5 \). Suppose \( Q \) is at the center of the middle rectangle of dimensions \( 2c \times b \). Then the dose \( D_Q \) at 10-cm depth is given by:

\[
\frac{1}{2} [D_Q(40 \times 10) - D_Q(10 \times 10)]
\]

If \( D_Q \) is normalized to \( D_{\text{max}} \) at \( P \), one gets the percent depth dose at \( Q \) or \( \%D_Q \).

\[
\%D_Q = \frac{1}{\text{BSF}(15 \times 15)} \cdot \frac{1}{2} (\text{BSF}(40 \times 10) - \text{BSF}(10 \times 10) \times \%\text{DD}(10 \times 10))
\]

Thus for a \( \text{\textsuperscript{60}Co} \) beam at SSD = 80 cm,

\[
\%D_Q = \frac{1}{1.052} \cdot \frac{1}{2} [1.054 \times 58.8 - 1.036 \times 55.6] = 2.1
\]

Again for higher-energy beams, the above procedure is simplified by assuming BSF = 1. Also, if \( S_p \) values are known instead of BSF, the above calculation can be performed by substituting \( S_p \) for BSF.

D. Point Under the Block

As discussed earlier, the dose distribution in a blocked field is best determined by Clarkson's method of irregular field dosimetry. However, if the blocked portion of the field is approximated to a rectangle, a simpler method known as negative field method may be used. The concept of negative field has been described in the literature (25,26). In this method, the dose at any point is equal to the dose from the overall (unblocked) field minus the dose expected if the entire field were blocked, leaving the shielded volume open. In other words, the blocked portion of the field is considered a negative field and its contribution is subtracted from the overall field dose distribution.

A computerized negative field method not only is a fast method of calculating isodose distribution in blocked fields but is very convenient for manual point dose calculation. Its practical usefulness is illustrated by Example 8.

Example 8

A patient is treated with a split field of overall size 15 \( \times \) 15 cm, blocked in the middle to shield a region of size 4 \( \times \) 15 cm on the surface (Fig. 10.11). Calculate (a) the treatment time to deliver 200 cGy (rad) at a 10-cm depth at point \( P \) in the open portion of the field.
and (b) what percentage of that dose is received at point Q in the middle of the blocked area, given $^{60}\text{Co}$ beam, SSD = 80 cm, dose rate free space for a $15 \times 15$-cm field at 80.5 cm = 120 rads/min, lead block thickness = 5 cm with primary beam transmission of 5%, and shadow tray (or block tray) transmission = 0.97.

(a) Approximate equivalent field at point P ≈ $5.5 \times 15$, assuming negligible scatter contribution to P from the other open portion of the field across the blocked area.

\[
A/P(5.5 \times 15) = 2.01
\]

Equivalent square = $4 \times A/P = 8 \times 8$ cm

\[
\%DD(10, 8 \times 8, 80) = 54.0
\]

BSF = 1.029

\[
\text{Treatment time} = \frac{200 \times 100}{120 \times 1.029 \times 54.0 \times 0.97} = 3.09 \text{ min}
\]

(b) 

\[
D_Q = D_Q(15 \times 15) - D_Q(4 \times 15) \times (1 - T)
\]

where $T$ is the transmission factor for the lead block.

\[
D_Q(15 \times 15) = (\text{dose rate free space} \times \text{time}) \times \text{BSF} \times \%DD
\]

\[
= 120 \times 3.09 \times 1.052 \times 58.4
\]

\[= 227.8 \text{ cGy}
\]

\[
D_Q(4 \times 15) = 120 \times 3.09 \times 1.023 \times \frac{52.3}{100} [A/P(4 \times 15) = 1.58.
\]

\[\text{Equivalent square} = 6.3 \times 6.3 \text{ cm}]

\[= 198.4 \text{ cGy}
\]

Thus,

\[
D_Q = 227.8 - 198.4(1 - 0.05)
\]

\[= 39.3 \text{ cGy}
\]

Since $D_p = 200 \text{ cGy} \text{ (given)}$

\[
D_Q \text{ as a percentage of } D_p = \frac{D_Q}{D_p} \times 100 = \frac{39.3}{200} \times 100
\]

\[= 20\%
\]

**Alternative**

Let us project all fields at depth = 10 cm.

\[
\text{Magnification} = \frac{80 + 10}{80} = 1.125
\]

Projected fields.

\[(15 \times 15) \text{ cm} \times 1.125 = 17 \times 17 \text{ cm}\]

\[(4 \times 15) \text{ cm} \times 1.125 = 4.5 \times 17 \text{ cm} = 7 \times 7 \text{ cm} \text{ equivalent square}\]

\[(5.5 \times 15) \text{ cm} \times 1.125 = 6.2 \times 17 \text{ cm} = 9 \times 9 \text{ cm} \text{ equivalent square}\]
10. A System of Dosimetric Calculations

Alternative

Since

\[
\frac{D_Q}{D_p} = \frac{\text{TAR}(10, 17 \times 17) - [\text{TAR}(10, 7 \times 7)](1 - T)}{\text{TAR}(10, 9 \times 9)}
\]

\[
= \frac{0.771 - 0.667(1 - 0.05)}{0.694}
\]

\[
= 0.20 \text{ or } 20\%
\]

Although the primary transmission through the lead block is only 5%, the dose at a 10-cm depth under the block in the middle is about 20% of the dose in the open portion. This increase in dose is a result of the internal scatter contributed by the open areas of the field to point Q. Of course, the dose under the block depends on the extent of the blocked area, overall field size, block thickness, depth, and location of point Q.

APPENDIX TO CHAPTER

A. Derivation of \( S_p \)

\( S_p(r) \), as defined in section 10.1B, is the ratio of dose rate for the given field (\( r \)) at a reference depth to the dose rate at the same point for the reference field size (\( r \)), with the same collimator opening. This is illustrated in Fig. 10.12. The given field in Fig. 10.12A is blocked down to the size of the reference field in Fig. 10.12B without changing the collimator opening. Thus both arrangements have the same collimator scatter factor, \( S_p(r) \).

\[
\frac{D_Q}{D_p} = \frac{0.733 \times 1.02 - 0.651 \times 0.989(1 - 0.05)}{0.672 \times 0.997}
\]

\[
= 0.20 \text{ or } 20\%
\]

FIG. 10.12. Diagrams to illustrate definition of \( S_p \). A: Dose in phantom at reference depth for a given field. B: Dose at the same point for a reference field with the same collimator opening. (From Khan FM, Sewchand W, Lee J, et al. Revision of tissue-maximum ratio and scatter-maximum ratio concepts for cobalt 60 and higher energy x-ray beams. Med Phys 1980;7:230, with permission.)
but different phantom scatter. Let $D_f$ and $D_{\text{max}}$ be the free space dose rate and $D_{\text{max}}$ dose rate, respectively. Then, at the reference depth of maximum dose,

\[
S_p(r) = \frac{D_{\text{max}} \text{ in arrangement A}}{D_{\text{max}} \text{ in arrangement B}} = \frac{D_f(r_0) \cdot S_c(r) \cdot \text{BSF}(r)}{D_f(r_0) \cdot S_c(r) \cdot \text{BSF}(r)} = \frac{\text{BSF}(r)}{\text{BSF}(r_0)}
\]

which is the same as Equation 10.1.

Equation A1 can also be written as:

\[
S_p(r) = \frac{D_f(r) \cdot \text{BSF}(r)}{D_f(r_0) \cdot \text{BSF}(r_0) \cdot S_c(r)} = \frac{D_{\text{max}}(r)}{D_{\text{max}}(r_0) \cdot S_c(r)} = S_{p, r}(r)
\]

where $S_{p, r}(r)$ is the total scatter correction factor defined as the ratio of $D_{\text{max}}$ dose rate for a given field to the $D_{\text{max}}$ dose rate for the reference field (Fig. 10.1 B).

### B. Derivation of TMR

In Fig. 10.2, let $D_1$ and $D_2$ be the doses at depths $d$ and $d_0$ (reference depth of maximum dose), respectively. Let $r$, $r_0$, and $r_d$ be the field sizes at distances $f$, $f + f_0$, and $f + d$ from the source, respectively. Then, by definition:

\[
\text{TMR}(d, r_d) = \frac{D_1}{D_2}
\]

and

\[
D_1 \frac{D_2}{D_2} = \frac{P(d, r, f)}{100}
\]

where $D(d_0, r_0, f)$ is the dose at depth $d_0$, field size $r_0$, and SSD = $f$.

\[
\frac{D_2}{D_2} = \frac{S_p(r_d)}{S_p(r_0)} \left( \frac{f + d}{f + d_0} \right)^2
\]

Combining Equations A4, A5, and A6,

\[
\text{TMR}(d, r_d) = \frac{P(d, r, f)}{100} \left( \frac{f + d}{f + d_0} \right)^2 \frac{S_p(r_0)}{S_p(r_d)}
\]

### C. Derivation of SMR

Referring to Fig. 10.2, let $D_1(d, r_d)$ be the dose at point 1 and $D_2(d_0, r_d)$ be the dose at point 2 for field size $r_d$. Let $D_1(d, 0)$ and $D_2(d_0, 0)$ be the corresponding doses for $0 \times 0$ field with the same collimator opening. Then:

\[
\text{SMR}(d, r_d) = \frac{D_1(d, r_d) - D_1(d, 0)}{D_2(d_0, r_d)}
\]
or
\[
\text{SMR}(d, r_d) = \frac{D_1(d, r_d)}{D_2(t_0, r_d)} \times \frac{D_2(t_0, r_0)}{D_2(t_0, r_0)} = \frac{D_1(d, 0)}{D_2(t_0, 0)}
\] (A9)

where \( r_0 \) is the reference field \((10 \times 10 \text{ cm})\) for normalizing \( S_p \), since:
\[
\text{TMR}(d, r_d) = \frac{D_1(d, r_d)}{D_2(t_0, r_d)} \quad \text{and} \quad \frac{D_1(d, 0)}{D_2(t_0, 0)}
\]
\[
\text{TMR}(d, 0) = \frac{D_1(d, 0)}{D_1(t_0, 0)} \quad \text{and} \quad \frac{D_1(d, 0)}{D_2(t_0, r_0)} \quad \text{(same collimator opening)}
\]

and
\[
S_p(r_d) = \frac{D_2(t_0, r_0)}{D_2(t_0, r_0)} \quad \text{(same collimator opening)}
\]

Equation A9 becomes:
\[
\text{SMR}(d, r_d) = \text{TMR}(d, r_d) \cdot \frac{S_p(r_d)}{S_p(0)} - \text{TMR}(d, 0) \quad \text{(A10)}
\]

REFERENCES

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The central axis depth dose distribution by itself is not sufficient to characterize a radiation beam that produces a dose distribution in a three-dimensional volume. In order to represent volumetric or planar variation in absorbed dose, distributions are depicted by means of *isodose curves*, which are lines passing through points of equal dose. The curves are usually drawn at regular intervals of absorbed dose and expressed as a percentage of the dose at a reference point. Thus the isodose curves represent levels of absorbed dose in the same manner that isotherms are used for heat and isobars, for pressure.

**11.1. ISODOSE CHART**

An *isodose chart* for a given beam consists of a family of isodose curves usually drawn at equal increments of percent depth dose, representing the variation in dose as a function of depth and transverse distance from the central axis. The depth-dose values of the curves are normalized either at the point of maximum dose on the central axis or at a fixed distance along the central axis in the irradiated medium. The charts in the first category are applicable when the patient is treated at a constant source-to-surface distance (SSD) irrespective of beam direction. In the second category, the isodose curves are normalized at a certain depth beyond the depth of maximum dose, corresponding to the axis of rotation of an isocentric therapy unit. This type of representation is especially useful in rotation therapy but can also be used for stationary isocentric treatments. Figure 11.1 shows both types of isodose charts for a 60Co γ-ray beam.

Examination of isodose charts reveals some general properties of x- and y-ray dose distributions.

1. The dose at any depth is greatest on the central axis of the beam and gradually decreases toward the edges of the beam, with the exception of some Linac x-ray beams, which exhibit areas of high dose or “horns” near the surface in the periphery of the field. These horns are created by the flattening filter, which is usually designed to overcompensate near the surface in order to obtain flat isodose curves at greater depths.

2. Near the edges of the beam (the penumbra region), the dose rate decreases rapidly as a function of lateral distance from the beam axis. As discussed in Chapter 4, the width of geometric penumbra, which exists both inside and outside the geometrical boundaries of the beam, depends on source size, distance from the source, and source-to-diaphragm distance.

3. Near the beam edge, falloff of the beam is caused not only by the geometric penumbra but also by the reduced side scatter. Therefore, the geometric penumbra is not the best measure of beam sharpness near the edges. Instead the term *physical penumbra* may be used. The *physical penumbra* width is defined as the lateral distance between two specified isodose curves at a specified depth (e.g., lateral distance between 90% and 20% isodose lines at the depth of $D_{max}$).
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4. Outside the geometric limits of the beam and the penumbra, the dose variation is the result of side scatter from the field and both leakage and scatter from the collimator system. Beyond this collimator zone, the dose distribution is governed by the lateral scatter from the medium and leakage from the head of the machine (often called therapeutic housing or source housing).

Figure 11.2 shows the dose variation across the field at a specified depth. Such a representation of the beam is known as the beam profile. It may be noted that the field

---

**FIG. 11.1.** Example of an isodose chart. A: SSD type, $^{60}$Co beam, SSD = 80 cm, field size = $10 \times 10$ cm at surface. B: SAD type, $^{60}$Co beam, SAD = 100 cm, depth of isocenter = 10 cm, field size at isocenter = $10 \times 10$ cm. (Data from University of Minnesota Hospitals, Eldorado $8$ Cobalt Unit, source size = 2 cm.)

**FIG. 11.2.** Depth dose profile showing variation of dose across the field. $^{60}$Co beam, SSD = 80 cm, depth = 10 cm, field size at surface = $10 \times 10$ cm. Dotted line indicates geometric field boundary at a 10-cm depth.
size is defined as the lateral distance between the 50% isodose lines at a reference depth. This definition is practically achieved by a procedure called the beam alignment in which the field-defining light is made to coincide with the 50% isodose lines of the radiation beam projected on a plane perpendicular to the beam axis and at the standard SSD or source-to-axis distance (SAD).

Another way of depicting the dose variation across the field is to plot isodose curves in a plane perpendicular to the central axis of the beam (Fig. 11.3). Such a representation is useful for treatment planning in which the field sizes are determined on the basis of an isodose curve (e.g., 90%) that adequately covers the target volume.

11.2. MEASUREMENT OF ISODOSE CURVES

Isodose charts can be measured by means of ion chambers, solid state detectors, or radiographic films (Chapter 8). Of these, the ion chamber is the most reliable method, mainly because of its relatively flat energy response and precision. Although any of the phantoms described in Chapter 9 may be used for isodose measurements, water is the medium of choice for ionometric measurements. The chamber can be made waterproof by a thin plastic sleeve that covers the chamber as well as the portion of the cable immersed in the water.

As measurement of isodose charts has been discussed in some detail in the International Commission on Radiation Units and Measurements (ICRU) (1), only a few important points will be discussed here. The ionization chamber used for isodose measurements should be small so that measurements can be made in regions of high dose gradient, such as near the edges of the beam. It is recommended that the sensitive volume of the chamber be less than 15 mm long and have an inside diameter of 5 mm or less. Energy independence of the chamber is another important requirement. Because the X-ray beam spectrum changes with position in the phantom owing to scatter, the energy response of the chamber should be as flat as possible. This can be checked by obtaining the exposure calibration of the chamber for orthovoltage (1 to 4 mm Cu) and $^{60}$Co beams. A variation of approximately 5% in response throughout this energy range is acceptable.
Automatic devices for measuring isodose curves have been developed for rapid mapping of the isodose curves. These systems are designed to be either stand alone or computer driven. Basically, the apparatus (Fig. 11.4) consists of two ionization chambers, referred to as the detector A (or probe) and the monitor B. Whereas the probe is arranged to move in the tank of water to sample the dose rate at various points, the monitor is fixed at some point in the field to monitor the beam intensity with time. The ratio of the detector to the monitor response \( A/B \) is recorded as the probe is moved in the phantom. Thus the final response \( A/B \) is independent of fluctuations in output. In the stand-alone system, the probe searches for points at which \( A/B \) is equal to a preset percentage value of \( A/B \) measured at a reference depth or the depth of maximum dose. The motion of the probe is transmitted to the plotter which records its path, the isodose curve.

In the computer-driven models, the chamber movement of the probe is controlled by a computer program. The probe-to-monitor ratio is sampled as the probe moves across the field at preset increments. These beam profiles are measured at a number of depths, determined by computer program. The data thus measured are stored in the computer in the form of a matrix that can then be transformed into isodose curves or other formats allowed by the computer program.

### A. Sources of Isodose Charts

Acquisition of isodose charts has been discussed (1). Atlases of premeasured isodose charts for a wide range of radiation therapy equipment are available from the sources listed in the literature (2-4). In addition, isodose distributions may also be obtained from manufacturers of radiation generators or from other institutions having the same unit. However, the user is cautioned against accepting isodose charts from any source and using them as basis for patient treatment without adequate verification. The first and most important check to be performed is to verify that the central axis depth-dose data correspond with percent depth-dose data measured independently in a water phantom. A deviation of 2% or less in local dose is acceptable up to depths of 20 cm. The edges of the distribution should be checked by measuring beam profiles for selected field sizes and depths. An agreement within 2 mm in the penumbra region is acceptable.

Besides direct measurements, isodose charts can also be generated by calculations using various algorithms for treatment planning (5-9). More current algorithms are discussed in...
Part III of this book. Some of these programs are commercially available with treatment planning computers. Again, the applicability of the computer-generated isodose curves to the user's machine must be carefully checked.

11.3. PARAMETERS OF ISODOSE CURVES

Among the parameters that affect the single-beam isodose distribution are beam quality, source size, beam collimation, field size, SSD, and the source-to-diaphragm distance (SDD). A discussion of these parameters will be presented in the context of treatment planning.

A. Beam Quality

As discussed previously, the central axis depth dose distribution depends on the beam energy. As a result, the depth of a given isodose curve increases with beam quality. Beam energy also influences isodose curve shape near the field borders. Greater lateral scatter associated with lower-energy beams causes the isodose curves outside the field to bulge out. In other words, the absorbed dose in the medium outside the primary beam is greater for low-energy beams than for those of higher energy.

Physical penumbra depends on beam quality as illustrated in Fig. 11.5. As expected, the isodose curves outside the primary beam (e.g., 10% and 5%) are greatly distended in the case of orthovoltage radiation. Thus one disadvantage of the orthovoltage beams is the increased scattered dose to tissue outside the treatment region. For megavoltage beams, on the other hand, the scatter outside the field is minimized as a result of predominantly forward scattering and becomes more a function of collimation than energy.

B. Source Size, Source-to-Surface Distance, and Source-to-Diaphragm Distance—The Penumbra Effect

Source size, SSD, and SDD affect the shape of isodose curves by virtue of the geometric penumbra, discussed in Chapter 4. In addition, the SSD affects the percent depth dose and therefore the depth of the isodose curves.

As discussed previously, the dose variation across the field border is a complex function of geometric penumbra, lateral scatter, and collimation. Therefore, the field sharpness at depth is not simply determined by the source or focal spot size. For example, by using penumbra trimmers or secondary blocking, the isodose sharpness at depth for $^{60}$Co beams with a source size less than 2 cm in diameter can be made comparable with higher-energy Linac beams, although the focal spot size of these beams is usually less than 2 mm. Comparison of isodose curves for $^{60}$Co, 4 and 10 MV in Fig. 11.5 illustrates the point that the physical penumbra width for these beams is more or less similar.

C. Collimation and Flattening Filter

The term *collimation* is used here to designate not only the collimator blocks that give shape and size to the beam but also the flattening filter and other absorbers or scatterers in the beam between the target and the patient. Of these, the flattening filter, which is used for megavoltage x-ray beams, has the greatest influence in determining the shape of the isodose curves. Without this filter, the isodose curves will be conical in shape, showing markedly increased x-ray intensity along the central axis and a rapid reduction transversely. The function of the flattening filter is to make the beam intensity distribution relatively uniform across the field (i.e., "flat"). Therefore, the filter is thickest in the middle and tapers off toward the edges.

The cross-sectional variation of the filter thickness also causes variation in the photon spectrum or beam quality across the field owing to selective hardening of the beam by the filter. In general, the average energy of the beam is somewhat lower for the peripheral areas
compared with the central part of the beam. This change in quality across the beam causes the flatness to change with depth. However, the change in flatness with depth is caused by not only the selective hardening of the beam across the field but also the changes in the distribution of radiation scatter as the depth increases.

Beam flatness is usually specified at a 10-cm depth with the maximum limits set at the depth of maximum dose. By careful design of the filter and accurate placement in the beam, it is possible to achieve flatness to within ±3% of the central axis dose value at a 10-cm depth. This degree of flatness should extend over the central area bounded by at least 80% of the field dimensions at the specified depth or 1 cm from the edge of the field. The above specification is satisfactory for the precision required in radiation therapy.

To obtain acceptable flatness at 10-cm depth, an area of high dose near the surface may have to be accepted. Although the extent of the high dose regions, or horns, varies with the design of the filter, lower-energy beams exhibit a larger variation than higher-energy beams. In practice, it is acceptable to have these “superflat” isodose curves near the surface.
D. Field Size

Field size is one of the most important parameters in treatment planning. Adequate dosimetric coverage of the tumor requires a determination of appropriate field size. This determination must always be made dosimetrically rather than geometrically. In other words, a certain isodose curve (e.g., 90%) enclosing the treatment volume should be the guide in choosing a field size rather than the geometric dimensions of the field.

Great caution should also be exercised in using field sizes smaller than 6 cm in which a relatively large part of the field is in the penumbra region. Depending on the source size, collimation, and design of the flattening filter, the isodose curves for small field sizes, in general, tend to be bell-shaped. Thus treatment planning with isodose curves should be mandatory for small field sizes.

The isodose curvature for $^{60}$Co increases as the field size becomes overly large unless the beam is flattened by a flattening filter. The reason for this effect is the progressive reduction of scattered radiation with increasing distance from the central axis as well as the obliquity of the primary rays. The effect becomes particularly severe with elongated fields such as cranial spinal fields used in the treatment of medulloblastoma. In these cases, one needs to calculate doses at several off-axis points or use a beam-flattening compensator.

11.4. WEDGE FILTERS

Frequently, special filters or absorbing blocks are placed in the path of a beam to modify its isodose distribution. The most commonly used beam-modifying device is the wedge filter. This is a wedge-shaped absorber that causes a progressive decrease in the intensity across the beam, resulting in a tilt of the isodose curves from their normal positions. As shown in Fig. 11.6, the isodose curves are tilted toward the thin end, the degree of tilt depends

![Figure 11.6: Isodose curves for a wedge filter. A: normalized to $D_{max}$. B: normalized to $D_{max}$ without the wedge. $^{60}$Co, wedge angle = 45 degrees, field size = 8 x 10 cm, SSD = 80 cm.](image)
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FIG. 11.7. Photograph of a 45 degree wedge filter for a 4-MV x-ray Linac (ATC 400).

on the slope of the wedge filter. In actual wedge filter design, the sloping surface is made either straight or sigmoid in shape; the latter design is used to produce straighter isodose curves.

The wedge is usually made of a dense material, such as lead or steel, and is mounted on a transparent plastic tray which can be inserted in the beam at a specified distance from the source (Fig. 11.7). This distance is arranged such that the wedge tray is always at a distance of at least 15 cm from the skin surface, so as to avoid destroying the skin-sparing effect of the megavoltage beam.

Another class of wedges (not discussed here) are the dynamic wedges. These wedges are generated electronically by creating wedged beam profiles through dynamic motion of an independent jaw within the treatment beam. Dynamic wedges do not offer significant clinical advantages over the traditional metal wedges. Moreover, all wedges and compensators are now superseded by the new technology using dynamic multileaf collimators in conjunction with the intensity-modulated radiation therapy (IMRT).

A. Wedge Isodose Angle

The term wedge isodose angle (or simply wedge angle) refers to "the angle through which an isodose curve is titled at the central ray of a beam at a specified depth" (11). In this definition, one should note that the wedge angle is the angle between the isodose curve and the normal to the central axis, as shown in Fig. 11.6. In addition, the specification of depth is important since, in general, the presence of scattered radiation causes the angle of isodose tilt to decrease with increasing depth in the phantom. However, there is no general agreement as to the choice of reference depth. Some choose depth as a function of field size (e.g., 1/2 or 2/3 of the beam width) while others define wedge angle as the angle between the 50% isodose curve and the normal to the central axis. The latter choice, however, becomes impractical when higher-energy beams are used. For example, the central axis depth of the 50% isodose curve for a 10-MV beam lies at about 18 cm for a 10 x 10-cm field and 100-cm SSD. This depth is too large in the context of most wedge filter applications. As will be discussed in section 11.7, the wedge filters are mostly used for treating superficial tumors, for example, not more than 10 cm deep. Therefore, the current recommendation is to use a single reference depth of 10 cm for wedge angle specification (11).
B. Wedge Transmission Factor

The presence of a wedge filter decreases the output of the machine, which must be taken into account in treatment calculations. This effect is characterized by the wedge transmission factor (or simply wedge factor), defined as the ratio of doses with and without the wedge, at a point in phantom along the central axis of the beam. This factor should be measured in phantom at a suitable depth beyond the depth of maximum dose (e.g., 10 cm).

In cobalt-60 teletherapy, the wedge factor is sometimes incorporated into the isodose curves, as shown in Fig. 11.6B. In this case, the depth dose distribution is normalized relative to the $D_{\text{max}}$ without the wedge. For example, the isodose curve at depth of $D_{\text{max}}$ is 72%, indicating that the wedge factor is already taken into account in the isodose distribution. If such a chart is used for isodose planning, no further correction should be applied to the output. In other words, the machine output corresponding to the open beam should be used.

A more common approach is to normalize the isodose curves relative to the central axis $D_{\text{max}}$ with the wedge in the beam. As see in Fig. 11.6A, the 100% dose is indicated at the depth of $D_{\text{max}}$. With this approach, the output of the beam must be corrected using the wedge factor.

C. Wedge Systems

Wedge filters are of two main types. The first may be called the individualized wedge system, which requires a separate wedge for each beam width, optimally designed to minimize the loss of beam output. A mechanism is provided to align the thin end of the wedge with the border of the light field (Fig. 11.8A). The second system uses a universal wedge, that is, a single wedge serves for all beam widths. Such a filter is fixed centrally in the beam while the field can be opened to any size. As illustrated in Fig. 11.8B, only a small part of this wedge, i.e., $ABC$, is effective in producing the given wedge angle. The rest ($ACDE$), being unwedged, does not contribute to the isodose tilt but unnecessarily reduces the beam intensity. Since the individualized system economizes on the beam output, it is preferred for use in cobalt teletherapy. The universal wedge, on the other hand, is useful for linear accelerator beams where the output is plentiful. From the set-up and treatment planning points of view, the universal wedge is simpler to use than the individualized filter.

D. Effect on Beam Quality

In general, the wedge filter alters the beam quality by preferentially attenuating the lower-energy photons (beam hardening) and, to a lesser extent, by Compton scattering, which results in energy degradation (beam softening). For the $^{60}$Co beam, because the primary beam is essentially monoenergetic, the presence of the wedge filter does not significantly alter the central axis percent depth dose distribution. For x-rays, on the other hand, there can be some beam hardening (12), and consequently, the depth dose distribution can be somewhat altered, especially at large depths.

Although the wedge filters produce some change in beam quality, as noted above, the effect is not large enough to alter other calculation parameters such as the backscatter factor or the equivalent square, which may be assumed to be the same as for the corresponding open beams. Even central axis percent depth doses, tissue-air ratios or tissue maximal ratios may be assumed unchanged for small depths (e.g., less than 10 cm). The error caused by this assumption is minimized if the wedge transmission factor has been measured at a reference depth close to the point of interest.

E. Design of Wedge Filters

The design of wedge filters for megavoltage beams has been described by many authors (13–16). Here I will briefly present the design of a universal wedge filter following the
technique of Aron and Scapicchio (16). The principle of this method is to determine the ratio of percent depth doses at various points for wedged and nonwedged fields. The thickness of the wedge filter material at these points is then determined from these ratios and the knowledge of the half-value layer or the attenuation coefficient of the given beam for the filter material.

Figure 11.9 illustrates the design of a wedge filter. A line is drawn at a selected depth across the nonwedged field at right angles to the central axis. This depth should correspond to the reference depth used for the wedge angle definition. Fan lines, representing rays from the source, are drawn at fixed intervals (e.g., 1 cm) on both sides of the central axis. A series of parallel lines is drawn making an angle with the central axis equal to the complement of the given wedge angle and intersecting the central axis at the same points of intersection as the nonwedged isodose lines. A table is constructed that includes the percentage depth doses at the points of intersection of the fan lines and the reference depth line for the nonwedged isodose curves and the wedged isodose lines (sloping lines). The ratio of the wedged to nonwedged values is calculated as shown in Table 11.1. These ratios are normalized to the highest value within the field (excluding the penumbra region) to give the relative transmission ratio along the designated fan lines. A wedge filter of a given material can then be designed to provide these transmission ratios.
11.5. COMBINATION OF RADIATION FIELDS

Treatment by a single photon beam is seldom used except in some cases in which the tumor is superficial. The following criteria of acceptability may be used for a single field treatment: (a) the dose distribution within the tumor volume is reasonably uniform (e.g., within ±5%); (b) the maximum dose to the tissues in the beam is not excessive (e.g., not more than 110% of the prescribed dose); and (c) normal critical structures in the beam do not receive doses near or beyond tolerance. Whereas single fields of superficial x-rays are routinely used for treating skin cancers which are confined to a depth of a few millimeters, single megavoltage beams are used only in rare cases for which a combination of beams is either technically difficult or results in unnecessary or excessive irradiation of the normal tissues. Examples of a few treatments that use single megavoltage beams include the supraclavicular region, internal mammary nodes (anterior field), and the spinal

| TABLE 11.1. TRANSMISSION RATIOS FOR THE CONSTRUCTION OF WEDGE FILTER |
|--------------------|---|---|---|---|---|---|---|---|---|
|                    | A | B | C | D | E | F | G | H | I |
| Nonwedge isodose   | 40| 55| 62| 65| 67| 68| 68| 67| 65 |
|                    |   |   |   |   |   |   |   |   |   |
| Wedge isodose      | 35| 39| 41| 47| 53| 60| 68| 76| 86 |
| Ratio (wedge/nonwedge) | 0.875| 0.710| 0.660| 0.720| 0.790| 0.880| 1.00| 1.12| 1.28 |
| Transmission ratio |   |   |   |   |   |   |   |   |   |
| mm Pb              |   |   |   |   |   |   |   |   |   |

From Aron BS, Scapicchio M. Design of universal wedge filter system for a cobalt 60 unit. AJR 1966;96:70, with permission.
cord (posterior field). Although the dose distribution is not ideal, the single-field technique in these cases results in simplicity of set-up without violating the above criteria of acceptability.

For treatment of most tumors, however, a combination of two or more beams is required for an acceptable distribution of dose within the tumor and the surrounding normal tissues. Although radiation fields may be combined in many ways, the discussion here will be confined to the basic principles that are useful in treating tumors involving different sites.

**A. Parallel Opposed Fields**

The simplest combination of two fields is a pair of fields directed along the same axis from opposite sides of the treatment volume. The advantages of the parallel opposed fields are the simplicity and reproducibility of set-up, homogeneous dose to the tumor, and less chances of geometrical miss (compared with angled beams), given that the field size is large enough to provide adequate lateral coverage of the tumor volume. A disadvantage is the excessive dose to normal tissues and critical organs above and below the tumor.

A composite isodose distribution for a pair of parallel opposed fields may be obtained by adding the depth dose contribution of each field (Fig. 11.10). The manual procedure consists of joining the points of intersection of isodose curves for the individual fields which sum to the same total dose value. The resultant distribution shows the combined isodose

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**FIG. 11.10.** Composite isodose distribution for a pair of parallel opposed fields. A: Each beam is given a weight of 100 at the depth of $D_{max}$. B: Isocentric plan with each beam weighted 100 at the isocenter.
distribution normalized to the individual beam weights. The beams are usually weighted in dose units of 100 at the depth of $D_{\text{max}}$ in the case of SSD techniques or at the isocenter for the isocentric techniques. For the example shown in Fig. 11.10A, the minimum percent isodose surrounding the tumor is 110. This means that the minimum dose to the tumor (with a generous margin) is 110 rads if 100 rads are delivered at the depth of $D_{\text{max}}$ by each field. Thus, if the tumor dose were to be specified at this isodose level, one could calculate the $D_{\text{max}}$ dose and the treatment time for each field. For the isocentric plan shown in Fig. 11.10B, the beam weights refer to doses delivered to the isocenter. Thus the 100% isodose curve represents the specified minimum dosage level if each beam delivered 100 rads to its isocenter. Once the isocenter dose is calculated, one can determine the treatment time or monitor units as described in section 10.2.

A.1. Patient Thickness Versus Dose Uniformity

One advantage of equally weighted parallel opposed beams is that the dose distribution within the irradiated volume can be made uniform. However, the uniformity of distribution depends on the patient thickness, beam energy, and beam flatness. In general, as the patient thickness increases or the beam energy decreases, the central axis maximum dose near the surface increases relative to the midpoint dose. This effect, called tissue lateral effect, is shown in Fig. 11.11 in which two opposing beams are placed 25 cm apart with the midpoint dose normalized to 100. The curves for cobalt-60 and 4 MV show that for a patient of this thickness parallel opposed beams would give rise to an excessively higher dose to the subcutaneous tissues compared with the tumor dose at the midpoint. As the energy is increased to 10 MV, the distribution becomes almost uniform and at 25 MV it shows significant sparing of the superficial tissues relative to the midline structures.

The ratio of maximum peripheral dose to midpoint dose is plotted in Fig. 11.12 as a function of patient thickness for a number of beam energies. Such data are useful in choosing the appropriate beam energy for a given patient thickness when using parallel opposed fields. For example, acceptable uniformity of dose, that is, within ±5%, is achievable with cobalt-60 or 4- to 6-MV beams for thicknesses of about 15 cm or less (e.g., head, neck, and extremities). However, for thicknesses of 20 cm or greater (e.g., thorax, abdomen, and pelvis), 10-MV or higher energies must be used to spare the normal subcutaneous tissues.

![Fig. 11.11. Depth dose curves for parallel opposed field normalized to midpoint value. Patient thickness = 25 cm, field size = 10 x 10 cm, SSD = 100 cm.](image)
II. Classical Radiation Therapy

A.2. Edge Effect (Lateral Tissue Damage)

When treating with multiple beams, the question arises whether one should treat one field per day or all fields per day. Wilson and Hall (17) have discussed this problem in terms of cell survival curves and Ellis's time-dose-fractionation formula (18,19). For parallel opposed beams, they have shown that treating with one field per day produces greater biologic damage to normal subcutaneous tissue than treating with two fields per day, despite the fact that the total dose is the same. Apparently, the biologic effect in the normal tissue is greater if it receives alternating high- and low-dose fractions compared with the equal but medium-size dose fractions resulting from treating both fields daily. This phenomenon has been called the edge effect, or the tissue lateral damage (20). The problem becomes more severe when larger thicknesses (e.g., ≥20 cm) are treated with one field per day using a lower-energy beam (e.g., ≤6 MV). In such cases, the dose per fraction to the subcutaneous tissues, although delivered on alternate days, becomes prohibitively high.

A.3. Integral Dose

One way of comparing dose distributions for different quality beams is to calculate the integral dose for a given tumor dose. Integral dose is a measure of the total energy absorbed in the treated volume. If a mass of tissue receives a uniform dose, then the integral dose is simply the product of mass and dose. However, in practice, the absorbed dose in the tissue is nonuniform so rather complex mathematical formulas are required to calculate it.

For a single beam of x- or γ radiation, Mayneord (21) formulated the following expression:

\[ \sum = 1.44D_0\,A\,d_{1/2}(1 - e^{-0.693d/d_{1/2}}) \left( 1 + \frac{2.88d_{1/2}}{SSD} \right) \]  

(11.1)

where \( \sum \) is the integral dose, \( D_0 \) is the peak dose along the central axis, \( A \) is the geometric area of the field, \( d \) is the total thickness of patient in the path of the beam, \( d_{1/2} \) is the half-value depth or the depth of 50% depth dose and SSD is the source-surface distance. The term \( 1 + \frac{2.88d_{1/2}}{SSD} \) is a correction for geometric divergence of the beam.
Treatment Planning

I. Isodose Distribution

Beam Energy [MV]

Integral Dose [kg rad]

FIG. 11.13. Integral dose as a function of photon beam energy, when 1,000 rad are delivered at a midpoint of a 25-cm-thick patient. Field size, 10-cm diameter at an SSD of 100 cm. (Redrawn from Podgorsak EB, Rawlinson JA, Johns HE. X-ray depth doses for linear accelerators in the energy range from 10 to 32 MeV. Am J Roentgenol 1975;123:182.)

Because integral dose is basically the product of mass and dose, its unit is the gram-rad or kilogram-gray or simply joule (since 1 Gy = 1 J/kg). Figure 11.13 shows the integral dose as a function of the energy of radiation for a tumor dose of 1,000 rad (1 rad = 10^{-2} Gy) at a depth of 12.5 cm in the patient of 25-cm thickness treated with parallel opposed beams (22). The curve shows a useful result, namely, the higher the photon energy the lower the integral dose.

Although it is generally believed that the probability of damage to normal tissue increases with the increase in the integral dose, this quantity is seldom used clinically to plan dosages or predict treatment outcome. However, it does provide qualitative guidelines for treatment planning for selecting beam energy, field sizes, and multiplicity of fields. As a general rule, one should keep the integral dose to a minimum, provided the adequacy of tumor irradiation and the sparing of critical organs are not compromised.

B. Multiple Fields

One of the most important objectives of treatment planing is to deliver maximum dose to the tumor and minimum dose to the surrounding tissues. In addition, dose uniformity within the tumor volume and sparing of critical organs are important considerations in judging a plan. Some of the strategies useful in achieving these goals are (a) using fields of appropriate size; (b) increasing the number of fields or portals; (c) selecting appropriate beam directions; (d) adjusting beam weights (dose contribution from individual fields); (e) using appropriate beam energy; and (f) using beam modifiers such as wedge filters and compensators. Although obtaining a combination of these parameters that yields an optimal plan is time-consuming if done manually, treatment-planning computers are now available that can do the job quickly and accurately. Some of these systems are highly interactive so that the user can almost instantly modify, calculate, and examine various plans to select one that is clinically superior.

In section 11.5A, I discussed the case of two parallel opposed fields. Although the technique results in uniform irradiation of the tumor, there is little sparing of the surrounding normal tissue. In fact, the dose to the peripheral tissues can be significantly higher than the midline dose. Reduction of dose to subcutaneous tissue and normal tissue surrounding the tumor can be achieved by using a combination of three or more fields. Figure 11.14 illustrates various multiple-field arrangements in which the beam enters the patient from various directions, always directed at the tumor. Thus, by using multiple fields, the ratio of the tumor dose to the normal tissue dose is increased. Figure 11.15A,B shows typical examples of multiple fields, one used for treatment of the esophagus and the other, for the prostate gland. Figure 11.15C illustrates a fixed SSD-type technique in which the beam...
weights are delivered to $D_{\text{max}}$ points. In actual practice, one may use a combination of parallel opposed fields and multiple fields to achieve the desired dose distribution.

Although multiple fields can provide good distribution, there are some clinical and technical limitations to these methods. For example, certain beam angles are prohibited because of the presence of critical organs in those directions. Also, the set-up accuracy of a treatment may be better with parallel opposed than with the multiple angled beam arrangement. It is, therefore, important to realize that the acceptability of a treatment plan depends not only on the dose distribution on paper but also on the practical feasibility, set-up accuracy, and reproducibility of the treatment technique.

11.6. ISOCENTRIC TECHNIQUES

Most modern machines are constructed so that the source of radiation can rotate about a horizontal axis. The gantry of the machine is capable of rotating through 360 degrees with the collimator axis moving in a vertical plane. The isocenter is the point of intersection of the collimator axis and the gantry axis of rotation.

A. Stationary Beams

The isocentric technique of irradiation consists of placing the isocenter of the machine at a depth within the patient and directing the beams from different directions. The distance
FIG. 11.15. Examples of multiple field plans. A: Three-field isocentric technique. Each beam delivers 100 units of dose at the isocenter; 4 MV, field size = 8 x 8 cm at isocenter, SAD = 100 cm.
B: Four-field isocentric technique. Each beam delivers 100 units of dose at the isocenter; 10 MV, field size = 8 x 8 cm at isocenter, SAD = 100 cm. C: Four-field SSD technique in which all beams are weighted 100 units at their respective points of Dmax; 10 MV, field size = 8 x 8 cm at surface, SSD = 100 cm.

of the source from the isocenter, or the SAD, remains constant irrespective of the beam direction. However, the SSD in this case may change, depending on the beam direction and the shape of the patient contour. For any beam direction, the following relationship holds:

$$SSD = SAD - d$$

(11.2)

where $d$ is the depth of the isocenter. Knowing the depth and position of isocenter from one direction such as the anterior posterior, the SSD can be calculated according to Equation...
11.2 and set up from that direction. Then the positioning of subsequent fields simply requires moving the gantry and not the patient.

Although all techniques for which SSD ≤ SAD can be carried out isocentrically, the major advantage of this method is the ease with which multiple field set-ups (three or more) can be treated when all fields are treated the same day. This technique not only dispenses with the setting up of SSD for each beam direction but relies primarily on the accuracy of machine isocentricity and not on the skin marks which are unreliable points of reference in most cases.

The treatment calculations for isocentric treatments have been presented in section 10.2A.2. Figure 11.15A,B shows examples of isodose distribution for isocentric techniques.

### B. Rotation Therapy

Rotation therapy is a special case of the isocentric technique in which the beam moves continuously about the patient, or the patient is rotated while the beam is held fixed. Although this technique has been used for treating tumors of the esophagus, bladder, prostate gland, cervix, and brain, the technique offers little advantage over the isocentric technique using multiple stationary beams. For example, the esophagus can be treated equally well with three fields; the prostate gland and bladder, with four fields (sometimes combined with parallel opposed fields); and the brain, with two or three fields or with wedges, depending on the size and location of the tumor. Many times it is a matter of individual preference, although one technique may offer particular advantages over the other in regard to patient positioning, blocking, and the size of volume to be irradiated. Especially when intricate blocking is required, rotation therapy should not be attempted.

Rotation therapy is best suited for small, deep-seated tumors. If the tumor is confined within a region extending not more than halfway from the center of the contour cross-section, rotation therapy may be a proper choice. However, rotation therapy is not indicated if (a) volume to be irradiated is too large, (b) the external surface differs markedly from a cylinder, and (c) the tumor is too far off center.

Calculation for rotation therapy can be made in the same way as for the stationary isocentric beams, except that a reasonably large number of beams should be positioned around the patient contour at fixed angular intervals. The dose rate at the isocenter is given by:

\[
D_{iso} = D_{ref} \times T
\]

where \(D_{ref}\) is the reference dose rate related to the quantity \(T\) which may be average tissue-to-air ratio (TAR) or tissue maximal ratio (TMR) (averaged over all depths at the selected angles). In the case of TARs, \(D_{ref}\) is the dose rate in free space for the given field at the isocenter. A method of manual calculations based on this system was discussed in section 9.4D. If the TMRs are used, \(D_{ref}\) is the \(D_{max}\) dose rate for the given field at the SAD. Using the TMR system discussed in Chapter 10,

\[
D_{iso} = D_0 \times S_c \times S_p \times TMR
\]

where \(D_0\) is the \(D_{max}\) dose rate for a 10 x 10-cm field at the SAD, and \(S_c\) and \(S_p\) are the collimator and phantom scatter correction factors for the given field size at the isocenter. In the case of a linear accelerator, \(D_0\) is the monitor unit (MU) rate (assuming 1 MU = 1 rad (cGy) at the isocenter for a depth of \(D_{max}\) for a 10 x 10-cm field).

**Example**

A patient is to receive 250 rad at the isocenter by rotation therapy, using 4-MV x-rays, 6 x 10-cm field at the isocenter, and a SAD of 100 cm. If TMR calculated according to the procedure in section 9.4D is 0.746, calculate the number of monitor units to be set on
the machine if the machine output is set at 200 MU/min and given $S_t \times G_t = 0.98$ and $S_p (G \times 10) = 0.99$. From Equation 11.4,

$$D_{iso} = D_0 \times S_t \times S_p \times TMR$$

or

$$D_{iso} = 200 \times 0.98 \times 0.99 \times 0.746$$

$$= 144.8 \text{ rad/min}$$

Treatment time = \frac{250 \text{ rad}}{144.8 \text{ rad/min}} = 1.73 \text{ min}

Total MU to be set = 200 (MU/min) \times 1.73 \text{ min}

= 345 \text{ MU}

Gantry rotation speed is set so that 345 MU are delivered at the conclusion of the rotation. Some machines perform only one rotation, whereas others can perform a specified number of arcs or rotations in a pendulum manner. Most modern machines allow for automatic adjustment of rotation speed to deliver a preset number of monitor units by the end of a single rotation.

The determination of complete isodose curves for rotation therapy by manual means is very time-consuming. It is essentially the same procedure as used in multiple fixed beams, but with a large number of beams. The isocentric isodose chart (Fig. 11.1B) in which isodoses are normalized to a point at depth on the central axis is used with the isocenter placed at the point of normalization. By summing the isodose values at selected points while the chart is placed at different angles, the dose distribution can be determined relative to the isocenter. Because of the tedium involved in the procedure, this task is ideally suited for computer application. Such programs are available with commercial treatment planning computers.

Figure 11.16 shows three examples of isodose distribution for rotation therapy: (a) 100 degree arc rotation; (b) 180 degree arc rotation; and (c) full 360 degree rotation. It should be noted that whereas the maximum dose for the 360-degree rotation occurs at the isocenter, for the partial arcs it is displaced toward the irradiated sector. This illustrates an important principle that in arc therapy or when oblique fields are directed through one side of a patient, they should be aimed a suitable distance beyond the tumor area. This is sometimes referred to as past pointing. The extent of past pointing required to bring the maximum dose to the tumor site depends on the arc angle and should be determined for an individual case by actual isodose planning.

11.7. WEDGE FIELD TECHNIQUES

Relatively superficial tumors, extending from the surface to a depth of several centimeters, can be irradiated by two "wedged" beams directed from the same side of the patient. Figure 11.17A shows isodose distribution of two angled beams with no wedge in the beams. It is seen that in the region of overlap of the beams, the dose distribution is quite nonuniform. The dose is highest in the superficial or region of overlap and falls off to lower values toward the deeper areas. By inserting appropriate wedge filters in the beam and positioning them with the thick ends adjacent to each other, the angled field distribution can be made fairly uniform (Fig. 11.17B). Each wedged beam in this case has a reduced dose in the superficial region relative to the deeper region so that the dose gradient in the overlap region is minimized. The dose falls off rapidly beyond the region of overlap or the "plateau" region, which is clinically a desirable feature.

There are three parameters that affect the plateau region in terms of its depth, shape, and dose distribution: $\theta$, $\phi$, and $S$, where $\theta$ is the wedge angle (section 11.4A), $\phi$ is the hinge angle, and $S$ is the separation. These parameters are illustrated in Fig. 11.18. The hinge angle
is the angle between the central axes of the two beams and the separation $S$ is the distance between the thick ends of the wedge filters as projected on the surface. Cohen and Martin (3) have discussed in detail how $\theta$, $\phi$, and $S$ can be adjusted to achieve a desired plateau.

There is an optimum relationship between the wedge angle $\theta$ and the hinge angle $\phi$ which provides the most uniform distribution of radiation dose in the plateau:

$$\theta = 90^\circ - \frac{\phi}{2} \quad (11.5)$$

This equation is based on the principle that for a given hinge angle the wedge angle should be such that the isodose curves from each field are parallel to the bisector of the hinge angle (Fig. 11.18). Under these conditions, when the isodoses are combined, the resultant distribution is uniform.

The Equation 11.5, although helpful in treatment planning, may not yield an optimum plan for a given patient contour. The relationship assumes that the wedge isodose curves are not modified by the surface contour. In practice, however, contours are usually curved.
or irregular in shape and thus modify the isodose distribution for the wedged beams. As a result, the isodose curves for the individual fields are no longer parallel to the bisector of the hinge angle, thus giving rise to a nonuniform distribution in the overlap region. This problem can be solved by using compensators (discussed in Chapter 12), which make the skin surface effectively flat and perpendicular to each beam. An alternative approach is to modify the wedge angle (using a different wedge angle filter from that given by Equation 11.5) so that a part of the wedge angle acts as a compensator and the rest as a true wedge filter. The main objective is to make the isodose curves parallel to the hinge angle bisector.

Although the latter approach obviates the need for a compensator, the determination of an
optimum wedge angle may not be easy if planning is done manually. The former method, on the other hand, is well-suited for manual calculations since all you need is a compensator and an atlas of precalculated isodose distributions for a variety of $\theta$, $\phi$, and $S$ values. This method, however, becomes technically difficult to implement if complicated secondary blocking is required in addition to the compensator and the wedge filter.

Equation 11.5 suggests that for each hinge angle one should use a different wedge angle. However, in practice, selected wedge angles, i.e., 15 degrees, 30 degrees, 45 degrees, and 60 degrees, are adequate over a wide range of hinge angles.

In modern radiation therapy, complex treatment techniques are frequently used, which may involve wedge filters, compensators, field blocking, and field reductions, all for the same patient. Manual treatment planning is difficult for such cases. For this reason, in many institutions, all complex treatments, including wedged fields, are planned by computer as a matter of standard practice.

A. Uniformity of Dose Distribution

Because wedge pair techniques are normally used for treating small, superficial tumor volumes, a high-dose region (hot spot) of up to +10% within the treatment volume is usually acceptable. These hot spots occur under the thin ends of the wedges and their magnitude increases with field size and wedge angle. This effect is related to the differential attenuation of the beam under the thick end relative to the thin end.

Generally, the wedge filter technique is suitable when the tumor is approximately from 0 to 7 cm deep and when it is necessary to irradiate from one side of the skin surface. The most desirable feature of this technique is the rapid dose falloff beyond the region of overlap. This falloff can be exploited to protect a critical organ such as the spinal cord. Although wedge filters are invaluable in radiotherapy, some of these techniques are being replaced by electron beam techniques (Chapter 14).

B. Open and Wedged Field Combinations

Although wedge filters were originally designed for use in conjunction with the wedge-pair arrangement, it is possible to combine open and wedged beams to obtain a particular dose distribution. One such arrangement which uses an open field anteriorly and wedged field laterally in the treatment of some tumors is shown in Fig. 11.19A. The anterior field is weighted to deliver 100 units to the lateral 15 units to the isocenter (these beams could be weighted in terms of $D_{\text{max}}$ in the SSD technique). The weights and wedge angle are usually adjusted for an individual case to obtain an acceptable distribution. The principle of this technique is that as the dose contribution from the anterior field decreases with depth, the lateral beam provides a boost to offset this decrease. As seen in Fig. 11.19A, a wedged beam with the thick end positioned superiorly provides the desired compensation for the dose dropoff. Thus such a combination of open and wedged beams gives rise to a distribution which remains constant with depth within certain limits.

Figure 11.19B shows another technique in which the anterior open beam is combined with the two lateral wedged beams. Again, the beam weights and wedge angles are chosen to make the open beam distribution remain constant throughout the tumor volume.

11.8. TUMOR DOSE SPECIFICATION FOR EXTERNAL PHOTON BEAMS

The results of treatments can be meaningfully interpreted only if sufficient information is provided regarding the irradiation technique and the distribution of dose in space and time. In the absence of this information, recording of only the so-called tumor dose serves little purpose. Unfortunately, this important problem is often ignored. More often than not, treatment summaries and records are ambiguous and even incomprehensible
FIG. 11.19. Treatment plans using open and wedged field combinations. A: Isocentric plan with anterior open field weighted 100 and lateral wedged field weighted 15 at the isocenter. B: A combination of anterior open beam and two lateral wedged beams; 4 MV x-ray beam from ATC-400 Linac.
to other people. Therefore, one cannot overemphasize the need for a dose recording system which is sufficiently explicit and detailed to enable other centers to reproduce the treatment.

In 1978, the ICRU (23) recognized the need for a general dose-specification system which could be adopted universally. Although the system proposed by the ICRU has not been universally implemented, there is a substantial advantage in adopting a common method of dose specification. In this section, I present the highlights of the ICRU proposal. For details, the reader is referred to current documents: Report No. 50 and 62 (24,25).

Figure 11.20 is a schematic representation of various volumes that the ICRU Report no. 50 (24) recommends to be identified in a treatment plan. Delineation of these volumes is greatly facilitated by 3-D imaging but the concept is independent of the methodology used for their determination.

A.1. Gross Tumor Volume

The gross tumor volume (GTV) is the gross demonstrable extent and location of the tumor. It may consist of primary tumor, metastatic lymphadenopathy, or other metastases. Delineation of GTV is possible if the tumor is visible, palpable or demonstrable through imaging. GTV cannot be defined if the tumor has been surgically removed, although an outline of the tumor bed may be substituted by examining preoperative and postoperative images.

A.2. Clinical Target Volume

The CTV consists of the demonstrated tumor(s) if present and any other tissue with presumed tumor. It represents, therefore, the true extent and location of the tumor. Delineation of CTV assumes that there are no tumor cells outside this volume. The CTV must receive adequate dose to achieve the therapeutic aim.

A.3. Internal Target Volume

ICRU Report no. 62 (25) recommends that an internal margin (IM) be added to CTV to compensate for internal physiological movements and variation in size, shape, and position of the CTV during therapy in relation to an internal reference point and its corresponding
coordinate system. The volume that includes CTV with these margins is called the internal target volume (ITV).

A.4. Planning Target Volume

The volume that includes CTV with an IM as well as a set-up margin (SM) for patient movement and set-up uncertainties is called the planning target volume (PTV). To delineate the PTV, the IM and SM are not added linearly but are combined rather subjectively. The margin around CTV in any direction must be large enough to compensate for internal movements as well as patient-motion and set-up uncertainties.

A.5. Planning Organ at Risk Volume

The organ(s) at risk (OR) needs adequate protection just as CTV needs adequate treatment. Once the OR is identified, margins need to be added to compensate for its movements, internal as well as set-up. Thus, in analogy to the PTV, one needs to outline planning organ at risk volume (PRV) to protect OR effectively.

Figure 11.21 schematically illustrates the process of outlining PTV and PRV. This process is intended to make the radiation oncologist think methodically and analytically when outlining targets and organs at risk instead of taking a wild guess. Although absolute accuracy in either case cannot be assured, the objective of this approach is to minimize errors by paying attention to details.

It is also important to point out that there is a common tendency among practitioners to draw target volumes based on GTV with little margins to account for subclinical

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**FIG. 11.21.** Schematic representation of ICRU volumes and margins. (From ICRU, Prescribing, recording and reporting photon beam therapy [supplement to ICRU Report 50], ICRU Report 62. Bethesda, Maryland: International Commission on Radiation Units and Measurements, 1999.)
II. Classical Radiation Therapy

disease, organ motion, or set-up uncertainties. The so-called conformal radiation therapy is a double-edged sword—a high degree of plan conformity can create a high probability of geographical miss. Thus great caution must be exercised in designing PTV and PRV. It is just as important to know the limitations of the system as it is to know its capabilities.

A.6. Treated Volume

Additional margins must be provided around the target volume to allow for limitations of the treatment technique. Thus the minimum target dose should be represented by an isodose surface which adequately covers the PTV to provide that margin. The volume enclosed by this isodose surface is called the treated volume. The treated volume is, in general, larger than the planning target volume and depends on a particular treatment technique.

A.7. Irradiated Volume

The volume of tissue receiving a significant dose (e.g., ≥50% of the specified target dose) is called the irradiated volume. The irradiated volume is larger than the treated volume and depends on the treatment technique used.

A.8. Maximum Target Dose

The highest dose in the target area is called the maximum target dose, provided this dose covers a minimum area of 2 cm². Higher dose areas of less than 2 cm² may be ignored in designating the value of maximum target dose.

A.9. Minimum Target Dose

The minimum target dose is the lowest absorbed dose in the target area.

A.10. Mean Target Dose

If the dose is calculated at a large number of discrete points uniformly distributed in the target area, the mean target dose is the mean of the absorbed dose values at these points. Mathematically:

\[
\text{Mean target dose} = \frac{1}{N} \sum_{i,j} D_{ij}
\]

where \(N\) is the number of points in the matrix and \(D_{ij}\) is the dose at lattice point \(i,j\) located inside the target area (\(A_T\)).

A.11. Median Target Dose

The median target dose is simply the value between the maximum and the minimum absorbed dose values within the target.

A.12. Modal Target Dose

The modal target dose is the absorbed dose that occurs most frequently within the target area. If the dose distribution over a grid of points covering the target area is plotted as a frequency histogram, the dose value showing the highest frequency is called the modal dose. In Fig. 11.22 the modal dose corresponds to the peak of the frequency curve.
A.13. Hot Spots

A hot spot is an area outside the target that receives a higher dose than the specified target dose. Like the maximum target dose, a hot spot is considered clinically meaningful only if it covers an area of at least $2 \text{ cm}^2$.

B. Specification of Target Dose

The absorbed dose distribution in the target volume is usually not uniform. Although a complete dosimetric specification is not possible without the entire dose distribution, there is value in having one figure as the main statement of target dose. The use of the term tumor dose is not recommended (23). The quantity maximum target dose alone cannot be used for reporting, since it can conceal serious underdosages in some parts of the target volume. Although local tumor control depends on the minimum target dose, this quantity alone is not recommended by the ICRU (23), because it is difficult to determine the extent of the tumor, and therefore, the selection of the minimum target dose becomes difficult if not arbitrary. Moreover, if most of the target volume receives a dose that is appreciably different from the minimum, this may also reduce its clinical significance. A statement of both the maximum and minimum values is useful, but it is not always representative of the dose distribution. Furthermore, this would do away with the simplicity of having one quantity for reporting target dose.

The mean, median, and modal doses are not generally recommended, because they usually require elaborate calculations for their accurate determination and may not be feasible by institutions having limited computation facilities.

B.1. The ICRU Reference Point

The target dose should be specified and recorded at what is called the ICRU reference point. This point should satisfy the following general criteria (25):

1. The point should be selected so that the dose at this point is clinically relevant and representative of the dose throughout the PTV;
2. The point should be easy to define in a clear and unambiguous way;
3. The point should be selected where the dose can be accurately calculated;
4. The point should not lie in the penumbra region or where there is a steep dose gradient.
In most cases the ICRU reference point should lie well within the PTV, provided it generally meets the above mentioned criteria. Recommendations for simple beam arrangements are discussed below as examples.

B.1.1. Stationary Photon Beams
1. For a single beam, the target absorbed dose should be specified on the central axis of the beam placed within the PTV.
2. For parallel opposed, equally weighted beams, the point of target dose specification should be on the central axis midway between the beam entrances.
3. For parallel opposed, unequally weighted beams, the target dose should be specified on the central axis placed within the PTV.
4. For any other arrangement of two or more intersecting beams, the point of target dose specification should be at the intersection of the central axes of the beams placed within the PTV.

B.1.2. Rotation Therapy
For full rotation or arcs of at least 270 degrees, the target dose should be specified at the center of rotation in the principal plane. For smaller arcs, the target dose should be stated in the principal plane, first, at the center of rotation and, second, at the center of the target volume. This dual-point specification is required because in a small arc therapy, past-pointing techniques are used that give maximum absorbed dose close to the center of the target area. The dose at the isocenter in these cases, although important to specify, is somewhat less.

B.2. Additional Information
The specification of target dose is meaningful only if sufficient information is provided regarding the irradiation technique. The description of technique should include radiation quality, SSD or SAD, field sizes, beam-modification devices (wedges and shielding blocks, etc.), beam weighting, correction for tissue heterogeneities, dose fractionation, and patient positioning. Many of the above treatment parameters are listed with the treatment plan (isodose pattern) and can be attached to the patient chart. In vivo absorbed dose measurements can also provide useful information and should be recorded in the chart.

Finally, the main objectives of a dose specification and reporting system are to achieve uniformity of dose reporting among institutions, to provide meaningful data for assessing the results of treatments, and to enable the treatment to be repeated elsewhere without having recourse to the original institution for further information.

REFERENCES


Basic depth-dose data and isodose curves are usually measured in a cubic water phantom having dimensions much larger than the field sizes used clinically. Phantom irradiations for this purpose are carried out under standard conditions, for example, beams incident normally on the flat surface at specified distances. The patient's body, however, is neither homogeneous nor flat in surface contour. Thus the dose distribution in a patient may differ significantly from the standard distribution. This chapter discusses several aspects of treatment planning, including acquisition of patient data, correction for contour curvature, and tissue inhomogeneities and patient positioning.

12.1. ACQUISITION OF PATIENT DATA

Accurate patient dosimetry is only possible when sufficiently accurate patient data are available. Such data include body contour, outline, and density of relevant internal structures, location, and extent of the target volume. Acquisition of these data is necessary whether the dosimetric calculations are performed manually or with a computer. However, this important aspect of treatment planning is often executed poorly. For example, in a busy department there may be an inordinate amount of pressure to begin the patient's treatment without adequate dosimetric planning. In other cases, lack of sufficient physics support and/or equipment is the cause of this problem. In such a case, it must be realized that the final accuracy of the treatment plan is strongly dependent on the availability of the patient data and that great effort is needed to improve its quality.

A. Body Contours

Acquisition of body contours and internal structures is best accomplished by imaging (computed tomography [CT] and magnetic resonance imaging, etc.). The scans are performed specifically for treatment planning purposes, with the patient positioned the same way as for actual treatment. In 3-D treatment planning (Chapter 19) these data are all image-based and are acquired as part of the treatment planning process. However, for cases in which 3-D treatment planning is not considered necessary or if body contours are obtained manually for verification of the image-based contours, mechanical or electromechanical methods are used for contouring.

A number of devices have been made to obtain patient contours. Some of these are commercially available while others can be fabricated in the department machine shop. The most common and the simplest of the devices is a solder wire or a lead wire embedded in plastic. Because the wire may not faithfully retain the contour dimensions when transferring it from the patient to the paper, one must independently measure anteroposterior and/or lateral diameters of the contour with a caliper.
Another kind of simple device (1) consists of an array of rods, the tips of which are made to touch the patient's skin and then placed on a sheet of paper for contour drawing. Perhaps the most accurate of the mechanical devices is a pantograph type apparatus (Fig. 12.1) in which a rod can be moved laterally as well as up and down. When the rod is moved over the patient contour, its motion is followed by a pen that records the outline on paper.

Clarke (2) has described an electromechanical device in which motion of the rod over the patient contour is read by a sensing device and transferred to an X-Y recorder. Such a device can be used for digitizing the patient contour for direct input to the treatment planning computer. Optical (3) and ultrasonic (4) methods have also been devised to obtain the contour information.

Although any of the above methods can be used with sufficient accuracy if carefully used, some important points must be considered in regard to manual contour making.

(a) The patient contour must be obtained with the patient in the same position as used in the actual treatment. For this reason, probably the best place for obtaining the contour information is with the patient properly positioned on the treatment simulator couch.

(b) A line representing the tabletop must be indicated in the contour so that this horizontal line can be used as a reference for beam angles.

(c) Important bony landmarks as well as beam entry points, if available, must be indicated on the contour.

(d) Checks of body contour are recommended during the treatment course if the contour is expected to change due to a reduction of tumor volume or a change in patient weight.

(e) If body thickness varies significantly within the treatment field, contours should be determined in more than one plane.
B. Internal Structures

Localization of internal structures for treatment planning should provide quantitative information in regard to the size and location of critical organs or inhomogeneities. Although qualitative information can be obtained from diagnostic radiographs or atlases of cross-sectional anatomy, they cannot be used directly for precise localization of organs relative to the external contour. In order for the contour and the internal structure data to be realistic for a given patient, the localization must be obtained under conditions similar to those of the actual treatment position and on a couch similar to the treatment couch.

The following devices are used for the localization of internal structures and the target volume. A brief discussion regarding their operation and function will be presented.

B.1. Transverse Tomography

Conventional tomography pre-dates the introduction of CT. A transverse tomography unit (Fig. 12.2) consists of a diagnostic x-ray tube and a film cassette that rotates simultaneously with the x-ray tube. The tube is set in such a position that the central axis of the beam makes an angle of about 20 degrees with the film surface. The patient is positioned on the table so that the x-ray beam can pass through a desired body cross-section. As the gantry rotates, structures in only one transverse plane are “in focus.” The structures in regions other than that plane are blurred because of the motion of the x-ray tube and the film cassette.

The difference between conventional tomography and transverse tomography is the orientation of the plane in focus. Whereas a conventional tomographic image is parallel to the long axis of the patient, the transverse tomogram provides a cross-sectional image perpendicular to the body axis.

Transverse tomograms have been used in radiation therapy to provide cross-sectional information of internal structures in relation to the external contour (5,6). Although the

FIG. 12.2. Photograph of a conventional transverse tomography unit (right) in combination with a treatment simulator. (Courtesy of Toshiba Medical Systems, Chicago, IL.)
method can be used for localizing bone, lung, air cavities, and organs with contrast media, the tomograms have poor contrast and spatial resolution. In addition, the images are marred by artifacts and require interpretation.

**B.2. Computed Tomography**

The main disadvantage of conventional transverse tomography is the presence of blurred images resulting from structures outside the plane of interest. In CT, the x-rays used for reconstructing the image enter only the layer under examination, so that unwanted planes are completely omitted. Basically, a narrow beam of x-rays scans across a patient in synchrony with a radiation detector on the opposite side of the patient. If a sufficient number of transmission measurements are taken at different orientations of the x-ray source and detector (Fig. 12.3A), the distribution of attenuation coefficients within the layer may be determined. By assigning different levels to different attenuation coefficients, an image can be reconstructed that represents various structures with different attenuation properties. Such a representation of attenuation coefficients constitutes a CT image.

Since CT scanning was introduced about 30 years ago, there has been a rapid development in both the software and hardware. Most of the improvements in hardware had to do with the scanner motion and the multiplicity of detectors to decrease the scan time. Figure 12.3B shows a modern scanner in which the x-ray tube rotates within a circular array of detectors.
array of 1,000 or more detectors. With such scanners, scan times as fast as 1 sec or less are achievable. Figure 12.4 shows a typical CT image.

Spiral or helical CT scanners were introduced in the early 1990s in which the x-ray tube rotates continuously as the patient is slowly translated through the CT aperture. Helical CTs are faster and provide better visualization of anatomy and target volumes.

The reconstruction of an image by CT is a mathematical process of considerable complexity, generally performed by a computer. For a review of various mathematical approaches for image reconstruction, the reader is referred to a paper by Brooks and Di Chiro (7). The reconstruction algorithm generates what is known as CT numbers, which are related to attenuation coefficients. The CT numbers range from $-1,000$ for air to $+1,000$ for bone, with that for water set at 0. The CT numbers normalized in this manner are called Hounsfield numbers ($H$).

$$H = \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}} \times 1,000$$

(12.1)

where $\mu$ is the linear attenuation coefficient. Thus a Hounsfield unit represents a change of 0.1% in the attenuation coefficient of water.

Because the CT numbers bear a linear relationship with the attenuation coefficients, it is possible to infer electron density (electrons cm$^{-3}$) as shown in Fig. 12.5. Although CT numbers can be correlated with electron density, the relationship is not linear in the entire range of tissue densities. The nonlinearity is caused by the change in atomic number of tissues, which affects the proportion of beam attenuation by Compton versus photoelectric interactions. Figure 12.5 shows a relationship that is linear between lung and soft tissue but nonlinear between soft tissue and bone.

Atomic number information can also be obtained if attenuation coefficients are measured at two different x-ray energies (8). It is possible to transform the attenuation coefficients measured by CT at diagnostic energies to therapeutic energies (9). However, for low atomic number materials such as fat, air, lung, and muscle, this transformation is not necessary for the purpose of calculating dose distributions and inhomogeneity corrections (9).

Application of CT in radiation therapy treatment planning has been the subject of many papers (9–21). The CT information is useful in two aspects of treatment planning: (a) delineation of target volume and the surrounding structures in relation to the external contour; and (b) providing quantitative data (in the form of CT numbers) for tissue
heterogeneity corrections. From a practical point of view, the first aspect is more important than the second. Accurate delineation of surface contour, internal structures, and target volume is not only crucial for optimizing a treatment technique but also necessary for accurate calculation of dose distribution. Even the tissue heterogeneity corrections for megavoltage photon beams can be made with acceptable accuracy by using the CT cross-sectional image to determine the extent of the inhomogeneity and employing published values of electron density. To quote Sontag et al. (12), "The most severe errors in computing the dose distribution are caused by inaccurate delineation of the geometric outlines of tissue inhomogeneities. Less severe errors in the dose calculation are caused by using an inaccurate relative electron density for the inhomogeneity, provided the outline is accurate." Similar observations were also made by Geise and McCullough (11).

From these investigations, it seems that more sophisticated inhomogeneity correction algorithms such as those using "pixel-by-pixel" CT data produce only small improvements in dose accuracy compared with the traditional methods using equivalent depth, provided the extent of the inhomogeneity is accurately known. However, greater precision in the inhomogeneity outline and electron density may be required in regions where severe dose gradients exist in the direction of the beam. For example, the electron density information may be critical for some cases of electron beam therapy and, for high energy photons, in regions where electronic equilibrium is not established. In such instances, methods using pixel-by-pixel correction methods may be required (16,22,23).
There are several commercially available computer treatment planning systems that allow display and use of CT images for treatment planning. Once the CT image has been produced, the data can be transferred to the treatment-planning computer either directly or the outlines of the contour and internal structures can be traced by hand and then entered into the computer. In direct systems, the CT scan is displayed in gray-scale mode on the computer monitor of the treatment planning system and relevant structures can be outlined on the basis of CT number distribution. After the treatment plan is finalized, it can be superimposed on the CT image for visual display.

The use of CT scans in treatment planning is now an established procedure. Comparative studies of treatment planning with and without CT have demonstrated significantly improved accuracy of target delineation, field shaping and normal tissue exclusion from the field when treatments are designed with the aid of CT scans. A review of CT applications in radiotherapy is presented in a book edited by Ling et al. (24). More current applications are discussed in part III of this book.

Although external contour and internal structures are well-delineated by CT, their use in treatment planning requires that they be localized accurately with respect to the treatment geometry. Diagnostic CT scans obtained typically on a curved tabletop with patient position different from that to be used in treatment have limited usefulness in designing technique and dose distribution. Special treatment planning CT scans are required with full attention to patient positioning and other details affecting treatment parameters.

Some of the common considerations in obtaining treatment planning CT scans are the following: (a) a flat tabletop should be used, usually a flat wooden board can be designed to provide a removable insert for the diagnostic CT couch; (b) a large diameter CT aperture (e.g., \( \geq 70 \) cm) can be used to accommodate unusual arm positions and other body configurations encountered in radiation therapy; (c) care should be taken to use patient-positioning or immobilization devices that do not cause image artifacts; (d) patient-positioning, leveling, and immobilization should be done in accordance with the expected treatment technique or simulation if done before CT; (e) external contour landmarks can be delineated, using radiopaque markers such as plastic catheters; (f) sufficiently magnified images for digitization can be obtained if radiographs on film are to be used for drawing target and other structures; and (g) image scale should be accurate both in the X and Y directions.

### Three-dimensional Treatment Planning

Additional considerations go into CT scanning for 3-D treatment planning. Because the 3-D anatomy is derived from individual transverse scans (which are imaged in 2-D), the interslice distance must be sufficiently small to reconstruct the image in three dimensions. Depending on the tumor site or the extent of contemplated treatment volume, contiguous scans are taken with slice thickness ranging from 2 to 10 mm. The total number of slices may range from 30 to 80 mm. This requires fast scan capability to avoid patient movement or discomfort.

Delineation of target and critical organs on each of the scans is necessary for the 3-D reconstruction of these structures. This is an extremely time-consuming procedure, which has been a deterrent to the adoption of 3-D treatment planning on a routine basis. Efforts have been directed toward making this process less cumbersome such as automatic contouring, pattern recognition, and other computer manipulations. However, the basic problem remains that target delineation is inherently a manual process. Although radiographically visible tumor boundaries can be recognized by appropriate computer software, the extent of target volume depends on grade, stage, and patterns of tumor spread to the surrounding structures. Clinical judgment is required in defining the target volume. Obviously, a computer cannot replace the radiation oncologist! At least, not yet.

Besides the time-consuming process of target localization, 3-D computation of dose distribution and display requires much more powerful computers in terms of speed and
store capacity than the conventional treatment-planning systems. However, with the phenomenal growth of computer technology this is not perceived to be a significant barrier to the adoption of routine 3-D planning. The biggest deterrent, however, is going to be the cost of treatment, including equipment and personnel.

Whereas 3-D planning for every patient may not be presently realistic, it has already been found to be quite useful and practical for certain tumors or tumor sites (head and neck, lung, prostate). Treatment of well-localized small lesions (e.g., less than 4 cm in diameter) in the brain by stereotactic radiosurgery has greatly benefited by 3-D planning. In this procedure, the target volume is usually based on the extent of radiographically visible tumor (with contrast), thus obviating the need for manual target delineation on each CT slice. The 3-D display of dose distribution to assess coverage of the target volume confined to a relatively small number of slices is both useful and practical. Similarly, brachytherapy is amenable to 3-D planning because of the limited number of slices involving the target.

The next best thing to full-fledged 3-D planning (e.g., 3-D computation and display) is to do 2-D planning, using a limited number of CT scans, selected to obtain a reasonably adequate perspective of the dose distribution in three dimensions. For example, targets and other critical structures may be drawn on 5 to 10 CT cuts, spanning the volume of interest. This can be done either by drawing targets and structures on the CT films or directly on the computer screen by using a cursor or a light pen. Treatment-planning software is available whereby margins around the target volume can be specified to set the field boundaries. After optimizing field margins, beam angles, and other plan parameters relative to the central CT cut, the dose distributions can be viewed in other slices either individually or simultaneously by serial display on the screen. Beam’s eye view (BEV) display in which the plan is viewed from the vantage point of the radiation source (in a plane perpendicular to the central axis) is useful in providing the same perspective as a simulator or port film. In addition, a BEV outline of the field can be obtained to aid in the drawing of custom blocks on the simulator film. More discussion on CT-based treatment planning is provided in Chapter 19.

6.3. Magnetic Resonance Imaging

Magnetic resonance imaging (MRI) has developed, in parallel to CT, into a powerful imaging modality. Like CT, it provides anatomic images in multiple planes. Whereas CT provides basically transverse axial images (which can be further processed to reconstruct images in other planes or in three dimensions) MRI can be used to scan directly in axial, sagittal, coronal, or oblique planes. This makes it possible to obtain optimal views to enhance diagnostic interpretation or target delineation for radiotherapy. Other advantages over CT include not involving the use of ionizing radiation, higher contrast, and better imaging of soft tissue tumors. Some disadvantages compared with CT include lower spatial resolution; inability to image bone or calcifications; longer scan acquisition time, thereby increasing the possibility of motion artifacts; technical difficulties due to small hole of the magnet and magnetic interference with metallic objects; and current unavailability of many approved MRI contrast agents.

The previous cursory comparison between CT and MRI shows that the two types of imaging are complementary.

Basic physics of MRI involves a phenomenon known as nuclear magnetic resonance (NMR). It is a resonance transition between nuclear spin states of certain atomic nuclei when subjected to a radiofrequency (RF) signal of a specific frequency in the presence of an external magnetic field. The nuclei that participate in this phenomenon are the ones that intrinsically possess spinning motion, i.e., have angular momentum. These rotating charges act as tiny magnets with associated magnetic dipole moment, a property that gives a measure of how quickly the magnet will align itself along an external magnetic field. Because of the spinning motion or the magnetic dipole moment, nuclei align their spin axes along the external magnetic field (H) as well as orbit or precess around it (Fig. 12.6).
Protons behave as magnetic dipoles and align in the direction of a strong magnetic field.

Magnetic field orientation

Protons rotate about their own axes...

...and simultaneously orbit (precess) around the direction of the magnetic field.

Magnetic field orientation

The frequency of precession is called the Larmor frequency. A second alternating field is generated by applying an alternating voltage (at the Larmor frequency) to an RF coil. This field is applied perpendicular to H and rotates around H at the Larmor frequency. This causes the nuclei to precess around the new field in the transverse direction. When the RF signal is turned off, the nuclei return to their original alignment around H. This transition is called relaxation. It induces a signal in the receiving RF coil (tuned to the Larmor frequency) which constitutes the NMR signal.

The turning off of the transverse RF field causes nuclei to relax in the transverse direction ($T_2$ relaxation) as well as to return to the original longitudinal direction of the magnetic field ($T_1$ relaxation). This is schematically illustrated in Fig. 12.7. The relaxation times, $T_1$ and $T_2$, are actually time constants (like the decay constant in radioactive decay) for the exponential function that governs the two transitions.

The signal source in MRI can be any nucleus with nonzero spin or angular momentum. However, certain nuclei give larger signal than the others. Hydrogen nuclei (protons), because of their high intrinsic sensitivity and high concentration in tissues, produce signals of sufficient strength for imaging. Work with other possible candidates, $^{31}$P, $^{23}$Na, $^{19}$F,
12. Treatment Planning 11: Patient Data, Corrections, and Set-up

A radiofrequency wave burst is applied at right angles to the magnetic field, causing deflection of the protons.

When the radiofrequency current is turned off, the protons realign themselves in their original direction.

The protons then realign with the magnetic field, during T1 (longitudinal) relaxation.

13C, and 2H is continuing. Currently, routine MRI is based exclusively on proton density and proton relaxation characteristics of different tissues.

Localization of protons in a 3-D space is achieved by applying magnetic field gradients produced by gradient RF coils in three orthogonal planes. This changes the precession frequency of protons spatially, because the MR frequency is linearly proportional to field strength. Thus by the appropriate interplay of the external magnetic field and the RF field gradients, proton distribution can be localized. A body slice is imaged by applying field gradient along the axis of the slice and selecting a frequency range for a readout. The strength of the field gradient determines the thickness of the slice (the greater the gradient, thinner the slice). Localization within the slice is accomplished by phase encoding (using back-to-front Y-gradient) and frequency encoding (using transverse X-gradient). In the process, the computer stores phase (angle of precession of the proton at a particular time) and frequency information and reconstructs the image by mathematical manipulation of the data.

Most MR imaging uses a spin echo technique in which a 180-degree RF pulse is applied after the initial 90-degree pulse, and the resulting signal is received at a time that is equal to twice the interval between the two pulses. This time is called the echo time (TE). The time between each 90-degree pulse in an imaging sequence is called the repetition time (TR). By adjusting TR and TE, image contrast can be affected. For example, a long TR and short TE produces a proton (spin) density-weighted image, a short TR and a short TE produces a T1-weighted image, and a long TR and a long TE produces a T2-weighted image.

image. Thus differences in proton density, $T_1$, and $T_2$ between different tissues can be enhanced by a manipulation of TE and TR in the spin echo technique.

Figure 12.8 shows examples of MR images obtained in the axial, sagittal, and coronal planes. By convention, a strong MR signal is displayed as white and a weak signal is displayed as dark on the cathode ray tube or film.

B.4. Ultrasound

Ultrasonic imaging for delineating patient contours and internal structure is becoming widely recognized as an important tool in radiation therapy. Tomographic views provide cross-sectional information that is always helpful for treatment planning. Although in most cases the image quality or clinical reliability is not as good as that of the CT, ultrasonic procedure does not involve ionizing radiation, is less expensive, and in some cases, yields data of comparable usefulness.
Ultrasound can provide useful information in localizing many malignancy-prone structures in the lower pelvis, retroperitoneum, upper abdomen, breast, and chest wall (25). A detailed review of these techniques in the context of radiation therapy planning has been presented by Carson et al. (25,26).

An ultrasound (or ultrasonic) wave is a sound wave having a frequency greater than 20,000 cycles per sec or hertz (Hz). At this frequency, the sound is inaudible to the human ear. Ultrasound waves of frequencies 1 to 20 MHz are used in diagnostic radiology.

Ultrasound may be used to produce images either by means of transmission or reflection. However, in most clinical applications, use is made of ultrasonic waves reflected from different tissue interfaces. These reflections or echoes are caused by variations in acoustic impedance of materials on opposite sides of the interfaces. The acoustic impedance \( Z \) of a medium is defined as the product of the density of the medium and the velocity of ultrasound in the medium. The larger the difference in \( Z \) between the two media, the greater is the fraction of ultrasound energy reflected at the interface. For example, strong reflections of ultrasound occur at the air-tissue, tissue-bone, and chest wall-lung interfaces due to high impedance mismatch. However, because lung contains millions of air-tissue interfaces, strong reflections at the numerous interfaces prevents its use in lung imaging.

Attenuation of the ultrasound by the medium also plays an important role in ultrasound imaging. This attenuation takes place as the energy is removed from the beam by absorption, scattering, and reflection. The energy remaining in the beam decreases approximately exponentially with the depth of penetration into the medium, allowing attenuation in different media to be characterized by attenuation coefficients. As the attenuation coefficient of ultrasound is very high for bone compared with soft tissue, together with the large reflection coefficient of a tissue-bone interface, it is difficult to visualize structures lying beyond bone. On the other hand, water, blood, fat, and muscle are very good transmitters of ultrasound energy.

Ultrasound waves are generated as well as detected by an ultrasonic probe or transducer. A transducer is a device that converts one form of energy into another. An ultrasonic transducer converts electrical energy into ultrasound energy, and vice versa. This is accomplished by a process known as the piezoelectric effect. This effect is exhibited by certain crystals in which a variation of an electric field across the crystal causes it to oscillate mechanically, thus generating acoustic waves. Conversely, pressure variations across a piezoelectric material (in response to an incident ultrasound wave) result in a varying electrical potential across opposite surfaces of the crystal.

Although the piezoelectric effect is exhibited by a number of naturally occurring crystals, most common crystals used clinically are made artificially such as barium titanate, lead zirconium titanate, and lead metaniobate. The piezoelectric effect produced by these materials is mediated by their electric dipole moment, the magnitude of which can be varied by addition of suitable impurities.

As the ultrasound wave reflected from tissue interfaces is received by the transducer, voltage pulses are produced that are processed and displayed on the cathode ray tube (CRT), usually in one of three display modes: A (amplitude) mode, B (brightness) mode, and M (motion) mode. A mode consists of displaying the signal amplitude on the ordinate and time on the abscissa. The time, in this case, is related to distance or tissue depth, given the speed of sound in the medium. In the B mode, a signal from a point in the medium is displayed by an echo dot on the CRT. The \((x, y)\) position of the dot on the CRT indicates the location of the reflecting point at the interface and its proportional brightness reveals the amplitude of the echo. By scanning across the patient, the B-mode viewer sees an apparent cross-section through the patient. Such cross-sectional images are called ultrasonic tomograms.

In the M mode of presentation, the ultrasound images display the motion of internal structures of the patient's anatomy. The most frequent application of M mode scanning is echocardiography. In radiotherapy, the cross-sectional information used for treatment planning is exclusively derived from the B scan images (Fig. 12.9). The use of ultrasound in brachytherapy (e.g., ultrasound-guided prostate implants) is discussed in Chapter 23.
12.2. TREATMENT SIMULATION

A treatment simulator (Fig. 12.10) is an apparatus that uses a diagnostic x-ray tube but duplicates a radiation treatment unit in terms of its geometrical, mechanical, and optical properties. The main function of a simulator is to display the treatment fields so that the target volume may be accurately encompassed without delivering excessive irradiation to surrounding normal tissues. By radiographic visualization of internal organs, correct positioning of fields and shielding blocks can be obtained in relation to external landmarks. Most commercially available simulators have fluoroscopic capability by dynamic visualization before a hard copy is obtained in terms of the simulator radiography.

Specifications of a treatment simulator must closely match those of the treatment unit. Several authors (27–31) have discussed these aspects. For a comprehensive discussion, the reader is referred to the paper by McCullough and Earl (31).
The need for simulators arises from four facts: (a) geometrical relationship between the radiation beam and the external and internal anatomy of the patient cannot be duplicated by an ordinary diagnostic x-ray unit; (b) although field localization can be achieved directly with a therapy machine by taking a port film, the radiographic quality is poor because of very high beam energy, and for cobalt-60, a large source size as well; (c) field localization is a time-consuming process which, if carried out in the treatment room, could engage a therapy machine for a prohibitive length of time; (d) unforeseen problems with a patient set-up or treatment technique can be solved during simulation, thus conserving time within the treatment room.

Although the practical use of simulators varies widely from institution to institution, the simulator room is increasingly assuming the role of a treatment-planning room. Besides localizing treatment volume and setting up fields, other necessary data can also be obtained at the time of simulation. Because the simulator table is supposed to be similar to the treatment table, various patient measurements such as contours and thicknesses, including those related to compensator or bolus design, can be obtained under appropriate set-up conditions. Fabrication and testing of individualized shielding blocks can also be accomplished with a simulator. To facilitate such measurements, modern simulators are equipped with accessories such as laser lights, contour maker, and shadow tray.

Some simulators have a tomography attachment in which the image from the image intensifier is analyzed and reconstructed using either analogue\(^1\) or digital\(^2\) processing. Because of the poor image quality, this technology cannot compete with CT-based virtual simulation.

An exciting development in the area of simulation is that of converting a CT scanner into a simulator. A commercial device\(^3\) known as CT-SIM uses a CT scanner to localize the treatment fields on the basis of the patient's CT scans. A computer program, specifically written for simulation, automatically positions the patient couch and the laser cross hairs to define the scans and the treatment fields. The software provides automatic outlining of external contours and critical structures, interactive portal displays and placement, display of isodose distribution, and review of multiple treatment plans. Such an integrated approach to treatment planning has the potential of becoming the simulator cum treatment-planning system of the future.

12.3. TREATMENT VERIFICATION

A. Port Films

The primary purpose of port filming is to verify the treatment volume under actual conditions of treatment. Although the image quality with the megavoltage x-ray beam is poorer than with the diagnostic or the simulator film, a port film is considered mandatory not only as a good clinical practice but as a legal record.

As a treatment record, a port film must be of sufficiently good quality so that the field boundaries can be described anatomically. However, this may not always be possible due to either very high beam energy (10 MV or higher), large source size (cobalt-60), large patient thickness (>20 cm), or poor radiographic technique. In such a case, the availability of a simulator film and/or a treatment diagram with adequate anatomic description of the field is helpful. Anatomic interpretation of a port film is helped by obtaining a full-field exposure on top of the treatment port exposure.

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\(^1\)Oldelft Corporation of America, Fairfax, Virginia.

\(^2\)Varian Associates, Palo Alto, California.

\(^3\)Theratronics International Ltd., Kanata, Ontario, Canada. Also, Picker International, Inc., St. Davida, Pennsylvania.
Radiographic technique significantly influences image quality of a port film. The choice of film and screen as well as the exposure technique is important in this regard. Droege and Bj"{a}rn~gard (32) have analyzed the film screen combinations commonly used for port filming at megavoltage x-ray energies. Their investigation shows that the use of a single emulsion film with the emulsion adjacent to a single lead screen between the film and the patient is preferable to a double emulsion film or a film with more than one screen. Thus, for optimum resolution, one needs a single emulsion film with a front lead screen and no rear screen. Conventional nonmetallic screens are not recommended at megavoltage energies. Although thicker metal screens produce a better response, an increase in thickness beyond the maximum electron range produces no further changes in resolution (32).

Certain slow-speed films, ready packed but without a screen, can be exposed during the entire treatment duration. A therapy verification film such as Kodak XV-2 is sufficiently slow to allow an exposure of up to 200 cGy without reaching saturation. In addition, such films can be used to construct compensators for both the contours and tissue heterogeneity (33).

B. Electronic Portal Imaging

Major limitations of port films are (a) viewing is delayed because of the time required for processing; (b) it is impractical to do port films before each treatment; and (c) film image is of poor quality especially for photon energies greater than 6 MV. Electronic portal imaging overcomes the first two problems by making it possible to view the portal images instantaneously, i.e., images can be displayed on computer screen before initiating a treatment or in real-time during the treatment. Portal images can also be stored on computer discs for later viewing or archiving.

On-line electronic portal imaging devices (EPIDs) are currently being clinically used in several institutions, and some of them are commercially available. Many of the systems are video based; the beam transmitted through the patient excites a metal fluorescent screen, which is viewed by a video camera using a 45 degree mirror (34–37) (Fig. 12.11).

---

4 Such a sheet of lead acts as an intensifying screen by means of electrons ejected from the screen by photon interactions. These electrons provide an image on the film which reflects the variation of beam intensity transmitted through the patient.
The camera is interfaced to a microcomputer through a frame-grabber board for digitizing the video image. The images are acquired and digitized at the video rate of 30 frames per second. An appropriate number of frames is averaged to produce a final image. Depending on the computer software, the image data can be further manipulated to improve the image quality or perform a special study.

One problem with mirror-based EPIDs is the large size of the mirror, which can pose practical problems. Wong et al. (38) have developed a system that replaces the mirror with a fiberoptics system to direct the fluorescent light to the video camera. The fiberoptics channels consist of thin clear polystyrene columns encased by a thin acrylic cladding. Because of the difference in the refractive indices of the two plastics, it is possible to conduct light without significant loss of intensity. This “light piping” is accomplished by the process of total internal reflection at the cladding interface.

Another class of EPIDs consists of a matrix of liquid ion chambers used as detectors (39, 40). These devices are much more compact than the video-based systems and are comparable in size to a film cassette, albeit a little heavier. Figure 12.12 shows a system developed at The Nederlands Kanker Institute. The system consists of a matrix of 256 x 256 ion chambers containing an organic fluid and a microcomputer for image processing. Figure 12.13 shows an image obtained with such a device. Besides imaging, another potential use of this device is on-line patient dose monitoring. Further work is needed to develop this application.

Yet another type of EPID uses solid-state detectors. One approach employs a scanning linear array of silicon diodes. Another uses a linear array of zinc tungstate (ZnWO₄) scintillating crystals attached to photodiodes. An excellent review of these developments is provided by Boyer et al. (41).
A variety of technologies are being explored to develop new EPIDS or refine the existing ones. For example, Varian Medical Systems has recently introduced PortalVision aS500 featuring an array of image detectors based on Amorphous Silicon (a-Si) technology. Within this unit a scintillator converts the radiation beam into visible photons. The light is detected by an array of photodiodes implanted on an amorphous silicon panel. The photodiodes integrate the light into charge captures. The sensitive area of the EPID is $40 \times 30$ cm$^2$ with $512 \times 384$ pixels, spatial resolution is 0.78 mm and the read-out image has about 200,000 pixels. This system offers better image quality than the previous system using liquid ion chambers.

**12.4. CORRECTIONS FOR CONTOUR IRREGULARITIES**

As mentioned at the beginning of this chapter, basic dose distribution data are obtained under standard conditions, which include homogeneous unit density phantom, perpendicular beam incidence, and flat surface. During treatment, however, the beam may be obliquely incident with respect to the surface and, in addition, the surface may be curved.
or irregular in shape. Under such conditions, the standard dose distributions cannot be applied without proper modifications or corrections.

Contour corrections may be avoided by using a bolus or a compensator (to be discussed in section 12.4) but under some circumstances, it is permissible or even desirable, to determine the actual dose distribution by calculation. The following three methods are recommended for angles of incidence of up to 45 degree for megavoltage beams and of up to 30 degree from the surface normal for orthovoltage x-rays (42). Although all computer treatment-planning algorithms are capable of correcting for contour irregularities, these methods are discussed below to illustrate the basic principles.

### A. Effective Source-to-Surface Distance Method

Consider Fig. 12.14 in which source-to-surface is an irregularly shaped patient contour. It is desired to calculate the percent depth dose at point $A$ (i.e., dose at $A$ as a percentage of $D_{\text{max}}$, dose at point $Q$). The diagram shows that the tissue deficit above point $A$ is $h$ cm and the depth of $D_{\text{max}}$ is $d_{\text{max}}$. If we note that the percent depth dose does not change rapidly with SSD (provided that the SSD is large), the relative depth dose distribution along the line joining the source with point $A$ is unchanged when the isodose chart is moved down by the distance $h$ and positioned with its surface line at $S'\rightarrow S''$. Suppose $D_A$ is the dose at point $A$. Assuming beam to be incident on a flat surface located at $S'\rightarrow S''$,

$$D_A = D'_{\text{max}} \cdot P'$$

(12.2)

where $P'$ is percent depth dose at $A$ relative to $D'_{\text{max}}$ at point $Q'$. Suppose $P_{\text{corr}}$ is the correct percent depth dose at $A$ relative to $D_{\text{max}}$ at point $Q$. Then,

$$D_A = D_{\text{max}} \cdot P_{\text{corr}}$$

(12.3)

![Diagram illustrating methods of correcting dose distribution under an irregular surface such as S-S. The solid isodose curves are from an isodose chart which assumes a flat surface located at S'-S'. The dashed isodose curves assume a flat surface at S''-S'' without any air gap.](image-url)
From Equation 12.2 and 12.3,
\[
P_{\text{corr}} = P' \cdot \left( \frac{D'_{\text{max}}}{D_{\text{max}}} \right)
\]  
(12.4)

Because, when the distribution is moved, the SSD is increased by a distance \( h \), we have:
\[
\frac{D'_{\text{max}}}{D_{\text{max}}} = \left( \frac{\text{SSD} + d_m}{\text{SSD} + h + d_m} \right)^2
\]  
(12.5)

Therefore,
\[
P_{\text{corr}} = P' \left( \frac{\text{SSD} + d_m}{\text{SSD} + h + d_m} \right)^2
\]  
(12.6)

Thus the effective SSD method consists of sliding the isodose chart down so that its surface line is at \( S' - S \)'s end, reading off the percent dose value at \( A \) and multiplying it by the inverse square law factor to give the corrected percent depth dose value.

The above method applies the same way when there is excess tissue above \( A \) instead of tissue deficit. In such a case, the isodose chart is moved up so that its surface line passes through the point of intersection of the contour line and the ray line through \( A \). The value of \( h \) is assigned a negative value in this case.

**B. Tissue-air (or Tissue-maximum) Ratio Method**

This method depends on the principle that the tissue-air, or tissue-maximum, ratio does not depend on the SSD and is a function only of the depth and the field size at that depth. Suppose, in Fig. 12.14, the surface is located at \( S'' - S'' \)'s end and the air space between \( S - S \) and \( S'' - S'' \) is filled with tissue-like material. Now, if a standard isodose chart for the given beam and SSD is placed with its surface at \( S'' - S'' \), the percent depth dose value at \( A \) will correspond to the depth \( d + h \). But the actual value at \( A \) is greater than this as there is a tissue deficit. The correction factor can be obtained by the ratio of tissue-air or tissue-maximum ratios for depths \( d \) and \( d + h \).

\[
\text{Correction factor (CF)} = \frac{T(d, r_A)}{T(d + h, r_A)} \tag{12.7}
\]

where \( T \) stands for tissue-air ratio or tissue-maximum ratio and \( r_A \) is the field size projected at point \( A \) (i.e., at a distance of SSD + \( d + h \) from the source).

Thus if the uncorrected value of percent depth dose at \( A \) with the surface line of the isodose chart at \( S'' - S'' \) is \( P'' \), then the corrected value \( P_{\text{corr}} \) is given:
\[
P_{\text{corr}} = P'' \cdot \text{CF}
\]

**C. Isodose Shift Method**

The preceding methods are useful for making individual point dose calculations. However, for manual treatment planning, it is convenient to correct the entire isodose chart for contour irregularities. This can be done by an empirical method, known as the isodose shift method. The procedure is illustrated in Fig. 12.15. Suppose \( S - S \) is the patient contour drawn on a transparent paper and \( S' - S' \) is a flat surface line passing through the point of intersection of the central axis with the contour. From the line \( S' - S' \), draw vertical grid lines, parallel to the central axis and spaced about 1 cm apart, to cover the full field width. Place the standard isodose chart underneath this paper and align the central line of the chart with that of the grid. Mark the percent depth dose values on the central axis. For each grid line, slide the isodose chart up or down, depending on whether there is tissue
12. Treatment Planning II: Patient Data, Corrections, and Set-up

excess or deficit along that line, by an amount \( k \times h \) where \( k \) is a factor less than 1 (given in Table 12.1). Then mark the isodose values at points of intersection of the given grid line and the shifted isodose curves. After all the isodose positions along all the grid lines have been marked, new isodose curves are drawn by joining the marked points having the same isodose values.

The factor \( k \) depends on the radiation quality, field size, depth of interest, and SSD. Table 12.1 gives approximate values recommended for clinical use when manual corrections are needed.

Of the three methods discussed above, the tissue-air or tissue-maximum ratio method gives the most accurate results. The first two methods are especially useful in computer treatment planning.

**TABLE 12.1. ISODOSE SHIFT FACTORS FOR DIFFERENT BEAM ENERGIES**

<table>
<thead>
<tr>
<th>Photon Energy (MV)</th>
<th>Approximate Factor ( k )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Up to 1</td>
<td>0.8</td>
</tr>
<tr>
<td>( {\gamma}_{Co-5} )</td>
<td>0.7</td>
</tr>
<tr>
<td>5-15</td>
<td>0.6</td>
</tr>
<tr>
<td>15-30</td>
<td>0.5</td>
</tr>
<tr>
<td>Above 30</td>
<td>0.4</td>
</tr>
</tbody>
</table>

Example 1

For point A in Fig. 12.14, \( h = 3 \) cm and \( d = 5 \) cm. Calculate the percent depth dose at point A using (a) the effective SSD method and (b) the tissue-air ratio method.

Given \(^{60}\text{Co}\) beam, \( \text{TAR}(5, 11 \times 11) = 0.910 \) and \( \text{TAR}(8, 11 \times 11) = 0.795 \) and SSD = 80 cm:

(a) using solid isodose curve lines in Fig. 12.14,

\[
\text{Percent depth dose at } A = 78.1
\]

Inverse square law factor \( = \frac{80 + 0.5}{80 + 3 + 0.5} \)

\( = 0.929 \)

Corrected percent depth dose at A \( = 78.1 \times 0.929 \)

\( = 72.6 \)

(b) Field dimension projected at A \( = 10 \times \frac{88}{80} = 11 \) cm. Thus field size at A \( = 11 \times 11 \) cm:

\[
\text{CF} = \frac{\text{TAR}(5, 11 \times 11)}{\text{TAR}(8, 11 \times 11)} = \frac{0.910}{0.795} = 1.145
\]

Using dashed isodose lines in Fig. 12.14, uncorrected percent depth dose \( = 65.2 \).

Corrected percent depth dose \( = 65.2 \times 1.145 \)

\( = 74.6 \)

Comparing the results for (a) and (b), the agreement between the two methods is within 3%.

12.5. CORRECTIONS FOR TISSUE INHOMOGENEITIES

Applications of standard isodose charts and depth dose tables assume homogeneous unit density medium. In a patient however, the beam may transverse layers of fat, bone, muscle, lung, and air. The presence of these inhomogeneities will produce changes in the dose distribution, depending on the amount and type of material present and on the quality of radiation.

The effects of tissue inhomogeneities may be classified into two general categories: (a) changes in the absorption of the primary beam and the associated pattern of scattered photons and (b) changes in the secondary electron fluence. The relative importance of these effects depends on the region of interest where alterations in absorbed dose are considered. For points that lie beyond the inhomogeneity, the predominant effect is the attenuation of the primary beam. Changes in the associated photon scatter distribution alters the dose distribution more strongly near the inhomogeneity than farther beyond it. The changes in the secondary electron fluence, on the other hand, affects the tissues within the inhomogeneity and at the boundaries.

For x-ray beams in the megavoltage range, where Compton effect is a predominant mode of interaction, the attenuation of the beam in any medium is governed by electron density (number of electrons per cm\(^3\)). Thus an effective depth can be used for calculating transmission through nonwater-equivalent materials. However, close to the boundary or interface, the distribution is more complex. For example, for megavoltage beams, there may be loss of electronic equilibrium close to the boundaries of low-density materials or air cavities. For orthovoltage and superficial x-rays, the major problem is the bone. Absorbed dose within the bone or in the immediate vicinity of it may be several times higher than the dose in the soft tissue in the absence of bone. This increased energy absorption is caused
by the increase in the electron fluence arising from the photoelectric absorption in the mineral contents of the bone.

A. Corrections for Beam Attenuation and Scattering

Figure 12.16 is a schematic diagram showing an inhomogeneity of electron density $\rho_e$ relative to that of water. $P$ is the point of dose calculation.

Three methods of correcting for inhomogeneities are illustrated with reference to Fig. 12.16.

A.1. Tissue-air Ratio Method

The following $CF$ applies to the dose at $P$ if the entire phantom was water equivalent:

$$CF = \frac{T(d', r_d)}{T(d, r_d)}$$

where $d'$ is the equivalent water depth, i.e., $d' = d_1 + \rho_e d_2 + d_3$, and $d$ is the actual depth of $P$ from the surface; $r_d$ is the field size projected at point $P$.

The above correction method does not take into account the position of the inhomogeneity relative to point $P$. In other words, the correction factor will not change with $d_3$ as long as $d$ and $d'$ remain constant.

A.2. Power Law Tissue-air Ratio Method

Batho (43) and Young and Gaylord (44) have proposed a method in which the ratio of the tissue-air ratios is raised to a power. Referring again to Fig. 12.16, the correction factor at
Here $\rho_s$ is the electron density (number of electrons/cm$^3$) of the heterogeneity relative to that of water.

As seen in Equation 12.9, the correction factor does depend on the location of the inhomogeneity relative to point $P$ but not relative to the surface. This formulation is based on theoretical considerations assuming Compton interactions only. It does not apply to points inside the inhomogeneity or in the build-up region. Experimental verification of the model has been provided for $^{60}$Co $\gamma$ beams (43,44).

A more general form of the power law method is provided by Sontag and Cunningham (45) that allows for correction of the dose to points within an inhomogeneity as well as below it. This is given by:

$$CF = \left[ \frac{T(d_s + d_3, r_d)}{T(d_s, r_d)} \right]^{p_3 - p_2}$$

where $p_3$ is the density of the material in which point $P$ lies and $d_3$ is its depth within this material. $p_2$ is the density of the overlying material, and $(d_2 + d_3)$ is the depth below the upper surface of it. It may be pointed out that Equation 12.10 reduces to Equation 12.9 if $P$ lies in a unit density medium as shown in Fig. 12.16.

### A.3. Equivalent Tissue-air Ratio Method

The use of water equivalent depth in Equation 12.8 appropriately corrects for the primary component of dose. However, the change in scattered dose is not correctly predicted because the effect of scattering structures depends on their geometric arrangement with respect to point $P$. Sontag and Cunningham (21) accounted for these geometric factors through the scaling of the field size parameter. Their method using "equivalent" tissue-air ratios (ETARs) is given by:

$$CF = \frac{T(d', r')}{T(d, r)}$$

where $d'$ is the water equivalent depth, $d$ is the actual depth, $r$ is the beam dimension at depth $d$, $r' = r \cdot \bar{\rho}$ = scaled field size dimension, and $\bar{\rho}$ is the weighted density of the irradiated volume.

The weighted density $\bar{\rho}$ can be determined by the averaging procedure:

$$\bar{\rho} = \frac{\sum_i \sum_j \sum_k \rho_{jk} \cdot W_{jk}}{\sum_i \sum_j \sum_k W_{jk}}$$

where $\rho_{jk}$ are the relative electron densities of scatter elements (e.g., pixels in a series of CT images of the irradiated volume) and $W_{jk}$ are the weighting factors assigned to these elements in terms of their relative contribution to the scattered dose at the point of calculation.

The weighting factors are calculated using Compton scatter cross-sections and integrating scatter over the entire irradiated volume for each point of dose calculation. A more practical approach is to "coalesce" all of the density information from individual slices into a single "equivalent" slice, thus reducing the volume integration to an integration over a plane. Details of this procedure are discussed by Sontag and Cunningham (21).

An alternative approach to the ETAR method is to calculate scattered dose separately from the primary dose by summation of the scatter contribution from individual scatter elements in the irradiated heterogeneous volume. This method is known as the differential scatter-air ratio (DSAR) method (46,47). More advanced computer-based methods such
TABLE 12.2. ISODOSE SHIFT FACTORS* FOR INHOMOGENEITIES

<table>
<thead>
<tr>
<th>Inhomogeneity</th>
<th>Shift Factor n*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air cavity</td>
<td>-0.6</td>
</tr>
<tr>
<td>Lung</td>
<td>-0.4</td>
</tr>
<tr>
<td>Hard bone</td>
<td>0.5</td>
</tr>
<tr>
<td>Spongy bone</td>
<td>0.25</td>
</tr>
</tbody>
</table>

*Approximate factors, determined empirically for 60Co and 4-MV x-rays.

as delta volume (DV) (47,48), dose spread array (DSA) (49) and differential pencil beam (DPB) (50) methods have been proposed to take into account multiple scattering of photons and electron transport to predict dose more accurately as well as in the regions where electronic equilibrium does not exist. A discussion of model-based algorithms using dose kernels (e.g., convolution/superposition algorithms) and Monte Carlo techniques is presented in Chapter 19.

A.4. Isodose Shift Method

This method, proposed by Greene and Stewart (51) and Sundblom (52), is convenient for manually correcting isodose charts for the presence of inhomogeneities. The isodose curves beyond the inhomogeneity are moved by an amount equal to $n$ times the thickness of the inhomogeneity as measured along a line parallel to the central axis and passing through the point of interest. The shift is toward the skin for bone and away from the skin for lung or air cavities. Table 12.2 gives experimentally determined values of $n$ which apply to 60Co radiation and 4-MV x-rays. The factors are believed to be independent of field size.

A.5. Typical Correction Factors

None of the methods discussed above can claim an accuracy of ±5% for all irradiation conditions encountered in radiotherapy. The new generation of algorithms that take account of the 3-D shape of the irradiated volume and the electron transport are expected to achieve that goal but are still under development. Most commercial systems use one-dimensional methods in which bulk density-based inhomogeneity corrections are applied along ray lines, disregarding the extent of inhomogeneities in the other dimensions.

Tang et al. (53) have compared a few commonly used methods, namely, the TAR, the ETAR, and the generalized Batho against measured data using a heterogeneous phantom containing layers of polystyrene and cork. Their results show that for the geometries considered (a) the TAR method overestimates the dose for all energies, (b) the ETAR is best suited for the lower-energy beams (≤6 MV), and (c) the generalized Batho method is the best in the high-energy range (≥10 MV). Thus the accuracy of different methods depend on the irradiation conditions, e.g., energy, field size, location and extent of inhomogeneity, and location of point of calculation.

Table 12.3 gives some examples of increase in dose beyond healthy lung for various beam energies. These correction factors have been calculated by using Equation 12.10, assuming $d_1 = 6$ cm, $d_2 = 8$ cm, and $d_3 = 3$ cm, relative $\rho$, for lung = 0.25, and field size = 10 × 10 cm. The values were rounded off to represent approximate factors for typical lung corrections. More detailed tables of the beyond-lung and in-lung correction factors have been calculated by McDonald et al. (54) for several representative beam energies and field sizes.
II. Classical Radiation Therapy

**TABLE 12.3. INCREASE IN DOSE TO TISSUES BEYOND HEALTHY LUNG**

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Correction Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Orthovoltage</td>
<td>+10%/cm of lung</td>
</tr>
<tr>
<td>$^{60}$Co γ-rays</td>
<td>+4%/cm of lung</td>
</tr>
<tr>
<td>4-MV x-rays</td>
<td>+3%/cm of lung</td>
</tr>
<tr>
<td>10-MV x-rays</td>
<td>+2%/cm of lung</td>
</tr>
<tr>
<td>20-MV x-rays</td>
<td>+1%/cm of lung</td>
</tr>
</tbody>
</table>

*Approximate values calculated with Equation 12.10 for typical clinical situations.

Table 12.4 gives the decrease in dose beyond bone that might be expected with beams of different energies. These are approximate values because the shielding effect of bone depends on the size of the bone, field size, and other parameters that affect scattering. The shielding effect of bone diminishes quite rapidly as the beam energy increases. The shielding effect of bone for x-rays generated between 500 kV and 4 MV is entirely due to its greater electron density (electrons per cm$^3$), as all the attenuation is due to the Compton process. In the megavoltage range, the corrections for bone attenuation in most clinical situations are small and are usually neglected. However, as the x-ray energy increases beyond 10 MV, the shielding effect begins to increase because pair production becomes significant. Recall that the absorption of radiation as a result of pair production depends on the atomic number.

**B. Absorbed Dose within an Inhomogeneity**

As mentioned earlier, the absorbed dose within an inhomogeneity or in the soft tissues adjacent to it is strongly influenced by alterations in the secondary electron fluence. For example, for x-rays generated at potentials less than 250 kVp, there is a substantial increase in absorbed dose inside bone because of increased electron fluence arising from photoelectric absorption. Spiers (55,56) has made a comprehensive study of absorbed dose within mineral bone as well as within soft tissue components of bone. The interested reader is referred to the original work or to Johns and Cunningham (57) for details. Some practical aspects of the problem will be discussed in this section.

**B.1. Bone Mineral**

Under the conditions of electronic equilibrium, the ratio of absorbed doses in different media, for a given photon energy fluence, is given by the ratio of their energy absorption coefficients (see Chapter 8). Because the rad/R or the f factor is proportional to the energy

**TABLE 12.4. REDUCTION IN DOSE BEYOND 1 CM OF HARD BONE**

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Correction Factor (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 mm Cu HVL</td>
<td>-15$^b$</td>
</tr>
<tr>
<td>3 mm Cu HVL</td>
<td>-7</td>
</tr>
<tr>
<td>$^{60}$Co</td>
<td>-3.5</td>
</tr>
<tr>
<td>4 MV</td>
<td>-3</td>
</tr>
<tr>
<td>10 MV</td>
<td>-2</td>
</tr>
</tbody>
</table>

*Approximate values calculated with Equation 12.8 for typical clinical situations. Assumed electron density of bone relative to water = 1.55.

absorption coefficient relative to air, the ratio of \( f \) factors also reflects the relative absorbed dose. Thus, for a given quality radiation and the energy fluence, the absorbed dose in bone mineral relative to absorbed dose in muscle is the ratio:

\[
\frac{f_{\text{bone}}}{f_{\text{muscle}}} \propto \left( \frac{\mu_{\text{bone}}}{\rho_{\text{muscle}}} \right)
\]

under electronic equilibrium conditions.

Figure 12.17A shows a plot of absorbed dose as a function of depth for an orthovoltage beam incident on a composite phantom containing 2-cm thick bone. Because for this quality radiation \( f_{\text{bone}}/f_{\text{muscle}} = 1.9/0.94 = 2.0 \), the dose in the first layer of bone will be about twice as much as in soft tissue. In the subsequent layers, the dose will drop from this value due to increased attenuation by bone (Table 12.4). Figure 12.17B compares the

![Graph A](image1.png)

![Graph B](image2.png)

**FIG. 12.17.** Percentage depth dose as a function of depth in a phantom containing 2 cm of bone. A: HVL = 1 mm Cu; SSD = 50 cm; field size = 10 \times 10 \text{ cm}. B: \(^{60}\text{Co} \gamma\text{ ray beam; SSD} = 80 \text{ cm; field size} = 10 \times 10 \text{ cm.}**
TABLE 12.5. ABSORBED DOSE TO BONE RELATIVE TO SOFT TISSUE FOR DIFFERENT ENERGY BEAMS

<table>
<thead>
<tr>
<th>Radiation Quality</th>
<th>HVL* Approximate Effective Energy</th>
<th>Bone Mineral¹</th>
<th>Soft Tissue in Bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 mm Al</td>
<td>20 keV</td>
<td>4.6</td>
<td>5.0</td>
</tr>
<tr>
<td>3 mm Al</td>
<td>30 keV</td>
<td>4.8</td>
<td>5.3</td>
</tr>
<tr>
<td>1 mm Cu</td>
<td>80 keV</td>
<td>2.1</td>
<td>3.8</td>
</tr>
<tr>
<td>2 mm Cu</td>
<td>110 keV</td>
<td>1.4</td>
<td>2.4</td>
</tr>
<tr>
<td>3 mm Cu</td>
<td>135 keV</td>
<td>1.2</td>
<td>1.6</td>
</tr>
<tr>
<td>10.4 mm Pb (60Co γ rays)</td>
<td>1.25 MeV</td>
<td>0.96</td>
<td>1.03</td>
</tr>
<tr>
<td>11.8 mm Pb (4-MV x-rays)</td>
<td>1.5 MeV</td>
<td>0.96</td>
<td>1.03</td>
</tr>
<tr>
<td>14.7 mm Pb (10-MV x-rays)</td>
<td>4 MeV</td>
<td>0.98</td>
<td>1.05</td>
</tr>
<tr>
<td>13.7 mm Pb (20-MV x-rays)</td>
<td>8 MeV</td>
<td>1.02</td>
<td>1.09</td>
</tr>
<tr>
<td>12.3 mm Pb (40-MV x-rays)</td>
<td>10 MeV</td>
<td>1.04</td>
<td>1.11</td>
</tr>
</tbody>
</table>

HVL, half-value layer.
*HVL and approximate effective energies calculated using attenuation coefficients (chapter 7).

situation with ⁶⁰Co beam. Since \( f_{\text{bone}}/f_{\text{muscle}} = 0.955/0.957 = 0.96 \) for this energy, the dose to bone mineral for a ⁶⁰Co beam is slightly less than that expected in the soft tissue. Beyond the bone, the dose is reduced due to the shielding effect of bone because the electron density of bone is higher than that of the muscle tissue.

Table 12.5, column 3, gives the change in dose expected in the bone mineral for different energy beams. These calculations are made on the basis of the \( f \) factor ratios of bone to muscle or the ratio of energy absorption coefficients. For orthovoltage beams, these values represent the maximal enhancement in dose occurring just inside bone on the entrance side of the beam.

B.2. Bone-tissue Interface

Soft Tissue in Bone

The bone discussed in section B.1 is the inorganic bone (bone mineral). Of greater importance biologically, however, is the dose to soft tissue embedded in bone or adjacent to bone. The soft tissue elements in bone may include blood vessels (the Haversian canals), living cells called osteocytes and bone marrow. These structures may have very small thicknesses, ranging from a few microns to a millimeter. When the thickness of a soft tissue structure in bone is small compared with the range of the electrons traversing it, it may be considered as a Bragg-Gray cavity (see Chapter 8), containing soft tissue embedded in the bone medium. Under these conditions photon interactions in the cavity can be ignored and the ionization in the cavity is considered entirely due to electrons (photo-, Compton-, or pair-production electrons) originating from the surrounding material. The dose \( D_{STB} \) to a very small volume of soft tissue embedded in bone, assuming no perturbation of the photon or electron fluences, is given by:

\[
D_{STB} = D_{B} \cdot \left( \frac{\bar{S}}{\rho} \right)_{ST}^{B}
\]  

(12.13)

where \( D_{B} \) is the dose to the surrounding bone matrix and \( \left( \frac{\bar{S}}{\rho} \right)_{ST}^{B} \) is the ratio of average mass collision stopping power of soft tissue to bone for the electrons.

As discussed earlier in section B.1, the dose at a point in the bone mineral is related to the dose \( (D_{ST}) \) at the same point if the bone is replaced by a homogeneous medium of soft tissue:

\[
D_{B} = D_{ST} \cdot \left( \frac{\bar{\mu}_{en}}{\rho} \right)_{ST}^{B}
\]  

(12.14)
From equations 12.13 and 12.14, we get:

$$D_B = D_{ST} \cdot (\bar{\mu}_m/\rho)^B_{ST} \cdot (\bar{S}/\rho)^{ST}_{B}$$

(12.15)

The ratio $\gamma$ of dose to a soft tissue element embedded in bone to the dose in a homogeneous medium of soft tissue, for the same photon energy fluence, is given by:

$$\gamma = D_{STB} / D_{ST} = (\bar{\mu}_m/\rho)^B_{ST} \cdot (\bar{S}/\rho)^{ST}_{B}$$

(12.16)

Calculated values of $\gamma$ for different energy beams are given in column 4 of Table 12.5. These data show that for the same photon energy fluence, soft tissue structures inside the bone will receive higher dose than the dose to the bone mineral or the dose to soft tissue in the absence of bone. There are two reasons for this increase in dose: (a) $\bar{\mu}_m/\rho$ is greater for bone than soft tissue in the very low energy range because of the photoelectric process and in the very high energy range because of the pair production. However, in the Compton range of energies, $\bar{\mu}_m/\rho$ for bone is slightly less than that for soft tissue; (b) $\bar{S}/\rho$ is greater for soft tissue at all energies because it contains greater number of electrons per unit mass than the bone (Table 5.1). The combined effect of (a) and (b) gives rise to a higher dose to the soft tissue embedded in bone than the surrounding bone mineral or the homogeneous soft tissue in the absence of bone.

In a clinical situation, the dose to a small tissue cavity inside a bone may be calculated by the following equation:

$$D_{STB} = D_{ST} \cdot \gamma \cdot \text{TMR}(\sigma_T + \rho_B \cdot t_b) / \text{TMR}(\sigma_T + t_b)$$

(12.17)

where $\sigma_T$ and $t_b$ are thicknesses of soft tissue and bone, respectively, traversed by the beam before reaching the point of interest; $\rho_B$ is the relative electron density of bone; and TMR is the tissue-maximum ratio (or similar attenuation function) for the given field size.

**Soft Tissue Surrounding Bone**

On the entrance side of the photon beam, there is a dose enhancement in the soft tissue adjacent to the bone. In the megavoltage range of energies, this increase in dose is primarily due to the electron backscattering. Das and Khan (58) have shown that the magnitude of the backscatter is nearly the same for all photon energies from $^{60}$Co to 24 MV. For bone, the dose enhancement due to backscatter is approximately 8% in the above energy range. Because of the very short range of the backscattered electrons, the enhancement effect is limited only to a few millimeters (Fig. 12.18). For instance the dose enhancement drops from 8% to less than 2% within 2 mm upstream from the interface.

On the transmission side of the beam, the forward scatter of electrons from bone and the buildup of electrons in soft tissue give rise to a dose perturbation effect which depends on photon energy (59). Figure 12.19 shows this energy dependence. For energies up to 10 MV, the dose at the interface is initially less than the dose in a homogeneous soft tissue medium but then builds up to a dose that is slightly greater than that in the homogeneous case. For higher energies, there is an enhancement of dose at the interface because of the increased electron fluence in bone due to pair production. The effect decreases with distance and lasts up to the range of the electrons.

Most patients are treated with parallel-opposed beams. Also, dose distributions are normally not corrected for the presence of bone when using megavoltage photon beams. The following discussion analyzes the bone dosage problem in a practical clinical situation.

Figure 12.20 shows examples of depth dose distributions expected in a patient treated with parallel-opposed beams. Doses are normalized to the midpoint dose expected in a homogeneous soft tissue medium. The distribution corrected for increased bone attenuation (shielding effect) alone shows dose reduction throughout. The magnitude of this reduction depends on bone thickness relative to the soft tissue thickness, bone density and beam energy. The actual distribution in the presence of bone includes both bone attenuation and bone-tissue interface effects discussed earlier. These effects in the megavoltage range of
II. Classical Radiation Therapy

FIG. 12.18. Backscatter dose factor (BSDF) for various energy photon beams plotted as a function of distance, toward the source, from the bone-polystyrene interface. BSDF is the ratio of dose at the interface with bone to that without bone. (From Das IJ, Khan FM. Backscatter dose perturbation at high atomic number interfaces in megavoltage photon beams. Med Phys 1989;16:367, with permission.)

FIG. 12.19. Forward dose perturbation factor (FDPF) for various energy photon beams plotted as a function of distance, away from the source, from bone-polystyrene interface. FDPF is the ratio of dose at the interface with bone to that without bone for the same photon energy fluence. (From Das IJ. Study of dose perturbation at bone-tissue interfaces in megavoltage photon beam therapy. [Dissertation.] University of Minnesota, 1988:119, with permission.)
energies cause an increase in dose to soft tissue adjacent to bone, but the net increase is not significant at lower energies (≤10 MV). However, as the pair production process becomes significant at higher energies and the electron range increases, appreciable enhancement in dose occurs at the bone tissue interfaces. This is seen in Fig. 12.20 and Table 12.6.

**B.3. Lung Tissue**

Dose within the lung tissue is primarily governed by its density. As discussed in section 12.5A, lower lung density gives rise to higher dose within and beyond the lung. Figure 12.21
## Table 12.6. Dose Enhancement at Bone-Tissue Interface for Parallel-Opposed Beams

<table>
<thead>
<tr>
<th>Thickness of Bone (cm)</th>
<th>6 MV</th>
<th>10 MV</th>
<th>18 MV</th>
<th>24 MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>1.01</td>
<td>1.02</td>
<td>1.03</td>
<td>1.04</td>
</tr>
<tr>
<td>1.0</td>
<td>1.01</td>
<td>1.02</td>
<td>1.03</td>
<td>1.05</td>
</tr>
<tr>
<td>2.0</td>
<td>1.00</td>
<td>1.01</td>
<td>1.03</td>
<td>1.05</td>
</tr>
<tr>
<td>3.0</td>
<td>0.99</td>
<td>1.00</td>
<td>1.03</td>
<td>1.05</td>
</tr>
</tbody>
</table>

*Dose to soft tissue adjacent to bone relative to midpoint dose in a homogeneous soft tissue; total thickness = 20 cm; field size = 10 \times 10 \text{ cm}; SSD = 100 \text{ cm}.

From Das IJ, Khan FM, Kase KR. Dose perturbation at high atomic number interfaces in parallel opposed megavoltage photon beam irradiation [abstract]. Phys Med Biol 1988;33[Suppl 1]:121, with permission.

This gives the increase in lung dose as a function of depth in the lung for selected energies using a 10 \times 10 \text{ cm} field. But in the first layers of soft tissue beyond a large thickness of lung, there is some loss of secondary electrons (60). This gives rise to a slight decrease in dose relative to that calculated on the basis of lung transmission.

Kornelson and Young (61) have discussed the problem of loss of lateral electronic equilibrium when a high-energy photon beam traverses the lung. Because of the lower density of lung, an increasing number of electrons travel outside the geometrical limits of the beam. This causes the dose profile to become less sharp. For the same reason there is a greater loss of laterally scattered electrons, causing a reduction in dose on the beam axis. The effect is significant for small field sizes (<6 \times 6 \text{ cm}) and higher energies (>6 \text{ MV}). Clinically, when treating a tumor in the lung there is a possibility of underdosage in the periphery of the tumor if small fields and high-energy beams are used. However, considering the fact that most protocols in this country require no lung correction in dose prescription, consideration of this effect in dosimetry becomes rather academic.

### B.4. Air Cavity

The most important effect of air cavities in megavoltage beam dosimetry is the partial loss of electronic equilibrium at the cavity surface. The actual dose to tissue beyond and in front of the cavity may be appreciably lower than expected. This phenomenon of dose buildup at the air cavities has been extensively studied by Epp et al. (62,63). The most significant decrease in dose occurs at the surface beyond the cavity, for large cavities (4 cm deep) and the smallest field (4 \times 4 \text{ cm}). Epp et al. (62) have estimated that in the case of $^{60}$Co the reduction in dose in practical cases, such as the lesions located in the upper

![FIG. 12.21. Percentage increase in lung dose as a function of depth in the lung for selected energies. Field size = 10 \times 10 \text{ cm}. (From McDonald SC, Keller BE, Rubin P. Method for calculating dose when lung tissue lies in the treatment field. Med Phys 1976;3:210, with permission.]

**% Increase in Lung Dose**

<table>
<thead>
<tr>
<th>Depth in Lung (cm)</th>
<th>0</th>
<th>4</th>
<th>8</th>
<th>12</th>
<th>16</th>
</tr>
</thead>
<tbody>
<tr>
<td>4 MV</td>
<td>0</td>
<td>10</td>
<td>20</td>
<td>30</td>
<td>40</td>
</tr>
<tr>
<td>6 MV</td>
<td>0</td>
<td>15</td>
<td>30</td>
<td>45</td>
<td>60</td>
</tr>
<tr>
<td>10 MV</td>
<td>0</td>
<td>20</td>
<td>40</td>
<td>60</td>
<td>80</td>
</tr>
</tbody>
</table>
12. Treatment Planning II: Patient Data, Corrections, and Setup

12.6. TISSUE COMPENSATION

A radiation beam incident on an irregular or sloping surface produces skewing of the isodose curves. Corrections for this effect were discussed in section 12.2. In certain treatment situations, however, the surface irregularity gives rise to unacceptable nonuniformity of dose within the target volume or causes excessive irradiation of sensitive structures such as the spinal cord. Many techniques have been devised to overcome this problem, including the use of wedged fields or multiple fields and the addition of bolus material or compensators. Areas having a smaller thickness of tissue can also be blocked for the last few treatments to reduce the dose in these areas.

Bolus is a tissue-equivalent material placed directly on the skin surface to even out the irregular contours of a patient to present a flat surface normal to the beam. This use of bolus should be distinguished from that of a bolus layer, which is thick enough to provide adequate dose buildup over the skin surface. The latter should be termed the build-up bolus.

Placing bolus directly on the skin surface is satisfactory for orthovoltage radiation, but for higher-energy beams results in the loss of the skin sparing advantage. For such radiations, a compensating filter should be used, which approximates the effect of the bolus as well as preserves the skin-sparing effect. To preserve the skin-sparing properties of the megavoltage photon beams, the compensator is placed a suitable distance (≥20 cm) away from the patient's skin. Yet the compensator is so designed that its introduction in the beam gives rise to isodose curves within the patient that duplicate, as closely as possible, those for the bolus.

A. Design of Compensators

Figure 12.22 illustrates schematically the use of a compensator to provide the required beam attenuation that would otherwise occur in the "missing" tissue when the body surface is

![FIG. 12.22. Schematic representation of a compensator designed for an irregular surface. (From Khan FM, Moore VC, Burns DJ. The construction of compensators for cobalt teletherapy. Radiology 1970;96:187, with permission.)](image)
irregular or curved. Because the compensator is designed to be positioned at a distance from the surface, the dimensions and shape of the compensator must be adjusted because of (a) the beam divergence, (b) the relative linear attenuation coefficients of the filter material and soft tissues, and (c) the reduction in scatter at various depths when the compensator is placed at a distance from the skin rather than in contact with it. To compensate for this scatter, the compensator is designed such that the attenuation of the filter is less than that required for primary radiation only. These considerations and others have been discussed in the literature (64–70).

Minification of the compensating material for geometric divergence of the beam has been achieved in many ways. One method (64,66–68) constructs the compensator out of aluminum or brass blocks, using a matrix of square columns corresponding to the irregular surface. The dimension of each column is minified according to the geometric divergence correction, which is calculated from the SSD and the filter-surface distance. Khan et al. (71) described an apparatus that uses thin rods duplicating the diverging rays of the therapy beam (Fig. 12.23). The rods move freely in rigid shafts along the diverging paths and can be locked or released by a locking device. The apparatus is positioned over the patient so that the lower ends of the rods touch the skin surface. When the rods are locked, the upper ends of the rods generate a surface that is similar to the skin surface but corrected for divergence. A plastic compensator can then be built over this surface (69). Beck et al. (72) and Boge et al. (73) have described Styrofoam cutters (Fig. 12.24) that work on a pantographic principle and use a heating element or a routing tool mechanism for the hollowing of the Styrofoam. The cavity thus produced is a minified version of the patient surface, which can be filled with the compensator material.

FIG. 12.23. An apparatus for the construction of 3-D compensator in one piece. (From Khan FM, Moore VC, Burns DJ. An apparatus for the construction of irregular surface compensators for use in radiotherapy. Radiology 1968;90:593, with permission.)
A tissue equivalent compensator designed with the same thickness as that of the missing tissue will overcompensate, i.e., the dose to the underlying tissues will be less than that indicated by the standard isodose chart. This decrease in depth dose, which is due to the reduction in scatter reaching a point at depth, depends on the distance of the compensator from the patient, field size, depth, and beam quality. To compensate for this decrease in scatter, one may reduce the thickness of the compensator to increase the primary beam transmission. The compensator thickness should be such that the dose at a given depth is the same whether the missing tissue is replaced with the bolus in contact or with the compensator at the given distance from the skin surface. The required thickness of a tissue-equivalent compensator along a ray divided by the missing tissue thickness along the same ray may be called the density ratio or thickness ratio (69) ($h'/h$ in Fig. 12.22). Figure 12.25
II. Classical Radiation Therapy

gives a plot of thickness ratio, \( r \), as a function of compensator-surface distance, \( d \). \( r \) is unity at the surface and decreases as \( d \) increases.

Thickness ratio depends, in a complex way, on compensator-surface distance, thickness of missing tissue, field size, depth, and beam quality. However, a detailed study of this parameter has shown that \( r \) is primarily a function of \( d \) (for \( d \leq 20 \text{ cm} \)) and that its dependence on other parameters is relatively less critical (69,74). Thus a fixed value of \( r \), based on a given \( d \) (usually 20 cm), \( 10 \times 10 \text{ cm} \) field, 7-cm depth, and a tissue deficit of 5 cm can be used for most compensator work.

The concept of thickness ratios also reveals that a compensator cannot be designed to provide absorbed dose compensation exactly at all depths. If, for given irradiation conditions, \( r \) is chosen for a certain compensation depth, the compensator overcompensates at shallower depths and undercompensates at greater depths. Considering the limitations of the theory and too many variables affecting \( r \), we have found that an average value of 0.7 for \( r \) may be used for all irradiation conditions provided \( d \geq 20 \text{ cm} \). The same value has been tested to yield satisfactory results (errors in depth dose within \( \pm 5\% \)) for \(^{60}\text{Co} \), 4-MV and 10-MV x-rays (74).

In the actual design of the compensator, the thickness ratio is used to calculate compensator thickness \( (t_c) \) at a given point in the field:

\[
t_c = \frac{TD}{r} \times (t/\rho_c)
\]

where TD is the tissue deficit at the point considered and \( \rho_c \) is the density of the compensator material.

A direct determination of thickness \( (t/\rho_c) \) for a compensator system may be made by measuring dose at an appropriate depth and field size in a tissue equivalent phantom (e.g., polystyrene) with a slab of compensator material placed in the beam at the position of the compensator tray. Pieces of phantom material are removed from the surface until the dose equals that measured in the intact phantom, without the compensator. The ratio of compensator thickness to the tissue deficit gives the thickness ratio.

It may be mentioned that a term compensator ratio (CR) has also been used in the literature to relate tissue deficit to the required compensator thickness (75). It is defined as the ratio of the missing tissue thickness to the compensator thickness necessary to give the dose for a particular field size and depth. The concepts of compensator ratio and the thickness ratio are the same, except that the two quantities are inverse of each other, i.e.,

\[
\text{CR} = \frac{TD}{t_c} = \frac{\rho_c}{r}
\]

B. Two-dimensional Compensators

Designing a 3-D compensator is a time-consuming procedure. In a well-equipped mold or machine shop, a trained technician can probably construct such compensators routinely with a reasonable expenditure of time. In the absence of such facilities and personnel, however, most situations requiring compensation can be handled satisfactorily with simple 2-D compensators. In many treatment situations, the contour varies significantly in only one direction: along the field width or length. In such cases, a compensator can be constructed in which the thickness varies only along this dimension. For example, if anterior and posterior fields are incident on a sloping mediastinum, compensation is usually not changed in the lateral direction but only in the craniocaudal direction.

One simple way of constructing a two-dimensional compensator is to use thin sheets of lead (with known thickness ratio or effective attenuation coefficient) and gluing them together in a stepwise fashion to form a laminated filter. The total thickness of the filter at any point is calculated to compensate for the air gap at the point below it. Another method, used routinely at the University of Minnesota, is to construct the compensator in one piece from a block of Lucite. The patient contour is taken showing body thickness at at least three reference points: central axis, inferior margin, and superior margin of the field. Tissue deficits, \( \Delta t \), are calculated by subtracting thicknesses at the reference points from the maximum thickness. A thickness minification factor is calculated by dividing the
thickness ratio \( r \) by the electron density \((\text{e}^{-}/\text{per cm}^3)\) of Lucite relative to that of tissue. The geometric minification factor is calculated by \((f - d)/f\) where \(f\) is the SSD at the point of maximum thickness and \(d\) is the filter-surface distance. The compensator dimensions can now be drawn by multiplying the \(A_t\) values with the thickness minification factor and the spacing between the reference points with the geometric minification factor. A Lucite block is then machined and glued on a thin Lucite plate for placement in the beam. The same method may be used to construct a compensator by stacking Lucite plates in a stepwise fashion and attaching them together firmly with pieces of Scotch tape.

C. Three-dimensional Compensators

Early 3-D compensator systems were mechanical devices to measure tissue deficits within the field in both the transverse and the longitudinal body cross-sections. Examples of these systems include Ellis type filters (64,66), rod boxes (68,69) and pantographic devices (72,73). More recent devices include Moiré camera, 3-D magnetic digitizers, CT-based compensator programs and electronic compensation using Multileaf collimators (Chapter 20).

C.1. Moiré Camera

A specially designed camera system allows topographic mapping of the patient body surface and provides tissue deficit data necessary for the design of a 3-D compensator. The principle of operation of the camera has been discussed by Boyer and Goitein (76). The camera can be mounted on a simulator without interfering with the simulator's normal use. Moiré fringes observed on the patient's surface represent iso-SSD lines from which the tissue deficit data can be deduced. The data can be used to drive a pantographic cutting unit. A commercial version of this system is manufactured by DCD, S&S PAR Scientific (Brooklyn, NY).

C.2. Magnetic Digitizer

A handheld stylus containing a magnetic field sensor is used to digitize the position of the sensor as it is scanned over the patient's surface in the presence of a low-strength, low-frequency magnetic field. Tissue deficit data are calculated by the computer from the sensor coordinates and used to drive a Styrofoam cutter. Cavities corresponding to the tissue deficit are then filled with an appropriate compensator material to design a compensator. A commercially available system, known as Compuformer is manufactured by Huestis Corporation (Bristol, RI).

C.3. Computed Tomography-based Compensator Systems

Three-dimensional radiotherapy treatment planning systems that use multilevel CT scans have sufficient data available to provide compensation not only for the irregular surface contours but also for the tissue inhomogeneities. There are two commercial systems that provide software for the design of compensating filters: the Target (G.E. Medical Systems, Milwaukee, WI) and the TheracomHEK (Theratronics Ltd., Ontario, Canada). These systems extract the tissue deficit data from the CT scans, which are then used to cut the Styrofoam mold using a drill bit or a heated wire loop. Although any compensator material of known compensator ratio may be cast into the filter molds, it is desirable to use medium-density materials rather than heavier materials such as Cerrobend. The main reason for this is to minimize error in dose distribution when small errors are made in cutting the mold.

There are several other compensator systems that have not been discussed here. For a detailed review of this topic the reader is referred to Reinstein (77).
D. Compensating Wedges

For oblique beam incidence or curved surfaces for which the contour can be approximated with a straight line, standard compensating wedges are very convenient (69,70). Compensating wedges (C-wedges) are fabricated from a metal such as copper, brass, or lead. They are designed to compensate for a "missing" wedge of tissue, using the same design principles as discussed in section 12.4B.

Distinction needs to be made between a wedge filter and a compensating wedge. Although a wedge filter can be used effectively as a compensator, it is primarily designed to tilt the standard isodose curves through a certain wedge angle in conjunction with the wedge-pair technique (Chapter 11). The wedge filter isodose curves must be available and used to obtain the composite isodose curves before the filter is used in a treatment set-up. The C-wedge, on the other hand, is used just as a compensator, so that the standard isodose charts can be used without modification. In addition, no wedge transmission factors are required for the C-wedges.

An important advantage of C-wedges over wedge filters used as compensators is that the C-wedges can be used for partial field compensation, that is, the C-wedge is used to compensate only a part of the contour, which is irregular in shape. A wedge filter, in this case, could not be used as a compensator because it is designed to be placed in the field in a fixed position.

E. Other Applications

Compensating filters can be designed to compensate for tissue heterogeneity. Most of this work was done by Ellis and his coworkers (33) in which compensators were designed from the knowledge of cross-sectional anatomy using transaxial tomography or a photographic film. More recently, Khan et al. (78) have described compensators for total body irradiation including compensation for lungs.

Compensators have also been used to improve dose uniformity in the fields where nonuniformity of the dose distribution arises from sources other than contour irregularity: reduced scatter near the field edges and unacceptable high dose regions or "horns" in the beam profile. Leung et al. (79) have discussed the design of filters for the mantle technique in which the compensator is designed on the basis of calculated dose distribution in the absence of a compensator. Boge et al. (80) have described a special compensator filter to reduce the horns present in large fields of a 4-MV linear accelerator.

F. Compensator Set-up

As mentioned earlier, the compensator should be placed at a distance of 20 cm or more away from the skin surface to preserve the skin-sparing properties of the megavoltage beams. Because the dimensions of the compensator are reduced (compared to the bolus) in the plane perpendicular to the beam axis to allow for beam divergence, the filter must be placed at the filter-surface distance for which it is designed. In addition, the nominal SSD should be measured from the plane perpendicular to the beam axis, containing the most elevated point on the contour included in the field (Fig. 12.22). For isocentric treatments, it is most convenient to use field dimensions projected at the isocenter in compensator design. Accordingly, the depth of the isocenter is measured from the level of the most elevated point on the contour to be compensated.

12.7. PATIENT POSITIONING

Availability of isocentric treatment machines, simulators, CT scanners, and computers has made it possible to achieve a high degree of precision in radiation therapy. However, one of the weakest links in the treatment-planning process is the problem of patient positioning
and immobilization. It is frequently observed that some of the treatment techniques in current practice are outdated or do not take advantage of the accuracy available with the modern equipment. For example, the patients are treated in less than a stable position, are moved between different fields, and set up primarily by marks inked or tattooed on the skin surface. But, as any experienced observer knows, such practices are prone to serious errors. Skin marks are vulnerable to variation in skin sag and body position on the treatment table.

The problem of precise patient positioning and immobilization has been addressed by a number of investigators (81–86), including a recent review by Reinstein (87). But this problem still remains the area of greatest variance in actual treatment. The following ideas are presented to focus attention on this important area and offer some guidelines for precise patient positioning.

A. General Guidelines

1. Treatments should be set up isocentrically. The principal advantage of isocentric technique over SSD technique is that the patient is not moved between fields. Once the isocenter is positioned accurately within the patient, the remaining fields are arranged simply by gantry rotation or couch movement, not by displacing the patient relative to the couch.

2. Isocenter position within the patient can be established using the treatment simulator. This is usually accomplished by anterior and lateral radiographs, using the radiographically visible structures to define the target volume.

3. To accurately define the patient’s position, thick pads or mattresses should not be used on the simulator table or the treatment table. This is essential for accurate measurement of set-up parameters as well as reproducibility.

4. For head and neck treatments, flexible head rests, such as pillows or sponges, should be avoided. The head should rest on a rigid surface such as a block of hard Styrofoam or a plastic “head-neck” support (Fig. 12.26).

5. Many methods of head immobilization are available such as partial body casts (85), bite block system (88), nose bridges, head clamps, or simple masking tape. Choice of any of the above will depend on the location of the treatment fields.

6. As far as possible, the patient should be treated in the supine position. An overhead

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3 Such devices are commercially available, for example, from Radiation Products Design, Inc., Buffalo, Minnesota and Med-Tec, Orange City, Iowa.
II. Classical Radiation Therapy

Sagittal laser line is useful in aligning the sagittal axis of the patient with the axis of gantry rotation.

7. For head and neck treatments, the chin extension should be defined anatomically, for example, distance between the sternal notch and the chin bone. This measurement should be accurately made after the head position has been established on the basis of stability and field localization.

8. During simulation as well as treatment, the depth of isocenter should be defined by either the set-up SSD (usually measured anteriorly or posteriorly) or by setting the distance between the tabletop distance and lateral beam axis. Side laser lights may also be used for this purpose. In the latter case, the laser lights should be checked frequently for alignment accuracy, because these lights are known to drift presumably by expansion and contraction of the walls on which they are mounted.

9. Skin marks should not be relied on for daily localization of the treatment field. The field boundaries should be defined relative to the bony landmarks established during simulation. Do not force the field to fit the skin marks!

10. For lateral portals, the Mylar section of the couch or tennis racket should be removed and the patient placed on a solid surface to avoid sag. These should be used only for anteroposterior (AP) treatments for which skin sparing is to be achieved. For example, if the four-field pelvis technique is used, one can use two fields a day in which case AP treatments are given isocentrically on a Mylar window using anterior or posterior set-up SSD, and lateral fields are treated on a flat tabletop section using the tabletop distance to lateral beam axis. Or if four fields are treated the same day, the posterior field can be treated through the rigid Plexiglas section of the couch instead of the Mylar window. Or AP treatments can be given on the Mylar window and then the window can be replaced by the Plexiglas section for the lateral treatments. The last alternative involves two separate set-ups, one for the AP and the other for the lateral fields. It should be used only when skin dose from the posterior fields is to be reduced to a minimum.

11. For isocentric techniques, field sizes must be defined at the isocenter which will, in most cases, be at the center of the treatment volume and not on the skin surface. Physicians who are accustomed to using standard field sizes (e.g., pelvic fields) defined at the skin surface should make adjustments in field sizes so that the fields encompass the same irradiated volume.

Some institutions have developed elaborate casting techniques to immobilize patients during treatments. This requires a well-equipped mold room as well as personnel trained in mold technology. Some of these techniques have been shown to be quite effective in minimizing patient motion (85,87). However, patients are known to move within a cast especially if the fit is not good or if there is a change in the surfaces contour due to regression of the tumor or weight loss.

Detection of patient motion is possible by using small dots of reflective tape on the patient with a pencil light ray and photocell device. Laser localization lights can also be used for this purpose. The signal received from the photocell can be further processed to activate an interlock to interrupt treatment or sound an alarm if pertinent motion exceeds a preset limit. Thus a good motion detection system can complement patient positioning and immobilization techniques by monitoring the stability of patient position as well as the effectiveness of immobilization.

B. The XYZ Method of Isocenter Set-up

In the isocentric technique, the isocenter is placed inside the patient, usually at the center of the target volume. Once this point has been localized by simulation, a good treatment set-up should reproduce it quickly and accurately. The following steps outline a procedure, hereby called the XYZ method, for the localization of this important point.
B.1. Simulation Procedure

1. The patient is positioned on the simulator couch following the general guidelines discussed in section 12.7A.
2. The patient is leveled using the side lasers (or a bubble level) and the sagittal laser beam to define the sagittal axis of the patient. The patient is then constrained from movement by a suitable immobilization device. For head and neck positioning, chin extension (distance between chin bone and the sternal notch) should be accurately measured.
3. The treatment fields are simulated using anterior and lateral radiographs and the isocenter is established according to the treatment plan.
4. A reference anatomic point is chosen on the sagittal axis, somewhere in the neighborhood of the treatment area, to represent a stable anatomic landmark. For example, nasion for head and neck, sternal notch for neck and thorax, tip of xiphoid for thorax and abdomen, and bottom of pubic ramus or tip of coccyx for pelvis, can be chosen as reasonably stable reference points.
5. The coordinates of the treatment isocenter are represented by \((X, Y, Z)\) where \(X\) is the lateral distance and \(Y\) is the longitudinal distance (along patient axis) of the isocenter from the reference point and \(Z\) is the tabletop to isocenter distance (Fig. 12.27). Beam angle \(\theta\) is recorded.

B.2. Treatment Set-up

1. Position and level the patient on the treatment couch as in simulation.
2. With the gantry vertical, place the central axis at the reference anatomic point and mark it with ink.
3. Move the couch: up or down to obtain \(Z\) using the side laser; laterally through \(X\) and

![Diagram of X, Y, Z coordinates to a patient set-up.](image-url)
longitudinally through distance Y. Rotate the gantry through angle $\theta$. This gives the required central axis of the field and the isocenter location.

4. Make secondary checks according to the field diagram such as SSD, location of field borders, etc.

5. For isocentric set-up, other fields are positioned by simply rotating the gantry and positioning it at predetermined angles.

One potential advantage of this method is that the set-up parameters X, Y, Z, and $\theta$ could be computer controlled, thereby decreasing the set-up time and minimizing human errors. The therapist, in this case, will position the patient as usual and place the central axis vertically at the reference point. Then, with a switch on the hand pendant, the computer control could be initiated to move the couch and the gantry to the X, Y, Z, and $\theta$ coordinates. Such a method could be adopted by some of the existing treatment monitoring systems that are capable of moving the couch and the gantry.

Even manually, the XYZ method can greatly economize set-up time as well as enhance set-up precision. Most modern couches are motor driven and equipped with motion-sensing devices. Videographic display of the couch motions could be conveniently used to position the couch. A reset switch for the X, Y, and Z coordinates would make it easier to move the couch through the X, Y, and Z distances.

REFERENCES

12. Treatment Planning II: Patient Data, Corrections, and Set-up

Shielding of vital organs within a radiation field is one of the major concerns of radiation therapy. Considerable time and effort are spent in shaping fields not only to protect critical organs but also to avoid unnecessary irradiation of the surrounding normal tissue. Attendant to this problem is its effect on skin dose and the buildup of dose in the subcutaneous tissue. Skin sparing is an important property of megavoltage photon beams, and every effort should be directed to maintaining this effect when irradiating normal skin.

Another problem frequently encountered in radiation therapy is the matching of adjacent fields. This situation arises when radiation fields available with the equipment are not large enough to encompass the entire target volume. In some cases, the target volume is divided into two parts so treatment to the second part does not commence until the treatment course to the first part has been completed. Such a scheme is designed to avoid toxicity due to irradiating an excessive volume of tissue. Multiple adjacent fields are also used when tumor distribution or patient anatomy does not allow coplanar fields (fields with central axes in the same plane). The main problem with these techniques is the possibility of extreme dose inhomogeneity in the junctional region. Because radiation beams are divergent, adjacent fields can overlap at depth and give rise to regions of excessive dose or hot spots. Overlaps can be avoided by separating the fields, but this in turn can give rise to areas of reduced dose or "cold spots."

This chapter on treatment planning focuses on the above problems and discusses their possible solutions.

13.1. FIELD BLOCKS

The shaping of treatment fields is primarily dictated by tumor distribution—local extensions as well as regional metastases. Not only should the dose to vital organs not exceed their tolerance but the dose to normal tissue, in general, should be minimized. As long as the target volume includes, with adequate margins, the demonstrated tumor as well as its presumed occult spread, significant irradiation of the normal tissue outside this volume must be avoided as much as possible. These restrictions can give rise to complex field shapes, which require intricate blocking.

The frequency and complexity of field shaping vary from institution to institution. However, if complex techniques involving elaborate blocking are used often, it is necessary to establish a rational system of field shaping.

A. Block Thickness

Shielding blocks are most commonly made of lead. The thickness of lead required to provide adequate protection of the shielded areas depends on the beam quality and the allowed transmission through the block. A primary beam transmission of 9% through the
block is considered acceptable for most clinical situations. If \( n \) is the number of half-value layers to achieve this transmission,

\[
\frac{1}{2^n} = 0.05
\]

or

\[
2^n = \frac{1}{0.05} = 20
\]

or

\[
n \log 2 = \log 20
\]

or

\[
n = \frac{\log 20}{\log 2} = 4.32
\]

Thus a thickness of lead between 4.5 and 5.0 half-value layers would give less than 5% primary beam transmission and is, therefore, recommended for most clinical shielding.

Shielding against primary radiation for superficial and orthovoltage beams is readily accomplished by thin sheets of lead that can be placed or molded on to the skin surface. However, as the beam energy increases to the megavoltage range, the thickness of lead required for shielding increases substantially. The lead blocks are then placed above the patient supported in the beam on a transparent plastic tray, called the shadow tray. Table 13.1 gives the recommended lead shield thicknesses for various quality beams.

Although the primary beam transmission can be reduced further by using extra thick blocks, the reduction in dose in the shielded region may not be that significant due to the predominance of scattered radiation from the adjoining open areas of the field.

### B. Block Divergence

Ideally, the blocks should be shaped or tapered so that their sides follow the geometric divergence of the beam. This minimizes the block transmission penumbra (partial transmission of the beam at the edges of the block). However, divergent blocks offer little advantage for beams with large geometric penumbra. For example, in the case of \(^{60}\)Co, the sharpness of the beam cutoff at the block edge is not significantly improved by using divergent blocks. Also, for some clinical situations this sharpness is not critical or worth the time required for making divergent blocks, which have to be invariably custom designed for a given treatment set-up. Therefore, most institutions keep a stock of straight-cut blocks of various shapes and dimensions.

#### TABLE 13.1. RECOMMENDED MINIMUM THICKNESS OF LEAD FOR SHIELDING\(^a\)

<table>
<thead>
<tr>
<th>Beam Quality</th>
<th>Required Lead Thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0 mm Al HVL</td>
<td>0.2 mm</td>
</tr>
<tr>
<td>2.0 mm Al HVL</td>
<td>0.3 mm</td>
</tr>
<tr>
<td>3.0 mm Al HVL</td>
<td>0.4 mm</td>
</tr>
<tr>
<td>1.0 mm Cu HVL</td>
<td>1.0 mm</td>
</tr>
<tr>
<td>3.0 mm Cu HVL</td>
<td>2.0 mm</td>
</tr>
<tr>
<td>4.0 mm Cu HVL</td>
<td>2.5 mm</td>
</tr>
<tr>
<td>(^{137})Cs</td>
<td>3.0 cm</td>
</tr>
<tr>
<td>(^{60})Co</td>
<td>5.0 cm</td>
</tr>
<tr>
<td>4 MV</td>
<td>6.0 cm</td>
</tr>
<tr>
<td>6 MV</td>
<td>6.5 cm</td>
</tr>
<tr>
<td>10 MV</td>
<td>7.0 cm</td>
</tr>
<tr>
<td>25 MV</td>
<td>7.0 cm</td>
</tr>
</tbody>
</table>

HVL, half-value layer.

\(^a\)Approximate values to give ≤5% primary transmission.
Divergent blocks are most suited for beams having small focal spots. Because the sides of these blocks follow beam divergence, one can reduce the lateral dimensions by designing the shields for smaller source-to-block distances without increasing the block transmission penumbra.

13.2. FIELD SHAPING
A. Custom Blocking

Although a number of systems have been used for field shaping (1–8), the one introduced by Powers et al. (1) is most commonly used in radiation therapy. This system uses a low melting point alloy, Lipowitz metal (brand name, Cerrobend), which has a density of 9.4 g/cm³ at 20°C (~83% of lead density). This material consists of 50.0% bismuth, 26.7% lead, 13.3% tin, and 10.0% cadmium (1). The main advantage of Cerrobend over lead is that it melts at about 70°C (compared with 327°C for lead) and, therefore, can be easily cast into any shape. At room temperature, it is harder than lead.

The minimum thickness of Cerrobend blocks required for blocking may be calculated from Table 13.1 using its density ratio relative to lead (e.g., multiply lead thickness by 1.21). In the megavoltage range of photon beams, the most commonly used thickness is 7.5 cm, which is equivalent to about 6 cm of pure lead.

The procedure for constructing Cerrobend blocks starts with a simulator radiograph or a port film on which the radiotherapist draws the outline of the treatment field indicating areas to be shielded. The film is then used to construct divergent cavities in a Styrofoam block that are used to cast Cerrobend blocks. Figure 13.1 shows a Styrofoam-cutting device that consists of an electrically heated wire which pivots about a point simulating the source or the x-ray target. The film, the Styrofoam block, and the wire apparatus are so adjusted

![FIG. 13.1. Photograph of block cutter. (Courtesy of Huestis Machine Corp., Bristol, RI.)](image-url)
that the actual treatment geometry (same source to film and source to block distances) is obtained. The lower end of the wire traces the outline on the film.

If "positive" blocks such as lung blocks are to be made, cavities are cut in the Styrofoam with the heated segment of the wire and subsequently filled with melted Cerrobend. If a "negative" block with central area open and peripheral areas blocked is desired, an inner cut is first made to outline the field opening. An outer rectangular cut is then made to define the collimator field with 1- to 2-cm margin. The three Styrofoam pieces thus made are placed on a Lucite plate and carefully aligned relative to the central axis. The intermediate piece, corresponding to the areas to be shielded, is then removed and Cerrobend is poured into the cavity.

It is important that the Cerrobend is poured slowly to prevent formation of air bubbles. Also, the Styrofoam block should be pressed tightly against a rubber pad at the bottom to avoid leakage of the liquid metal. The inside walls of the cavity may be sprayed with silicone for easy release of the Styrofoam pieces from the block.

The blocks can be mounted on a Lucite plate or blocking tray, which is premarked with the central axis cross hairs. Blocks can also be placed on a template made on a clear film by tracing the outline of the field at the shadow tray position while the port film outline is placed at the distance at which the radiograph was taken.

Figure 13.2 shows examples of Cerrobend blocks, one constructed for shielding lungs and the other for a head and neck field.

B. Independent Jaws

Asymmetric fields are sometimes used to block off a part of the field without changing the position of the isocenter. Although blocking is often used to generate irregular field shapes, rectangular blocking can be easily done by independently movable collimators, or
jaws. This feature is very convenient when matching fields or beam splitting. In the latter case, the beam is blocked off at the central axis to remove divergence. Whereas half-beam blocks have been used as beam splitters in the past, this can now be done simply by moving in the independent jaws.

Most modern machines are equipped with independently movable jaws. Some machines have one independent jaw, others have two independent pairs, and some have all four jaws as independent. Operationally, the independent jaw option is interlocked to avoid errors in the setting of symmetric fields, in which case the opposite jaws open or close symmetrically.

One of the effects of asymmetric collimation is the change in the physical penumbra (defined in section 11.1) and the tilt of the isodose curves toward the blocked edge (Fig. 13.3). This effect is simply the result of blocking, which eliminates photon and electron scatter from the blocked portion of the field, thereby reducing the dose near the edge. The same effect would occur on the isodose curves if the blocking were done with a lead or Cerrobend block on a tray.

FIG. 13.3. Comparison of isodose distribution with half the beam blocked by an independent jaw versus a block on a tray. Notice close agreement as well as the tilt of the isodose curves toward the blocked edge.
When asymmetric fields are used, special considerations must be given to the beam flatness and the dosimetric parameters used to calculate monitor units. Khan et al. (9) have proposed a system of dose calculation for fields generated by asymmetric collimators, which was discussed in Chapter 10.

C. Multileaf Collimators

A multileaf collimator (MLC) for photon beams consists of a large number of collimating blocks or leaves that can be driven automatically, independent of each other, to generate a field of any shape (Fig. 13.4). Typical MLC systems consist of 80 leaves (40 pairs) or
more. The individual leaf has a width of 1 cm or less as projected at the isocenter. The leaves are made of tungsten alloy ($\rho = 17.0$ to 18.5 g/cm$^3$) and have thickness along the beam direction ranging from 6 cm to 7.5 cm, depending on the type of accelerator. The leaf thickness is sufficient to provide primary x-ray transmission through the leaves of less than 2% (compared with about 1% for jaws and 3.5% for Cerrobend blocks). The interleaf (between sides) transmission is usually less than 3%. The primary beam transmission may be further minimized by combining jaws with the MLC in shielding areas outside the MLC field opening.

Some MLC systems have double-focused leaves, that is, the leaves form a cone of irregular cross-section diverging from the source position and move on a spherical shell centered at the source. The rationale behind a double-focused MLC is to provide a sharp beam cutoff at the edge. However, for high-energy beams this objective is achieved only to a limited extent, because the dose falloff at the edge is largely determined by scattered photons and electrons. Because double-focused MLCs are difficult to manufacture, some systems have been designed with rounded leaf edges and directions of travel perpendicular to the central ray. The purpose of rounded edges is to provide constant beam transmission through a leaf edge, regardless of its position in the field.

An important consideration in the use of MLCs for stationary fields is the conformity between the planned field boundary, which is continuous, and the jagged stepwise boundary created by the MLC. The degree of conformity between the two depends not only on the projected leaf width but also on the shape of the target volume and the angle of rotation of the collimator. Optimization of MLC rotation and setting has been discussed by Brahme (10). His analysis shows that the best orientation of the collimator is when the direction of motion of the leaves is parallel with the direction in which the target volume has the smallest cross-section.

The physical penumbra (section 11.1) with MLC is larger than that produced by the collimator jaws or the Cerrobend blocks (Fig. 13.5). This is usually not a serious drawback except for the treatment of small fields or when blocking is required close to critical structures. Also, jaggedness of the field edges makes it difficult to match adjacent fields.

The use of MLC in blocking and field shaping is ideally suited for treatments requiring large numbers of multiple fields because of automation of the procedure thus resulting in a significant reduction of set-up time. MLC can practically eliminate the use of Cerrobend blocking except for shaping small fields or "island" blocking in which an area within the open portion of the fields needs to be blocked.

The importance of MLC is not just the replacement of Cerrobend blocking. The greater impact of this technology is in the automation of field shaping and modulation of beam intensity. Modern radiotherapy techniques such as 3-D conformal radiation therapy (Chapter 19) and intensity-modulated radiation therapy (Chapter 20) are dependent on the dynamically controlled MLC. Other applications include dynamic wedges and electronic compensation. For further details on MLC designs and applications, the reader is referred to a review by Boyer (11).

### 13.3. Skin Dose

When a patient is treated with a megavoltage beam, the surface dose or skin dose can be substantially lower than the maximum dose that occurs in the subcutaneous tissues. In contrast to lower energy beams (e.g., superficial and orthovoltage x-rays), which give rise to maximum ionization at or close to the skin surface, the megavoltage beams produce an initial electronic buildup with depth, resulting in a reduced dose at the surface and maximum dose at the equilibrium depth. This phenomenon of dose buildup was discussed in Chapter 9.

Skin sparing is one of the most desirable features of high-energy photon beams. However, this effect may be reduced or even lost if the beam is excessively contaminated.
with secondary electrons. In the following sections, the sources of this contamination and the methods used to reduce it will be discussed.

A. Electron Contamination of Photon Beams

Surface dose is the result of electron contamination of the incident beam as well as the backscattered radiation (both electrons and photons) from the medium. It is well-known that all x-ray and y-ray beams used in radiation therapy are contaminated with secondary electrons. These electrons arise from photon interactions in the air, in the collimator,
and in any other scattering material in the path of the beam. If a shadow tray is used to support beam-shaping blocks, secondary electrons produced by photon interactions in the tray and the air column between the tray and the skin surface significantly increase skin dose. The shadow tray is usually thick enough to absorb most of the electrons incident on the tray.

There has been a controversy as to the relative contribution of secondary electrons vs. low energy scattered photons to the dose in the build-up region. It is well known that as the field size increases, the depth dose in the build-up region increases resulting in a shift in the depth of maximum dose, $d_{\text{max}}$, to increasingly shallower depths (12–14). Specifically, the cause of the $d_{\text{max}}$ shift with field size has been studied by several investigators (15–17). Current evidence favors the hypothesis that the effect is predominantly caused by the secondary electrons.

**B. Measurement of Dose Distribution in the Build-up Region**

Because of the steep dose gradient in the build-up region, the size of the dosimeter along the beam direction should be as small as possible. Extrapolation chambers (see Chapter 6) are the instruments of choice for these measurements. However, few institutions have these instruments available. Instead, fixed-separation plane-parallel ionization chambers are most commonly used for this purpose. Although these chambers are very suitable for measurements in regions of severe dose gradients, their response is dependent, in a complex manner, on their design. Several papers have discussed the inaccuracies in the measurement of dose in the build-up region when using fixed-separation plane-parallel chambers. These inaccuracies arise primarily as a result of electron scattering from the side walls of the chamber (18–20). These may be minimized by using a smaller plate separation and wider guard ring in the design of the chamber. Furthermore, the chambers may exhibit a significant polarity effect in the build-up region, which may be corrected by averaging the readings obtained with the positive and negative polarities. Gerbi and Khan (21) have studied several commercially available plane-parallel chambers and found that they overrespond in the build-up region. The errors were more severe at the surface and for the lower beam energies (e.g., $^{60}\text{Co}$). The magnitude of overresponse at the surface for a $^{60}\text{Co}$ beam ranged from 9% to 20% for the chambers studied.

Thin layers (<0.5 mm) of thermoluminescent dosimeter (TLD) material can also be used for measuring dose distribution in the build-up region. The TLD phosphor (e.g., LiF) can be in the form of chips, crystals embedded in plastic, or powder layers (18, 22, 23). The surface dose may be obtained by extrapolating the depth dose distribution curve to zero depth. In vivo measurements of surface dose can also be made by placing thin TLD chips directly on the skin surface. Such measurements are useful in checking dosimetry if an unacceptable degree of skin reaction develops.

**C. Skin Sparing as a Function of Photon Energy**

Studies have shown that the dose distribution in the build-up region depends on many variables such as beam energy, SSD, field size, and configuration of secondary blocking tray (18, 22–26). Table 13.2 gives values for different energies. These data are presented here as an example and should not be considered universal for all machines, especially for depths less than 2 mm. Reasonable agreement between different machines has been shown to exist for greater depths.

Examination of Table 13.2 would also indicate that for all energies the dose increases rapidly within the first few millimeters and then gradually achieves its maximum value at the depth of peak dose. For example, in the case of 4 MV, the percent depth dose increases from 14% to 74% in the first 2 mm, reaches 94% at a 5-mm depth and achieves its maximum value at a 10-mm depth. A practical application of this phenomenon is the case in which build-up bolus (Chapter 12) is used intentionally to maximize the dose on the
skin (e.g., covering a scar with a strip of bolus). A tissue equivalent bolus of 5- to 6-mm of thickness is usually adequate for 4 MV. Thus the thickness of bolus required to achieve 90% to 95% build-up of dose is substantially less than the depth of maximum dose.

Although the skin sparing depends on many conditions, as mentioned earlier, the effect, in general, becomes more and more pronounced as photon energy increases. For higher-energy beams, significant sparing can be achieved not only for the skin surface but also for the subcutaneous tissues.

### D. Effect of Absorber-skin Distance

The electron contamination with no absorber placed in the beam is mainly caused by the secondary electron emission from the collimator (including source, flattening filter, and air). When an absorber of thickness greater than the range of secondary electrons (equilibrium thickness) is introduced in the beam, the collimator electrons are almost completely absorbed but the absorber itself becomes the principal source of electron contamination of the beam. By increasing the distance between the tray and the surface, the electron fluence incident on the skin is reduced because of divergence as well as absorption and scattering of electrons in the air. Thus skin sparing is enhanced by placing the shadow tray farther away from the skin. In the case of a $^{60}$Co γ ray beam, it has been shown (27,28) that for small fields an air gap of 15 to 20 cm between the scatterer and the skin is adequate to keep the skin dose to an acceptable level (<50% of the $D_{max}$). This has been found to be true for higher-energy beams as well (17).

Figure 13.6 shows the effect on dose distribution in the build-up region as a Lucite shadow tray is placed in the beam at various distances from the phantom surface. Not only does the relative surface dose increase with decreasing tray-to-surface distance but the point of maximum dose buildup moves closer to the surface.

Figure 13.6 also illustrates the principle of what is known as the "beam spoiler." A low atomic number absorber, such as a Lucite shadow tray, placed at an appropriate distance from the surface, can be used to modify the build-up curve. Doppke et al. (29) have discussed the treatment of certain head and neck cancers with 10-MV x-rays using a beam spoiler to increase the dose to the superficial neck nodes.
E. Effect of Field Size

The relative skin dose depends strongly on field size. As the field dimensions are increased, the dose in the build-up region increases. This increase in dose is due to increased electron emission from the collimator and air. Figure 13.7 is a plot of relative surface dose as a function of field size for $^{60}$Co, 4-MV, and 10-MV beams. These data show that skin sparing is significantly reduced for the larger field sizes.

Saylor and Quillin (24) have discussed the relative importance of field size and tray-to-skin distance for $^{60}$Co $\gamma$ rays. They have shown that the optimum skin sparing occurs for an $h/r$ value of about 4, where $h$ is the tray-to-surface distance and $r$ is the radius of an

![Diagram](image)

**FIG. 13.7.** Percent surface dose as a function of field size. $^{60}$Co, Theratron 80, SSD = 80 cm, SDD = 59 cm, 4 MV, Clinac 4, SSD = 80 cm, 10 MV, LMR 13, SSD = 100 cm, SDD = 50 cm, $^{60}$Co and 4-MV. (Data are from Velkley DE, Manson DJ, Purdy JA, et al. Buildup region of megavoltage photon radiation sources. Med Phys 1975;2:14. 10-MV data are from Khan FM, Moore VC, Levitt SH. Effect of various atomic number absorbers on skin dose for 10-MeV x-rays. Radiology 1973;109:209.)
equivalent circular field. This ratio can be easily achieved for the $5 \times 5$ cm field, because it requires a distance of 12 cm; however, for the $30 \times 30$ cm field, the corresponding absorber-surface distance is 67 cm, which is hardly possible for isocentric treatments.

When using large fields with a tray-to-skin distance of 15 to 20 cm, it becomes necessary to use electron filters to maintain the skin-sparing effect. These are discussed in the next section.

**F. Electron Filters**

The skin dose can be reduced by using gamma ray absorbers of medium atomic number ($Z$ in the range of $30-80$). Such absorbers are commonly known as electron filters, because their introduction in the photon beam reduces the secondary electron scatter in the forward direction. Hine (30,31) studied the scattering of electrons produced by gamma rays in materials of various atomic numbers. He showed that the medium atomic number absorbers give less electron scatter in the forward direction than either the low or the very high $Z$ materials. Khan (22) and Saylor and Quillin (24) applied the results of Hine's study to the design of electron filters for the purpose of improving skin dose for $^{60}$Co teletherapy. Later it was shown that such filters not only reduce the surface dose but also improve the build-up characteristics of large fields (32).

Figure 13.8 is a plot of relative surface dose as a function of log ($Z + 1$). These data are plotted in this manner to show agreement with the theoretical relationship discussed by Hine (30,31). As $Z$ increases, the surface dose falls to a shallow minimum due to increased electron scattering in the absorbers. Further increases in $Z$ result in increased surface dose due to increased production of photoelectrons and electron pairs in addition to the Compton electrons. The minimum occurs at about $Z = 50$, which is the atomic number of tin. These results qualitatively agree with those obtained for $^{60}$Co gamma rays (24,30,31).

Effectiveness of tin in reducing skin dose is demonstrated in Fig. 13.9. Greater reduction is possible by increasing filter-skin distance as discussed previously. To preserve the light field, Saylor and Quillin (24) have suggested the use of leaded glass as an electron filter. However, breakability of leaded glass may pose a serious problem. We have used a tin sheet mounted on a pressed wood sheet that could be slipped under the Plexiglas tray at the end of the treatment set-up. In this arrangement, the tin filter must face the patient surface.
The thickness of an electron filter, in theory, should be at least equal to the maximum range of secondary electrons. For $^{60}$Co, this thickness is about 0.5 g/cm$^2$ or 0.9 mm of tin (assuming $\rho_{\text{tin}} = 5.75$ g/cm$^3$). For higher energies, thicknesses less than the maximum range of electrons may be used for practical reasons.

### G. Skin Sparing at Oblique Incidence

Skin dose has been shown to increase with increasing angle of incidence (33–39). Clinically, brisk reactions have been observed in patients when the beam is incident at near glancing angles. Jackson (35) has explained the increase in skin dose with increasing angle of incidence through the concept of electron range surface (ERS). The ERS is a 3-D representation of secondary electron range and distribution produced by a pencil beam of photons interacting with the medium (Fig. 13.10). Electrons generated inside the ERS volume will reach $P$ and contribute to the dose there, whereas those generated outside, because of their inadequate range, make no contribution. The ERS for $^{60}$Co $\gamma$ rays is in the shape of an ellipsoid with axial dimensions of $5 \times 2.4$ mm (35). As illustrated in Fig. 13.10, the increase in the angle of incidence of the photon beam results in additional surface dose at $P$ because of electron contribution from the portion of the ERS, which appears below the phantom surface (hatched curve). For tangential beam incidence, since half of the ERS is below the phantom surface, an upper estimate of the dose to the skin may be obtained by the following relationship (35,39):

$$\text{Percent skin dose} = \frac{1}{2}(100\% + \text{entrance dose}) \quad (13.1)$$
where the entrance dose represents the surface dose for normal incidence expressed as a percentage of $D_{\text{max}}$. The skin dose for other angles of incidence will lie between the values for the normal and the tangential incidence.

Gerbi et al. (40) did a systematic study of dose buildup for obliquely incident beams as a function of energy (6–24 MV), angle, depth, field size, and SSD. A quantity obliquity factor (OF) was defined as the dose at a point in phantom on central axis of a beam incident at angle $\theta^\circ$, with respect to the perpendicular to the surface, divided by the dose at the same point and depth along central axis with the beam incident at angle $0^\circ$. The obliquity factor, therefore, represents dose enhancement due to beam obliquity for the same depth. Figure 13.11 shows that the obliquity factor at the surface increases with the increase in the angle of incidence, first gradually and then dramatically beyond 45 degrees. Thus the surface dose at large oblique angles can be significantly higher than at normal incidence. At tangential or grazing incidence, the surface dose approaches the value given by Equation 13.1.

Another important effect associated with oblique angles is that as the surface dose increases with the angle of incidence, the depth of maximum buildup decreases. The dose reaches its maximum value faster at glancing angles than at normal incidence. As a result, the dose build-up region is compressed into a more superficial region. Under these conditions, a high skin reaction becomes much more likely. Jackson (35) has discussed the possibility that if the sensitivity of the skin extends to the first or second millimeter below the surface, at glancing angles skin sparing is practically lost for the cobalt unit and greatly reduced for higher-energy beams.

### 13.4. SEPARATION OF ADJACENT FIELDS

Adjacent treatment fields are commonly employed in external beam radiation therapy, such as the "mantle" and "inverted-Y" fields for the treatment of Hodgkin's disease. In some cases, the adjacent fields are orthogonal, such as the craniospinal fields used in the
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At Phantom Surface

![Graph showing obliquity factor at the surface plotted as a function of beam angle for various energy beams. Jackson formula for tangential beam incidence is based on Equation 13.1. (From Gerbi BJ, Meigooni AS, Khan FM. Dose buildup for obliquely incident photon beams. Med Phys 1987;14:393, with permission.)]

- 6-MV, Varian 6/100
- 6-MV, Varian 6-2500
- 10-MV, Philips SL 75-20
- 18-MV, Philips SL 75-20
- 24-MV, Varian 2500
- Using Jackson formula

FIG. 13.11. Obliquity factor at the surface plotted as a function of beam angle for various energy beams. Jackson formula for tangential beam incidence is based on Equation 13.1. (From Gerbi BJ, Meigooni AS, Khan FM. Dose buildup for obliquely incident photon beams. Med Phys 1987;14:393, with permission.)

Another example is the irradiation of head and neck tumors when the lateral neck fields are placed adjacent to the anterior supraventricular field. In each of these situations, there is a possibility of introducing very large dosage errors across the junction. Consequently, this region is at risk for tumor recurrence if it is underdosed or severe complications if it is overdosed.

The problem of adjacent fields has been extensively studied (41-53). A number of techniques have been devised to achieve dose uniformity in the field junction region. Some of the more commonly used techniques are illustrated in Fig. 13.12. Figure 13.12A has been described by Lance and Morgan (41); here fields are angled away from a common line of abutment to avoid overlap of the fields due to their geometric divergence. Figure 13.12B illustrates the methods in which the fields are separated at the skin surface to provide dose uniformity at a desired depth. The separation or gap between the fields is calculated on the basis of geometric divergence (53) or isodose curve matching (42,43). A technique using split beams (49,53) is illustrated in Fig. 13.12C. In this method, the beam is split along the plane containing the central axis by using a half-beam block or a beam-splitter, thus removing the geometric divergence of the beams at the split line. Figure 13.12D uses penumbra generators or spoilers (46,47). These lead wedges are custom designed to provide satisfactory dose distribution across the field junction.

In clinical practice, the fields are usually abutted at the surface if the tumor is superficial at the junction point. Care is however taken that the hot spot created due to the overlap of the beams at depth is clinically acceptable, considering the magnitude of the overdosage and the volume of the hot spot. In addition, the dosage received by a sensitive structure such as the spinal cord must not exceed its tolerance dose.

For the treatment of deep-seated lesions such as in the thorax, abdomen, and pelvis, the fields can be separated on the surface. It is assumed in this case that the cold spots created by the field separation are located superficially where there is no tumor.
A. Methods of Field Separation

As stated earlier, the field separation can be accomplished geometrically or dosimetrically.

A.1. Geometric

If the geometric boundary of the field is defined by the 50% decrement line (line joining the points at depth where the dose is 50% of the central axis value at the same depth), the dose at the point of junction between the beams will add up to be 100%. The dose distribution laterally across the junction is more or less uniform, depending on the interfield scatter contribution and the penumbra characteristics of the beam.

If the two fields are incident from one side only and made to junction at a given depth (Fig. 13.13), the dose above the junction will be lower and below the junction higher than the junction dose. In the case of four fields when two fields are incident from one side and two from the parallel opposed direction (Fig. 13.14), the fields are usually made to junction at the midline depth (e.g., mantle and inverted Y fields). Such an arrangement
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FIG. 13.13. Geometry of two adjacent beams, separated by a distance $S_1 + S_2$ on the surface and junctioning at depth $d$.

FIG. 13.14. Two pairs of parallel opposed fields. Adjacent fields are separated on the surface so that they all join at a point on the midline. A: Ideal geometry in which there is no three-field overlap. B: Arrangement in which there are two regions (shaded) of three-field overlap.
can be used to obtain almost a uniform distribution at the midline, but cold spots are created above and below the junction point.

Figure 13.13 shows the geometry of two adjacent beams that are allowed to join at a given depth \( d \). Let \( L_1 \) and \( L_2 \) be the field lengths and \( SSD_1 \) and \( SSD_2 \) be the source-surface distances. Since triangles \( ABC \) and \( CDE \) are similar:

\[
\frac{CD}{DE} = \frac{BC}{AB}
\]

or

\[
\frac{S_1}{d} = \frac{L_1}{2} \cdot \frac{1}{SSD_1}
\]

giving

\[
S_1 = \frac{L_1}{2} \cdot \frac{d}{SSD_1}
\] (13.2)

Similarly,

\[
S_2 = \frac{L_2}{2} \cdot \frac{d}{SSD_2}
\] (13.3)

Thus the total separation \( S \) on the surface is given by:

\[
S = S_1 + S_2 = \frac{L_1}{2} \cdot \frac{d}{SSD_1} + \frac{L_2}{2} \cdot \frac{d}{SSD_2}
\] (13.5)

Figure 13.14A shows an ideal geometry in which there is no overlap between a field and its adjacent opposing neighbor. The arrangement shown in Fig. 13.14B, on the other hand, creates regions of "three-field overlap" (shaded areas) where the bigger fields diverge into the opposing smaller fields. Consequently, the total dose there may exceed the central axis dose at the same depth. This will be of concern if a significant portion of the spinal cord is in the three-field overlap region.

The maximum length of three-field overlap (\( \Delta S \)) occurs on the surface and is given by:

\[
\Delta S = S_1 - S_2 \] (13.6)

\( \Delta S \) can be made equal to zero if:

\[
\frac{L_1}{L_2} = \frac{SSD_1}{SSD_2}
\] (13.7)

Thus if the field lengths are different, the SSDs can be adjusted to eliminate the three-field overlap. Also, if the geometrically calculated gap (\( S_1 + S_2 \)) is increased by \( \Delta S \), the three-field overlap is eliminated at the expense of a cold spot at the midline. As a compromise, one could increase the gap (\( S_1 + S_2 \)) by an amount \( \Delta S' \) just enough to eliminate the three-field overlap in a specific region such as the spinal cord. \( \Delta S' \) can be calculated geometrically:

\[
\Delta S' = \Delta S \cdot \frac{d'}{d} - \frac{d'}{d}
\] (13.8)

where \( d' \) is the depth of the cord from the anterior surface and \( d \) is the midline depth.

The three-field overlap in Fig. 13.14B can also be avoided by using the same length and SSD for all the four fields and blocking the second pair (e.g., paraaortic or inverted Y fields) caudally as needed. This technique is more convenient when the accelerator is equipped with asymmetric collimators that can be moved independently of each other.
Example 1

A patient is treated with parallel-opposed mantle and paraortic fields of lengths 30 and 15 cm, respectively. Calculate (a) the gap required on the surface for the beams to intersect at a midline depth of 10 cm and (b) the gap required to just eliminate the three-field overlap on the cord assumed to be at a depth of 15 cm from the anterior surface, given SSD = 100 cm for all the fields:

\[ a. S_1 = \frac{d}{SSD} = \frac{1}{2} \times 30 \times \frac{10}{100} = 1.5 \text{ cm} \]

\[ b. S_2 = \frac{d}{SSD} = \frac{1}{2} \times 15 \times \frac{10}{100} = 0.75 \text{ cm} \]

Total gap required = 1.5 + 0.75 = 2.3 cm

\[ b. \Delta S = S_1 - S_2 = 1.5 - 0.75 = 0.75 \text{ cm} \]

Length of three-field overlap on the cord:

\[ \Delta S' = \Delta S - \frac{d'}{d} = 0.75 \times \frac{15 - 10}{10} = 0.4 \text{ cm} \]

New gap required = \( S_1 + S_2 + \Delta S' = 2.7 \text{ cm} \)

Although the previous geometric considerations provide useful criteria for field separation, one must be aware of their limitations. For example, the actual dose distribution may present a different picture than the predictions based on pure geometry of beam divergence. Patient positioning, beam alignment, field penumbra, and radiation scatter are all relevant factors that make this problem one of the most complex in radiation therapy.

Figure 13.15 shows the dose distribution for the cases discussed in Example 1. The expected three-field hot spot is seen in Fig. 13.15A when the beams intersect at the midline. This hot spot is eliminated when the gap is increased from 2.3 to 3.0 cm (\( S_1 + S_2 + \Delta S' \)) (Fig. 13.15B). However, the dose in the junction region has dropped considerably. Such a procedure will be justified only if the junction region is tumor free. Figure 13.15C shows the distribution when the gap is just enough to eliminate the three-field overlap at the cord, i.e., gap = 2.7 cm. This reduces the dose to the cord but also cools down the midjunction area by about 10%.

In practice, the choice between the options shown in Fig. 13.15 should be based on physical, clinical, and technical considerations. As usual, the guiding principles are that the tumor must receive adequate dosage and sensitive structures must not be treated beyond tolerance. If these conditions are not satisfied, other methods of field matching, discussed earlier in this chapter, may be considered.

A.2. Dosimetric

The separation of fields can be determined by optimizing the placement of fields on the contour so that the composite isodose distribution is uniform at the desired depth and the hot and cold spots are acceptable. The accuracy of this procedure depends on the accuracy of the individual field isodose curves especially in the penumbra region.

B. Orthogonal Field Junctions

Orthogonal fields denote an arrangement in which the central axes of the adjacent fields are orthogonal (i.e., perpendicular to each other). For example, orthogonal fields are used for the treatment of medulloblastoma in which the craniospinal irradiation is accomplished by lateral parallel-opposed brain fields coupled with a posterior spine field. Another common
example is treatment of the neck by bilateral fields while an orthogonally adjacent anterior field is used to treat the supraclavicular areas.

The problem of matching orthogonal fields has been discussed by several investigators (48–52). For superficial tumors such as in the head and neck areas, it may be inadvisable to separate the adjacent fields unless the junction area is over a tumor-free region. If separation is not possible, one may use beam splitters and abut the fields along or close to their central axes (50). The matching line should be drawn each time before treatment to avoid overlap of the fields. If a sensitive structure such as the spinal cord exists in the junction region, one may additionally block an appropriate segment of the cord anteriorly or laterally, provided there is no tumor in the shielded region.
As stated previously, field separation is possible for deep-seated tumors if there is no tumor in the superficial junction region. A geometrical method of orthogonal field separation has been described by Werner et al. (52). According to this method, one pair of opposing fields, defined by the collimating light, is allowed to diverge on the skin and the point of intersection of the field borders is marked. From this point, a distance $S$ is calculated to separate the orthogonal fields. The separation $S$ is given by:

$$S = \frac{\gamma_1 \cdot L \cdot d}{SSD}$$

where $d$ is the depth at which the orthogonal fields are allowed to join. A general diagram for orthogonal field separation is illustrated in Fig. 13.16A.

**B.1. Craniospinal Fields**

Craniospinal irradiation involves a complex technique in which orthogonal junctions are created between the lateral brain fields and a posterior spine field. The spinal field, because of its large length, may be split into two spinal fields with a junction gap calculated according to Equation 13.5. The junction between the cranial and the spinal fields can be accomplished in several ways (48–50,52,54,55).
Technique A

Figure 13.16B presents an example showing bilateral cranial fields adjacent to a spinal field. The cranial light fields are allowed to diverge on the skin and their inferior borders meet at a point midway on the posterior neck surface. From this point, the spinal field is separated by a distance $S$, which is calculated from Equation 13.9 by substituting depth $d$ of spine (from the posterior surface), length $L$, and SSD for the spinal field. In this diagram, the solid line represents the light field on the surface. The dashed line shows the field projected at the depth of the spinal cord. Figure 13.16C is the lateral view of Fig. 13.16B.

Technique B

The patient is positioned prone with the forehead resting on a rigid head support and the chest and abdomen resting on hard Styrofoam blocks (Fig. 13.17A). Some institutions use a half-shell plaster body cast under the patient for immobilization of head and neck relative to thorax (54,55). The spine field is simulated with the cephalad margin on the neck but without exiting through the mouth. By opening the light field, the diverging boundary of the cephalad margin of the spinal field is displayed on the lateral aspect of the neck. This boundary is marked on the patient’s skin to provide a match line for the lateral cranial fields. The cranial fields are set up so that their caudad field margins are parallel with the diverging cephalad margin of the spinal field. This is accomplished by rotating the collimator of the cranial fields through an angle $\theta_{\text{coll}}$ (Fig. 13.17B).

If the cranial fields were nondivergent, the rotation of the cranial fields through $\theta_{\text{coll}}$ would be sufficient to provide the desired geometric match between the cranial and the spinal fields. However, to match the diverging cranial fields with the diverging spinal field, the couch must also be rotated through $\theta_{\text{couch}}$, in addition to the rotation of the cranial fields through $\theta_{\text{coll}}$ (Fig. 13.17C). The two angles $\theta_{\text{coll}}$ and $\theta_{\text{couch}}$ can be calculated as:

$$\theta_{\text{coll}} = \arctan \left( \frac{\sqrt[3]{L_1 \cdot 1}}{\text{SSD}} \right)$$

$$\theta_{\text{couch}} = \arctan \left( \frac{\sqrt[3]{L_2 \cdot 1}}{\text{SAD}} \right)$$

where $L_1$ is the length of the posterior spinal field, $L_2$ is the length of the lateral cranial field, SSD is the source-to-surface distance for the spinal field, and SAD is the source-to-axis distance for the cranial fields, assuming the SSD technique is used for the spinal field and the SAD technique for the cranial fields. The couch is rotated toward the side the cranial field enters the head.

An alternative approach to rotating the couch is to eliminate cranial field divergence by using a half-beam block or an independent jaw to split the fields at the craniospinal junction line (Fig. 13.17D). The beam splitter is positioned at the central axis or close to it, thereby eliminating divergence of the rays at the junction line. The collimator of the cranial fields is still tilted through $\theta_{\text{coll}}$ as discussed earlier.

The technique of using independent jaw and $\theta_{\text{coll}}$ to match the craniospinal fields, has two advantages: (a) orthogonal field matching is achieved with no overlaps between the cranial and spinal fields at any depth, and (b) the independent jaw can be conveniently used to move the craniospinal junction line caudally by about one centimeter each week during the treatment course to smear out the junctional dose distribution. As long as the independent jaw splits the cranial fields within a few centimeters of the central axis, the divergence of the cranial fields into the spinal field at the matching line will be minimal.

C. Guidelines for Field Matching

1. The site of field matching should be chosen, insofar as possible, over an area that does not contain tumor or a critically sensitive organ.
2. If the tumor is superficial at the junction site, the fields should not be separated because a cold spot on the tumor will risk recurrence. However, if the diverging fields abut on the skin surface, they will overlap at depth. In some cases, this may be clinically acceptable, provided the excessive dosage delivered to the underlying tissues does not exceed their tolerance. In particular, the tolerances of critical structures such as the spinal cord must not be exceeded. In the case of a superficial tumor with a critical organ located at depth, one may abut the fields at the surface but eliminate beam divergence using a beam splitter or by tilting the beams.

3. For deep-seated tumors, the fields may be separated on the skin surface so that the junction point lies at the midline. Again, care must be taken in regard to a critical structure near the junction region.
4. The line of field matching technique must be verified by actual isodose distributions before it is adopted for general clinical use. In addition, beam alignment with the light field and the accuracy of isodose curves in the penumbra region are essential prerequisites.

REFERENCES

22. Khan FM. Use of electron filter to reduce skin dose in cobalt teletherapy. AJR 1971;111:180.