CHAPTER 7

Clinical Dose Calculations

7-1. Introduction

Computing absorbed doses in a patient using data measured in a phantom has been the standard of practice in radiotherapy. This is because direct measurement of absorbed doses in a patient is impractical and often impossible. Machine data are typically measured in phantom at nominal source-to-surface distance (SSD). Several dosimetric functions have been introduced to relate absorbed doses measured in a phantom to absorbed doses in a patient. These include the percent depth dose (P), tissue-air ratio (TAR), backscatter factor (BSF), tissue-phantom ratio (TPR) and tissue-maximum ratio (TMR). The percent depth dose was introduced to calculate doses for treatments delivered using fixed treatment distance machines. Later, the tissue-air ratio was introduced to provide a means of calculating doses for rotational therapy. This term was introduced by John et al.[1] and originally called "tumor-air ratio". With the introduction of isocentric treatment machines, the concept of TAR was extended to isocentric treatments. However, the concept of TAR was found to be inadequate for high-energy photon beams because of the need for a large "mini-phantom" for in-air measurements. A new dosimetric function referred to as tissue-phantom ratio (TPR) was introduced by Karzmark et al.[2] to circumvent this inadequacy. The introduction of these dosimetric functions were based on their specific needs and applied accordingly. The inter-relationships between these functions were not fully appreciated until recently.[3] In this chapter, we shall learn how these dosimetric functions are inter-related and the procedure of moving from one dosimetric function to another. Next, we shall examine the properties of these dosimetric functions. Lastly, we shall learn how to compute the treatment times or monitor units using these dosimetric functions as applied to data measured in phantom. In addition to the dosimetric functions, parameters relevant for computing the treatment times or monitor units including machine outputs, field size factors, and transmission factors of materials placed between the beam and the patient shall be briefly discussed.

7-2. Equivalent Square

The concept of equivalent square field size was introduced to reduce the amount of dosimetric data acquisitions and simplify dosimetric calculations. **A beam field size is said to be equivalent to a square field size if the scattering effects and depth doses are identical**. The numerous possible combinations of field length and width that can be used to form a treatment field to conform to any target shape would present a huge task of dosimetric data collections. This could be avoided through the use of the equivalent field size technique. There are several ways of obtaining equivalent field size. A method of obtaining equivalent field size is through the use of published data in the British Journal of Radiology (BJR) Supplement 25 published in 1996.[4] This is based on lookup tables, which is widely implemented in the clinics. Part of the equivalent table is given below in Table 7-1.

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[1] John HE, Whitmore GF, Watson TA. A system of dosimetry for rotational therapy with typical rotational distributions. J Can Assn Radiol 1953: 4; 1.

[2] Karzmark CJ, Deubert, A, Loevinger R. Tissue-phantom ratios - an aid to treatment planning. Br J Radiol 1965: 38; 158.

[3] Saw CB, Ayyangar KM, Hussey DH. Review of dosimetric functions for meterset calculations. Med Dosm 2000: 25; 55-60.

[4] BJR Supplement No. 25. Central axis depth dose data for use in radiotherapy: 1996. London (England): British Institute of Radiology; 1996.

Table 7-1. Equivalent squares of rectangular fields.[5]

Long Axis (cm)	Short Axis (cm) 1	2	3	4	5	6	7	8	9	10	11	12
1	1.0											
2	1.4	2.0										
3	1.6	2.4	3.0									
4	1.7	2.7	3.4	4.0								
5	1.8	2.9	3.8	4.5	5.0							
6	1.9	3.1	4.1	4.8	5.5	6.0						
7	2.0	3.3	4.3	5.1	5.8	6.5	7.0					
8	2.1	3.4	4.5	5.4	6.2	6.9	7.5	8.0				
9	2.1	3.5	4.6	5.6	6.5	7.2	7.9	8.5	9.0			
10	2.2	3.6	4.8	5.8	6.7	7.5	8.2	8.9	9.5	10.0		
11	2.2	3.7	4.9	6.0	6.9	7.8	8.5	9.3	9.9	10.5	11.0	
12	2.2	3.7	5.0	6.1	7.1	8.0	8.8	9.6	10.3	10.9	11.5	12

EXAMPLE 7-1. Using the lookup table (Table 7-1), find the equivalent square field of a rectangular field of 5 cm x 10 cm.
SOLUTION:

1. Using the long axis, find the 10 cm row
2. Use the short axis, find the 5 cm column.
3. The intersection of these row and column is the equivalent square value, which is 6.7 cm

A simple rule-of-thumb method of equating a rectangular field to a square field was proposed by Sterling et al.[6] Sterling's formula states that a rectangular field size is equivalent to a square field size if the two field sizes have the same area to perimeter ratio. If A is the area and P is the perimeter of a rectangular field and s is the equivalent side square then Sterling relationship can be written as

$$s = 4 \times \left(\frac{A}{P}\right) \qquad (7\text{-}1)$$

EXAMPLE 7-2. Compute the side of an equivalent square for a rectangular field of 5 cm x 10 cm using Sterling approximation.
SOLUTION:

$$s = 4 \times \left(\frac{5 \times 10}{2(5+10)}\right) \text{cm}$$

$$= 4 \times \left(\frac{50}{30}\right) \text{cm} = 6.7 \text{ cm}$$

[5] BJR Supplement No. 25. Central axis depth dose data for use in radiotherapy: 1996. London (England): British Institute of Radiology; 1996.
[6] Sterling TD et al. Derivation of a mathematical expression for the percent depth dose surface of a cobalt 60 beams and visualization of multiple field dose distributions. Br J Radiol 1964: 37; 544-550.

The above example shows that the Sterling's approximation predicts the equivalent square well. Note that this approximation is not valid for circles or irregularly shaped fields indicating there are limitations to the use of Sterling approximation rule.

7-3. Dosimetric Functions

Several dosimetric functions are available to assist in the computation of absorbed doses in a patient based on data measured in a phantom. The dosimetric functions describe the doses at various points in space "filled" with a reference phantom, a phantom (simulating patient setup condition) or a mini-phantom in air. For clarification purposes, we shall set these phantoms side by side to examine their dosimetric functions and their relationships as illustrated in Figure 7-1.

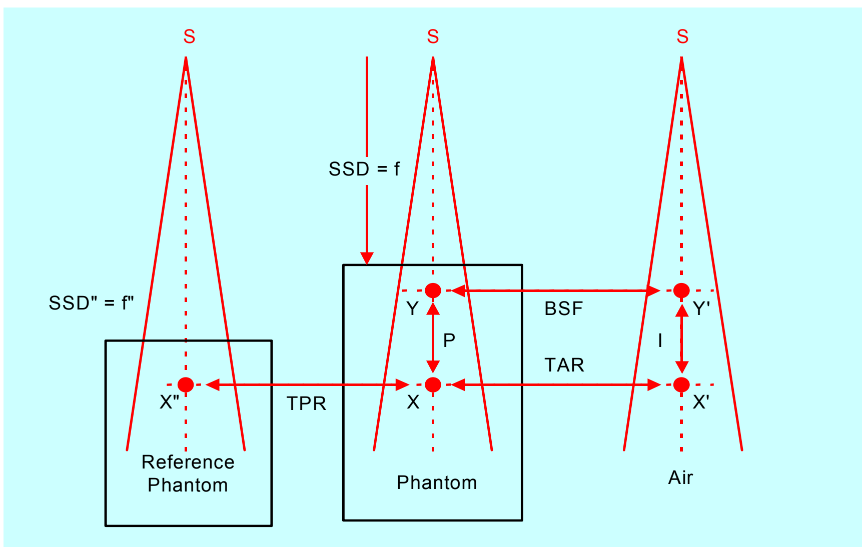

Figure 7-1. Dose points associated with dosimetric functions.

In Figure 7-1, the schematics show the dose points relative to the radiation source (S) and also relative to a reference phantom, a phantom, and mini-phantom in air. The points X, X', and X" are all located along the beam axes at the same distance from the source. However, the point X is located at a depth of d in phantom with SSD=f while the point X" at a reference depth d" in the reference phantom with SSD"=f". Both points Y and Y' are also along the beam axes at the same distance but closer to the source compared to points X, X' or X". **In this notation, the single quote represents dose points in air while the double quotes dose points in reference phantom and without quote for dose points in a phantom**.

We say that the **percent depth dose (P)** or depth dose at point X is the ratio of the dose (D_X) at point X to the dose (D_Y) at point Y as

$$P(d) = \frac{D_X}{D_Y} x 100 \qquad (7\text{-}2)$$

where d denotes the depth from the phantom surface. The notation P(d) is used to indicate that the depth dose depends on the depth of the point of interest from the surface of the phantom. As presented, the percent depth dose is a ratio of doses at two points inside a phantom. The depth of point Y from the phantom surface is called reference depth, usually is chosen to be the depth of maximum dose (dm).

We say that the dose ($D_{X'}$) at point X′ is related to the dose ($D_{Y'}$) at point Y′ via the **inverse square law (I)**. The inverse square law is actually the ratio of beam intensities at two points in air, which is inversely proportional to the square of their distances from the source as

$$I(d, d_m, f) = \frac{D_{X'}}{D_{Y'}} = \left(\frac{f + dm}{f + d}\right)^2 \qquad (7\text{-}3)$$

where f is the source-to-surface distance and dm is the depth of maximum dose in phantom. Both point X′ and point Y′ must be in air medium.

The **tissue-air ratio (TAR)** is the ratio of the dose (D_X) at point X to the dose ($D_{X'}$) at point X' as

$$TAR(d) = \frac{D_x}{D_{X'}} \qquad (7\text{-}4)$$

Note that point X is in a phantom while point X' is in air. TAR therefore relates a dose in phantom to a dose in air.

In the special case, where the point in phantom is at the depth of maximum dose such as point Y, the TAR is given a special name called **backscatter factor (BSF)** expressed as

$$BSF(dm) = \frac{D_Y}{D_{Y'}} \qquad (7\text{-}5)$$

The dose at any depth in a phantom can be computed using these dosimetric functions.

EXAMPLE 7-3. Express depth dose in terms of TAR.
SOLUTION:

Looking at Figure 7-1, we set up dose points from percent depth dose to pass through the TAR dosimetric function as follows:

$$P = \frac{D_X}{D_Y} x\,100$$

$$= \frac{D_X}{D_{X'}} x \frac{D_{X'}}{D_{Y'}} x \frac{D_{Y'}}{D_Y} x100$$

$$= TAR\ x \left(\frac{f + dm}{f + d}\right)^2 x \frac{1}{BSF} x100$$

$$= \frac{TAR}{BSF} x \left(\frac{f + dm}{f + d}\right)^2 x\ 100$$

Depending on the treatment technique, the use of one of these dosimetric functions is more convenient compared to the other. Typically the percent depth dose is more convenient if the treatment setup involves **fixed treatment distance (or SSD) technique**. For **isocentric (or SAD) technique**, the TAR function is preferred because it is independent of the distances from the source-to-surface distance of the patient. This is very convenient when a plan is made to deliver radiation doses from a number of gantry angles.

Another function, the tissue-phantom ratio was introduced for the dosimetry of high-energy photon beams. The **tissue-phantom ratio (TPR)** is the ratio of the dose (D_X) at point X in phantom to the dose ($D_{X''}$) at point X" in a reference phantom. It is written as

$$TPR(d) = \frac{D_X}{D_{X''}} \tag{7-6}$$

Here, the point X″ is defined to be at a reference depth (d") in a reference phantom. TPR relates the dose in a phantom that simulates a patient setup to the dose in a reference phantom. Since the two points are at the same distance from the source, we say that the TPR is independent of the SSD. Like TAR, this dosimetric function should be convenient for use with SAD treatment technique. For the special case when the reference depth is exactly equal to the depth of maximum dose (dm) in the reference phantom, the TPR becomes the **tissue-maximum ratio** (TMR) as

$$TMR(d, dm'') = \frac{D_X}{D_{X''}} \tag{7-7}$$

where dm″ represents the depth of maximum dose in the reference phantom.

EXAMPLE 7-4. Relate TPR to TAR.
SOLUTION:

From Figure 7-1, we set up the dose points from TPR to pass through the TAR function as follows:

$$TPR = \frac{D_X}{D_{X''}}$$

$$= \frac{D_X}{D_{X'}} \times \frac{D_{X'}}{D_{X''}}$$

$$= TAR \times \frac{1}{BSF}$$

The BSF is used not to confuse with the second TAR relating the dose at point X' to the dose at point X". The BSF is valid only if X" is at depth of maximum dose.

The TPR can be viewed as an extension to TAR where measurement in air at point X′ is replaced with measurement in a reference phantom at point X″. It can also be viewed as an extension of percent depth dose in the case where the source is at infinite distance and the inverse square term is dropped as examined in EXERCISE 7-1.

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*__EXERCISE 7-1__. Discuss EXAMPLE 7-3 and 7-4 and show that TPR is an extension of percent depth dose.

The manipulation of these six dosimetric functions has been presented above. These dosimetric functions can be summarized as the expressed relationships between doses at various points inside and outside the designated phantoms. In the next few sections we shall examine the properties of these dosimetric functions and the factors that affect their values.

7-4. Percent Depth Dose

The percent depth dose (P) expressed as a percentage, at any depth is the ratio of the dose at that depth to the dose at a reference depth. The depth (dm) of maximum dose is usually chosen as the reference depth. Mathematically, the equation for the depth dose of a photon beam quality is

$$P(d, A_{f+dm}, f) = \frac{D}{D_m} \times 100 \qquad (7\text{-}8)$$

where D represents the dose at depth (d) in phantom, D_m is the maximum dose. The notation $P(d, A_{f+dm}, f)$ indicates that the percent depth dose depends not only on the depth of the point of interest, that is, point X in phantom but also the beam size, A_{f+dm} and the source-to-surface distance (f) of the phantom. The subscript f+dm on A indicates that the beam size is measured at f+dm from the source, at point Y in phantom. The percent depth dose also depends on the photon beam quality, which is not explicitly expressed in the equation. For a single field, the maximum dose along the beam axis is sometimes called **incident dose (ID)** or **given dose (GD)**.[7]

Percent depth dose data are typically collected using motorized water scanning equipment.[8] It consists of a large water tank with facility to attach radiation detector on motorized rails. Since the rails restrict the movement of the detector in the direction aligned with the water tank, proper water tank alignment is important so that the radiation detector can track along the beam axis to allow for depth dose data collection. For practical reason, the water surface is set at the nominal SSD where the beam size is defined, even though by definition, the **beam size is defined at reference depth**.[9]

EXERCISE 7-2. In the percent depth dose definition, the field size is defined at the depth of maximum dose. Estimate the amount of error when the field size is set at the phantom surface for practical purpose for a kilovoltage beam, a 6 MV photon beam and a 23 MV photon beam.

For a given field size, data are collected along the beam axis from the water surface to a depth of 30-40 cm. The data are then normalized to the maximum dose and expressed as percentage. During commissioning, the depth dose data are collected for a range of field sizes from 3-cm x 3-cm to 40-cm x 40-cm for a particular photon beam quality with setup at nominal source-to-surface distance of the phantom. Depth doses for field sizes smaller that 3-cm x 3-cm are generally not collected in this way.

EXAMPLE 7-5. Calculate the given dose if 200 cGy is delivered to a depth of 10-cm using a 6 MV photon beam. The percent depth dose at a depth of 10 cm is 67%.
SOLUTION:

The given dose is

$$GD = \frac{D}{P/100}$$

$$= \frac{200}{0.67} = 299 \text{ cGy}$$

The depth dose for a photon beam is characterized by an initial buildup of dose reaching a maximum and thereafter decreases exponentially as a function of depth. The region between the surface and point of maximum dose is referred to as the **dose buildup region**. For kilovoltage x-ray beam,

[7] ICRU Report No. 24. Determination of absorbed dose in a patient irradiated by beam of X or gamma rays in radiotherapy procedures. Washington (DC): ICRU; 1976.
[8] Welhofer Scanning System. IBA Advanced Radiotherapy, Barlett, TN 38133.
[9] ICRU Report No. 24. Determination of absorbed dose in a patient irradiated by beam of X or gamma rays in radiotherapy procedures. Washington (DC): ICRU; 1976.

there is no buildup region since the maximum dose occurs at phantom surface.

As stated above, the depth dose or beam penetration also depends on the beam energy, depth in phantom, field size, and SSD to a phantom. The depth dose increases with increasing beam energy, field size, and SSD while decreasing with depth. These characteristics have been discussed in Chapter 6. If the energy of the photon beam is increased, the surface dose decreases, the dm shifts deeper into the phantom and the depth dose increases. The decrease in surface dose is referred to as the **skin sparing effect,** which permits larger doses to be delivered to deep-seated target or tumor without exceeding the tolerance dose of the skin. Besides the energy of the photon beam, skin sparing also improves (surface dose decreases) as the field size decreases. When the field size increases, the surface dose increases, the dm shifts towards the surface of the phantom, and the depth dose increases. When the SSD increases, the depth dose also increases. The effect of extended SSD and the use of **Mayneord factor** to correct the depth dose will be discussed in Section 7-8.

7-5. Inverse Square Law

The radiation intensity varies inversely as the square of the distance from the source. This phenomenon is called the **inverse square law**. The **inverse square law is only valid if the source is a point source and there is no attenuating material or scattering material present**. The source is considered a point source if the distance of the point of measurement is significantly larger than the size of the source, which is the case for teletherapy.

The dose at point p from a point source r cm away can be written as

$$D(p) = \frac{k}{r^2} \tag{7-9}$$

If the dose at a particular point r_1 cm from a point source is determined to be D_1, and D_2 at a distance r_2 cm from the same source, then the ratio of their doses is equal to the inverse ratio of the square of their distances as

$$\frac{D_1}{D_2} = \left(\frac{r_2^2}{r_1^2}\right) \tag{7-10}$$

This relationship can be derived by taking the ratio of the beam intensities at two points away from a source.

EXAMPLE 7-6. What is the dose rate in air of a Cobalt-60 unit at 89.5 cm if the dose rate at 80.0 cm is 105.5 cGy/min.

SOLUTION:

Using equation (7-10), we have

$$D_2 = D_1\left(\frac{r_1^2}{r_2^2}\right)$$

$$= (105.5\ \frac{cGy}{min})\left(\frac{80^2}{89.5^2}\right)$$

$$= (105.5\ \frac{cGy}{min})\ (0.799)$$

$$= 84.29\ \frac{cGy}{min}$$

Looking at Figure 7-1, the inverse square law relates the doses in air at points X′ and Y′. The distances of these points from the source are deduced based on their relative positions to the phantom. As defined, f is the distance from the source to the surface of the phantom (that simulates the patient setup), dm is the depth inside the phantom to point Y, and d is the depth to the point X. Since points X and X' are at the same distance from the source, the distance of X' from the source is f+d. Likewise we can deduce the distance from the source to point Y' as f+dm. Substituting these values into equation (7-10), the equation for the inverse square is

$$I(d, dm, f) = \frac{D_{X'}}{D_{Y'}} = \left(\frac{f + dm}{f + d}\right)^2 \qquad (7\text{-}11)$$

We say that the inverse square is dependent on the SSD but is independent of beam quality. Experimental measurements of intensities from a linear accelerator show it follows the inverse square function. However, the source appears to be 1-2 cm closer to the point of measurement than its physical location due to radiation scatter in the treatment head.

7-6. Tissue-Air Ratio and Backscatter Factor

The tissue-air ratio (TAR) at any depth in phantom is defined as the ratio of dose (D_{Tissue}) at that depth in phantom to dose (D_{Air}) in air at the same distance from the source. The equation for TAR is

$$TAR(d, A_{f+d}) = \frac{D_{Tissue}}{D_{air}} \qquad (7\text{-}12)$$

By definition, **TAR is said to be independent of the SSD** since both points in tissue and in air are at the same distance from the source. The TAR depends on depth d into the phantom and the field size, A_{f+d}, which is determined at that depth. The definition of field size is different for TAR compared to the depth dose where the field size is defined at dm or (f+dm from the source). The TAR is also dependent on beam quality although not explicitly indicated in equation (7-12). A partial list of TAR for cobalt-60 unit taken from BJR Suppl 25 is given in Table 7-2.

Table 7-2. TAR for cobalt-60 gamma rays.[10]

Depth	Equivalent Square Field Size (cm x cm)										
(cm)	0	4x4	5x5	6x6	7x7	8x8	9x9	10x10	12x12	15x15	20x20
0.5	1.000	1.032	1.036	1.040	1.043	1.048	1.052	1.054	1.060	1.068	1.078
1	0.968	1.016	1.022	1.029	1.034	1.039	1.043	1.048	1.054	1.063	1.073
2	0.906	0.978	0.989	0.999	1.006	1.012	1.017	1.023	1.031	1.042	1.055
3	0.849	0.936	0.949	0.961	0.970	0.978	0.985	0.992	1.002	1.014	1.031
4	0.795	0.893	0.908	0.921	0.931	0.942	0.950	0.957	0.970	0.984	1.003
5	0.744	0.847	0.864	0.880	0.892	0.904	0.913	0.921	0.936	0.953	0.974
6	0.697	0.801	0.819	0.835	0.850	0.862	0.873	0.884	0.900	0.920	0.942
7	0.652	0.756	0.777	0.794	0.809	0.823	0.835	0.845	0.863	0.886	0.909
8	0.611	0.715	0.734	0.751	0.768	0.782	0.794	0.805	0.825	0.849	0.876
9	0.572	0.672	0.692	0.712	0.728	0.742	0.755	0.769	0.789	0.813	0.843
10	0.536	0.631	0.654	0.671	0.688	0.704	0.719	0.731	0.751	0.779	0.809

EXAMPLE 7-7. A parallel-opposed field (AP-PA) plan was implemented to deliver 200 cGy to midplane of a patient whose thickness is 20 cm using an 80-cm SSD Cobalt-60 unit. Compute the timer setting per field if an isocentric treatment technique was used. The TAR at 10 cm depth is 0.751. Assume that the dose rate at isocenter is 180 cGy/min and no timer error.
SOLUTION:

The interest in this problem is to compute the timer. This time multiply by the output would give the dose delivered. For Cobalt-60 unit, the output is specified at point X′. From Figure 7-1, the dose point of interest is X′ related to dose point X as

$$D_{X'} = \frac{D_{X'}}{D_X} \times D_X$$

$$t \times O = \frac{1}{TAR} \times D$$

or

$$t = \frac{D}{TAR} \times \frac{1}{O}$$

$$= \frac{100\ cGy}{0.751} \times \frac{1}{180\frac{cGy}{min}} = 0.74\,min$$

***EXERCISE 7-3**. Explain why the BSF is not needed in EXAMPLE 7-7?

[10] BJR Supplement 25. Central axis depth dose data for use in radiotherapy: 1996. London (England): British Institute of Radiology; 1996.

The TAR data are often not measured but derived from the depth dose data. The relationship between TAR and depth dose will be derived in Section 7-8. In theory, a TAR value involves two measurements, one in the phantom medium at a particular depth and the other in air. To perform these measurements, the distance from the source, the field size and beam quality must be the same. The in-air measurement requires a buildup cap covered over the detector. The TAR should be measured at SAD where the field size is precisely defined by the collimator settings. The requirement that the detector must be in the same position placed significant challenges in measuring the TAR. If solid water is used as the phantom material, the measurement procedure requires a measurement in the solid water followed by in-air measurement. This process has to be repeated for each TAR value. A different data collection technique would be to use a water scanning equipment where the water level can be easily changed. If the detector is fixed at the SAD, after each measurement, water can be pumped out to create a condition for in-air measurements. The sequences on how data should be collected for example all in-phantom data first followed by all in-air measurements will depend on the ingenuity of the person collecting the data.

The TAR decreases with increasing depth into the phantom as shown in Figure 7-2. The TAR is also dependent on the field size, increasing with increasing field size also demonstrated in Figure 7-2.

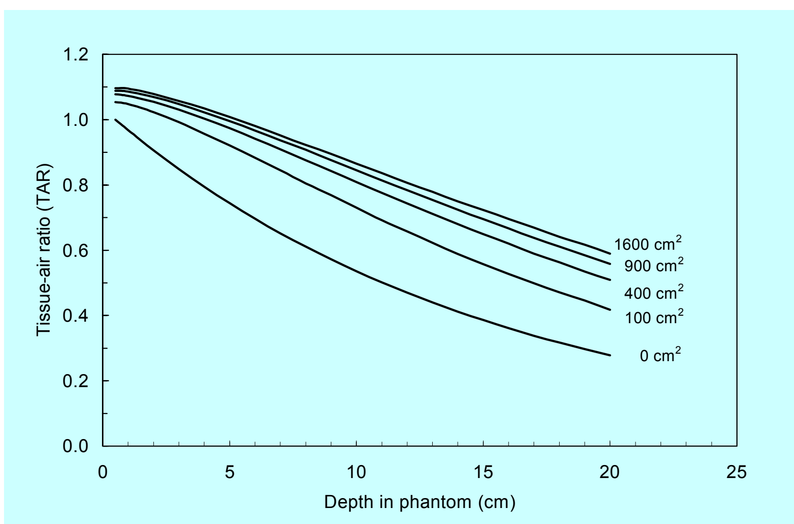


Figure 7-2. Variation of TAR with depth for differenat field area for cobalt-60. Data taken from BJR 25.

Not shown in Figure 7-2 is the effect of beam quality on TAR. The TAR increases with increasing photon beam energy.

At the depth (dm) of maximum dose, the TAR becomes the **backscatter factor** (BSF), i.e.,

$$BSF(dm, A_{f+dm}) = TAR(dm, A_{f+dm}) \qquad (7\text{-}13)$$

Like the TAR, the BSF is also independent of SSD and increases with field size as shown in Figure 7-3.

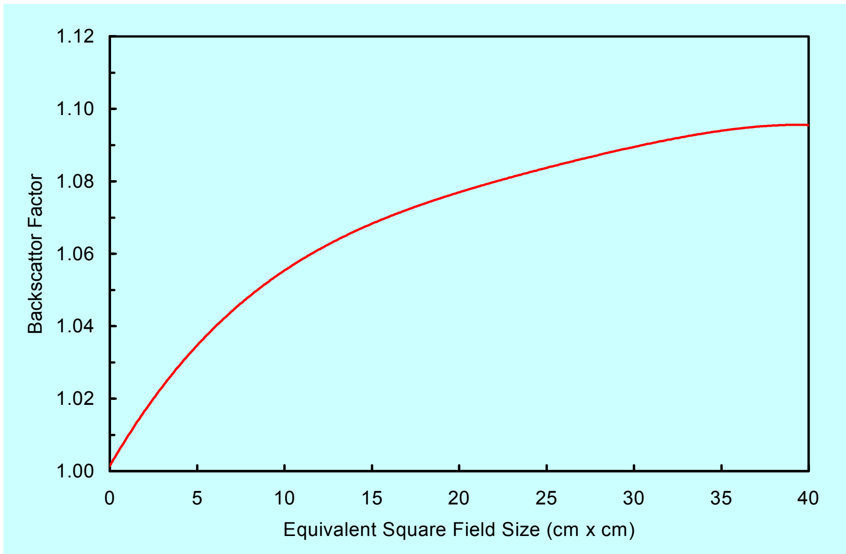


Figure 7-3. BSF of Cobalt-60 as a function of field size from BJR 25.

As presented in Chapter 6, the BSF increases and reaches a maximum value with beam quality of about 0.8 mm Cu HVL, and thereafter decreases with beam energy. For higher photon beam energy, the BSF decreases to nearly unity. This decrement is understandable since the interaction of higher energy photon beam with matter results in producing radiation that is scattered in the forward direction.

EXAMPLE 7-8. If the patient in EXAMPLE 7-7 is in the supine position, compute the cord dose if the point of interest is 5 cm towards the source for the PA field. The TAR at 5 cm depth is 0.936.

SOLUTION:

The cord dose is at a different location from the prescribed dose. In this problem we shall relate the prescribed dose to the cord dose. As such, we shall use point X as the prescribed dose point and point Y as the cord dose point. From Figure 7-1, we can set the relationship between the two points as:

$$D_Y = \frac{D_Y}{D_{Y'}} \text{ x } \frac{D_{Y'}}{D_{X'}} \text{ x } \frac{D_{X'}}{D_X} \text{ x } D_X$$

$$= TAR(d = 5) \text{ x } \left(\frac{80}{75}\right)^2 \text{ x } \frac{1}{TAR(d = 10)} \text{ x } 100\,cGy$$

$$= 0.936 \text{ x } 1.138 \text{ x } \frac{1}{0.751} \text{ x } 100$$

$$= 141.8\ cGy$$

EXERCISE 7-4. In EXAMPLE 7-8 the TAR at 5 cm depth is taken directly from Table 7-2 for a 12 cm x 12 cm field size creating a small error. Compute the actual field size that should be used?

Since most radiation is scattered in the forward direction for high-energy photon beams, the term peakscatter factor (PSF) is used instead of backscatter factor. The **peakscatter factor** is defined as the ratio of radiation output in a phantom to the radiation output in air measured with a buildup cap. As defined, the peakscatter factor has little theoretical significance compared to backscatter factor. Again, this is because measurements using a large "mini-phantom" for high-energy photon beams do not represent the theoretical concept of a small mini-phantom.

7-7. Scatter-Air Ratio (SAR)

Analysis of the concept of TAR shows that the dose at a point in phantom can be broadly divided as the dose contributions from the primary beam and the scattered beam. The in-air measurement defines the contribution from the primary beam while the in-phantom measurement is a treatment of both primary and scattered components. A simple way of determining the contribution of the primary beam is to allow the field size to become very small and approaches zero. As the field size approaches zero, the scattered component becomes very small and goes to zero. As such, the primary beam is defined by the condition at which the field size is zero. Then the scattered dose is the difference between the dose of a normal field minus the dose of a zero field size. In terms of TAR, the scatter-air ratio (SAR) is defined as

$$SAR(d, A_{f+d}) = TAR(d, A_{f+d}) - TAR_o \qquad (7\text{-}14)$$

where TAR_o is the component for zero field size. This primary component does not change but the scatter component increases with field size as illustrated in Figure 7-4.

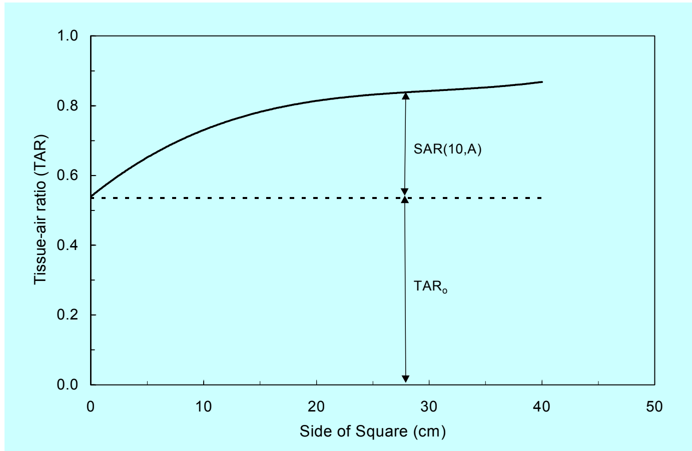


Figure 7-4. The relationship between TAR and SAR for a cobalt-60 photon beam.

The zero-area TAR is a mathematical abstraction and cannot be physically measured. It is determined through extrapolation of TAR for different field size at depth to zero field size. This method of extracting the zero-TAR is not valid for high-energy photon beams because the TAR increases instead of decreases as the field size goes to zero.[11] This is inconsistent with our assumption that the change is scattered radiation is proportional to the field size. This extrapolation is not precise but its shape can be deduced since it is expected to be identical to the narrow beam attenuation curve, which can be measured.

The application of scatter-air ratio to the calculations of irregular field will be discussed in Section 7-13.

7-8. Tissue-Air Ratio and Percent Depth-Dose Relationship

Tables of depth doses and TARs for a photon beam as a function of field size and depth are available in most clinics. The depth dose tables are used for dosimetric computation for patient set up following the fixed treatment distance or SSD technique. If the treatment uses isocentric technique, the TAR tables are used. It is instructive to relate the depth dose to TAR. Using Figure 7-1, we note that the TAR relationship can be written as

$$\mathrm{TAR}(d,A)=\frac{D_X}{D_{X'}}=\frac{D_X}{D_Y}\times\frac{D_Y}{D_{Y'}}\times\frac{D_{Y'}}{D_{X'}}$$

In analyzing the terms, we find that the first term is the depth dose, the second is the backscatter factor, and the last term is the inverse square. Substituting these terms into the above equation, we have

[11] Saw CB, Berta C, Wu A. High energy photon beam percent depth dose curves. Med Dosm 1992: 17; 35-36.

$$TAR(d, A_{f+d}) = \frac{P(d, A_{f+dm}, f)}{100} \times BSF(dm, A_{f+dm}) \times \left(\frac{f+d}{f+dm}\right)^2 \quad (7\text{-}15)$$

The TAR is related to the depth dose through the BSF and the inverse square relationship.

Another derivation that is of interest is the Mayneord factor used to adjust the percent depth dose for the effect of changing SSD. Assume that the depth doses at $SSD_1 = f_1$ have been measured and tabulated and $SSD_2 = f_2$ is a clinical setup where $f_1 < f_2$. The percent depth dose at f_2 can be derived using point dose relationship as shown in Figure 7-5.

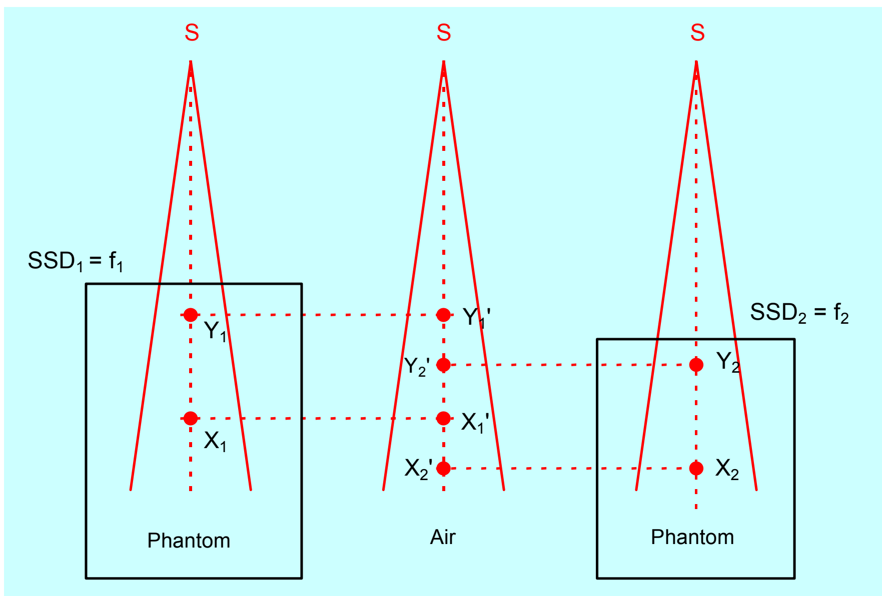


Figure 7-5. Dose points for extended SSD treatment technique.

The new depth dose is given as

$$P(d, f_2) = \frac{TAR_2 \times TAR_1^{-1}}{BSF_2 \times BSF_1^{-1}} \times M \times P(d, f_1) \quad (7\text{-}16)$$

with

$$M = \left(\frac{f_1 + d}{f_2 + d}\right)^2 / \left(\frac{f_1 + d_m}{f_2 + d_m}\right)^2 \quad (7\text{-}17)$$

and M is called **Mayneord factor**. The Mayneord factor is greater than 1 if $f_2 > f_1$. It is less than 1 if $f_2 < f_1$.

***EXERCISE 7-5**. Derived new percent depth dose at extended SSD.

It is obvious that the ratio of TARs and the ratio of BSFs are not one since the field sizes at the two SSDs are different. However, it can be approximated that they are close to unity provided the depth and dm are small compared to the SSD, which is always the case. The depth dose at extended SSD can then be approximated as

$$P(d, f_2) = M \times P(d, f_1) \quad (7\text{-}18)$$

EXAMPLE 7-9. Compute the percent depth dose at 10 cm for 120 cm SSD if the depth dose at 10 cm at 100 cm SSD is 72.1% for a 6 MV photon beam (dm = 1.5 cm)
SOLUTION:

$f_1 = 100$ cm and $f_2 = 120$

$$M = \left(\frac{100+10}{120+10}\right)^2 / \left(\frac{100+1.5}{120+1.5}\right)^2$$

$$= (0.846)^2 / (0.835)^2$$

$$= 0.7160 / 0.6979$$

$$= 1.0256$$

$P(f_2,d) = MP(f_1,d) = (1.0256)(72.1\%) = 74.0\%$

7-9. Tissue-Phantom Ratio and Tissue-Maximum Ratio

As alluded to earlier, the tissue-phantom ratio was introduced to circumvent the issue of the inadequacy of TAR for high-energy photon beam. The TAR requires a large buildup cap, which is impractical and also cannot be fully irradiated for small treatment fields. It replaces measurements in-air with measurements in a reference phantom. The **tissue-phantom ratio (TPR)** at any depth in phantom is defined as the ratio of dose at that depth in a phantom to the dose in reference phantom at the same distance from the source. The equation for TPR is

$$TPR(d, A_{f+d}) = \frac{D_T(d, A_{f+d})}{D_P(d'', A_{f''+d''})} \quad (7\text{-}19)$$

where D_P (d", $A_{f''+d''}$) represents the dose at fixed reference depth d" with beam area, $A_{f''+d''}$ in a reference phantom and D_T (d, A_{f+d}) is the dose at the depth d with beam area, A_{f+d} in tissue (represented by the phantom). We should note that the beam area is the same just a different notation since both points X and X″ are at the same distance from the source. The field size for TPR is defined at depth just like TAR. The TPR is also independent of SSD since the two measurement points are at the same distance from the source. As such, we say that the concept of TPR is an extension of TAR. Since the value of TPR can be greater than one, this concept of TPR is favored because the dm varies with field size. Most often a reference depth is selected and measured doses are normalized to the dose at that reference depth.

If the reference depth in the reference phantom is chosen to be the depth of maximum dose, the TPR is given a special name tissue-maximum ratio (TMR). The TMR is a special case of TPR introduced by Holt et al.[12] The equation for TMR is

$$TMR(d, A_{f+d}) = \frac{D_T(d, A_{f+d})}{D_P(dm, A_{f''+dm})} \tag{7-20}$$

It is instructive to examine how TMR is related to TAR. From Figure 7-1, we see that TMR is the ratio of the dose at point X to the dose at point X" as

$$\begin{aligned} TMR(d, A_{f+d}) &= \frac{D_X}{D_{X''}} = \frac{D_X}{D_{X'}} x \frac{D_{X'}}{D_{X''}} \\ &= TAR \; x \; \frac{1}{BSF} = \frac{TAR(d, A_{f+d})}{BSF(dm, A_{f''+dm})} \end{aligned} \tag{7-21}$$

We see that the TMR is related to TAR through BSF. Like TAR, TMR is independent of SSD. The TMR increases with increasing photon beam energy and increasing field size as demonstrated in Figure 7-6.

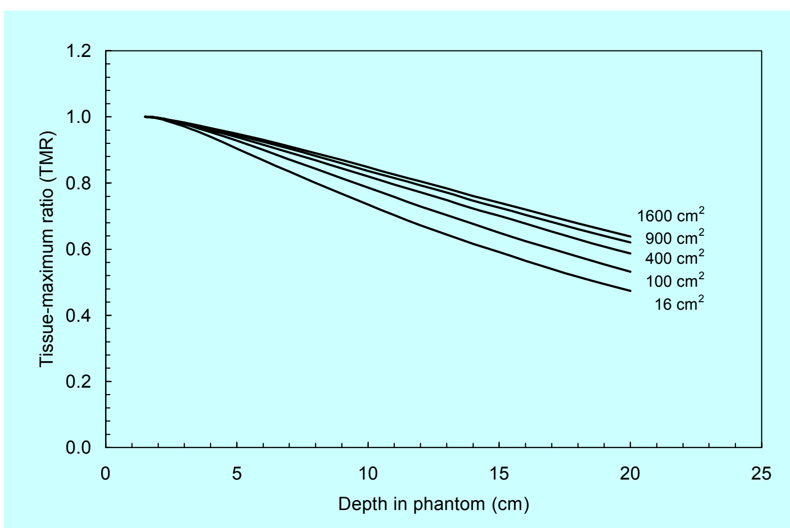


Figure 7-6. The variation of TMR as a function of depth for different field size

[12] Holt JG, Laughlin JS, Moroney JP. The extension of the concept of tissue-air ratios (TAR) to high energy x-ray beams. Radiol 1970: 96; 437

For 45 MV photon beam, the maximum dose for TMR value was found to be 2 cm deeper than the maximum dose for percent depth dose.[13] This depth of maximum dose is also independent of field size.

EXAMPLE 7-10. A 4 MV photons beam was calibrated using a 10 cm x 10 cm field size at a depth of 5 cm in phantom at 80 cm SAD. The measurement was 92 cGy for 100 MU. What was the output at dm if the TMR is equal to 0.904? Additional data provided were $C_\lambda = 0.94$, T = 22 C, P = 775 mmHg, $C_f = 1.05$.

SOLUTION:

The dose at 5 cm deep is

$$D = M \cdot C_{TP} \cdot C_f \cdot C_\lambda$$

$$= (92\ cGy / 100\ MU)(0.987)(1.05)(0.94)$$

$$= 0.896\ cGy / MU$$

The dose at dm is

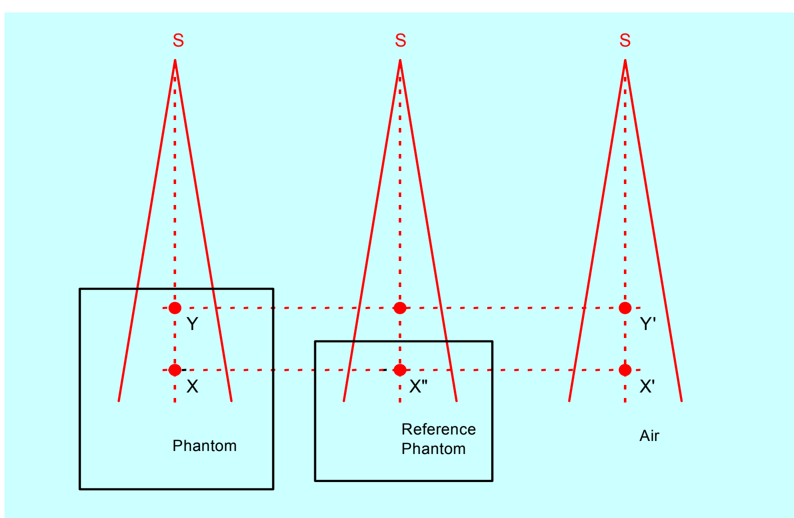
$$D(dm,10x10) = \frac{D(d,10x10)}{TMR}$$

$$\frac{0.896\ cGy / MU}{0.904}$$

$$= 0.991\ cGy / MU$$

7-10. Tissue-Maximum Ratio and Percent Depth Dose Relationships

Instead of TAR, tables of TMR are provided for dosimetric calculation of high-energy photon beams. The TMR is related to the depth dose as shown below in Figure 7-7.



13 Suntharalingam, 1977

Figure 7-7. Relationship between TPR and percent depth dose.

Using the definition of TMR, we set the dose points to involve the depth dose as follows:

$$TMR(d,A) = \frac{D_X}{D_{X''}} = \frac{D_X}{D_Y} \times \frac{D_Y}{D_{Y'}} \times \frac{D_{Y'}}{D_{X'}} \times \frac{D_{X'}}{D_{X''}}$$

Analyzing the terms, we find that the first term is the depth dose, the second is the backscatter factor, the third is the inverse square and the last is the backscatter factor at depth. Substituting each term into the above equation, we get

$$TMR(d, A_{f+d}) = \frac{P(d, A_{f+dm})}{100} \times BSF(dm, A_{f+dm}) \times \left(\frac{f+d}{f+dm}\right)^2 \times \frac{1}{BSF(d, A_{f+d})} \qquad (7\text{-}22)$$

Analysis of the equation (7-22) shows that the field size of the two backscatter factors have to be evaluated at two distances f+dm and f+d from the source. As such, they should have different values. However for high energy photon beams, each backscatter factor is close to unity and as such, they can be dropped from the equation reducing it to the relationship involving only the depth dose and inverse square functions.

7-11. Rotational Field Calculations

The concept of TAR is most useful for performing dose calculations that involves rotational field technique. Rotational or arc therapy is a type of isocentric irradiation where the source moves continuously around the patient. The MU for the treatment is determined by taking the prescribed dose to the target volume divided by the average TAR.

$$MU = \frac{D}{TAR} \times \frac{1}{S_{cp}} \qquad (7\text{-}23)$$

where D is the prescribed dose and S_{cp} is the collimator and phantom scatter factor. As a general rule, rotational field does not involve the use of tray or transmission factor. Depending on the design of a plan, wedge may be used but rarely.

The average TAR is determined by initially taking patient contour and drawing radial segments out from the isocenter as illustrated in Figure 7-8.

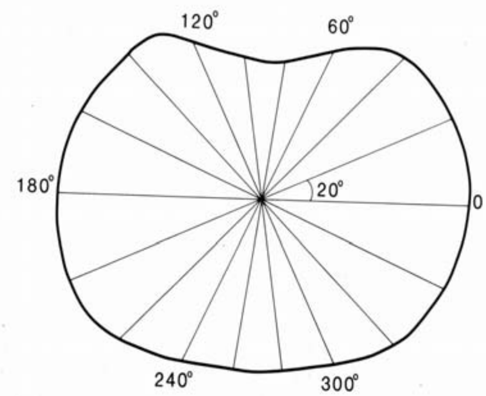


Figure 9.10. Contour of patient with radii drawn from the isocenter of rotation at 20° intervals. Length of each radius represents a depth for which TAR is determined for the field size at the isocenter. (See Table 9.3.)

Figure 7-8. Radial segment in dose calculation for rotational therapy.

Typically the radial segments are created at regular angular interval of 10 degree to intersect the patient contour. The distance from this intersection to the isocenter is measured and used as the depth to identify the TAR value from the TAR table for a given field size. The average TAR is determined by adding all the TAR values divided by the total number of segments. It should be noted that the outcome would be different if the average distance is determined first and then looking up the TAR from the lookup tables.

EXERCISE 7-6. Use an example to show that the averaging TAR results in a different value from averaging the distances and then looking up the TAR value for a given field size.

7-12. Blocked Field Calculations

The need to shield normal tissue from radiation leads to the introduction of blocks. The advantage of blocks is to permit field shaping, and thereby limits the delivery of radiation dose to only the target volume and its surrounding margin. When a block is introduced into a radiation beam, the primary beam is not affected, but the scattering condition is greatly modified. The modification of the scattering component affects the dosimetric functions such as TAR or depth dose. Various methods have been used to account for the proportion of scattered contributions from the reduced field. These methods are based on the proportion of open field areas to the collimation size.

A simple method of determining the effective TAR is to derive the proportionality on the SAR, which is affected by blocks. The procedure is to compute the amount of scatter from the reduced field and added to the TAR_o to obtain the effective TAR. The proportional amount of SAR is determined based on the fraction of open area. This technique is illustrated in EXAMPLE 7-11 for a simple half-beam block field depicted in Figure 7-9.

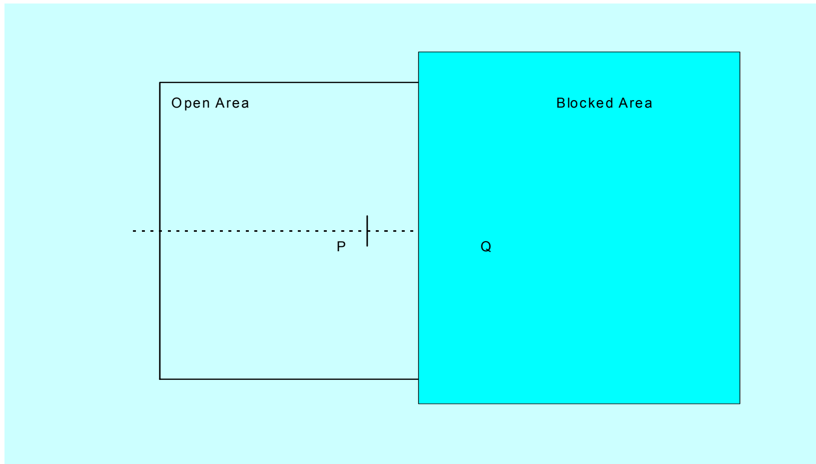


Figure 7-9. Determination of SAR of half-beam block field.

EXAMPLE 7-11. Assume that we have to determine the dose at point P, 10 cm deep about 2 cm from edge of the block. The field size used is 10 cm x 20 cm, half of which is blocked to produce an open field of 10 cm x 10 cm for a cobalt beam. What is the effective TAR?
SOLUTION:

TAR (d = 10, A = 10x20) is 0.760 from TAR table
TAR_o(d = 10) is 0.536 from TAR table.

$$SAR(10,10x20) = TAR(10,10x20) - TAR_O(10)$$
$$= 0.760 - 0.536$$
$$= 0.224$$

Since the open field is only half after the block, we divide the SAR by half to account for half the scatter component of the total open field. The effective TAR becomes

$$TAR(10,10x10) = TAR_O(10) + SAR(10,10x20)/2$$
$$= 0.536 - 0.224/2$$
$$= 0.563 + 0.112$$
$$= 0.675$$

The dose at a point under a block is determined in a similar fashion as the dose at a point in the beam. The SAR is determined in the same way as used for determining the dose in irregular fields. However, the primary beam is attenuated by the thickness of the block. The block transmission is typically between 3% to 5%. The effective TAR is computed as shown in EXAMPLE 7-12.

EXAMPLE 7-12. Compute the effective TAR for the point Q under a half-beam block as depicted in Figure 7-9.

SOLUTION:

Assume the block transmission is 3%, the TAR becomes

$$TAR(10,10x20) = TAR(10,0) + SAR(10,10x20)$$
$$= (0.536x0.03) + 0.112$$
$$= 0.128$$

The above method is crude since the change in SAR is not directly proportional to the area of the open field as shown in Figure 7-4.

EXERCISE 7-7. Explain why the above method is crude. What does it meant by the statement that the change in SAR is not directly proportional to the area of open field?

Another method of accounting for the scattered component is to determine the effective field size of the reduced open field. The reduced open field area is determined by computing the open field area minus the block area. The side of the square of effective field size is determined by taking the square root of the difference. This side of the square is used to obtain the effective TAR as demonstrated in EXAMPLE 7-13.

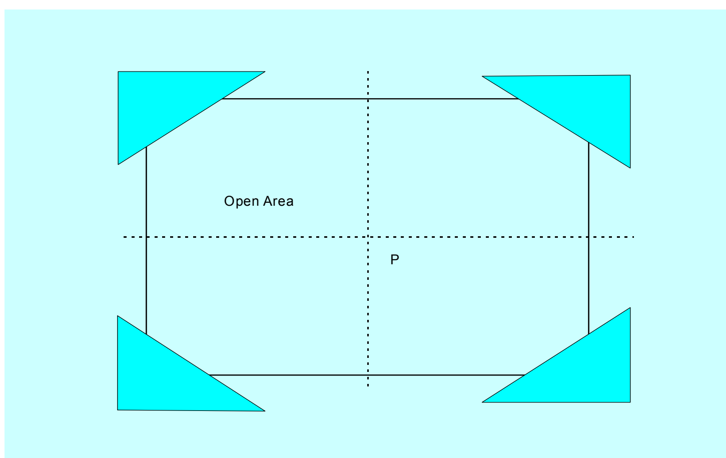


Figure 7-10. Block field using four corner blocks.

EXAMPLE 7-13. Compute the equivalent square given that a field is blocked at four corners as shown in Figure 7-10. The field size is 10 cm x 20 cm, and each triangular area is 1.5 cm x 2.5 cm adjacent sides.

SOLUTION:

The total block areas given as

$$A_{block} = 4 \cdot (\frac{1}{2} x\, 1.5 \,x\, 2.5)$$
$$= 4 \cdot (1.875)$$
$$= 7.5\ cm^2$$

The open area of blocked field is

$$\begin{aligned} A_{diff} &= A_{open} - A_{block} \\ &= 10x20 - 7.5 \\ &= 200 - 7.5 \\ &= 192.5\ cm^2 \end{aligned}$$

The side square of the open area of the blocked field is

$$x = \sqrt{A_{diff}}$$

$$= 13.9\ cm$$

As a general rule, if the blocked area is small compared to the total unblocked, the change in the scattered component is small.

7-13. Irregular Field Calculations

The above two methods of computing the proportional amount of reduced scattered are less sophisticated as the **Clarkson method**. The Clarkson method also called **irregular field calculation** method is based on the relation of scattered component to the beam area.

In the Clarkson approach, a beam area is divided into pie shape slices or segments radiating from the point at which the dose is to be determined. The average radius of each segment is used to determine the scatter-air ratio (SAR). The average SAR is determined by summing all the SAR values divided by the total number of segments. The TAR for the shape field is determined as the sum of TAR(d,0) and average scattered-air ratio as

$$TAR(d, A) = TAR(d,0) + \frac{\Delta\theta}{2\pi}\sum_{i=1}^{N} SAR(d, A_i) \qquad (7\text{-}24)$$

where $\Delta\theta$ is the angle subtended by each pie segment. Equation (7-24) may look cumbersome but as stated above, the average SAR is the sum of all the SAR divided by the total number of segments.

As an example, we examine the mantle field where the pie segments involve both open and block field as shown in Figure 7-11.

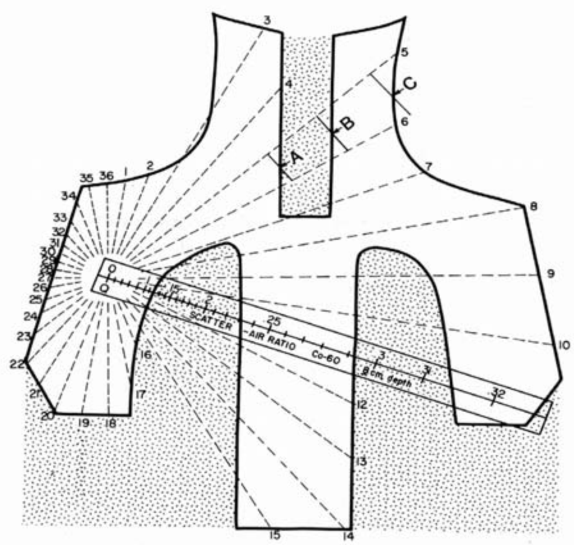


Figure 9.11. Outline of mantle field in a plane perpendicular to the beam axis and at a specified depth. Radii are drawn from point Q, the point of calculation. Sector angle = 10°. Redrawn from American Association of Physicists in Medicine. Dosimetry workshop: Hodgkin's disease. Chicago, IL, MD Anderson Hospital, Houston, TX, Radiological Physics Center, 1970.

Figure 7-11. Application of Clarkson Method to a mantle field [From Johns and Cunningham text.

Here, the dose to point Q is to be determined. From point Q, the field is divided into pie segments. In each pie segment, the resultant SAR for each segment that pass through a block and open area is equal to the sum of SAR of open area minus the block area. For example, the SAR for segment C is equal to $SAR_{radius=OC}$ - $SAR_{radius=OB}$ + $SAR_{radius=OA}$. The effective TAR is computed as

$$(TAR)_{average} = TAR_o + SAR_{average} \qquad (7\text{-}25)$$

Since the point Q is not on the beam axis. The average TAR should be multiplied by an **off-axis factor (OAF)**. This factor accounts for the beam profile at point Q. As discussed in Chapter 6, the beam profile is normalized to the beam axis. Once the TAR is determined, it can be transformed using the dosimetric functions discussed in Section 7-8 to depth dose for use with fixed treatment distance technique.

7-14. Time and Monitor Unit Calculation Techniques

Meterset (times or monitor units) calculations relating the absorbed dose at a point in a patient to machine settings are performed using the dosimetric functions. If the radiation machine is turned on for the time set on the timer of a Cobalt-60 unit or the monitor units (MU) set primary monitor of a linear accelerator, the machine is expected to deliver the desired dose to the designated point in the patient under the treatment condition. These machine settings are derived based on machine data that have been collected at nominal SSD. Typically the percent depth doses as a function of field size are measured in water with its surface at nominal SSD. Although it is possible to measure TAR or TPR, these dosimetric parameters are typically extracted from the percent depth dose data through mathematical manipulations. The properties of these parameters have been discussed in details in the previous sections. Besides the percent depth dose, other dosimetric data such as the field size factor and beam modify factors are also measured. Lastly, the treatment unit has to be calibrated to relate a machine unit (MU or time) to the dose delivered to a calibration point in phantom or in air.

The **field size factor** or collimator and phantom scatter factor (S_{cp}) refers to the relative change of radiation output as a function of field size. The radiation output is normalized to a 10-cm x 10-cm field. The field size factor can be separated into a **collimator scatter factor (S_c)** and a **phantom scatter factor (S_p)**. The separation is represented as the product,

$$S_{cp}(A) = S_c(A) \times S_p(A) \tag{7-26}$$

for a given field size area A. The collimator scatter factor is a relative value normalized to a 10-cm x 10-cm field relates the scattered radiation produced from the inside of the radiation-producing machine. On the other hand, phantom scatter factor is also a relative value normalized to a 10-cm x 10-cm field size. It represents the scattered radiation from the phantom or patient tissue. Although it is relatively easy to conceptually separate the scattered component into S_c and S_p, the practical measurements of these factors are difficult. A typical method of obtaining these values is to measure S_{cp} and S_c. The S_{cp} is measured with full phantom for a range of field sizes. However, the measurement of S_c requires a buildup cap. Depending on the size of the buildup cap, the values of S_c can change significantly. Again, it is a problem to measure S_c for small field sizes. Once both parameters are measured for a range of field sizes, the phantom scatter is obtained by dividing S_{cp} into S_c. This scatter formalism is only useful when the treatment involves the use of blocks that set the open area much smaller that the collimator settings. Under such situation, the collimator scatter factor will depend on the collimator setting while the phantom scatter will depend on the open field of the block.

For Cobalt-60 machine, the **output factor** is used instead of the field size factor. The output factor is the product of the field size factor and the machine output (cGy/min) as a function of field size. Thus output factors incorporate both dose rate and field size effects.

Beam modifying factors are basically **transmission or attenuation factors** of materials placed between the source and the patient. These materials include tray, compensator, wedge, and attenuator. The beam modifying factors are used to adjust for reduced transmitted beam intensity. These factors are typically measured and expressed as a ratio of intensity with to without the modifier for a particular field size. Recently these factors have been found to vary with field sizes and as such have been implemented.

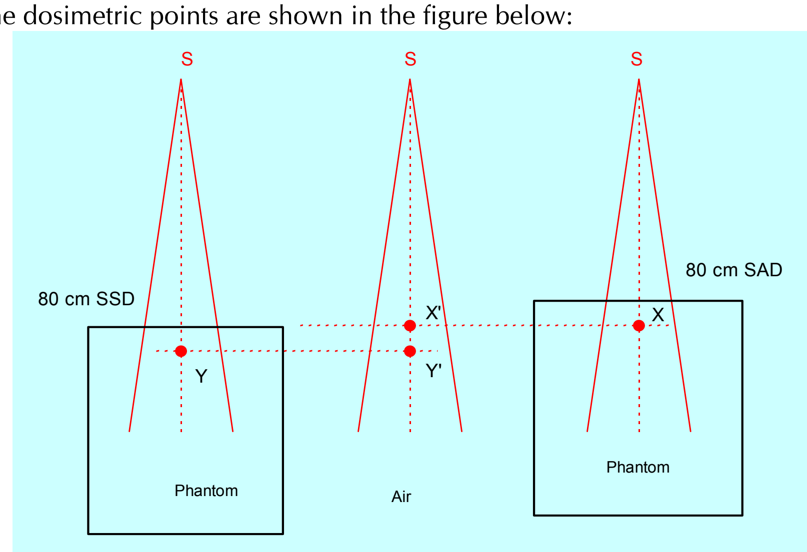
EXERCISE 7-8. Explain how you would measure the transmission factor for a (a) solid tray, (b) slotted tray, and (c) tray with star pattern holes.

Machine calibration refers to the process of specifying the dose delivered per unit time or monitor unit (MU). Since the radiation intensity vary with distance and field size, the geometric setup for calibration has to be clearly defined including the field size and the point of measurement called **reference point or calibration point**. This point is sometimes referred to as source-to-chamber distance (SCD) point. It should be noted that this point might not be the point where the ion chamber response is collected. For example, the calibration point is usually at 1.5-cm depth with 100 cm SSD for a 6 MV photon beam. However, the ion chamber response is measured at 5 cm or 7 cm deep. The field size is typically set at 10-cm x 10-cm called reference field size. The calibration point is typically set at SSD+dm from the source. For a Cobalt-60 unit with 80-cm SSD, the machine output is specified in air at the calibration point of 80 cm (isocenter). For 100-cm SSD linear accelerator, the calibration point is typically set at SSD+dm. Here, the phantom surface is typically set at nominal source-to-surface distance (SSD). The conversion of ion chamber response at a defined depth to the dose at the calibration point is carried out following the AAPM TG51 calibration protocol.

EXAMPLE 7-14. An accelerator is calibrated at a 1.0 cm deep in full scatter phantom at 80 cm SSD. If the measured dose is 125.7 cGy/100 MU, find the dose/100 MU at the 80 cm SAD for isocentric treatment. The backscatter factor is 1.040.

SOLUTION:

The dosimetric points are shown in the figure below:



The problem here is the find the dose at point X given the dose at point Y. To do that we set the dose point relationships as

$$D_X = \frac{D_X}{D_{X'}} \times \frac{D_{X'}}{D_{Y'}} \times \frac{D_{Y'}}{D_Y} \times D_Y$$

$$= BSF \times \left(\frac{80+1}{80}\right)^2 \times \frac{1}{BSF} \times D$$

$$= \left(\frac{80+1}{80}\right)^2 \times 125.7 cGy/100MU$$

$$= 128.9\ cGy/100MU$$

Patients are typically set up for either the SSD treatment technique or SAD treatment technique. For single field treatment, SSD technique is often used. For a treatment that uses radiation beams from a number of gantry angles, the SAD technique is preferred. The calibration point for a linear accelerator is typically chosen as nominal SSD+dm for any photon beam. The meterset calculations shall be derived based on the above setup conditions.

For **SSD treatment technique**, the patient's skin is positioned at a fixed distance from the source. The machine settings can be computed using the formalism described in Figure 7-12 following through the depth dose function.

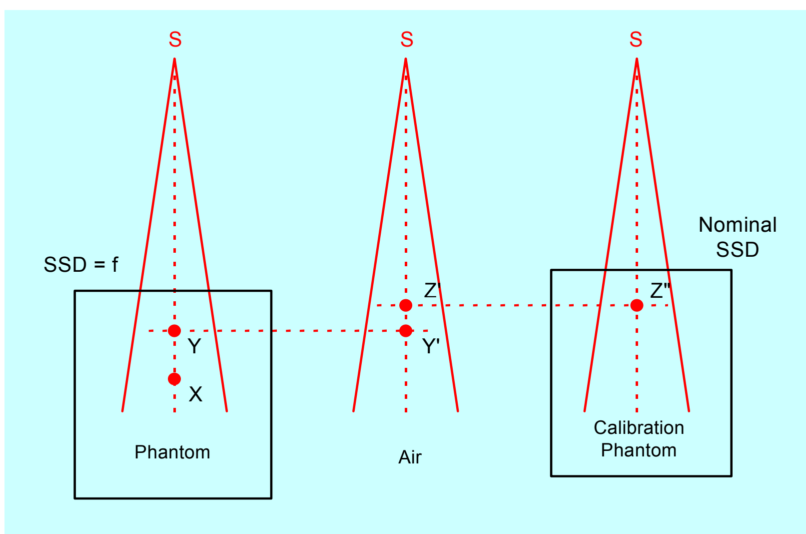


Figure 7-12. SSD setup to calibration point via depth dose formalism.

Assume that the machine is calibrated at point Z″ with source-to-calibration distance denoted as $SCD = SSD_N + dm$ where SSD_N is the nominal SSD. The dose at this point for a 10-cm x 10-cm field is adjusted to be 1 cGy per MU. A path relating the prescribed dose (D) at any point in the patient to the dose at this calibration point Z″ will allow the determination of monitor units (MU) or time to be set on the treatment unit. By setting and delivering the set MUs, the machine will deliver the prescribed dose to the designated location at point X. The relationship between the dose at any point X to the calibration point Z" can be written as follows:

$$D_{Z"} = \frac{D_{Z"}}{D_{Z'}} \times \frac{D_{Z'}}{D_{Y'}} \times \frac{D_{Y'}}{D_{Y}} \times \frac{D_{Y}}{D_{X}} \times D_X \qquad (7\text{-}27)$$

We can see that the first ratio is the peakscatter factor, the second ratio is the inverse square, the third ratio is the peakscatter factor and the last ratio is the depth dose. Substituting these terms into the above equation we have

$$D_{Z"} = PSF(dm, A_{SCD}) \times \left(\frac{SSD + dm}{SCD}\right)^2 \times \frac{1}{PSF(dm, A_{SSD+dm})} \times \frac{1}{P/100} \times D \qquad (7\text{-}28)$$

where SSD refers to the source-to-surface distance of the patient. If the SSD is the same as nominal SSD, then we can set $SCD = SSD + dm$. This results in the cancellation of the peakscatter factor terms and the inverse square terms. The above equation reduces to

$$D_{Z"} = \frac{1}{P/100} \times D \qquad (7\text{-}29)$$

Equation (7-29) is simple indicating that percent depth dose tables should be used for fixed distance treatment technique when the machine is calibrated with the MU specified at $SCD = SSD + dm$.

The dose at point X for any arbitrary field size is related to the dose of a calibrating field size, 10 cm x 10 cm through the scatter factor, i.e. S_{cp} and the transmission factors. The dose is equal to the monitor units adjusting for the effect of field size as

$$MU = \frac{D}{P/100} \times \frac{1}{S_{cp}} \times \frac{1}{WF \times TF} \qquad (7\text{-}30)$$

where WF is the wedge transmission factor and TF is the tray transmission factor. The first term, the prescribed dose D divided by the depth dose is the given dose (GD) or incident dose (ID). The S_{cp} is the scatter factor.

EXAMPLE 7-15. Compute the monitor unit required to deliver 200 cGy to a point 9.5 cm deep in a phantom from a 4 MV linear accelerator. A tray with tray factor of 0.961, and a plate with a factor of 0.988 were used. The percent depth dose at a depth of 9.5 cm for this field size is 58%.

SOLUTION:

The monitor units is

$$MU = \frac{D_T}{P/100} \times \frac{1}{TF \times PF}$$

$$= \frac{200}{0.58} \times \frac{1}{0.961 \times 0.988}$$

$$= 363\ MU$$

Instead of MU, the timer (t) is required from a Cobalt-60 beam. The timer can be calculated as follows:

$$(t + \varepsilon) = \frac{D}{P/100} \times \frac{1}{O \times BSF} \times \frac{1}{S_{cp}} \times \frac{1}{WF \times TF} \qquad (7\text{-}31)$$

where O is the machine output specified in air expressed in cGy/min. The BSF converts dose-in-air to dose-in-phantom. The output of a Cobalt unit is also specified using the reference field size of 10 cm x 10 cm. A **timer error** ε, discussed in Section 2-11 has to be added to the timer to obtain the correct treatment time.

EXAMPLE 7-16. What is the time required to deliver 200 cGy to a tumor from an anterior field. Given that the percent depth dose is 70%, machine output is 85 R/min, backscatter factor is 1.02, and f-factor is 0.96.
SOLUTION:

Using equation (7-31), the treatment time is

$$t = \frac{TD}{P/100} \times \frac{1}{O \times BSF}$$

$$= \frac{200\ cGy}{.7} \times \frac{1}{(85\ R/min)(0.96\ R/cGy)(1.02)}$$

$$= 3.43\ min$$

For **SAD treatment technique**, the tumor or target volume is positioned at the isocenter. The dose at any point X in the patient (phantom) can be related to the calibration point Z″ through the TAR function as shown in Figure 7-13.

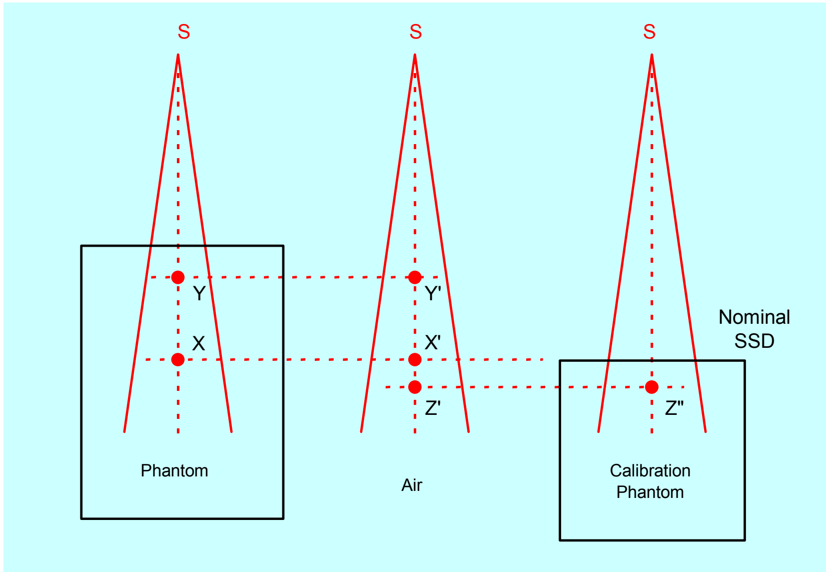


Figure 7-13. SAD setup relating to calibration point.

$$D_{Z''} = \frac{D_{Z''}}{D_{Z'}} \times \frac{D_{Z'}}{D_{X'}} \times \frac{D_{X'}}{D_X} \times D_X$$

Again, we identify the first term as the backscatter factor, the second as the inverse square, and the third as the TAR. Substituting these terms into the above equation, we have

$$D_{Z''} = BSF(dm, A_{SCD}) \times \left(\frac{SSD + d}{SCD}\right)^2 \times \frac{1}{TAR(d, A_{f+d})} \times D$$

The distance of point X' is determined by noting that it is at the same distance as point X. The distance of point X' from the source would be SSD + d, where SSD refers to the source to surface distance of the phantom.

The dose at point X for any arbitrary field size is related to the dose of a calibrating field size, 10 cm x 10 cm through the scatter factor, i.e. S_{cp} and the transmission factors. The dose is equal to the monitor units adjusting for the effect of field size as

$$MU = \frac{D}{TAR} \times BSF \times \frac{1}{S_{cp}} \times \frac{1}{WF \times TF} \times \left(\frac{SSD + d}{SCD}\right)^2 \qquad (7\text{-}32)$$

Compared to equation (7-30), the inverse square is present in the meterset calculation for SAD treatment technique. The TAR is commonly used for Cobalt-60 beam and low energy photon beams.

EXAMPLE 7-17. A portal with a 45 degree wedge is used to deliver 100 cGy to a tumor positioned at the isocenter at 10 cm depth. Given the following data, machine output is 1.04 cGy/MU; dm = 1.0 cm; wedge factor is 0.7; backscatter factor = 1.03; tissue-air-ratio is 0.75. Compute the number of monitor units required for this treatment.
SOLUTION:

Use the TAR technique to relate the point in tissue to the point in air, apply inverse square correction, and then relates the point back into the calibration setup condition using the backscatter factor. Lastly compute the MU by dividing it with the output factor and corrects for wedge transmission.

$$MU = \frac{200}{.75} \times 1.03 \times \frac{1}{1.04} \times \frac{1}{0.70} \times \left(\frac{80}{81}\right)^2$$

$$= 368$$

The timer for a Cobalt-60 beam using TAR function is:

$$t = \frac{D}{TAR} \times \frac{1}{O \times S_{cp}} \times \frac{1}{WF \times TF} \times \left(\frac{SSD + d}{SCD}\right)^2 \quad (7\text{-}33)$$

where the machine output O is expressed as dose rate (cGy/min) in air.

EXAMPLE 7-18. An 80-cm SSD cobalt-60 teletherapy unit produces an output of 140 cGy/min in air at the isocenter. What is the treatment time if this unit is used to deliver 150 cGy to a point 10 cm deep into a patient using isocentric setup? A compensator, which reduces the intensity by 6% is also used. The TAR at 10 cm deep is 0.731.
SOLUTION:

Equation (7-33) will be used to compute the treatment time,

$$t = \frac{D}{TAR} \times \frac{1}{O} \times \frac{1}{Comp} \times \left(\frac{80}{80}\right)^2$$

$$= \frac{150\ cGy}{0.731} \times \frac{1}{140\ cGy/min} \times \frac{1}{.94}$$

$$= 1.53\ min$$

In this particular case, the machine is calibrated at the isocenter and hence the inverse square factor dropped off.

Even though the isocentric treatment technique has been decided for the treatment of a particular patient, some radiation oncologists would prefer to know the given dose. In addition, older treatment chart requires the recording of the given dose. The equation for the given dose can be derived using Fig. 7-1. In this figure, the given dose is at point Y. From Figure 7-1, we see that the dose (GD) at point Y can be related to the dose (D) at point X, which is the prescription point as

$$D_Y = \frac{D_Y}{D_{Y'}} \times \frac{D_{Y'}}{D_{X'}} \times \frac{D_{X'}}{D_X} \times D_x$$

$$GD = BSF(dm, A_{SSD+dm)}) \times \left(\frac{SSD+d}{SSD+dm}\right)^2 \times \frac{1}{TAR(d, A_{SSD+d})} \times D$$

Rearranging the terms, the given dose (GD) is related to the TAR as follows:

$$GD = D \times \frac{BSF(dm, A_{SSD+dm})}{TAR(d, A_{SSD+d})} \times \left(\frac{SSD+d}{SSD+dm}\right)^2 \qquad (7\text{-}34)$$

The backscatter factor is computed at a distance of (SSD + dm), while the TAR is computed at (SSD + d) from the source. These different distances affect the field sizes, which these dosimetric functions depended upon. Although it is not correct, the effect of field size is often ignored to simplify the calculations.

EXAMPLE 7-19. Compute the given dose based on the problem in EXAMPLE 7-18 assuming the field size is 10 cm x 10 cm.
SOLUTION:

Using equation (7-34), we have

$$GD = D \times \frac{BSF}{TAR} \times \left(\frac{SAD}{SAD - d + d\max}\right)^2$$

$$= 150\text{ cGy} \times \frac{1.054}{0.731} \times \left(\frac{80}{80 - 10 + 0.5}\right)^2$$

$$= 216.28\text{ cGy} \times \left(\frac{80}{70.5}\right)^2$$

$$= 278.5\text{ cGy}$$

For high-energy photon beam with SAD treatment setup, the TPR dosimetric function is used in the meterset calculations. The dose at any point X in the phantom can be related to the calibration point Z″ through the TPR as shown in Figure 7-14.

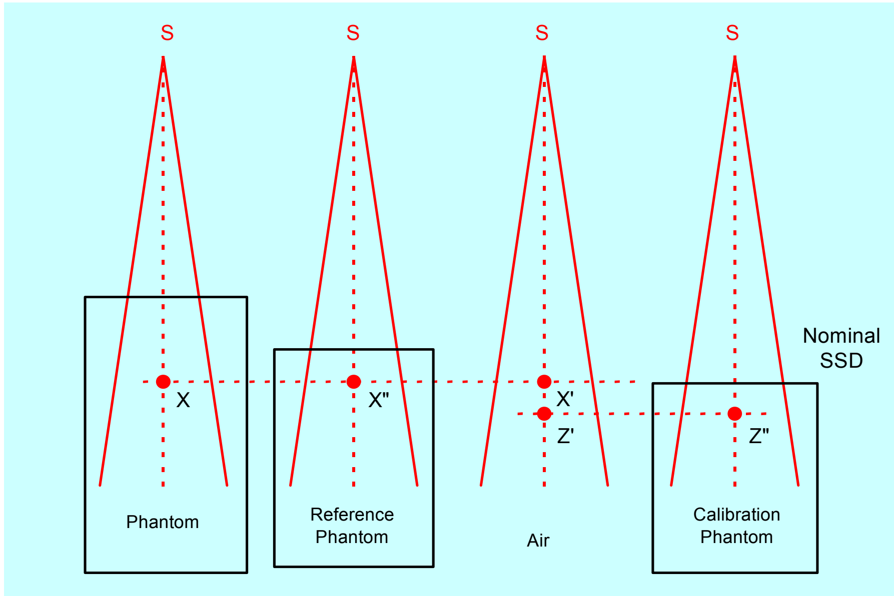


Figure 7-14. SAD setup using TPR formalism.

The dose point relationships are given as

$$D_{Z''} = \frac{D_{Z''}}{D_{Z'}} x \frac{D_{Z'}}{D_{X'}} x \frac{D_{X'}}{D_{X''}} x \frac{D_{X''}}{D_X} x\, D_X$$

We proceed by identifying the individual terms. The first term is the peakscatter factor, the second is the inverse square, the third is the peakscatter factor, and the forth is the TPR terms. After substituting in these terms, we have

$$D_{Z''} = PSF(dm, A_{SCD})\, x \left(\frac{SSD+d}{SCD}\right)^2 x \frac{1}{PSF(d, A_{SSD+d})} x \frac{1}{TPR(d, A_{SSD+d})} x\, D$$

This equation can be reduced if the peakscatter factors cancel other out. Actually, the peakscatter factors have different values due to the field size effect evaluated at different distances from the source. However, their difference of peakscatter factor due to the effect of field size is small and can be ignored.

The dose at point X for any arbitrary field size is related to the dose of a calibrating field size, 10 cm x 10 cm through the scatter factor, i.e. S_{cp} and the transmission factors. The dose is equal to the monitor units after adjusting for the effect of field size given as

$$MU = \frac{D}{TPR} x \frac{1}{S_{cp}} x \frac{1}{WF \text{ x } TF} x \left(\frac{SSD+d}{SCD}\right)^2 \qquad (7\text{-}35)$$

Again the inverse square factor is present. We should realize by now that the inverse square term reflects the way the machine is calibrated. If the machine is calibrated at the isocenter, the inverse square term will drop off. However, the meterset calculation for the SSD treatment technique has to be adjusted to reflect the calibration point is at the isocenter.

7-15. Point Dose Calculations

In some situations, it may be required to record the doses at other locations within a patient. Often this is the request of the radiation oncologists if the treatment involved irradiating areas near or through the spinal cord. For example, the dose is prescribed to the mid-plane of a parallel-opposed field while the dose to the spinal cord should be computed and recorded. The doses at other locations can be derived from either the prescribed dose or machine settings (time or MU). EXAMPLE 7-19 demonstrates that the given dose can be derived from the prescribed dose.

The dose (D_Q) at any point Q in a phantom can be related to another point X, also in the phantom through the three dosimetric functions, depth dose, TAR and TPR. The geometric relationship between the dose points is shown in Figure 7-15.

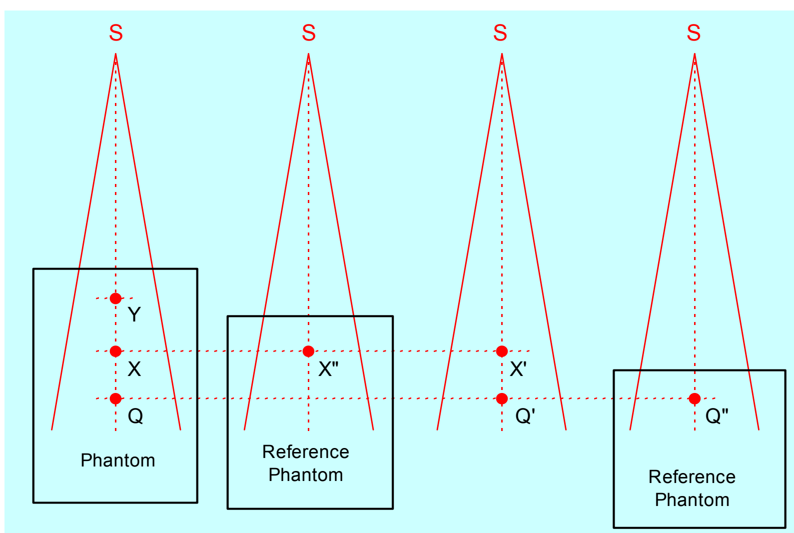


Figure 7-15. Geometric points relating to dosimetric functions.

A method of determining the dose at any point Q in a phantom based on the prescribed dose at point X, also in the phantom is through the use of the depth dose function. Referring to Figure 7-15, the dose point relationship can be set as

$$
\begin{aligned}
D_Q &= \frac{D_Q}{D_Y} x \frac{D_Y}{D_X} x D_X \\
&= (P/100)_Q x \frac{1}{(P/100)_X} x D_X \\
&= \frac{P_Q}{P_X} x D_X
\end{aligned}
\tag{7-36}
$$

This equation merely states the dose at point Q is related to the dose at point X via their ratio of their depth doses. The concern of this method is the need to determine the depth dose at different SSDs, which is different from the measurements made at nominal SSD. The depth dose at SSD other than the nominal SSD is obtained with the use of Mayneord factor.

EXAMPLE 7-20. A single field irradiation using a 23 MV photon beam was created to deliver 100 cGy to a depth of 10 cm. Find the dose at 15 cm deep using the percent depth dose method.

Parameters	Value	Value
Nominal SSD	100 cm	
Setup SSD	100 cm	
Field Size	15 cm x 15 cm	
Field Size Factor	1.034	
Depth	10 cm	15 cm
P	80.38%	66.35%
TPR	0.908	0.819
Prescribed Dose	100 cGy	
MU	120	

SOLUTION:

Using equation (7-36), we got

Dose at 15 cm deep is

$$
\begin{aligned}
D_{15} &= \frac{P_{15}}{P_{10}} x\ D_{10} \\
&= \frac{0.6635}{0.8038} x\ 100\ cGy \\
&= 82.5\ cGy
\end{aligned}
$$

From Figure 7-15, the dose at point Q can be related to the dose at point X via the TAR function. The relationship can be setup by moving from point Q, to point Q′, to point X′ and then to point X. The first relationship is the TAR and the next the inverse square and then the last relationship is BSF. If this process is carried out, the dose at point Q is given as

$$D_Q = \frac{TAR(d_Q, A_{SSD+dQ})}{BSF(d_X, A_{SSD+dX})} x \left(\frac{SSD + d_X}{SSD + d_Q} \right)^2 xD_X \quad (7\text{-}37)$$

We can also determine the dose at point Q relative to the dose at point X using the TPR function. From Figure 7-15, the relationship can be setup as moving in sequence from point Q, Q″, Q′, X′, X″, and then X. The dosimetric functions used in sequence from point Q are TPR, BSF, I, BSF, and finally TPR. Although the manipulation may look cumbersome, the equation can be simplified by noting that the backscatter function is near unity for high-energy photon beam. Under such condition, the equation reduces to

$$D_Q = \frac{TPR(d_Q, A_{SSD+dQ})}{TPR(d_X, A_{SSD+dX})} x \left(\frac{SSD + d_X}{SSD + d_Q} \right)^2 x\, D_X \quad (7\text{-}38)$$

where d_Q is the depth to point Q, d_X is the depth to point X.

EXAMPLE 7-21. In EXAMPLE 7-20, use the TPR method to find the dose at 15 cm deep.
SOLUTION:
Using equation (7-38), we got

$$D_{15} = \frac{TPR_{15}}{TPR_{10}} x \left(\frac{100 + 10}{100 + 15} \right)^2 x\ 100\ cGy$$

$$= \frac{0.819}{0.908} x \left(\frac{110}{115} \right)^2 x\ 100\ cGy$$

$$= 0.902\ x\ 0.915\ x\ 100\ cGy$$

$$= 82.5\ cGy$$

The determination of the dose at a point Q in a phantom based on the time or monitor units requires the reversed formalism of equations (7-30) through (7-35).

For **SSD treatment technique** that involves the depth dose, the dose at any point Q in a phantom can be derived from the MU of a linear accelerator as

$$D = MU\ x\ \frac{P}{100}\ x\ S_{c,p}\ x\ (WF\ x\ TF) \quad (7\text{-}39)$$

EXAMPLE 7-22. In EXAMPLE 7-20, use the MU and depth dose function to find the dose at 15 cm deep.
SOLUTION:
Using equation (7-39), we got

$$D_{15} = 120 \text{ MU x} \frac{66.35}{100} \text{x } 1.034 \text{ x} \frac{1 \text{ cGy}}{\text{MU}}$$
$$= 82.3 \text{ cGy}$$

For **SSD treatment technique** that involves the Cobalt-60 unit and depth dose, the dose at any point Q in a phantom can be derived from the time setting as

$$D = (t \text{ x } O) \text{ x BSF x } \frac{P}{100} \text{ x } S_{c,p} \text{ x (WF x TF)} \qquad (7\text{-}40)$$

For **SAD treatment technique** that involves the TAR, the dose at any point Q in a phantom can be derived from the MU setting on a linear accelerator as

$$D = \text{MU x} \frac{\text{TAR}(d, A_{SSD+d})}{\text{BSF}(dm, A_{SCD})} \text{ x } S_{c,p} \text{ x (WF x TF) x} \left(\frac{\text{SCD}}{\text{SSD}+d}\right)^2 \qquad (7\text{-}41)$$

For **SAD treatment technique** that involves the TAR and cobalt-60 unit, the dose at any point Q can be derived from the time settings as

$$D = (t \text{ x } O) \text{ x TAR x } S_{c,p} \text{ x (WF x TF) x} \left(\frac{\text{SCD}}{\text{SSD}+d}\right)^2 \qquad (7\text{-}42)$$

In equations (7-40) and (7-42), O represents the cobalt-60 unit's output measured in air at reference point in cGy/min. The calibration point is generally chosen to be at $SCD = SSD + dm$. The latter point is considered the traditional method of calibrating a beam. For a 6 MV photon beam, the calibration point using the SSD method for a linear accelerator with 100 cm SSD would be 101.5 cm.

For **SAD treatment technique** that involves the TPR, the dose at any point Q can be derived from the MU of the linear accelerator as

$$D = \text{MU xTPR x } S_{c,p} \text{ x (WF x TF) x} \left(\frac{\text{SCD}}{\text{SSD}+d}\right)^2 \qquad (7\text{-}43)$$

All these relationships can be derived using Figures 7-12 to 7-14.

EXAMPLE 7-20. A single field irradiation using a 23 MV photon beam was created to deliver 100 cGy to a depth of 10 cm using SAD technique. Find the dose at 15 cm deep using the TPR method.

Parameters	Value	Value

Nominal SSD	100 cm	
Setup SSD	90 cm	
Field Size	15 cm x 15 cm	
Field Size Factor	1.034	
Depth	10 cm	15 cm
P	79.46%	66.85%
TPR	0.909	0.818
Prescribed Dose	90 cGy	
MU	89	

SOLUTION:

Using equation (7-43), we got

$$D_{15} = 120\text{ MU x } 0.818 \text{ x } 1.034 \text{ x } \left(\frac{103.5}{105}\right)^2 \text{ x } \frac{1\text{ cGy}}{\text{MU}}$$

$$= 90\text{ MU x } 0.818 \text{ x } 1.034 \text{ x } 0.9716 \text{ x } \frac{1\text{ cGy}}{\text{MU}}$$

$$= 74.0\text{ cGy}$$

The dose at depth of 15 cm can also be checked using depth dose or TPR function relative to the prescribed dose. Using either of these functions would give a dose of 73.5 cGy.

Summary

7-1. Equivalent square refers to a square field, whose scattering effects and depth doses are identical to the shaped field of interest.

7-2. Percent depth dose at any depth is the ratio of dose at that depth to dose at a reference point expressed as a percentage. Usually, the reference point is taken at the depth of maximum dose.

7-3. Inverse square law expresses the relationship of doses to two points in air to the inverse square of their distance from the source.

7-4. Tissue-air ratio (TAR) at any depth is the ratio of dose in phantom at that depth to the dose in air at the same distance from the source. Hence TAR is independent of SSD.

7-5. Backscatter factor (BSF) is a special case of TAR where the depth is set the depth of maximum dose.

7-6. Peakscatter factor (PSF) refers to ratio of the radiation output measured in a phantom to the radiation output measured in air with a buildup cap. The PSF becomes the BSF for low kilovoltage photon beam.

7-7. Scatter-Air Ratio is the scatter component of the TAR obtained by subtracting zero field TAR from the TAR of any field.

7-8. Tissue-phantom ratio (TPR) at any depth is the ratio of dose in a phantom at that depth to the dose in a reference phantom at a reference depth. Tissue-Maximum Ratio

(TMR) is a special case of TPR with the reference depth is the depth of maximum dose.

7-9. The TAR for a rotational field is obtained by taking the average TAR of radial segments at isocenter.

7-10. Irregular field calculation also called Clarkson summation is a method of deriving the TAR at a point by adding the zero-field TAR and the average SAR.

7-11. There are two types of treatment technique referred to as the SSD or Fixed field technique and the SAD or Isocentric field technique. SSD technique general uses percent depth dose table while SAD technique uses TAR, TPR, or TMR tables.

7-12. Field size factor as a function of field size is the relative output factors normalized to the value of a reference field size, a 10-cm x 10-cm

7-13. The source of scattered radiation can be from the collimator of a linear accelerator or from the medium. The amount of scatter from the collimator is represented by the collimator scatter factor and the amount of scatter from the medium or phantom is represented by the phantom scatter factor.

7-14. The output factor expressed in cGy/min of a Cobalt-60 unit represents the product of the field size factor and the output of the machine as a function of field size.

7-15. General transmission factors include the factors of the block tray, wedge, compensator and attenuator.

7-16. The calibration point refers to the point where the output of the machine is known in relation to machine settings (monitor units or timer).

7-17. Normally the output of a linear accelerator is calibrated at a distance of nominal SSD plus dm using a reference field size of 10-cm x 10-cm. For cobalt-60 unit, the calibration is specified in air expressed in cGy/min. For higher photon beam energy, the calibration is specified in water.

7-18. The dose at another point (Q) within a phantom can be derived relative to a prescribed dose point (X) using the depth dose, TAR and TPR functions.

$$D_Q = \frac{P_Q}{P_X} \times D_X$$

or

$$D_Q = \frac{TPR(d_Q, A_{SSD+dQ})}{TPR(d_X, A_{SSD+dX})} x \left(\frac{SSD + d_X}{SSD + d_Q} \right)^2 x \; D_X$$

7-19. The dose at any point (Q) in a phantom can also be derived from the machine settings (timer or MU) as:

1. SSD technque:

$$D = MU \times \frac{P}{100} \times S_{c,p} \times (WF \times TF)$$

2. SAD technique:

$$D = MU \times TPR \times S_{c,p} \times (WF \times TF) \times \left(\frac{SCD}{SSD + d}\right)^2$$

Study Guide

7-1. Define in your own words the following terms:

- (a) equivalent square
- (b) Sterling's formula
- (c) percent depth dose
- (d) inverse square law
- (e) tissue-air ratio
- (f) backscatter factor
- (g) tissue phantom ratio
- (h) tissue-maximum ratio
- (i) given dose
- (j) peak scatter factor
- (k) scatter-air ratio
- (l) arc therapy
- (m) off axis factor
- (n) field size factor
- (o) collimator scatter factor
- (p) phantom scatter factor
- (q) irregular field calculation
- (r) Clarkson summation
- (s) fixed distance technique
- (t) isocentric technique

7-2. Identify two conditions that must be met before an arbitrary field is said to be equivalent square field?

7-3. List the two necessary conditions for the inverse square law to be valid.

7-4. List the three factors that changes percent depth dose.

7-5. List three factors that affect TAR.

7-6. Under what condition would the TAR be equal to the BSF?

7-7. List three factors that affect the backscatter factor.

7-8. Explain why the backscatter factor cannot be less than one?

7-9. Identify four factors that affect TPR?

7-10. What is the difference between TPR and TMR?

7-11. Identify the major difference between percent depth dose and TPR, making percent depth dose suitable for SSD technique and TPR for SAD technique.

7-12. What is the effect of increasing SSD on a) percent depth dose, b) dmax, c) backscatter factor, and d) geometric penumbra.

7-13. Explain how to obtain TAR for a rotational field.

7-14. Explain how to calculate the TAR for a blocked field.

7-15. Explain how to calculate the TAR for a irregular field

7-16. What is the purpose of separating the field size factor into collimator scatter factor and phantom scatter factor?

7-17. The chamber term in equation (7-32) drops out if we set SCD = SSD + d_m. What should the distance of the ion chamber be positioned from the source if the SSD of a 6 MV linear accelerator is 100 cm?

7-18. Under what circumstance the BSF is used in the TAR dose calculation formalism for isocentric technique?

7-19. In a bilateral head & neck treatment using 6 MV photon beam with weightings of 2:1, which dose points should be documented in the chart.

7-20. Explain the calibration condition in which the inverse square term dropped off in the a) fixed distance technique and b) isocentric technique.

Problems

7-1. Use Table 7-1, find the equivalent square of 4 cm x 7 cm field size.

7-2. Use Sterling approximation to determine the equivalent square of a 8 cm x 15 cm field size.

7-3. What is the area of a field size at the patient surface if a field size of 5 cm x 12 cm was set at a depth of 10 cm for an isocentric treatment. The source-to-axis distance of the treatment unit is 100 cm.

7-4. A treatment was designed to setup at 100 cm SSD on a 6 MV linear accelerator. Instead, it was mistakenly setup at 102 cm SSD. What is the approximate error in dose delivered to Dmax. It is an overdose or underdose?

7-5. The SSD was changed from 80 cm to 85 cm for a treatment using cobalt-60 unit. If the dose rate in air is 110 rads/min at 80 cm, what is the dose rate at 85 cm?

7-6. Compute the timer of a cobalt-60 unit if a patient is treated to 250 cGy at 5 cm depth at isocenter using a field size of 10 cm x 10 cm. The TAR is 0.921 and machine output at isocenter is 195 cGy/min in a mini-phantom.

7-7. A patient was treated using a cobalt-60 unit to a depth of 5 cm at isocenter using a 8 x 8 cm field for 180 cGy. Use Table 7-2, compute the given dose, neglecting the effect of field size on BSF.

7-8. A patient was planned for treatment at 110 cm SSD. If the percent depth dose at 100 cm SSD is 87%, what is the percent depth dose at 110 cm SSD?

7-9. A 4 MV linear accelerator was calibrated at a depth of 5 cm in a water phantom at 80 cm SAD. A reading of 91 R was recorded for 100 monitor units. What is the output at dmax if the TMR = 0.904, C_λ = 0.94, T = 22.5 °C, P = 775 mmHg, and Cf = 1.057

7-10. A patient SS was treated 180 cGy per fraction to the para-spinal location behind the heart with the following settings. What would the MU be if the dose per fraction is increased to 200 cGy?

Description	PA	R Lat	L Lat
Energy	24 MV	24 MV	24 MV
Technique	SAD	SAD	SAD
Wedge	0	30 deg	30 deg
Field Size	12x16	12x16	12x16
%Block	42	46	46
MU	75	95	94
Isocenter Dose	76	52	52

7-11. What is the dose rate at 10 cm depth in tissue when the output of 80 cm SSD cobalt-60 teletherapy unit is 110 R/min at 80.5 cm. Given the BSF is 1.06, percent depth dose is 60% and f-factor is 0.96.

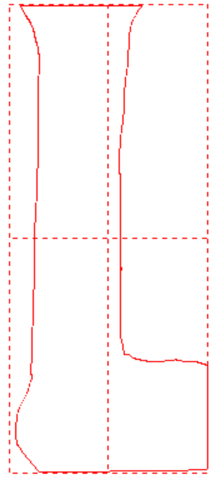
7-12. Compute the treatment time needed to deliver 200 cGy to the skin of a patient using orthovoltage unit. The cone size is 5 cm x 5 cm and BSF is 2.00. The machine output is 82 R/min and cGy/R is 0.75.

7-13. Compute the monitor units required to deliver 200 cGy to the midplane of a patient 15 cm thick using parallel opposed field SSD technique. The field size is 6 cm x 10 cm and percent depth dose is 50%. The calibration factor is 0.87 cGy per MU.

7-14. A 100 cm SSD technique was used to deliver 200 cGy to a depth of 9.5 cm in a patient using a 6 MV photon beam. The linear accelerator was calibrated such that it delivers 1 cGy per MU at isocenter. For this dose delivery, a collimator setting of 10 cm x 10 cm, a tray with tray factor of 0.971, and a wedge with wedge factor of 0.781 were used. The TMR at a depth of 9.5 cm for this field size is 0.766. Compute the monitor units needed for this dose delivery.

7-15. Compute the monitor units required to deliver a dose of 200 cGy at a depth of 9.5 cm from a 18 MV linear accelerator for an isocentric setup. The following setup was performed: a collimator setting of 10 cm x 10 cm; a tray with tray factor of 1.02, and a wedge with wedge factor of 3.69 were used. The TPR at a depth of 9.5 cm for this field size is 0.897. The linear accelerator was adjusted to deliver 1 cGy at isocenter per MU in a phantom.

7-16. A patient was treated AP/PA with jaw settings at 40 cm x 16 cm using the field outline as shown to the left femur isocentrically. A dose of 225 cGy was prescribed at the depth of 11 cm to the point marked with a green cross using 24 MV photon beam. Show that computed MU using hand calculation is 111 and irregular calculation is 110. [Assume a tray factor of 0.97 for 24 MV photon beam]. The same difference is due to the scatter contributions.



7-17. A single field irradiation was made using 6 MV photon beam with 12 cm x 12 cm field size. Find the dose at 5 cm if a dose of 100 cGy is prescribed to a depth of 10 cm.

Parameters	Value	Value
Nominal SSD	100 cm	
Setup SSD	90 cm	
Field Size	12 cm x 12 cm	
Field Size Factor	1.021	
Depth	10 cm	5 cm
Percent depth dose	66.43%	86.58%
TPR	0.793	0.933
Prescribed Dose	100 cGy	
MU	120	

*7-18. The front of a patient's brain was treated to 200 cGy/fraction using a plan prescribed to 100% isodose line. The SAD treatment technique is composed of two orthogonal AP and lateral wedged fields using the 6 MV photon beam from the SL/25.

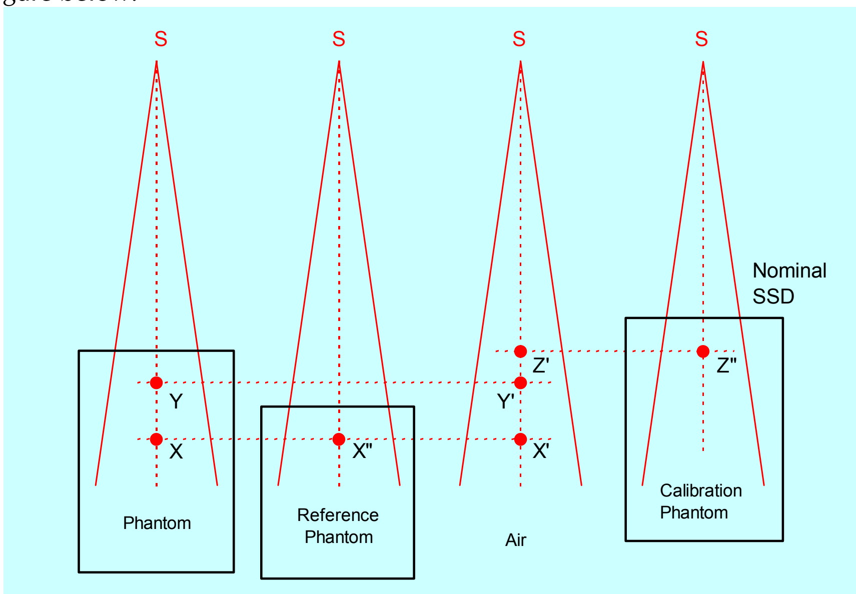
	AP field		Lateral field	
Parameters	Open	Wedged†	Open	Wedged†
Weight	0.29	0.21	0.29	0.21
SSD	96.6	96.6	95.7	95.7
Width	10	10	10	10
Length	12	12	12	12
MU	**232**	**45**	**230**	**45**

†Autowedge 60 degree with wedge factor or 0.27

After 11 fractions, the patient was found to have brain necrosis. Identify the problem associated with this treatment

*7-19. A patient was treated with two oblique fields (LPO, RPO) using the SAD technique to an average depth of 9.75 cm. The field size used was 8.5 cm x 7 cm. Now, the radiation oncologist decides to treat with a single AP field (SSD) to a depth of 8 cm. What is the maximum field size that can be used to match the field if the gap is 2.5 cm ?

*7-20. The schematics for the dose point relationships with extended SSD setup is shown in the figure below.



Show that the MU at extended fixed distance using TPR formalism is

$$MU = \frac{D}{TPR(d, A_{SSD+d})} \times \frac{1}{S_{cp}} \times \frac{1}{WFxTF} \times \frac{PSF(d_m, A_{SAD+d_m})}{PSF(d_m, A_{SSD+d})} \times \left(\frac{SSD + d}{SAD + d_m}\right)^2$$

where D is the prescribed dose, S_{cp} is the collimator factor and phantom scatter factor, WF is the wedge factor, TF is the tray factor, SSD is the source-to-surface distance and SAD is the source-to-surface distance in the calibration condition.

*7-21. In the meterset calculation, show that the given dose (GD) at extended fixed distance using TPR formalism is

$$GD = \frac{D}{TPR(d, A_{SSD+d})} \times \frac{PSF(A_{SSD+dm})}{PSF(A_{SSD+d})} \times \left(\frac{SSD + d}{SSD + dm}\right)^2$$

where D is the prescribed dose, SSD is the source-to-surface distance and d_m is the depth of given dose in phantom, d is the treatment depth in phantom.

Multiple Choice Questions

Choose the correct answer

7-1. Which of the following statement is true of equivalent square of a rectangular field
I. Square having the same percent depth dose as a rectangular field
II. Square having the same area as the rectangle
III. Square having side equal to twice the area divided by perimeter of the rectangle
IV. Square field having the same scatter dose as the rectangular field.

a) I and II

b) I and IV
c) II and III
d) II and IV
e) III and IV

7-2. The percent error in dose rate caused by an inaccuracy of 2 cm of the patient at a source to skin distance of 100 cm is
a) 1
b) 2
c) 4
d) 8
e) 10

7-3. The radiation dose at a given point inside a patient _____ as the distance of this point from the source increases.
a) increases
b) decreases
c) decreases and then increases
d) increases and then decreases
e) remains the same

7-4. The ratio of dose at any point in a phantom to the dose received by a small mass at the same point suspended in air is called
a) depth dose
b) tissue-air ratio
c) tissue-phantom ratio
d) tissue-maximum ratio
e) backscatter factor

7-5. Tissue-air-ratio (TAR) is a useful concept in dosimetry. Regarding TAR:
I. it is expressed as a ratio of dose in air
II. it is independent of treatment distance
III. it increases with increasing energy
IV. it decreases with increasing field size
V. it is not related to percent depth dose.

a) I, II, and III
b) II, III, and V
c) III, IV, and V
d) I, IV and V
e) II, III, and IV

7-6. The percent depth dose of a photon beam increases with
I. increasing SSD
II. increasing field size
III. increasing beam energy
IV. decreasing SSD
V. decreasing beam collimation

a) I, II, and III
b) I, III, and V
c) II, III, and IV
d) I and II

e) IV and IV

7-7. Which is not true of TAR
a) it increases with field size
b) it is independent of beam energy
c) it is independent of SSD
d) it depends on depth
e) none of the above

7-8. The following statements are related to backscatter factor. Which of these are true?

I. Backscatter factor cannot be less than 1
II. Its maximum value is about 1.5 at about 1 m Cu HVL beam
III. Backscatter factor increases with field size
IV. For high energy photon beam, the backscatter factor is about one

a) I and II
b) I and III
c) II and III
d) I, II, and III
e) I, II, III, and IV

7-9. Patient A with colon cancer was treated to a depth of 5 cm using cobalt-60 beam. If the treatment time is 2.4 min, what is the tumor dose if the percent depth dose is 83.5 %. Assume the backscatter factor is 1.037, f-factor is 0.957 and exposure rate is 95 R/min
a) 236 rads
b) 245 rads
c) 189 rads
d) 158 rads
e) 95 rads

7-10. The exposure dose rate at a defined point in a beam with the absence of a scattering medium is expected to be _____ the exposure dose rate in a scattering medium
a) less than for all energy photon beam
b) the same for all energy photon beam
c) greater than for all energy photon beam
d) greater than for high energy photon beam
e) the same for high energy photon beam